The Effects of 5 km Interval Running on the Anterior Cruciate Ligament Strain and Biomechanical Characteristic of the Knee Joint: Simulation and Principal Component Analysis

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Abstract: Interval running methodologies simulate competition and training conditions, with the aim of enhancing an athletes’ ability to cope with constant deceleration, acceleration, and sudden changes in direction, as associated athletic and performance challenges. Fifteen male athletes were recruited in this study, in which the anterior cruciate ligament was modeled as a nonlinear elastic passive soft tissue in OpenSim 4.2. Participants completed 5 km interval running training on a treadmill. Before and after the interval running, kinematics, kinetics, and electromyography activity of the lower leg during the cutting maneuvers were collected simultaneously. After running training, the anterior cruciate ligament strain demonstrated a decreasing trend when performing unexpected cutting maneuvers. Principal component analysis showed significant differences in knee moments during abduction-adduction; knee angles in flexion-extension, external-internal rotation, and abduction-adduction, as well as knee contact forces in the sagittal and coronal planes. The findings of the study highlight that athletes generate greater adduction moment at the onset of the cut, followed by greater abduction moment towards the end of the cut, which may have a substantial impact on the anterior cruciate ligament loading. Furthermore, athletes need to be mindful of changes in coronal plane contact forces.

Keywords: interval running; knee model; anterior cruciate ligament strain; cutting maneuver

1. Introduction

Soccer [1], basketball [2], and rugby [3] players are required to execute repetitive bouts of low and high-intensity physical activities. Furthermore, to achieve their competitive goals and meet the tactical requirements of their respective sports, players must perform specialized actions [4]. Typically, players are often required to cover a cumulative distance of 3 km, or even 5 km or more, involving constant changes of directions [2,5,6]. This rapidly changing direction of movements requires players to be well-coordinated and well-conditioned [7]. Interval running (IR) methods are a good training method to mimic game-specific loads during training, thereby developing the players’ ability to withstand fatigue and improving the skills needed for rapid changes in direction. The low and high-intensities, with high-frequency switching exercises, increase the likelihood of non-contact sports injuries [8–10]. Many researchers have focused on injuries of the knee joint in athletes, specifically the anterior cruciate ligament (ACL) [11,12], to study the mechanisms and biomechanical characteristics of overuse injuries. Renstrom et al.’s study revealed that the majority of ACL injuries in team sports were non-contact in nature [13].
Sugimoto et al. [14], supported this finding and reported an increased risk of ACL injury among fatigued athletes who changed their playing field position. A few conclusive studies found that alternating low- and high-intensity exercise lead to the ACL injury [15]. Young female athletes (professional and recreational) were subjected to 4–6 times higher risk of ACL injuries than their male counterparts when performing high-intensity activities such as jumping and side-cutting [16]. Although there are more studies on ACL injury in females than males [14], it is important to examine knee joint function in male athletes as well since the injury risk is likely to be not very different.

ACL injuries typically occur during high-impact tasks, specifically during decelerations, landings, and in continuous motions with changes of direction which are often called cutting maneuvers (CM) [17–19]. Previous studies have typically focused on alterations in knee abduction and internal rotation moment, as well as anterior tibial shear forces in athletes [12,20,21]. These parameters have been identified as load parameters that can potentially increase ACL strain [22]. In this study, experimental data combined with computer simulation have been utilized to evaluate ACL strains and forces, which can be a more intuitive and effective approach using appropriate mathematical and statistical methods to generate more scientifically rigorous and reliable assessments for athletes [23,24].

There is concern that an ACL injury is often accompanied by a meniscal tear [25], which can increase the risk of developing other musculoskeletal injuries. Therefore, it has been suggested that regular medical diagnosis and biomechanical observation of athletes are necessary measures for preventing non-contact ACL injuries in athletes [26,27]. Observation of ACL mechanics is difficult and direct measurements are only available via invasive methods, which are of limited use, because of the risk and ethical considerations these methods entail [28]. Because of the challenging nature of measuring ACL several studies use computer-generated models based on medical images and experimental data to simulate ACL mechanics during in vivo conditions [29,30]. Anne Schmitz and Davide Piovesan have created an open-source discrete element model that can be implemented and further developed within the OpenSim software using kinematic and kinetic experimental data [31,32]. This approach offers a valuable non-invasive research tool to investigate the effect of mechanical load on ACL stress and strain.

The main objective of this study was to determine if there was a significant increase in the risk of ACL injury during CM after performing a 5 km long continuous IR training. The second aim of this study was to explore which biomechanical characteristics of the lower limb can be a strong predictor for high risk of ACL injuries, using the outlier effect of principal component analysis (PCA). The hypothesis of this paper posits that subjects undergo notable variations in ACL strain when executing CM before and after IR training. Furthermore, PCA is employed to effectively differentiate the intrinsic distinctions in the biomechanical characteristics of the knee joint during CM, thereby offering a more accurate guidance for future investigations.

2. Materials and Methods

2.1. Participants

Fifteen male basketball and soccer players volunteered to participate in this study (Table 1). Each male player was identified as possessing a predominant right foot, a determination made based on a prior study [33]. Furthermore, these players were classified as amateur athletes in their respective sports. They were free from any lower extremity injuries or foot deformities in the past six months before the test. A medical expert inspected the participants’ knee ligaments to make sure that they all had a healthy ligament. Prior to the experiment, written consent was obtained from each participant, and they were thoroughly briefed on the experiment’s objectives, procedures, and requirements.
Table 1. Overview of Demographic Characteristics.

<table>
<thead>
<tr>
<th>Demographic</th>
<th>Results</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>24.70 ± 2.30</td>
</tr>
<tr>
<td>Height (m)</td>
<td>1.79 ± 3.12</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>77.52 ± 2.43</td>
</tr>
<tr>
<td>Training hours per week (h)</td>
<td>14.12 ± 4.30</td>
</tr>
<tr>
<td>Length of the ACL (mm)</td>
<td>32.44 ± 1.62</td>
</tr>
</tbody>
</table>

The Ethics Committee of the Ningbo University Research Institute approved this study (RAGH202209293312) which was performed in accordance with the Declaration of Helsinki. Furthermore, throughout the course of the experiment, the researcher diligently monitored and recorded the psychological changes exhibited by the subjects while adhering to the ethical guidelines set forth by the American Psychological Association.

2.2. Protocol and Instruments

2.2.1. Protocol

Figure 1 illustrates that the experiment comprises three primary components. First, participants were prepared for the measurements. They performed a standardized warm up including 5 min of running with a self-selected speed on a motorized treadmill and several stretching exercises. Before undertaking the CM test, each subject was required to perform three static standing tests. Initial standing position was on a force platform with their feet positioned shoulder-width apart, with their dominant right foot in place. Then the CM was performed to obtain the biomechanical characteristics of the lower limb during the task. At the corners of a six-meter sprint [34], participants exerted maximum effort to complete 90°, 135°, and 180° CM [35]. During the CM, participants utilized a forefoot strike running pattern with their dominant right foot [36]. Six meters behind the force plate and diagonally and on the left side, a marker was placed on the floor (Figure 1). A researcher (E.S.) experienced in coaching instructed the participants to perform sprint running to a given marker and back to the force plate then repeat a sprint toward a different marker in randomized order. Ten repetitions of sprints in each of the CM conditions were performed [37].

![Figure 1. (A) The laboratory collects CM data at angles of 90°, 135°, and 180°; (B) a flow chart was created to outline the data collection and processing.](image-url)
Then the participants performed a 5 km IR training on a treadmill. The level of intensity of the IR running was set based on the participants’ maximum sprinting speed during a 30 m sprint [38]. Thus, before the IR running, a 30-m sprint test was performed first [39]. IR running consisted of 2 or 3 maximum effort sprint runs on a motorized treadmill (h/p/cosmos sports and medical GmbH, Nussdorf-Traunstein, Germany). Each lasted approximately 10 s, with a five-minute rest period between each sprint, where participants were instructed to engage in a light jog [38]. During IR training, the timing of the sprint was determined by real-time monitoring of the participant’s heart rate using a Polar RS100 system (Polar Electro Oy, Woodbury, NY, USA). The total distance of jogging and sprint running combined during the IR training was 5 km for each participant.

After running training, the same procedures were repeated as before the running.

2.2.2. Kinetic, Kinematic and Electromyography Measurements

The marker trajectories of the reflective marker during CM movements were recorded with an eight-camera Vicon motion capture system (Vicon Metrics Ltd., Oxford, UK) at 1000 Hz sampling frequency. Ground reaction forces were measured with an embedded AMTI force platform (AMTI, Watertown, MA, USA) fixed in the middle of the path with a sampling frequency of 200 Hz. Following previous studies [40,41], 38 optical reflective markers (12.5 mm diameter) were carefully attached to the anatomical locations of each participant. Marker placement was consistently performed by the same operator. The cameras and lab setting were calibrated before each experimental session with a stable marker trajectory and minimal noise.

The duration of the CM trials was measured using single beam electronic timing gates (Brower Timing Systems, Draper, UT, USA) placed on a 1-m-high tripod at each marker location of the sprint test.

EMG of the rectus femoris (RECFEM), biceps femoris long head (BFLH), tibialis anterior (TIBANT), and gastrocnemius medialis (GASMED) were recorded at 1000 Hz sampling frequency using a Delsys EMG system (Delsys, Boston, MA, USA). EMG data was used to validate the OpenSim model [42].

2.3. ACL Modeling and Properties

As illustrated in Figure 2, this dataset includes markers located on both acromia, pelvis, bilateral thighs, bilateral knees, bilateral shanks, and multi-segment feet. This study employed an OpenSim full-body musculoskeletal model adapted from a recent study [40], we included updates on the mediolateral translations, adduction-abduction (AA) rotation, and internal-external (IE) rotation at both knees for this study. The hip joint was modelled as a ball-and-socket joint with three degree-of-freedoms (DOFs) in relation to the $x$-axis, $y$-axis, and $z$-axis. The ankle was modelled as a pin joint with dorsiflexion and plantarflexion in relation to the $z$-axis. Previous research has confirmed that ACL is fundamentally viscoelastic [43], with its tensile properties varying with strain rate or loading rate. In this study, the ACL was modeled as a non-linear, elastic, passive soft tissue [44], which was inserted into two fixation tunnels in the femur and tibia, as shown in Figure 2. The superior part of the ACL extends deep into the intercondylar space, while its inferior part attaches to the anterior tibial meniscus. The two primary fiber bundles of the ACL, namely the anteromedial (AM) and posterolateral (PL) bundles, were considered as a single entity in the model, as they were presumed to possess comparable characteristics and aligned orientation. The material properties of the ACL were established based on the criterion that the passive strains of the ACL during flexion-extension (FE), IE rotation, and AA rotation limits were below 15% [45]. The maximum strain was constrained to be less than 2.5–3.0% at the bone-ligament junction, which was the area of contact between the ligament and the bone. The contact area between the ligament and bone was considered, with the starting strain set at a very low value of 0.2% or less at isometric force. The ACL strain was found to be affected by the kinematic function of the knee, meaning that it was restricted by the prescribed limitations of the knee’s degrees of freedom. It should be noted
that the constant stiffness was only applicable in the region of lower strains (8–10%), with higher strains exhibiting increased stiffness according to the Gaussian law [46].

Figure 2. (A) The reflective markers were placed at specific locations to enable the CM; (B) The OpenSim platform was used to present the ACL as a model of nonlinear elastic passive soft tissue.

2.4. Data Processing and Statistical Analysis

The laboratory dataset underwent coordinate system conversion, low-pass filtering, data extraction, and formatting using MATLAB R2018b (The MathWorks, Natick, MA, USA) [42]. It was filtered using a 6-Hz and 30-Hz fourth-order, zero-phase lag Butterworth low-pass filter for marker trajectories and ground reaction forces [47]. The subsequent procedures were carried out in OpenSim 4.2 (Stanford University, Stanford, CA, USA) following a previously published workflow (Figure 1B) [42]. These steps encompassed anatomical scaling, inverse kinematics, residual reduction, computed muscle control, and forward dynamics.

Initially, the weight of the marked points in the model was manually fine-tuned. The model was subsequently scaled to align with the anthropometric characteristics of the participants, ensuring that the root mean square error between the marked points and the corresponding virtual marked points in the experiment remained below 0.02 m. The maximum error observed was below 0.04 m. Subsequently, the inverse kinematics algorithm was employed to compute the joint angles that yielded the minimum error between the marked points and the virtual marked points in the experiment. As per the official OpenSim guidelines, prior to implementing Computed Muscle Control (CMC) and Forward Dynamics (FD) analysis, the Residual Reduction Algorithm (RRA) was employed to mitigate the influence of modeling and marker data processing errors. The subtalar and metatarsophalangeal joints were locked during the data processing [40]. In the final stage, muscle activations computed from optimized excitations were used as input for FD, which computed the kinematics of all joints and advanced the motion by one step ahead in time. The FD simulation was executed in multiple steps to address significant perturbations that could cause diverging results.

2.4.1. Kinetic, Kinematic and EMG Measurements

Joint angles were calculated using inverse kinematics, then joint moments were calculated using inverse dynamics in OpenSim. The FE, IE, and AA rotation of the knee joint were exported for further statistical calculations. The ACL force and strain were obtained through the CMC and FD. The ACL fiber force computed by the CMC tool was normalized by dividing it by the subject’s body weight. The previously collected and calculated average ACL length of the subjects (Table 1) served as the resting length in the model, as well as the variation in ACL length derived from FD during CM. Utilizing the ACL length change
in relation to the resting length, ACL strain values were computed for each set of the CM. Additionally, the internal contact forces of the knee joint were determined through FD analysis [48]. The times series datasets were time normalized (0–101 data points) to the stance phase of each CM condition using a custom-written Python script [41].

2.4.2. Muscle Activation Analysis and Simulation Verification

Upon confirmation by a physician that all subjects are free of any lower extremity injury, the study proceeded with conducting maximum voluntary isometric contractions (MVICs). MVICs were performed separately on the RECFEM, BFLH, TIBANT, and GASMED muscles, following the protocols established in previous studies [49,50], with each segment of the muscle being tested three times. Following the MVICs testing, subjects underwent IR training and biomechanical data collection for the CM. This training and data acquisition took place before and after the training, specifically on the third day after completing the MVIC testing. Raw EMG data was filtered using a fourth-order Butterworth bandpass filter with a frequency range of 100–500 Hz. Root mean square values of the filter EMG signal were calculated using 50 ms window size. Each EMG dataset was normalized to the peak value measured during the maximum voluntary isometric contraction (MVIC) test (MVIC%) [50]. This is used to compare the results obtained from the CMC tool and validate the model [51]. We performed qualitative comparisons between the model-based predicted activations level and the processed experimental EMG data, accounting for appropriate physiological delays (30~40 ms) as per current recommendations [47]. This value ranges from 0 (completely inactive) to 1 (fully activated at 100%) and was used to quantify the level of muscle activation.

2.4.3. Principal Component Analysis

PCA can be generalized as correspondence analysis (CA) to handle qualitative variables and as multiple factor analysis (MFA) to handle heterogeneous sets of variables. Mathematically, PCA depends upon the eigen-decomposition of positive semi-definite matrices and the singular value decomposition (SVD) of rectangular matrices. In this study, the PCA was employed to reduce the dimensionality of the high-dimensional data and extract the principal components that capture the essential features of the variance [52]. Research studies have demonstrated that biomechanical attributes, which were linked to both knee kinematics and kinetic, may play a role in the risk of sustaining an ACL injury [14]. In this study, the PCA was used to identify the differences in biomechanical characteristics that result from variations in ACL strain values when performing the same CM direction. In particular, for the knee joint, 9 matrices were generated [53,54], encompassing 3 angles, 3 moments, and 3 contact forces. Within each of the matrices, the joint angle, moment, and contact force waveform data of the high ACL strain group was compared with those of the low ACL strain group in this study. This analysis was expected to reveal the particular biomechanical characteristics that are significantly affected when the ACL is subjected to high and low strains. In essence, our objective was to identify the primary biomechanical characteristics that exhibit a strong correlation with ACL strain through the utilization of PCA. The score plot of PCA employed two primary components to visually represent the joint angle, moment, and contact force datasets. The clustering criterion utilized for all PCAs in the score plot was a 95% confidence interval.

2.4.4. Statistical Analysis

The Shapiro–Wilk test (SPSS Inc., Chicago, IL, USA) was utilized to assess the normality of knee angles, moments, contact forces, and the ACL strain during the CM. Pearson’s correlation r was used to determine the magnitude and direction of the association between ACL force and strain. To compare the time series data of knee angles, moments, contact forces, and ACL strain before and after the 5 km running a one-dimensional statistical parametric mapping (SPM1d) based paired t-tests were performed. In the case of the non-normally distributed datasets, a non-parametric version of the SPM1d paired t-test
was used [55]. A Bonferroni correction was used to reduce the probability of the type I error because of the multiple comparisons. The statistical analyses were conducted in MATLAB R2018b (The MathWorks, Natick, MA, USA) and the alpha level was set to 5%.

3. Results

3.1. OpenSim Model Verification

As illustrated in Figure 3, a resemblance was observed between the muscle activation deduced using OpenSim’s CMC tool and the EMG data obtained during the present study.

![Figure 3](image_url)

**Figure 3.** The comparison of the experimentally measured (pink area) and the OpenSim-CMC (blue dashed line) simulated muscle activity for each muscle. The corresponding curves representing the mean values over the stance phase of CM at all three conditions (90°, 135°, 180°). Abbreviations: BFLH—biceps femoris long head, RECFEM—rectus femoris, GASMED—medial gastrocnemius, TIBANT—tibialis anterior.

The results from the FD analysis demonstrated a significant correlation between ACL strain and force (Figure 4). More specifically, the correlation between ACL force and ACL strain showed strong connection for each CM condition before the running training ($R^2 = 0.94, 0.92, \text{and} 0.97$ for 90°, 135°, and 180° CM). After completing the running training, the correlation between ACL force and ACL strain showed slightly stronger association as $R^2 = 0.97, p = 0.001$ for 90° CM, $R^2 = 0.98, p = 0.001$ for 135° CM, and $R^2 = 0.98, p = 0.001$ for 180° CM.

3.2. The One-Dimensional Statistical Nonparametric Mapping (SNPM1d) Comparison Results of Kinematics and Kinetics

In a case of non-Gaussian distribution of the data we use non-parametric test to compare the time series dataset of (knee angle, moment, and contact force, as well as ACL strain in three horizontal planes during 90°, 135° and 180° CM). We observed that ACL strain was lower after IR training during the 90° CM (Figure 5), and this difference was significant at 0–64%, $p = 0.011$ and 92–100%, $p = 0.010$ of the stance. Similarly, during the 135° CM, a lower ACL strain was observed after the IR training, but the difference was only significant at the beginning (0–8%, $p = 0.012$) of the stance phase. This was the case with 180 CM as well, however significant difference was not only found at the beginning of CM (0–8%, $p = 0.010$) but also at midstance phase as well (50–64%, $p = 0.010$).
Figure 4. Below are the correlations between ACL strain and force for the three CM directions, pre- and post-IR. (A–C) indicate the correlation between ACL strain and force in the knee joint while performing CM of 90°, 135° and 180°, respectively, before conducting the IR testing; (D–F) indicate the correlation between ACL strain and force in the knee joint while performing CM of 90°, 135° and 180°, respectively, post conducting the IR testing. Abbreviations: The intensity of the dots corresponds to the strength of the relationship between the two variables, with darker dots indicating a closer association.

Figure 5. The results of the SNPM1d test for ACL strain, as well as the heatmap results depicting mean clustering pre- and post-IR in various CM directions, were obtained. (A–C) demonstrates a comparison of ACL strain pre- and post-IR training during CM at 90°, 135° and 180°, respectively; (D) Displays the alterations in mean ACL strain values pre-IR training during CM at 90° (1), 135° (2) and 180° (3), as well as the changes in mean ACL strain values after IR training during CM at 90° (4), 135° (5) and 180° (6).
For the sake of brevity, the comparison of knee joint angle, moment, and contact force was only reported for 90° CM, as depicted in Figure 6. Before and after running training comparison of the external rotation moment of the knee revealed significantly lower moment at 65–97% ($p = 0.010$) of the stance phase (Figure 6B). A higher knee joint moment was observed during the beginning of the knee adduction (6–14%, $p = 0.015$) in the pre-training phase (Figure 6C). The magnitude of knee abduction moment was significantly higher at the 85–94% ($p = 0.010$) of the stance phase in pre-training compared to post-training. The SPNM1d test showed a significant increase in knee flexion angle (14–74% of the stance, $p = 0.020$) after training (Figure 6D). Knee extension angle was significantly greater during the 84–100% ($p = 0.012$) of the stance phase in pre-training than in post-training. The magnitude of IE rotation angle was significantly greater before the training compared to post-training, (Figure 6E). The adduction angle was significantly greater during the 14–74% ($p = 0.010$) of the stance phase after the running training than in post-training (Figure 6F). Knee joint contact forces in the sagittal plane increased significantly at 28–56% ($p = 0.023$) of the stance after training. The vertical direction knee joint contact force was markedly higher in the midstance of CM (34–68%, $p = 0.040$) during the pre-training trial than in the post training trial. We observed significant difference in the coronal plane knee joint contact forces at a longer percentage relative to the whole stance phase (10–12%, $p = 0.010$, 35–66%, $p = 0.012$ and 76–100%, $p = 0.020$) in pre training compared to the other planes.

**Figure 6.** The kinematic and kinetic means, as well as the standard deviation clouds of knee joint moments, angles, and joint contact forces, were compared pre- and post-IR training during 90° CM. (A) Demonstrates the changes in the moment of the knee FE stage; (B) demonstrates the change in the moment of the knee EI rotation stage; (C) demonstrates the change in the moment of knee AA stage; (D) demonstrates the change in angle of the knee FE stage; (E) demonstrates the change in the angle of the knee EI rotation stage; (F) demonstrates the change in the angle of knee AA stage; (G) demonstrates the change in knee joint contact force in the sagittal plane; (H) demonstrates the change in contact force of the knee joint in the vertical plane; (I) demonstrates the change in contact force of the knee joint in the coronal plane. (1) and (2) depict heat maps displaying the mean ACL strain values, with reference to Figure 5D; (1) A heat map was created to illustrate the change in mean ACL strain pre-IR training; (2) A heat map was created to illustrate the change in mean ACL strain post-IR training.
3.3. The Principal Component Analysis Results of Kinematics and Kinetics at 90° CM

The comparison of the first two principal components (PCs) for moment, angle, and contact force of the knee joint was carried out by generating score plots, as shown in Figure 7. The score plots comparing PC1 and PC2 reveal distinct groups of samples characterized by knee biomechanical features that exhibit high and low ACL strain. PCA was conducted on the joint moment, angle, and contact force obtained during 90° CM. The PCA score plots depicted the grouping of biomechanical characteristics, which were assessed based on the high and low ACL strain values. As demonstrated in Figure 7, a noteworthy disparity in the knee joint moment was observed during the AA rotation stage (Figure 7C). Notable dissimilarities in knee angle were evident during the FE (Figure 7D), EI rotation (Figure 7E), and AA rotation (Figure 7F) stages. A significant variance in knee contact force was observed in the sagittal plane (Figure 7G) and in the vertical plane as well (Figure 7I).

![Figure 7](image.png)

Figure 7. The results of PCA. PCA score plots demonstrate the biomechanical characteristics of 90° CM in pre-IR training. (A–C) demonstrates knee moments score based on the high and low levels of ACL strain during 90° CM; (D–F) demonstrates knee angles score based on the high and low levels of ACL strain during 90° CM; (G–I) demonstrates knee force score based on the high and low levels of ACL strain during 90° CM. The confidence interval of 95% was utilized as the clustering criterion for all PCA score plots.

4. Discussion

The main goal of this study was to investigate the effect of long-duration running training, which simulates the running load of a typical basketball match, on ACL strain during CM movements, which can indicate the risk of injury to ligaments. The second objective of this study was to investigate the changes in CM before and after the running training load to possibly identify specific lower extremity biomechanical characteristics that may be linked to high-risk ACL injuries. To explore changes in ACL strain we used a
model approach using OpenSim where ACL was presented as a nonlinear elastic passive soft tissue. To check the reliability of the model we also collected the electrical activity of the lower limb muscles to compare with CMC results.

According to Luque-Seron et al. [56], who demonstrated that human knee ligaments strain was typically up to 5% of the original ligament length during dynamic tasks. In contrast we observed twice as great elongation in the ACL at each condition. This can be explained by the different task and method to calculate ACL strain which was used in these studies. Furthermore, it has been suggested that micro-fiber damage and ACL rupture can occur when the strain reaches a percentage value between 9 and 15% [45]. ACL elongation in this study was simulated in OpenSim, which primarily calculates the length changes according to the kinematics of the knee combining with the ground reaction force data. The difference of CM characteristics in this study before and after the running training load is hard to compare with other studies because most reports used different protocol to induce fatigue, thus the outcome regarding the lower limb biomechanics is conflicting [57,58]. It has been suggested that incorporating targeted strength training could be a crucial factor in preventing ACL injuries [59]. We observed that ACL strain is lower (i.e., less elongation) after completing 5 km of IR training during each CM condition. It can be assumed that the highest strain was observed during the CM 90°, while the lowest strain was observed during CM 180°. Mechanical strain refers to the magnitude of elongation of a tissue relative to its resting length. Since ligaments cannot change their length by themselves, elongation of ligaments is induced by joint movement, and the amount of ligament lengthening depends on the amount of joint displacement and the magnitude of forces that stretch the ligaments. Therefore, the highest strain was observed during the initial ground contact when the ACL was elongated the most and during the takeoff phase where the knee extensor muscle generates high forces to accelerate the body.

The alterations in ACL strain following the running training load demonstrated a decrease in each CM condition, implying that fatigue may trigger an elevation in ligament rigidity that leads to diminished elongation. Conversely, the CM condition showed higher knee flexion following the training, which may have resulted in greater ACL elongation due to the observed increase in knee joint force during peak knee joint flexion. Thus, the lower strain can only be attributed to the changes in ligament stiffness induced by fatigue. Although the ACL strain was lower in CM 135° and CM 180°, the difference was not statistically significant during most of the stance. The multidirectional load placed on the ACL by CM necessitates a comprehensive examination of knee joint biomechanics in various planes for an improved understanding of ACL mechanics. Notably, significant differences in frontal and coronal plane joint displacement, forces, and moments were observed after the running training load, indicating a less stable knee joint that could increase the risk of knee injury. It has been recommended that strengthening the knee extensor and flexor muscles can help to reduce overload injury risk of ACL during the knee flexion and extension during CM [60]. The method presented in this study can be used for individual assessment of ACL mechanics during CM enabling to monitor changes in ACL strain over time. This can reflect the athlete’s improvement in neuromuscular control after a targeted training regimen, and enable to check how effectively can it reduce ACL strain during CM [61]. Most of knee extensor and flexor muscles are biarticular muscles thus fatigue of the muscles affects the hip joint as well. We also observed changes in hip kinematic characteristics, which may have an impact on knee joint mechanics and indirectly affect ACL mechanics. Therefore, targeted strength training of the hip joint muscles could potentially aid in reducing the risk of ACL injuries [62].

The PCA revealed notable distinctions in knee moment during the AA rotation stage, knee angle during FE (as shown in Figure 7D), and EI rotation (as depicted in Figure 7E), in addition to knee contact force in both the sagittal (Figure 7G) and coronal (Figure 7I) planes, when performing CM. This will hopefully serve as the key determinant in distinguishing between high and low levels of ACL strain. The primary function of the ACL is to resist anterior translation movement of the tibia with respect to the femur [63], but movements in
the frontal and transverse planes can also affect ACL loading [64–66]. Our results showed more internal knee moment at the beginning, which may have a substantial impact on ACL loading. During CM, the initial acceleration of the body, induced by its inertia, causes it to continue moving forward. However, the generation of more inward moment at the knee joint creates a counteracting force against the inertia, providing additional support to the body. The generated inward moment at the knee joint during cutting may also increase the risk of soft tissue injury due to insufficient internal force, as the knee joint is not able to counteract the generated moment effectively [67]. During change of direction the body is required to accelerate into a different direction, which can result in an increased knee abduction moment which may impose a significant load on the ACL [68]. Limiting the axial rotation of the tibia could potentially affect the knee’s natural movement when a valgus moment is applied. Therefore, in this study, we did not constrain the kinematics of the knee joint, which increased the accuracy of the model in calculating the ACL strain. The findings of this investigation reveal that variations in knee angle during all CM conditions across all three planes cause a significant influence on the mechanical loading of the ACL. The quick transition from internal to external rotation during athletic movements can lead to an imbalance in muscle function [69], which in turn may result in a great increase in knee joint loading and increase the risk of injury. This assumption can be supported by previous reports that also indicate that an increase in the knee abduction angle can have a significant impact on the risk of ACL injury [19,70]. While the relative importance of non-sagittal loads to ACL loading is still a topic of debate [71], it is widely accepted that anterior and posterior shear forces can lead to loading and unloading of the ACL, respectively [64,72]. Based on the results of our study, we recommend the inclusion of strength exercises performed in the coronal plane as part of knee injury prevention, which is important due to the distribution of muscles involved in completing cutting movements [12,73].

While our study has revealed some novel insights, we acknowledge that there are some methodological limitations that need to be addressed. It should be noted that this study had a limited sample size of only 15 healthy male athletes, which limits the generalization of our results. ACL in this study was modeled as a non-linear, elastic, passive soft tissue with homogenic material properties and shape, which does not accurately represent the individual differences. Although the model was scaled for everyone, the constraints of the model must be considered when interpreting the data. It is important to note that the model used in this investigation modeled the AM and PL bundles as individual entities. Assuming that the AM and PL bundles have similar characteristics and parallel orientations in the model applied in this investigation may have overlooked their constraints on each other. In future studies, it would be beneficial to establish AM and PL bundles separately to investigate if this produces distinct effects. The fatigue protocol we used in this study was the same for each person, but the fitness level of the individuals can influence the level of fatigue that occurred.

Since gender, age, sport-specific loads, and pathological conditions can influence ACL strain, it is recommended to collect further information about the effect of the abovementioned variances on ACL strain.

In this study, notable alterations in ACL strain were observed specifically at 90° CM. Consequently, the subsequent comparison and discussion of biomechanical characteristics in this paper solely focus on the results pertaining to the 90° case. However, in future research, our aim is to furnish more comprehensive insights into the patterns exhibited across all three cutting orientations, thereby providing a more informative analysis.

5. Conclusions

In this study, we investigated the changes of knee joint biomechanics during CM and compared ACL strain before and after 5 km of IR training. The study revealed that ACL strain become significantly lower after completing training. This difference is greatest during knee flexion and the late push-off phase of CM. We also demonstrated that different knee joint angle results in different ACL strain, pointing out that the 90° CM is potentially
subjecting the ACL to the highest risk of injury. Attention should be paid to the increased AA rotation moment of the body during confrontational maneuvers, as well as the increased contact forces in the coronal plane of the knee joint. Implementing training programs that aim to reduce the AA rotation moment and contact forces during CM could potentially play a vital role in preventing ACL strains, and subsequently reducing the overall risk of ACL injuries in athletes.

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