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Lower-Extremity Kinematics of Soccer Players with Chronic Ankle Instability during Running: A Case-Control Study

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Abstract: The purpose of this study was to clarify the characteristics of lower-extremity kinematics during the running of soccer players with chronic ankle instability (CAI) in comparison to those without CAI. Twenty-two male college soccer players participated in this study. Twelve players were assigned to the CAI group and ten players to the non-CAI group, and players were diagnosed according to the Cumberland Ankle Instability Tool. Kinematic data of the hip, knee, ankle, foot, and ground reaction force components during the stance phase of running were obtained using a three-dimensional motion analysis system. The results revealed that soccer players with CAI who landed with ankle inversion and other characteristic kinematics in their lower extremity during the stance phase of running were similar to those without CAI. These results show that running kinematics in soccer players are not affected by the presence or absence of CAI. Future studies based on the results of this study may contribute to the analysis of the risk of developing CAI during soccer and may also help prevent lateral ankle sprains.

Keywords: ankle sprain; biomechanics; motion analysis; joint instability

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1. Introduction

Soccer players experience high rates of ankle sprain injuries and re-injuries during practices and matches [1–3]. In addition, players with this type of injury require medical treatment and rehabilitation; moreover, it takes a long time for players to return to play with sufficient physical function [4]. Therefore, ankle sprains are considered to be extremely frequent injuries in soccer and a major factor preventing return to play, suggesting the importance of preventing ankle sprains in soccer players.

According to prospective studies, approximately 66–73% of ankle sprains most often involve the anterior talofibular ligament (ATFL) [1,5] with an excessive inversion and internal rotation of the hindfoot [6]. In addition, structural and functional failure of the ankle after a sprain results in the prolonged instability of the ankle joint [7,8]. This pathological condition, known as chronic ankle instability (CAI), has been defined as "repetitive bouts of lateral ankle instability resulting in numerous ankle sprains" [6,9]. It can occur either or both as functional ankle instability and mechanical (structural) ankle instability [6]. Most cases of CAI after an ankle sprain are due to the loosening of the static structure of the lateral collateral ligaments, including the ATFL, which are important for the joint stability of the ankle [6]. In a prospective cohort analysis, 40% of lateral ankle sprains presented CAI for a year after a first-time injury [10]. Ankle hyper-inversion, which is caused by lateral ankle instability, is likely to result in trauma to the lateral ankle ligaments when the evertor muscles cannot counteract the external inversion torque during landing [6,11]. Furthermore, recent studies have reported that individuals with CAI experienced a lack of dynamic postural control ability [12], landing kinematics [13], and weakness in their knee and hip muscles [14]. Therefore, taping and bracing have been widely used for individuals with CAI to prevent secondary injuries with an increase in CAI and the risk of recurrent ankle sprains.

Previous biomechanical studies have reported that CAI is associated with greater ankle inversion during the landing phase of gait [15–17]. Koldenhoven et al. (2019) reported that individuals with CAI demonstrated greater ankle inversion at the initial contact during gait; moreover, the ankle inversion angle increased as the gait speed increased [17]. In addition, it has been reported that ankles with CAI are controlled by the evertor muscles working concentrically, which may be the result of a change in the preprogrammed feed-forward motor control during the early stance phase of gait [15,16]. These kinematic characteristics of ankles with CAI may increase the mechanical stress applied to the ankle joint structures during gait; moreover, it may result in recurrent ankle sprains and consequent damage to ankle joint structures.

Soccer players are required to run a long total distance and a large number of sprints with high acceleration during a match [18,19]. For soccer players, running is the most frequent motion, and the total running distance per match is estimated to be approximately 9.4 km at an average speed of 91 m/min [20]. In addition, soccer players belonging to top-ranking teams ran a greater total distance than those belonging to bottom-ranking teams [19]. These findings suggest that, for soccer players, running is an indispensable motion during soccer because it contributes to increasing the possession rate of the ball and the chance of scoring. Furthermore, an inverted ankle position is one of the causes of sprained ankles upon initial contact during running. [21]. Therefore, recognizing the kinematic characteristics of running in soccer players with CAI is important for the prevention of ankle disorders, including recurrent ankle sprains and secondary injuries in the lower extremities. However, no study has yet investigated lower-extremity kinematics in soccer players with CAI during running. This study aimed to clarify the characteristics of lower-extremity kinematics during running in soccer players with CAI in comparison with those without CAI. We hypothesized that soccer players with CAI would land with ankle inversion or other characteristic kinematics in the lower extremities during running. The results of this study highlight the importance of preventing CAI after an ankle sprain in soccer players and the necessity of reconditioning them.

2. Materials and Methods

2.1. Participants

Twenty-two male college soccer players with more than eight years of soccer experience were included in this study. They were recruited between August 2019 and March 2020, and the study was conducted from January to March 2020 at the International University of Health and Welfare, Narita Campus, Chiba, Japan.

The preferred leg for kicking a ball for all participants was the right leg with the support of the left leg. Prior to inclusion in this study, the participants answered a questionnaire inquiring about their physical characteristics (age, height, and weight), current medical information, and medical history, including ankle sprains. The inclusion criteria were a soccer experience of >6 years, age $\geq$18 years, and absence of current pain in the lower extremities and lumber during daily activities and soccer practice. The exclusion criteria were any history of orthopedic surgery to the lower extremities and lumber. The study protocol was approved by the International University of Health and Welfare Ethics Committee (Approval No. 19-Io-59) and was conducted in accordance with the tenets of the Declaration of Helsinki. All participants provided written informed consent prior to testing.

2.2. Instrumentation and Measurement Protocols

All participants were required to answer the Cumberland Ankle Instability Tool (CAIT), which has been confirmed to be valid and reliable in the severity of the functional instability of the ankle [22,23]. The CAIT consists of nine items scored on a scale with a maximum of 30 points. In accordance with Hiller et al. (2011) and De Noronha et al. (2008), all participants were divided into non-CAI ($\geq$28 points) and CAI ($\leq$27 points with a history of ankle sprain) groups based on their nondominant legs [22,24].

Measurements were taken at the Motion Analysis Laboratory of the International University of Health and Welfare Narita Campus, Chiba, Japan. All participants wore closely fitted dark shorts and were barefoot for data collection. A three-dimensional motion analysis system with eight cameras (Vicon MX system, Oxford Metrics, Oxford, UK) and six AMTI force plates with an amplifier (MSA-6 Mini Amp, AMTI, Watertown, MA, USA) were used to record lower-extremity kinematic data and the ground reaction force (GRF) during testing at a sampling rate of 250 Hz and 1000 Hz, respectively. All kinematic data and GRF data were low-pass filtered at 16 Hz and 20 Hz using a fourth-order zero-lag Butterworth filter, respectively [25].

Reflective markers (14 mm) were placed to create the coordinate systems of the head, torso, arm, pelvis, and leg segments, according to a full-body Vicon Plug-in-Gait model (Oxford Metrics, Oxford, UK). Reflective markers (9.5 mm) were added to create the hindfoot and forefoot segments, according to the Oxford foot model [26,27]. These segment models were used to obtain the lower-extremity and foot kinematic data during testing. The angles of the lower-extremity joints and GRF component data were calculated using the Vicon Nexus 2 system (Oxford Metrics).

After a standardized warm-up, the participants performed the running task, which consisted of running five laps at a self-selected and comfortable speed on a 30-m runway on hard carpet with the presence of AMTI force plates. The participants were instructed by a research assistant to perform the running task. Trials in which the foot did not entirely strike the force plates were excluded and retaken until five successful trials were obtained. During all trials, participants were required to look straight ahead and not to aim to strike the force plates.

2.3. Data Analysis

The hip, knee, ankle, and foot angles, and the GRF component data (anteroposterior, mediolateral, and vertical) of the non-preferred leg, were obtained during the stance phase of running. The stance phase was defined as the duration from heel-strike, when the vertical GRF first exceeded 10 N, to toe-off, when GRF fell below 10 N [28]. For each participant, the angles of the hip, knee, ankle, foot, and GRF of the non-preferred leg were calculated during the stance phase of running for the middle three of the five successful running trials. The GRF component data were normalized to the body weight (kg) of the participants. These variables were converted to a percentage of the stance phase, with 0% representing heel contact with the force plate and 100% the toe-off. The output angles for the lower extremities were calculated from the YXZ Cardan angles derived by comparing the relative orientations of the two segments. In addition, the angles of the hindfoot with respect to the tibia (HFTBA) and forefoot with respect to the hindfoot (FFHFA) of the non-preferred leg were calculated.

2.4. Statistical Analysis

The Kolmogorov–Smirnov test was used to confirm that the kinematic data and GRF component data were normally distributed ($p < 0.05$). Depending on whether the physical characteristics, kinematics, and GRF component data were normally distributed, the Mann–Whitney U test or Student's t-test was used to identify differences in variables between the CAI and non-CAI groups. A probability (p) value of < 0.05 was considered statistically significant. Cohen's d effect sizes (ESs) were calculated for all analyses and showed the magnitude of differences between the CAI and non-CAI groups. Data analyses were conducted using IBM SPSS Statistics for Windows version 24.0. (IBM Corporation, Armonk, NY, USA) by an independent statistician. One-dimensional statistical parametric mapping (SPM[t]) unpaired t-tests were performed to compare each individual point of the angle and the GRF curves between the CAI and non-CAI groups. The threshold of significance was set at $\alpha = 0.05$ for all analyses. SPM(t) unpaired t-tests were conducted using the open-source SPM1d code (www.spm1d.org, accessed on 22 March 2022) in MATLAB_R2020b (The Mathworks Inc., Boston, MA, USA).

3. Results

Based on the CAIT score, all participants were assigned to groups, with 12 players in the CAI group (mean CAIT score, 23.6 ± 4.3; mean age, 20.6 ± 0.9 years; mean height, 173.7 ± 5.3 cm; mean weight, 66.5 ± 4.9 kg; mean playing experience, 9.8 ± 2.4 years) and 10 players in the non-CAI group (mean CAIT score, 29.6 ± 0.8; mean age, 29.3 ± 0.9 years; mean height, 172.1 ± 5.5 cm; mean weight, 64.5 ± 6.5 kg; mean playing experience, 10.7 ± 2.7 years) (Table 1).

Table 1. Physical characteristics of the CAI and the non-CAI groups.

	CAI Group	Non-CAI Group	*p* Value
	Mean ± SD [a]	Mean ± SD [a]	
Number	12	10	-
CAIT score	23.6 ± 4.3	29.6 ± 0.8	-
Age (years)	20.6 ± 0.9	20.8 ± 0.9	0.54
Height (cm)	173.7 ± 5.3	172.1 ± 5.5	0.46
Weight (kg)	66.5 ± 4.9	64.5 ± 6.5	0.12
Playing experience (years)	9.8 ± 2.4	10.7 ± 2.7	0.36

[a] SD; Standard deviation.

There was no difference in the peak GRF components between the CAI and non-CAI groups (Table 2, $p > 0.05$). In addition, there was no difference in the maximum values of the lower-extremity kinematics between the CAI and non-CAI groups during the stance phase of running (Table 3, $p > 0.05$).

The time-series data of the GRF components of the CAI and non-CAI groups are shown in Figure 1. The time-series data of hip, knee, ankle, and foot (HFTBA and FFHFA) kinematics of the CAI and non-CAI groups are shown in Figure 2. No between-group differences in these variables were observed at any time point during the stance phase of running.

Table 2. The maximum values of the peak GRF components of the CAI and the non-CAI groups.

	CAI	Non-CAI	Mean Difference (95% CI [b])		*t* Value	*p* Value	Effect Size Cohen's *d*
	Mean ± SD [a]	Mean ± SD [a]					
GRF [c], N/kg							
Anterior (−)/Posterior (+)	2.11 ± 0.37	1.94 ± 2.73	1.8	(−3.0–6.6)	0.78	0.45	0.35
Medial (−)/Lateral (+)	0.55 ± 0.24	0.46 ± 0.22	0.9	(−1.2–2.9)	0.89	0.38	0.41
Vertical	20.85 ± 1.70	19.54 ± 2.74	13.1	(−6.8–33.0)	1.37	0.19	0.59

[a] SD; Standard deviation, [b] CI; Confidence interval [c] GRF; Ground reaction force.

Table 3. The maximum values of the lower-extremity kinematics of the CAI and the Non-CAI groups.

	CAI	Non-CAI	Mean Difference (95% CI [b])		*t* Value	*p* Value	Effect Size Cohen's *d*
	Mean ± SD [a]	Mean ± SD [a]					
Hip, deg							
Flexion	30.5 ± 9.6	30.3 ± 12.5	0.2	(−9.6–10.0)	0.04	0.97	0.17
Adduction	8.5 ± 5.7	10.3 ± 2.6	−1.8	(−5.7–2.2)	−0.96	0.35	−0.38
Internal Rotation	18.4 ± 24.0	7.4 ± 21.8	11.0	(−9.5–31.6)	1.12	0.28	0.48
Knee, deg							
Flexion	33.2 ± 8.6	36.7 ± 6.9	−3.5	(−10.6–3.5)	−1.05	0.31	−0.45

Table 3. *Cont.*

	CAI	Non-CAI	Mean Difference (95% CI [b])		*t* Value	*p* Value	Effect Size Cohen's *d*
	Mean ± SD [a]	Mean ± SD [a]					
Varus	13.2 ± 14.6	6.2 ± 10.5	7.0	(−4.5–18.5)	1.27	0.22	0.54
Internal Rotation	2.9 ± 12.0	0.6 ± 16.7	2.2	(−10.6–15.0)	0.36	0.72	0.16
Ankle and Foot, deg							
Ankle Dorsi Flexion	24.4 ± 9.1	24.5 ± 5.5	−0.2	(−7.0–6.7)	−0.05	0.96	−0.02
HFTBA [c] Dorsi Flexion	9.4 ± 3.2	8.4 ± 5.2	1.0	(−2.8–4.8)	0.57	0.58	0.24
HFTBA [c] Inversion	−10.3 ± 2.6	−12.7 ± 4.9	2.4	(−1.5–3.2)	1.47	0.16	0.63
FFHFA [d] Dorsi Flexion	13.5 ± 4.2	14.6 ± 7.2	−1.1	(−6.2–4.0)	−0.44	0.66	−0.19
FFHFA [d] Inversion	−4.0 ± 1.8	−4.5 ± 3.5	0.5	(−3.9–4.7)	0.47	0.65	0.20

[a] SD; Standard deviation, [b] CI; Confidence interval; [c] HFTBA: Hindfoot with respect to tibia. [d] FFHFA: Forefoot with respect to hindfoot.

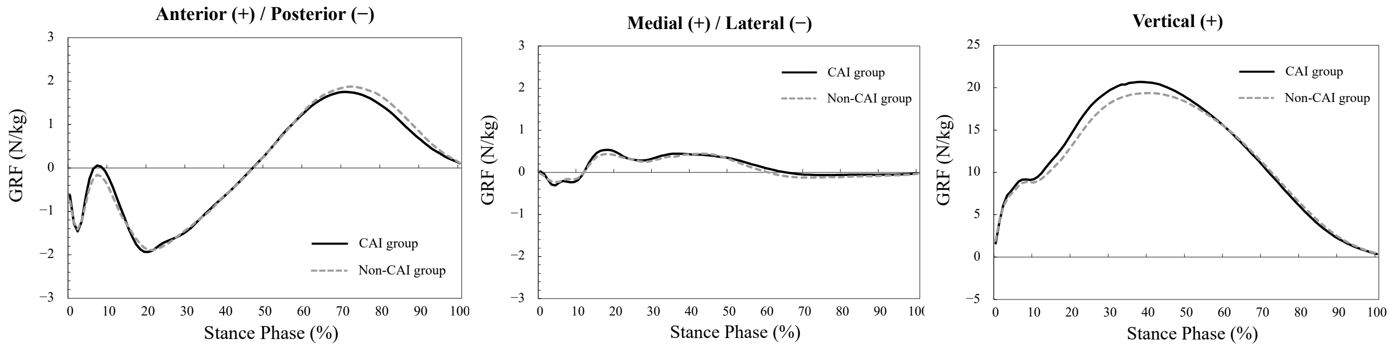

Figure 1. Time-series data of the three components of the GRF of the CAI and non-CAI groups during the stance phase of running.

Regarding the mean ± standard deviation values of the three components of the GRF in the CAI group (black lines) and non-CAI group (dotted gray lines), the stance phase (%), which is normalized to 100% of the stance phase, with 0% representing heel contact with the force plate and 100% toe-off, is presented on the x-axis.

Regarding the mean ± standard deviation values of the hip, knee, and foot angles in the CAI group (black lines) and non-CAI group (dotted gray lines), the stance phase (%), normalized to 100% of the stance phase, with 0% representing heel contact with the force plate and 100% toe-off, is presented on the x-axis. HFTBA and FFHFA represent the angles of the hindfoot with respect to the tibia and the angles of the forefoot with respect to the hindfoot, respectively.

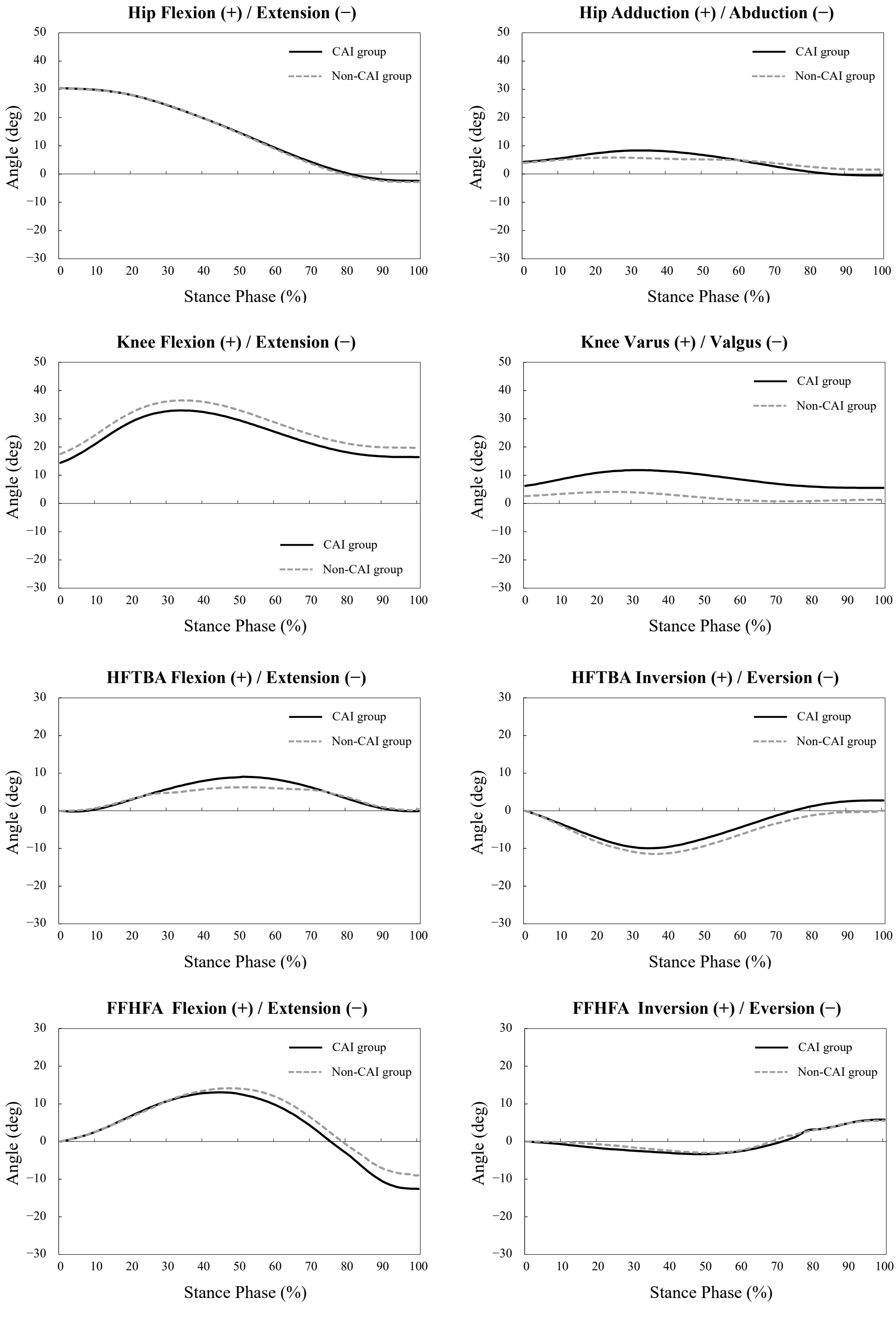

Figure 2. Time-series data of the hip, knee, and foot angles of the CAI and non-CAI groups during the stance phase of running.

4. Discussion

This case-control study aimed to clarify the characteristics of lower-extremity kinematics during running in soccer players with CAI in comparison to those without CAI. The results revealed that soccer players with CAI landed with ankle inversion and other characteristic kinematics in the lower extremity during the stance phase of running, similar to those without CAI. The research hypothesis that soccer players with CAI land with ankle inversion or other characteristic kinematics in the lower extremities during running was not validated in this study.

Delahunt et al. (2006) reported that individuals with functional ankle instability exhibited a more inverted ankle position immediately after heel strike during the post-heel strike phase of gait [16]. Moreover, Moisan et al. (2021) demonstrated that individuals with CAI exhibited greater ankle inversion angles, from 14 to 48% of the stance phase during the gait, compared to those without CAI [29]. In addition, a systematic review by Hoch and McKeon (2014) reported that the peroneus longus muscle in individuals with CAI presents reflex latency and an electromechanical delay of activation [30]. This was considered to be the cause of the lateral ankle sprain because the peroneus longus muscle is responsible for maintaining the ankle eversion position or for generating the ankle eversion moment [30]. Therefore, greater ankle inversion and delayed peroneus longus activity were characteristic kinematic changes in individuals with CAI during walking, suggesting that it may increase the risk of a recurrence of lateral ankle sprains.

Regarding running biomechanics in individuals with CAI, our results did not show an inverted ankle position in soccer players with CAI compared to those without CAI. This was in contrast to previous studies that reported ankle and foot kinematics in individuals with CAI during gait [16,29,30]. One of the reasons for this difference between gait and running is the GRF characteristic of the heel contact. In general, the GRF in the lateral direction at landing generates the ankle eversion moment, whereas, in the medial direction, it is the ankle inversion moment. During gait, the lateral GRF reaches the peak values (approximately 2.0% BW) in the early stance phase (0–6% gait cycle) [31]. In contrast, the lateral GRF during gait was lower in our study than that reported in previous studies [31,32]. Wannop et al. (2012) evaluated the GRF components while walking and running and clarified that the lateral GRF is smaller in the early stance phase during running than during gait [33]. These results indicate that the lower lateral GRF suppressed ankle inversion in soccer players with CAI. Furthermore, it has been reported that the mediolateral center of mass (COM) displacement during gait decreases as gait speed increases [34,35]. Therefore, the need for the ankle to control posture in the mediolateral direction may disappear during running as a result of stable COM associated with increased movement speed. This may be one of the reasons why the presence of CAI did not affect ankle and foot alignment during running in this study.

Koldenhoven et al. (2021) reported that women with CAI had an inverted ankle position upon initial contact during running, while those without CAI had an everted ankle position [21]. As this study was conducted on active females who had, or had recovered from, CAI, it was different from the male soccer players targeted in our study. Moreover, Previous studies have reported sex differences in lower-extremity kinematics and muscle activity characteristics during running [36–38]. Wilson and Madigan (2007) reported that the peroneus longus muscle reflex amplitude, which is related to lateral ankle sprain, decreased by 11.3% in men with fatigue [36]. Sinclair et al. (2012) showed differences between sexes in hip, knee, and ankle angles during running; in particular, males had a greater ankle inversion angle at toe-off and a lower peak eversion angle in the stance phase [37]. In addition, soccer players have higher physical fitness in sprinting, jump performance, and postural balance than untrained adolescents. Therefore, it should be noted that the results of this study are specific to male soccer players. This indicates that, even if soccer players have CAI after a lateral ankle sprain, they may have the ability to compensate with greater lower extremity function.

In this study, it was revealed that the lower-extremity kinematics of soccer players with CAI, during running, were not different from those without CAI. Soccer players are required to perform various complex movements such as jumping, cutting, and kicking the ball as well as running during matches and practices [39]. In particular, soccer includes the most frequent cutting movements (up to 800 per game) during running [39]. Therefore, in addition to running in a straight line, as in this study, the biomechanical characteristics of running with lateral movement and cutting should be investigated in soccer players with

CAI. This future study, along with the findings of this study, may contribute to an analysis of the risks of developing CAI during soccer and may help prevent lateral ankle sprains.

This study had some limitations. First, the presence of mechanical ankle instability or functional ankle instability in soccer players was not evaluated because the CAIT was used as a subjective evaluation tool to measure the severity of CAI, including mechanical ankle instability, functional ankle instability, or a combination of these two phenomena [6]. Therefore, it may be necessary to accurately evaluate mechanical and functional instability to determine the existence of CAI after an ankle sprain. Second, ankle and foot kinetic measurements with a multi-segment foot model, such as the Oxford foot model, are hampered by measurement limitations and modeling assumptions [40,41]. Joint moments of the ankle and foot were not evaluated during running in this study; therefore, we focused only on the lower extremities and foot kinematics. Finally, all running trials in this study were conducted barefoot to minimize measurement errors by shifting the reflective markers during running. In addition, all trials were conducted in the laboratory, and running speed was not specified, unlike the playing environment in daily soccer practice and games. Therefore, these results did not consider the effects of shock absorption from shoes or the protection/fixation of the foot shape.

5. Conclusions

Soccer players with CAI landed with ankle inversion and other characteristic kinematics in the lower extremity during the stance phase of running were similar to those without CAI. These results show that running kinematics in soccer players were not affected by the presence or absence of CAI. These findings were considered to be mainly due to the effects of the lower lateral GRF during running and the stable COM with a greater moving speed than gait. In addition to running in a straight line, as in this study, the biomechanical characteristics of running with lateral movement and cutting should be investigated in soccer players with CAI.

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Data Availability Statement: The data presented in this study are available on request from the corresponding author. The data are not publicly available due to privacy concerns.

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