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Research article

Biomechanical analysis of lower limb movements of badminton players before and after intense exercise

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Abstract: *Objective:* At present, there are few studies that systematically investigate the coordinated changes in the kinematics, dynamics, and overall stability control of multiple joints in the lower limbs under the fatigue program of badminton. This paper aimed to explore the effects of intense exercise on the left and right lunges of badminton players. *Methods:* Twenty badminton players from Shandong University of Political Science and Law completed three left lunge movements and three right lunge movements before and after intense exercise, with the right lower limb as the unified support under all conditions. Kinematic and dynamic data were collected simultaneously for two-factor repeated-measures analysis of variance (intense exercise × lunge direction) in SPSS 23.0. *Results:* The main effect of the lunge direction and the interaction effect of movement × direction did not reach statistical significance ($p > 0.05$) for all joint angles and torque indicators, indicating that the fatigue effect of intense exercise is consistent under left and right lunge conditions. Regarding the main effect of movement, the hip flexion angle significantly decreased ($p = 0.024$), and the hip adduction and external rotation both increased ($p = 0.008$, $p = 0.021$); the knee internal rotation increased ($p = 0.017$); the ankle eversion increased ($p = 0.034$); the hip extension torque ($p = 0.018$) and adduction torque ($p = 0.015$), the knee flexion torque ($p = 0.016$) and adduction torque ($p = 0.012$), and the ankle inversion torque ($p = 0.021$) all significantly decreased. The anterior–posterior velocity of the body's center of mass ($p = 0.005$), the vertical displacement range ($p = 0.014$), and the margin of stability ($p = 0.008$) were all significantly reduced. *Conclusion:* After intense exercise, the angles and torques of the lower limb joints of the subjects change. This may be associated with an increased risk of lower limb injuries under fatigue conditions, and it can provide a reference for the design of specialized gait training and injury prevention programs.

Keywords: badminton player; biomechanics; dynamics; gait movement; intense exercise; kinematics; sports injury

1. Introduction

Badminton is a non-contact sport; however, due to its high intensity [1] and asymmetry [2], the risk of sports injury is also very high. During competition, players need to perform numerous continuous changes of direction and jumps [3], which can induce exercise-induced fatigue, thereby affecting the completion of movements and body balance [4]. Fatigue caused by intense exercise has been extensively studied in the field of sports [5]. Badminton players use a lot of footwork during a game [6], among which the left/right lunge is the most common type. Continuous and steady lunges can provide a good position for players to hit the ball; at the same time, they also place a lot of pressure on the lower limb joints. Its landing cushioning, center of gravity shifting, and joint load characteristics are closely related to the load distribution and movement safety of the lower limbs. Therefore, exploring the biomechanical response patterns of the lower limbs during badminton's lunge movements under fatigue conditions has significant practical significance for injury prevention in the specific sport and scientific training.

There are relatively few available systematic studies on the biomechanics of badminton lunge. Lam et al. [7] pointed out in a systematic review on the badminton lunge that elite badminton players exhibited superior impact attenuation capacity, adopted more aggressive knee and ankle strategies (greater mechanical torque), and bore higher medial plantar load than amateur players. Valldecabres et al. [8] took 16 skilled right-handed badminton players as subjects and found that the knees exhibited significantly greater flexion force during the forehand lunge and greater internal rotation speed and extension torque during the backhand lunge. Huang et al. [9] explored the influence of direction and gender on the biomechanics of the lower limbs of badminton players and found that during the backhand lunge, the sagittal range of motion of the hip and knee joints and approximately 84% of the knee flexion torque during the standing stage were significantly higher than those during the forehand lunge.

Regarding the impact of fatigue on the biomechanics of lunges, Xie et al. [10] investigated the effects of gender, direction, and fatigue on the biomechanics of the lower limbs during lunges and pointed out that under fatigue conditions, the knee flexion angle of the participants decreased, and the stiffness of the legs increased, which might increase the load on the lower limbs. Du et al. [11] studied the effects of fatigue on postural control and the biomechanics of lunges and found that after fatigue, the peak torques and power of the knee and hip joints during the push-off phase were significantly reduced.

The shortcomings of the existing studies lie in the fact that there are many studies on forehand and backhand lunges, but there are relatively fewer studies on the left-front and right-front lunges completed by the same lower limb. It is impossible to rule out the interference caused by differences in the anatomical structure, muscle strength, and baseline of movement control of both lower limbs. Therefore, it is difficult to accurately reveal the independent effect of the lunge direction on the load on the lower limbs. Moreover, there is a lack of exploration of whether these two lunge patterns exhibit different biomechanical response modes after intense exercise. Therefore, this paper conducted a biomechanical analysis of left and right lunge movements before and after intense exercise. A uniform

experimental design was adopted, with the right lower limb serving as the sole supporting limb. The purpose was to abandon the traditional bilateral limb comparison model, avoid individual baseline differences in muscle strength, joint range of motion, and movement habits of the left and right lower limbs, and focus on the effect of intense exercise fatigue, achieving an interactive analysis of movement state × lunge direction. Compared with previous studies on forehand and backhand lunge movements and general fatigue lunge studies, this research controlled the supporting limb variable and focused on the mechanical laws of specialized lunge movements after intense exercise. The research perspective is more refined, and the specialized adaptability is also stronger. The purpose of this research is to analyze the possible relationship between mechanical parameters and lower extremity joint injuries, thereby providing some objective biomechanical basis for optimizing the specialized footwork of badminton players, correcting movements under fatigue conditions, and precisely preventing lower limb injuries. It also provides support for the scientific design of physical fitness and technical training in badminton.

2. Research subjects and methods

2.1. Subjects

The subjects were 20 male, right-handed badminton players from Shandong University of Political Science and Law, meeting the following criteria: no history of anterior cruciate ligament (ACL) injury, no lower limb injury within three months, and no intense exercise within 24 hours. All of them were informed of the experimental purpose and procedures and signed informed consent. These players had an average age of 20.12 ± 1.68 years, an average height of 178.23 ± 5.34 cm, an average weight of 74.56 ± 6.17kg, and an average training time of 4.33 ± 1.62 years.

The sample size of the study was pre-calculated using the G*Power 3.1.9.6 software. The experimental design was a fully within-subjects repeated-measures design with 2 exercise states (before intense exercise, after intense exercise) × 2 lunge directions (left front lunge, right front lunge). There was no inter-group grouping variable. The effect size was f = 0.35 (referring to the medium or large effect size standard used in similar studies on the biomechanics of badminton). The significance level, statistical power, and repeated-measures level were set to 0.05, $1-\beta=0.80$, and 2, respectively. The intra-group correlation coefficient followed the default value of 0.5 for similar studies on the biomechanics of badminton. Therefore, the minimum sample size required for the study was calculated to be 17 people. Considering factors such as sample dropout, the final sample size was determined to be 20 people. The sample size met the statistical test requirements.

2.2. Research methods

2.2.1. Experimental procedure

The flowchart of the research method is presented in Figure 1.

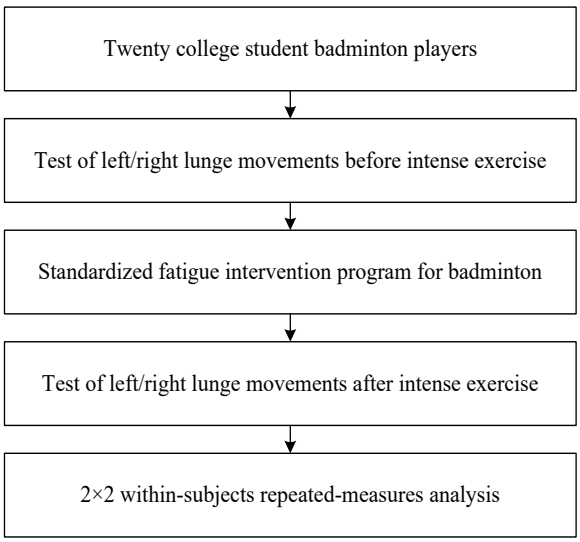


Figure 1. Experimental flowchart.

The test movement was the left and right lunges in badminton footwork. Each player held the racket with his right hand and stood in the center of the court, maintaining the ready position to receive the ball. Upon hearing the start command, he performed the following movements in sequence: a start step → a lunge to the left front to receive the ball and then push off the ground to return to the starting point → a start step → a lunge to the right front to receive the ball and push off the ground to return to the starting point.

Before the experiment, the players were informed of the content and procedures of the experiment and were then directed to a 5–10-min warm-up session. After these preparations, they were uniformly dressed and had reflective marker points affixed at the following positions by the same experimenter: left/right anterior superior iliac spine; left/right posterior superior iliac spine; medial condyle of the left/right femur; lateral condyle of the left/right femur; left/right lateral thigh; left/right lateral leg; lateral malleolus of the left/right fibula; medial malleolus of the left/right tibia; left/right heel; and left/right toe tip.

Kinematic data were collected using an eight-lens infrared motion capture system (100 Hz, Qualisys, Sweden), and dynamic data were collected using a three-dimensional force platform (1,000 Hz, Bertec, USA). The force platform was positioned in the left and right frontcourts of the badminton court. Each subject completed the collection of three complete movements before and after intense exercise.

The study employed a fatigue intervention program that closely mimics the high-intensity, repetitive movement characteristics of playing badminton. It differs from the general aerobic exercise fatigue model and is fully aligned with the specific power generation modes of specialized footwork. It ensures the specific adaptability and homogeneity of fatigue stimulation. The standardized fatigue intervention program is as follows: A single set consisted of 15 standardized movement cycles (left backcourt push-off jump → right frontcourt lunge to touch the ball→ return to position for preparation). The movement rhythm was controlled by a metronome, with a frequency of 120 beats per minute. After each set, a fixed 15-s break was taken between sets. All subjects completed three complete sets uniformly, with the total exercise duration and the total number of movements being consistent. There were no premature termination or excessive exercise situations to ensure that the fatigue stimulation of all subjects was the same.

The standard for collecting three valid movements is as follows: Based on the experimental video, invalid movements that included movement errors, abnormal gait, fluctuations in equipment signals, and significant deviation in the center of gravity were excluded. For all kinematic and dynamic parameters of valid movements, the average of the three test results was taken as the final original statistical data to reduce the random error of a single test and improve the stability and representativeness of the data.

The fatigue state in intense exercise was determined as follows: 1) Heart rate (HR) reached maximum HR (220 - age); 2) the subjective fatigue score rating of perceived exertion (RPE) was ≥18 points; 3) blood lactate concentration was ≥8 mmol/L; 4) it was difficult to complete the movements following the metronome, and obvious compensatory deformation occurred. If any of these four criteria were met, the subject was determined to have reached the fatigue state.

The HR was monitored in real time throughout the process using the polar H10 HR band. The instantaneous HR was recorded every 10 s in each set. The RPE score was obtained through subjective assessment within 30 s after the end of each set of interventions. Blood lactate was collected from a fingertip blood sample after the completion of all fatigue interventions and within 1 min of static rest. The measurement results are shown in Table 1.

To avoid the fatigue attenuation effect, after the collection of the indicators, a 2-min interval was set, and all subjects uniformly conducted a lunge movement capture test to maximize the retention of the biomechanical characteristics of exercise fatigue and to ensure the accuracy and comparability of the experimental data. For the two lunge directions (left front and right front), the same-side supporting leg (right leg) was employed to eliminate the confounding factors caused by differences in the anatomical structure and strength of the left and right legs.

Table 1. Status of players before and after intense exercise.

	Before intense exercise	After intense exercise	p value
HR/beat	77.84 ± 9.62	187.03 ± 6.71	<0.001
RPE/min	5.16 ± 0.04	18.62 ± 1.27	<0.001
Blood lactate (mmol/L)	3.33 ± 1.37	13.16 ± 3.34	<0.001

2.2.2. Mathematical statistics method

With reference to existing research on ACL injury in badminton, the moment of landing was defined as the moment when the ground reaction force first continuously exceeded 50 N [12]. The moment of leaving the ground was defined as the moment when the ground reaction force dropped to below 50 N. Kinematic data were processed using Visual 3D v6.1 software. Three-dimensional coordinates were smoothed using a Butterworth low-pass filter at a cutoff frequency of 12 Hz. Dynamic data were calculated using the inverse dynamics method [13]. After being smoothed by the Butterworth low-pass filter, the cutoff frequency was 50 Hz. Before the formal test, all subjects stood on the force platform in a natural standing posture, with their arms hanging down. They maintained a neutral and static standing position for 3 s to complete the full-body static calibration and establish the human coordinate system and lower limb segment model. The lower limb segments were established based on the following marker points using the 6-degree-of-freedom lower limb model built into Visual 3D: Pelvic segment: left and right anterior superior iliac spines, left and right posterior superior iliac spines; thigh segment: medial and lateral femoral condyles and lateral thigh; calf segment: lateral malleolus of the fibula, lateral malleolus of the tibia, and lateral calf; foot segment: heel and toe.

The three-dimensional coordinate systems of each segment were defined using the right-hand rule, in accordance with the standard for lower limb joint coordinate systems recommended by the International Society for Biomechanics. The inertial parameters of the body segments were taken from the default human inertial parameter library built into Visual 3D. All peak torques of joints and joint angle parameters were extracted from the complete support phase period (from the moment of landing to the moment of leaving the ground). The reported joint torques were the net internal joint torques, which can directly reflect the active force exerted by the lower limb muscle groups and the joint load level.

To ensure the consistency of statements throughout the whole paper, the positive and negative directions of joint angles of the lower extremity were defined uniformly. All joint angle descriptions in this paper followed the following conventions: On the sagittal plane, the flexion of the hip joint, knee joint, and ankle joint was defined as positive, and extension was negative; on the coronal plane, the joint adduction was defined as positive, and abduction was negative; on the horizontal plane, the internal rotation of the joint was defined as positive, and external rotation was negative; as to the frontal axis of the ankle joint, eversion was defined as positive, and inversion was negative.

The center of mass (COM) coordinates of the human body was calculated by the Visual 3D software based on the weighted calculation of the human body segment inertia parameter library. The margin of stability (MoS) was calculated using the static instantaneous stability margin and reflected the safety margin of the COM position relative to the boundary of the support surface at the landing moment. The calculation formula is: MoS = horizontal coordinate of the front edge of the support surface - horizontal coordinate of the body COM [14].

All results were recorded as the mean ± standard deviation. Independent variables (exercise state: before and after intense exercise; lunge direction: left and right) were within-subject repeated-measures variables. There was no between-groups grouping or mixed design.

During data processing, the three-times interquartile range method was employed to screen out outliers. Data exceeding the range of [Q1 − 3IQR, Q3 + 3IQR] were marked and reviewed. Combined with the experimental video, invalid abnormal data caused by operational errors or equipment fluctuations were eliminated. Then, the Shapiro–Wilk test checked the normality of all data. The result ($p > 0.05$) indicated that all data met the normal distribution and the data distribution requirements of repeated-measures variance analysis. Subsequently, the Levene's test was used to check the homogeneity of variance, and the result ($p > 0.05$) indicated that all assumptions of homogeneity of variance were met.

The statistical model employed a single-level within-subject repeated-measures analysis of variance. The significance threshold was set at $p < 0.05$.

3. Results and analysis

3.1. Lower extremity kinematic characteristics

The impact of intense exercise on hip joint angle at the moment of landing of the right lower limb is shown in Tables 2 and 3.

Table 2. Changes in hip joint angles (unit: °).

	Joint motion	Before intense exercise	After intense exercise
Left lunge	Flexion	53.74 ± 7.21	49.13 ± 6.56
Right lunge	Flexion	51.92 ± 7.12	49.22 ± 7.84
Left lunge	Abduction/adduction	−9.45 ± 24.16	−4.91 ± 17.62
Right lunge	Abduction/adduction	−22.75 ± 13.86	−16.13 ± 11.21
Left lunge	External/internal rotation	1.16 ± 14.21	−4.76 ± 19.33
Right lunge	External/internal rotation	−9.33 ± 10.54	−13.36 ± 9.92

Table 3. Analysis of the interaction and main effects of exercise state and lunge direction on hip joint angles.

		df	F	p	Partial η^2 (η^2p)
Flexion	Exercise state	(1, 19)	6.02	0.024	0.241
	Lunge direction	(1, 19)	0.11	0.742	0.006
	Exercise × direction	(1, 19)	0.33	0.572	0.017
Abduction/adduction	Exercise state	(1, 19)	8.76	0.008	0.316
	Lunge direction	(1, 19)	0.67	0.423	0.034
	Exercise × direction	(1, 19)	0.23	0.636	0.012
External/internal rotation	Exercise state	(1, 19)	6.31	0.021	0.249
	Lunge direction	(1, 19)	0.52	0.480	0.027
	Exercise × direction	(1, 19)	0.55	0.468	0.028

According to Tables 2 and 3, the hip flexion angles significantly decreased [$F(1,19) = 6.02$, $p = 0.024$]; the hip adduction significantly increased [$F(1,19) = 8.76$, $p = 0.008$]; the hip external rotation also significantly increased [$F(1,19) = 6.31$, $p = 0.021$]. The main effects of all joint directions were $p > 0.05$. The interaction effects of exercise × direction were not significant: all interaction effects were $p > 0.05$, indicating no significant differences in the exercise effect between left and right lunges, i.e., the influence of the exercise on both sides was consistent in direction.

The effect of intense exercise on knee angle at the moment of landing of the right lower limb is shown in Tables 4 and 5.

Table 4. Changes in knee joint angles.

	Joint motion	Before intense exercise	After intense exercise
Left	Extension/flexion	−15.87 ± 6.33	−10.64 ± 4.92
Right	Extension/flexion	−15.87 ± 6.56	−13.61 ± 5.22
Left	Abduction/adduction	−2.73 ± 6.81	−3.72 ± 2.83
Right	Abduction/adduction	3.12 ± 6.44	−0.61 ± 4.92
Left	External/internal rotation	−8.44 ± 21.07	1.71 ± 14.97
Right	External/internal rotation	−15.24 ± 11.86	−6.12 ± 18.55

Table 5. Analysis of the interaction and main effects of exercise state and lunge direction on knee joint angles.

		df	F	p	Partial η^2 (η^2p)
Extension/flexion	Exercise state	(1, 19)	1.78	0.198	0.086
	Lunge direction	(1, 19)	1.24	0.280	0.061
	Exercise × direction	(1, 19)	2.87	0.107	0.131
Abduction/adduction	Exercise state	(1, 19)	0.41	0.530	0.021
	Lunge direction	(1, 19)	1.88	0.186	0.090
	Exercise × direction	(1, 19)	3.11	0.094	0.141
External/internal	Exercise state	(1, 19)	6.79	0.017	0.263
rotation	Lunge direction	(1, 19)	1.95	0.179	0.093
	Exercise × direction	(1, 19)	0.05	0.825	0.003

According to Tables 4 and 5, the intense exercise showed no significant effects on the knee joint extension/flexion and abduction/adduction in terms of the main effect of the exercise, the main effect of the lunge direction, and the interaction effect ($p = 0.198, 0.530, 0.280, 0.186, 0.107, 0.094 > 0.05$). However, the knee internal rotation significantly increased, and the main effect of exercise was significant, with $F(1,19) = 6.79$, $p = 0.017$; the main effect of lunge direction was not significant, with $F(1,19) = 1.95$, $p = 0.179$; and the interaction effect was not significant, with $F(1,19) = 0.05$, $p = 0.825$.

The effect of intense exercise on the ankle angle at the moment of landing on the right lower limb is shown in Tables 6 and 7.

Table 6. Changes in ankle joint angles.

	Joint motion	Before intense exercise	After intense exercise
Left	Extension/flexion	15.21 ± 6.12	10.81 ± 10.81
Right	Extension/flexion	15.44 ± 7.03	11.56 ± 9.12
Left	Abduction/Adduction	1.52 ± 6.53	−1.02 ± 3.84
Right	Abduction/Adduction	−4.33 ± 6.16	0.33 ± 6.62
Left	Eversion/inversion	0.91 ± 4.11	2.62 ± 7.94
Right	Eversion/inversion	−3.52 ± 5.33	−0.83 ± 5.21

Table 7. Analysis of the interaction and main effects of exercise state and lunge direction on ankle joint angles.

		df	F	p	Partial η^2 (η^2p)
Extension/flexion	Exercise state	(1, 19)	1.87	0.188	0.090
	Lunge direction	(1, 19)	0.05	0.833	0.003
	Exercise × direction	(1, 19)	0.01	0.940	0.001
Abduction/Adduction	Exercise state	(1, 19)	0.32	0.576	0.017
	Lunge direction	(1, 19)	0.86	0.365	0.043
	Exercise × direction	(1, 19)	3.64	0.072	0.161
Eversion/inversion	Exercise state	(1, 19)	5.23	0.034	0.216
	Lunge direction	(1, 19)	2.18	0.156	0.103
	Exercise × direction	(1, 19)	0.72	0.406	0.037

According to Tables 6 and 7, the ankle eversion significantly increased, $F(1,19) = 5.23$, $p = 0.034$. The others were insignificant. After intense exercise, the ankle eversion angle increased during landing. This angle change may be accompanied by an increase in the passive tension of the lateral ligaments of the ankle joint and may be associated with a risk of overload of the lateral soft tissues.

3.2. Lower extremity dynamic characteristics

The effects of intense exercise on peak hip joint torque are presented in Tables 8 and 9.

Table 8. Changes in peak hip joint torque.

	Joint motion	Before intense exercise	After intense exercise
Left	Extension torque	2.24 ± 0.63	2.12 ± 0.25
Right	Extension torque	2.33 ± 0.41	1.83 ± 0.67
Left	Adduction torque	2.45 ± 0.43	2.21 ± 0.56
Right	Adduction torque	2.55 ± 0.43	2.16 ± 0.57

Table 9. Analysis of the interaction and main effects of exercise state and lunge direction on peak hip joint torque [unit: $(N \cdot m) \cdot kg^{-1}$].

		df	F	p	Partial η^2 (η^2p)
Extension torque	Exercise state	(1, 19)	6.71	0.018	0.261
	Lunge direction	(1, 19)	0.56	0.463	0.029
	Exercise × direction	(1, 19)	2.41	0.137	0.113
Adduction torque	Exercise state	(1, 19)	7.15	0.015	0.273
	Lunge direction	(1, 19)	0.69	0.416	0.035
	Exercise × direction	(1, 19)	0.55	0.468	0.028

According to Tables 8 and 9, the main effect of exercise on peak extension torque was significant, with $F(1,19) = 6.71$, $p = 0.018$; the main effect of lunge direction was not significant, with

$F(1,19) = 0.56$, $p = 0.463$; and the interaction effect was insignificant, with $F(1,19) = 2.41$, $p = 0.137$. The main effect of exercise on peak adduction torque was also significant, with $F(1,19) = 7.15$, $p = 0.015$; the main effect of lunge direction was insignificant, with $F(1,19) = 0.69$, $p = 0.416$; and the interaction effect was insignificant, with $F(1,19) = 0.55$, $p = 0.468$. This result suggests that the decrease in the net joint torque of the hip joint after intense exercise may be related to the reduced contribution of the extensor and adductor muscle groups or changes in the joint control strategy. The mechanical load environment of the soft tissues around the hip joint may also change accordingly.

The effects of intense exercise on peak knee torque are presented in Tables 10 and 11.

Table 10. Changes in peak knee torque.

	Joint motion	Before intense exercise	After intense exercise
Left	Flexion torque	1.87 ± 3.42	1.82 ± 3.26
Right	Flexion torque	2.02 ± 3.53	1.92 ± 3.56
Left	Adduction torque	0.33 ± 2.05	0.31 ± 1.97
Right	Adduction torque	0.33 ± 2.04	0.31 ± 1.98
Left	Internal rotation torque	0.16 ± 0.55	0.16 ± 0.86
Right	Internal rotation torque	0.17 ± 0.97	0.18 ± 0.84

Table 11. Analysis of the interaction and main effects of exercise state and lunge direction on peak knee joint torque [unit: $(N \cdot m) \cdot kg^{-1}$].

		df	F	p	Partial η^2 (η^2p)
Flexion torque	Exercise state	(1, 19)	6.95	0.016	0.268
	Lunge direction	(1, 19)	0.16	0.694	0.008
	Exercise × direction	(1, 19)	0.01	0.939	0.001
Adduction torque	Exercise state	(1, 19)	7.69	0.012	0.288
	Lunge direction	(1, 19)	0.00	0.984	<0.001
	Exercise × direction	(1, 19)	0.02	0.894	0.001
Internal rotation torque	Exercise state	(1, 19)	0.09	0.772	0.005
	Lunge direction	(1, 19)	0.02	0.884	0.001
	Exercise × direction	(1, 19)	0.04	0.849	0.002

According to Tables 10 and 11, for the peak flexion torque of the knee joint, the main effect of exercise was significant, $F(1,19) = 6.95$, $p = 0.016$; the main effect of lunge direction was not significant, $F(1,19) = 0.16$, $p = 0.694$; and the interaction effect was insignificant, $F(1,19) = 0.01$, $p = 0.939$. For the peak adduction torque, the main effect of exercise was significant, $F(1,19) = 7.69$, $p = 0.012$; the main effect of lunge direction was insignificant, $F(1,19) = 0.00$, $p = 0.984$; and the interaction effect was not significant, $F(1,19) = 0.02$, $p = 0.894$. For the peak internal rotation torque, the main effect of exercise was insignificant, $F(1,19) = 0.09$, $p = 0.772$; the main effect of lunge direction was not significant, $F(1,19) = 0.02$, $p = 0.884$; the interaction effect was insignificant, $F(1,19) = 0.04$, $p = 0.849$.

The effects of intense exercise on peak ankle torque are shown in Tables 12 and 13.

Table 12. Changes in ankle joint torque.

	Joint motion	Before intense exercise	After intense exercise
Left	Flexion torque	1.37 ± 0.52	1.33 ± 0.45
Right	Flexion torque	1.45 ± 0.44	1.45 ± 0.46
Left	Adduction torque	0.19 ± 0.08	0.18 ± 0.09
Right	Adduction torque	0.21 ± 0.13	0.21 ± 0.12
Left	Inversion torque	0.09 ± 0.53	0.06 ± 0.45
Right	Inversion torque	0.12 ± 0.45	0.07 ± 0.42

Table 13. Analysis of the interaction and main effects of exercise state and lunge direction on peak ankle joint torque [unit: $(N \cdot m) \cdot kg^{-1}$].

		df	F	p	Partial η^2 (η^2p)
Flexion torque	Exercise state	(1, 19)	0.52	0.478	0.027
	Lunge direction	(1, 19)	1.26	0.276	0.062
	Exercise × direction	(1, 19)	0.82	0.377	0.041
Adduction torque	Exercise state	(1, 19)	0.27	0.611	0.014
	Lunge direction	(1, 19)	1.01	0.328	0.050
	Exercise × direction	(1, 19)	0.10	0.757	0.005
Inversion torque	Exercise state	(1, 19)	6.33	0.021	0.250
	Lunge direction	(1, 19)	0.44	0.517	0.023
	Exercise × direction	(1, 19)	0.07	0.790	0.004

According to Tables 12 and 13, the main effect of exercise on the peak inversion torque of the ankle joint was significant, with $F(1,19) = 6.33$, $p = 0.021$; all other items were not significant. This result may suggest that the passive tension of the lateral ligament on the ankle joint increases at the initial stage of landing, and the torque contribution of the inversion muscle group changes, which may be associated with a risk of lateral ankle injury.

The COM movement and stability parameters at the landing moment before and after intense exercise are presented in Tables 14 and 15.

Table 14. Changes in the COM motion and stability parameters.

	Joint motion	Before intense exercise	After intense exercise
Left	Anterior–posterior velocity of COM ($m \cdot s^{-1}$)	1.93 ± 0.31	1.72 ± 0.28
Right	Anterior–posterior velocity of COM ($m \cdot s^{-1}$)	2.01 ± 0.34	1.78 ± 0.31
Left	Vertical displacement range of COM (cm)	16.83 ± 4.52	14.26 ± 4.11
Right	Vertical displacement range of COM (cm)	17.41 ± 4.83	14.78 ± 4.37
Left	MoS (anterior–posterior direction, cm)	8.22 ± 3.11	6.45 ± 2.87
Right	MoS (anterior–posterior direction, cm)	7.83 ± 3.24	5.92 ± 2.91

Table 15. Analysis of the interaction and main effects of exercise state and lunge direction on COM motion and stability parameters.

		df	F	p	Partial η^2 (η^2p)
Anterior–posterior velocity of COM	Exercise state	(1, 19)	9.82	0.005	0.341
	Lunge direction	(1, 19)	1.64	0.216	0.079
	Exercise × direction	(1, 19)	0.06	0.812	0.003
Vertical displacement range of COM	Exercise state	(1, 19)	7.24	0.014	0.276
	Lunge direction	(1, 19)	0.39	0.541	0.020
	Exercise × direction	(1, 19)	0.04	0.838	0.002
MoS	Exercise state	(1, 19)	8.76	0.008	0.316
	Lunge direction	(1, 19)	0.69	0.417	0.035
	Exercise × direction	(1, 19)	0.21	0.650	0.011

According to Tables 14 and 15, after intense exercise, the speed before and after intense exercise decreased significantly [$F(1,19) = 9.82$, $p = 0.005$], indicating that the horizontal momentum of the player during the lunge landing decreased. At the same time, the vertical displacement range of the COM significantly decreased [$F(1,19) = 7.24$, $p = 0.014$], suggesting that the players adopted a lower COM strategy after fatigue. MoS was significantly reduced in both lunge conditions [$F(1,19) = 8.76$, $p = 0.008$], indicating that the ability to resist disturbances decreased after fatigue.

4. Discussion

Previous research has mostly focused on biomechanical changes after running fatigue. The main innovation of this study is that it adopted a fatigue scheme more specific to badminton, closer to the fatigue patterns and movement requirements in actual competition scenarios. The research results showed that the hip flexion angle significantly decreased after intense exercise ($p = 0.024$), indicating that the hip joint tended to be in an extended position when landing; the hip adduction significantly increased ($p = 0.008$, decreased extension degree), and the external rotation significantly increased ($p = 0.021$). The knee internal rotation significantly increased after exercise ($p = 0.017$). Although showing an extension trend in description, it did not reach statistical significance ($p = 0.198$). The ankle eversion significantly increased ($p = 0.034$). Previous studies have shown that after fatigue, the maximum extension angle of the hip joint and the range of hip joint flexion and extension activities can significantly increase, which is consistent with the finding that the hip joint tended to be in an extended position in this study. Bayabayi et al. [15] pointed out in their research that the knees and hips are adversely affected by fatigue, while the ankle compensates for the instability caused by fatigue. In addition, differences in the subject group (exercise level, gender, running experience, etc.) may also affect the fatigue response of joint kinematics [16].

The main effect of the lunge direction (left/right) and the interaction effect between exercise and direction did not reach statistical significance ($p > 0.05$) for all joint kinematic indicators. This suggests that the impact of intense exercise on the angles of lower limb joints is consistent across different lunge

directions, and the fatigue effect does not show a clear preference for the left or right side. In other words, under the unified support leg design, the fatigue effect was consistent in both left and right lunge movements (the interaction effects between movement and direction were not significant), and the direction of lunge did not have any regulatory effect on the fatigue response.

In terms of dynamics, the peak extension and adduction torques of the hip joint decreased significantly; the peak flexion and adduction torques of the knee joint also decreased significantly, while the internal rotation torque did not change significantly. The peak inversion torque of the ankle joint decreased significantly, while the flexion and adduction torques did not change significantly. The decrease in joint torques may be related to the change in the co-activation pattern of synergistic muscles. The body may adjust its muscle force application strategy to form a new balance between torque output and joint stability [17].

The changes in the overall control indicators suggest that the body's ability to maintain stable posture may experience a systematic decline after intense exercise. Jiang et al. [18] also pointed out that the reduction in knee flexion angle and the decrease in load after fatigue might be due to compensation by the hip and ankle joints, which disrupts the overall stability control of the COM.

5. Research limitations

This study has the following limitations:

(1) Sample size limitation: Only 20 players were included, without considering the influence of exercise level and gender distribution.

(2) Lack of electromyography data: No muscle synergy analysis was conducted to explore the adaptive changes in the neuromuscular system after intense exercise.

(3) Inference of injury risk: Although a series of biomechanical changes related to known injury risk factors were found in the results, they cannot be directly equated to the actual injury occurrence rate. Longitudinal follow-up studies are required.

6. Conclusion

This paper compared the biomechanical characteristics of left and right lunges before and after intense exercise among 20 badminton players. It was found that the main effect of the lunge direction and the interaction effect of movement × direction did not reach statistical significance in all indicators, indicating that the fatigue effect of intense exercise was consistent across left and right lunge movements. After intense exercise, the hip flexion angle significantly decreased ($p = 0.024$), the hip adduction increased ($p = 0.008$), and the hip external rotation increased ($p = 0.021$); the knee internal rotation increased ($p = 0.017$); and the ankle eversion increased ($p = 0.034$). The peak torques of multiple joints decreased significantly, and the anterior–posterior velocity of the COM, the vertical displacement range, and MoS all significantly decreased ($p = 0.005$, 0.014, 0.008). The innovation of this study lies in systematically quantifying the fatigue response characteristics of the kinematics, dynamics, and execution control of the three joints of the lower limbs during lateral lunge, providing experimental evidence for inferring the lower limb neuromuscular control strategy of exercise-induced fatigue. Further studies can combine electromyography measurements with longitudinal injury tracking to further test the possible association between fatigue-related biomechanical changes and the risk of lower limb injuries.

Use of AI tools declaration

The author declares that he has not used Artificial Intelligence (AI) tools in the creation of this article.

Conflict of Interest

The author declares no conflict of interest.

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