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ORIGINAL RESEARCH

The alterations of leg joint angular displacements and muscle co-contraction at landing following various aerial catching movements

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Abstract

Background: Jump-landing is a major cause of lower limb injuries. This study investigated the effect of various aerial catching movements on the co-contraction index (CCI) of the knee and ankle muscles, as well as on joint angular displacements and joint moments during landing. **Methods**: Fifteen right-hand dominant collegiate basketball players (age: 21.0 ± 1.2 years; weight: 79.9 ± 7.9 kg; height: 180.9 ± 5.5 cm; training experience, 5.6 ± 3.5 years) performed maximal countermovement jumps under four conditions: no catching (NC), right (RULC), left (LULC) and bilateral (BULC) upper-limb catching. Electromyography of the rectus femoris (RF), biceps femoris (BF), lateral gastrocnemius (LG) and tibialis anterior (TA), along with kinematic and kinetic data, were recorded during landing. The co-contraction index (CCI) of knee and ankle muscles before and after landing, as well as angular displacements and peak joint moments were calculated. **Results**: When compared to the NC condition, all aerial catching movements resulted in reduced knee flexion (RULC, p = 0.004, d = −1.11; LULC, $p < 0.001$, d = −1.52; BULC, p = 0.013, d = −0.97) and increased ankle dorsiflexion (RULC, p = 0.002, d = −1.21; LULC, p = 0.004, d = −1.13; BULC, $p < 0.001$, d = −1.50) angular displacement after landing, along with significantly higher CCI of RF-BF (RULC, p = 0.018, d = 0.93; LULC, p = 0.033, d = 0.85; BULC, p = 0.042, d = 0.81) and LG-TA (RULC, p = 0.025, d = 0.88; LULC, p = 0.004, d = 1.12; BULC, p = 0.015, d = 0.95) before landing, the LULC condition led to greater knee abduction angular displacement (p = 0.002, d = 1.19) and moment (p = 0.001, d = 1.26), and lower RF-BF CCI after landing (p = 0.037, d = 0.83). Aerial catching movements increased lower limb muscle co-contraction before landing and led to greater knee stiffness after landing. However, LULC reduced knee co-contraction and increased frontal plane knee motion after landing, indicating decreased joint stability and higher injury risk. **Conclusions**: Injury prevention programs should incorporate upper limb coordination and perturbation training—especially for the left arm—to enhance motor control and joint stability during sport-specific tasks.

Keywords

Electromyography; Lower limb injuries; Biomechanics; Anterior cruciate ligament

1. Introduction

Jump-landing is a common movement frequently performed in various sports, particularly in basketball and volleyball, during actions such as shooting, blocking and receiving passes [1, 2]. However, the high frequency of jump-landings has been recognized as a significant contributing factor to lower limb injuries in athletes [1]. To more accurately assess injury risk under conditions that closely mimic real-world sports scenarios, recent studies have increasingly investigated the effects of sport-specific upper limb actions performed during the aerial phase such as catching a ball mid-air [3], executing trunk extension [4, 5] or performing various upper limb catching tasks [6]—on the biomechanics of lower limb landing. Although these actions are essential for successful skill execution, they may inadvertently increase postural control demands [4, 6], consequently altering landing mechanics. Nonetheless, existing studies have primarily focused on discrete variables such as peak joint angles and ground reaction forces, offering only a limited view of the underlying movement strategies. Little attention has been paid to the fact that indices such as joint angle displacements (*i.e.*, the total angular excursion of a joint across a movement phase) during the landing phase and co-contraction ratios of agonist and antagonist muscle groups can reflect the cumulative coordination of joint kinematic trajectories and neuromuscular over continuous time, which is essential for understanding joint stability and injury mechanisms.

Previous studies have shown that decreased knee flexion angular displacement in the sagittal plane [7, 8] and increased knee abduction angular displacement in the frontal plane [9] are key factors contributing to lower limb injuries during dynamic movements. In addition, both *in vivo* and *in vitro* studies have further demonstrated the underlying mechanical mechanisms contributing to these injury patterns. Specifically, the quadriceps are primarily responsible for generating anterior tibial translation, a movement closely associated with non-contact anterior cruciate ligament (ACL) injuries [10, 11]. In contrast, the hamstrings provide a counteracting force that helps resist quadriceps-induced anterior tibial translation [12, 13], thereby reducing the likelihood of ACL injury. Furthermore, co-contraction of the quadriceps and hamstrings is essential for stabilizing the knee joint and reducing tensile stress on the ACL [13].

Moreover, the same movement performed by the dominant and non-dominant arm may involve different neuromuscular control strategies [6, 14, 15]. Tasks involving the non-dominant arm may elicit greater neural interference [15], potentially impairing neuromuscular control of the lower limbs during landing and compromising muscle activation effectiveness. From a motor control perspective, aerial catching movements can introduce a dual-task scenario in which upper and lower limb coordination compete for shared central processing resources. This cognitive-motor interference may compromise neuromuscular activation strategies in the lower extremities, leading to an increased risk of injury. Muscle activation before landing, commonly referred to as feedforward activation, is a preprogrammed mechanism that prepares the lower extremities for impact absorption at ground contact [16, 17]. However, excessive co-contraction of the quadriceps and hamstrings before landing may reduce the knee flexion angle during the landing phase, potentially increasing injury risk [17]. Therefore, to better understand the underlying mechanisms by which upper limb movements during the aerial phase may influence injury risk during landing in real-world sports scenario, it is necessary to investigate the effects of different aerial catching movements on lower limb joint angular displacements and muscle co-contraction patterns during landing.

Previous studies [3–6] have rarely focused on lower limb muscle co-contraction or joint angular displacements during landing in aerial catching tasks, especially under the different upper limb movements. Accordingly, the present study aimed to investigate the effects of different aerial catching movements performed after take-off on the co-contraction index (CCI) of knee and ankle muscles, as well as on joint angular displacements and joint moments at the knee and ankle during landing. Based on previous evidence [3, 4, 6, 17] we hypothesize that these aerial catching tasks would lead to higher CCI before landing, reduced sagittal plane knee displacement and increased frontal plane displacement and joint moments, potentially elevating the risk of lower limb injuries particularly in conditions involving left upper limb movement.

2. Method

2.1 Experimental design

This study employed a cross-sectional design. Prior to the formal testing, all participants completed a standardized warm-up consisting of ten-minute of self-paced jogging on a treadmill followed by three-minute of dynamic stretching. Participants wore standardized athletic footwear and stood with their feet shoulder-width apart on two adjacent force platforms. The arms were flexed at approximately 90 degrees at the elbows and positioned close to the torso in a natural, ready stance. This starting posture was maintained across all test conditions to ensure consistency in movement initiation. Each participant performed a maximal countermovement jump (CMJ) while executing one of four upper limb catching movement conditions during the flight phase: no catching (NC), left upper limb catching (LULC), right upper limb catching (RULC) and bilateral upper limb catching (BULC). During the aerial phase, participants visually fixated on a target positioned 10 cm above their individual maximum CMJ reach height and attempted to reach it without making contact. Participants landed on the two force platforms, with no additional landing instructions or verbal cues provided. After completing 2–3 familiarization trials for each condition, participants rested for three minutes before the start of formal data collection. During the formal test, the four aerial movements were performed in a randomized order. A 60-second rest interval was provided between each CMJ trial to minimize fatigue. For each condition, data collection continued until a minimum of three valid trials were obtained.

2.2 Participants

Fifteen healthy male collegiate basketball athletes (age: 21.0 ± 1.2 years; weight: 79.9 ± 7.9 kg; height: 180.9 ± 5.5 cm; training experience, 5.6 ± 3.5 years; mean ± standard deviation), all right-hand dominant, were recruited to participate in this study. Participants were recruited between May and June 2024 through university sports teams. Inclusion criteria required participants to engage in physical training or competition at least four times per week, with each session lasting no less than 1.5 hours. Exclusion criteria included: history of ACL injury or other major lower limb injuries requiring surgical intervention; inability to participate in physical activity due to lower limb injury in the past six months; presence of cardiovascular, respiratory, neurological or other conditions limiting maximal physical effort; known allergy to adhesive materials. Sample size estimation was conducted using G*Power 3.1.9.2 software (Heinrich Heine University Düsseldorf, Düsseldorf, NRW, Germany) to determine the minimum required sample size. The analysis was based on a repeated-measures analysis of variance (ANOVA) (within-factors), with four measurements per participant, an assumed effect size of 0.25, an alpha level of 0.05, statistical power of 0.80 and a correlation among repeated measures of 0.71. The results indicated that a minimum of 14 participants was required to achieve sufficient statistical power for detecting significant effects. To ensure sufficient statistical power, 15 participants were recruited. Written informed consent was obtained from all participants following a thorough explanation of the experimental procedures, potential risks and the right to withdraw at any time. All procedures were conducted in accordance with the Declaration of Helsinki (1975), revised in 2013, and were approved by the Jeonbuk University Ethics Committee (JBNU2022-04-008-002, 01 April 2022).

2.3 Data collection

Fifty-seven reflective markers were attached to each participant to capture full-body motion. A motion capture system equipped with 13 infrared cameras (OptiTrack, Natural Point, Inc., Corvallis, OR, USA) was used to record kinematic data at a sampling rate of 120 Hz. Four wireless surface electromyography (EMG) sensors (Trigno Avanti, Delsys, Natick, MA, USA) were employed to record muscle activity, each fitted with Ag dual-differential bar electrodes (2.7 $\times$ 3.7 cm) and operating at a sampling frequency of 1200 Hz per channel. Sensors were placed on the dominant side of the following lower limb muscles: rectus femoris (RF), biceps femoris (BF), lateral gastrocnemius (LG) and tibialis anterior (TA). The electrodes were aligned parallel to the muscle fiber orientation [18], and placement locations were determined based on a muscle innervation zone atlas [19]. Prior to sensor placement, the skin was shaved and cleansed with alcohol wipes. After placement, surgical waterproof adhesive film (Tegaderm, 3M, St. Paul, MN, USA) was used to secure the sensors for signal stability and reduce motion artifacts during movement [16]. Ground reaction force (GRF) data were collected using two OR6-6-2000 force platforms (Advanced Mechanical Technology Inc., Watertown, MA, USA) at a sampling rate of 1200 Hz. Kinematic, EMG and GRF data were synchronized using the Motive 2.2.0 system (OptiTrack, Natural Point, Inc., Corvallis, OR, USA).

2.4 Data processing

The collected data were imported into Visual 3D (C-Motion, Inc., Germantown, MD, USA) to construct a full-body skeletal model for biomechanical analysis. Knee joint angles were defined as the orientation of the shank relative to the thigh, and ankle joint angles as the orientation of the foot relative to the shank. Joint angles for both the knee and ankle were calculated using an XYZ Cardan rotation sequence, corresponding to flexion/extension (X), abduction/adduction (Y), and internal/external rotation (Z). Joint moments were calculated via inverse dynamics and expressed in the coordinate system of the proximal segment. Joint angle and joint moment data were filtered using a low-pass filter (16 Hz, Butterworth filter, 4th order) [6], and GRF data were filtered using a low-pass filter (50 Hz, Butterworth filter, 4th order) [20]. The instant when vertical GRF first exceeded 10 N was defined as the initial ground contact [16]. As most lower limb injuries occur within the first 100 ms after initial contact [4, 6], the angular displacements (maximum joint angle during stance phase minus joint angle at initial ground contact) [21], as well as the peak external joint moments during this 100 ms window. These variables were calculated for knee and ankle joints in both the sagittal (flexion) and frontal (abduction) planes, and values were averaged across both lower limbs. Additionally, since muscle activation before landing represents a feedforward neuromuscular strategy for preparing the body to absorb impact, EMG signals were analyzed using a 200 ms time window spanning from 100 ms before to 100 ms after initial ground contact [16].

The extracted 200 ms EMG data were processed using MatlabR2021a (MathWorks, Natick, MA, USA). Raw EMG signals were band-pass filtered (20–450 Hz, Butterworth filter, 4th order) for filtering signal noise, full-wave rectified, and smoothed using a zero-lag low-pass filter (6Hz, Butterworth filter, 4th order) to obtain the linear envelope [22]. Each muscle's EMG data were then normalized to the peak activation observed during landing within the respective channel [22]. To assess the reproducibility of EMG recordings in this experiment, correlation analysis was performed on the filtered EMG data from the three valid landing trials under each aerial movement condition [16]. The two trials with the highest correlation ($r > 0.9$) were selected and averaged for further analysis of the CCI. The CCI for agonist and antagonist muscle pairs around the knee and ankle joints was calculated using the following formula [17, 23]:

$$CCI = \frac{2AB}{A + B} \times 100\%$$

Where A and B represent the EMG linear envelope areas of the agonist (RF or LG) and antagonist (BF or TA) muscles, respectively, and AB is the area of overlap between the two linear envelopes (Fig. 1). CCI values were calculated for two muscle pairs: RF and BF, and LG and TA, during both before and after landing phases.

2.5 Statistical analysis

All statistical analyses were performed using SPSS v.27 (IBM Corp., Armonk, NY, USA). To analyze the effects of the four aerial movement conditions on knee and ankle joint angular displacements and moments (flexion and abduction), as well as the CCI of the RF-BF and LG-TA muscles during landing, a linear mixed model (LMM) was applied. The fixed effect was the aerial movement condition (four levels: NC, LULC, RULC, BULC), and participant was included as a random intercept to account for within-subject variability. The significance level was set at $\alpha = 0.05$. When significant main effects were detected ($p < 0.05$), *post hoc* pairwise comparisons were conducted using Bonferroni correction. Effect sizes were reported using both partial eta squared (η^2) and Cohen's d. The thresholds for η^2 were interpreted as small ($\eta^2 \geq 0.01$), medium ($\eta^2 \geq 0.06$) and large ($\eta^2 \geq 0.14$) effects. Cohen's d was interpreted as trivial ($d < 0.2$), small ($0.2 \leq d < 0.5$), medium ($0.5 \leq d < 0.8$) and large ($d \geq 0.8$) Prior to analysis, the normality of residuals was verified for all models using the Shapiro-Wilk test ($p > 0.05$).

3. Results

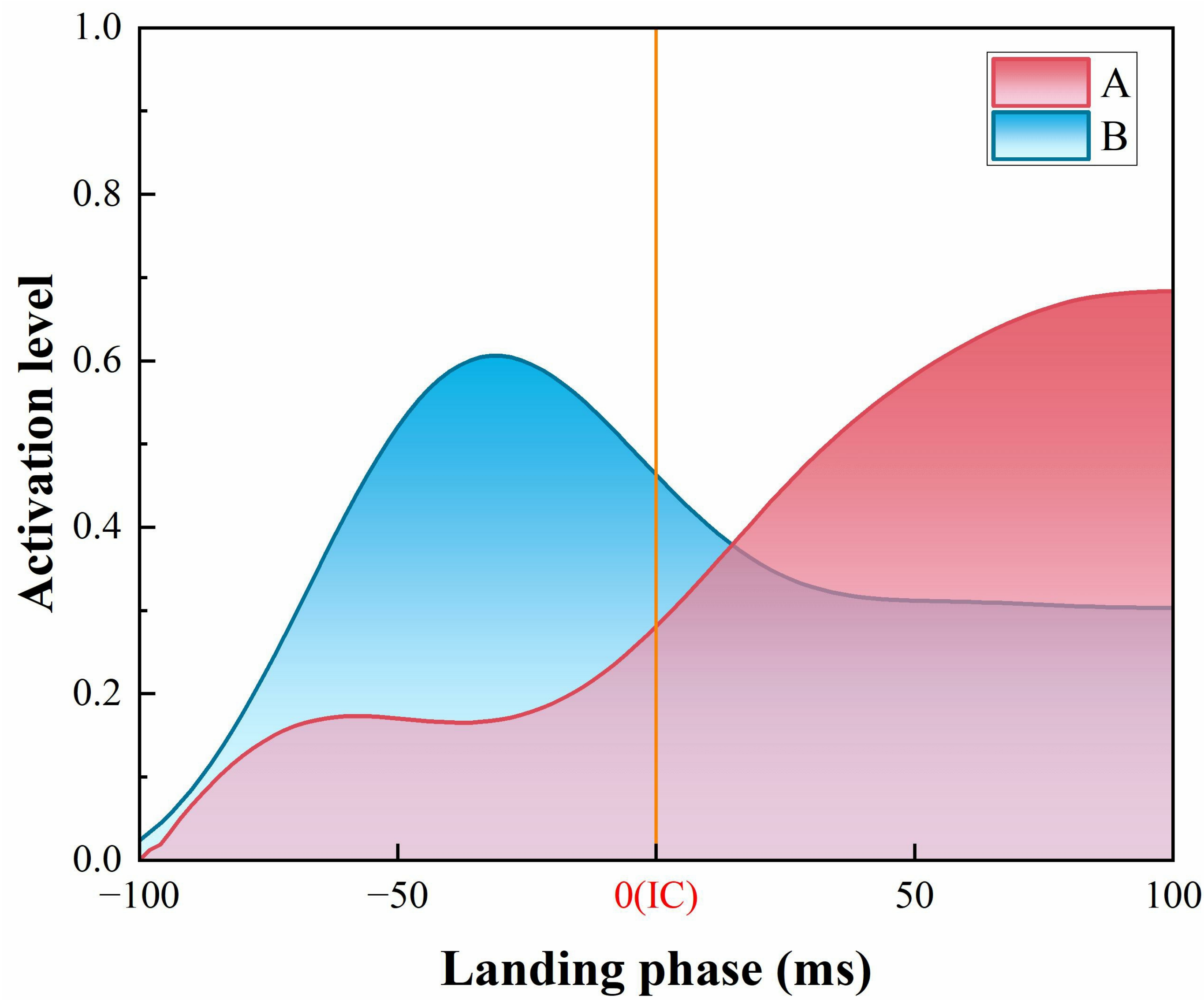

FIGURE 1. Schematic diagram of muscle co-contraction index calculation. A: EMG linear envelope areas of the agonist muscle; B: EMG linear envelope areas of the antagonist muscle. IC: Initial ground contact. Time 0 ms indicates initial contact.

3.1 Joint angular displacements

Significant differences were found in the angular displacement of the lower limb joints during landing among the four aerial upper limb movements (knee flexion $p < 0.001$, $\eta^2 = 0.391$; knee abduction $p = 0.001$, $\eta^2 = 0.315$; ankle dorsiflexion $p < 0.001$, $\eta^2 = 0.400$; ankle abduction $p < 0.001$, $\eta^2 = 0.385$) (Table 1). *Post-hoc* analysis showed that (Fig. 2) in knee flexion, the angular displacement of the aerial RULC ($p = 0.004$, $d = -1.11$, Confidential Interval (CI) (−1.44, −0.78)), LULC ($p < 0.001$, $d = -1.52$, CI (−1.85, −1.19)) and BULC ($p = 0.013$, $d = -0.97$, CI (−1.30, −0.64)) movements was lower than that of the aerial NC movement. In addition, in knee abduction, the angular displacement of the LULC movement was significantly greater than that of the aerial NC ($p = 0.002$, $d = -1.19$, CI (−1.52, −0.86)) and RULC ($p = 0.018$, $d = -0.93$, CI (−1.26, −0.60)) movements. In ankle dorsiflexion, the angular displacement of the aerial RULC ($p = 0.002$, $d = -1.21$, CI (−1.54, −0.88)), LULC ($p = 0.004$, $d = -1.13$, CI (−1.46, −0.80)) and BULC ($p < 0.001$, $d = -1.50$, CI (−1.83, −1.17)) movements was higher compared to the aerial NC movement. For ankle abduction, the angular displacement of the aerial RULC ($p < 0.001$, $d = -1.44$, CI (−1.77, −1.11)), LULC ($p < 0.001$, $d = -1.51$, CI (−1.84, −1.18)), and BULC ($p = 0.033$, $d = -0.84$, CI (−1.17, −0.51)) movements was lower compared to the aerial NC movement (Fig. 2).

3.2 Joint moments

There were also significant differences in the peak joint moments of the knee and ankle after landing among the four aerial movements (knee abduction $p = 0.001$, $\eta^2 = 0.321$; ankle dorsiflexion $p < 0.001$, $\eta^2 = 0.357$; ankle abduction $p = 0.046$, $\eta^2 = 0.171$) (Table 1). *Post-hoc* analysis showed that (Fig. 3) knee abduction moments were lower in the aerial NC ($p = 0.001$, $d = -1.26$, CI (−0.19, −0.06)), RULC ($p = 0.039$, $d = -0.82$, CI (−0.15, −0.01)) and BULC ($p = 0.007$, $d = -1.03$, CI (−0.13, −0.02)) movements compared to the aerial LULC movement. And the ankle dorsiflexion moments were significantly higher in the aerial BULC ($p = 0.011$, $d = 0.99$, CI (0.10, 0.95)) and LULC ($p = 0.032$, $d = 0.85$, CI (0.09, 0.94))

TABLE 1. ANOVA results for knee and ankle Angular displacement, joint moments and muscle ICCs (n = 15).

	Aerial Upper Limb Movements				Results of ANOVA	
	NC	BULC	RULC	LULC	p	η^2
Angular displacement of joints (°)						
Knee flexion	$51.91 \pm 4.20^{b,c,d}$	48.46 ± 6.25^{a}	48.09 ± 6.45^{a}	47.13 ± 4.87^{a}	<0.001**	0.391
Knee abduction	1.02 ± 0.68^{d}	0.91 ± 0.69	0.83 ± 0.78^{d}	$1.51 \pm 0.76^{a,c}$	0.001**	0.315
Ankle dorsiflexion	$38.74 \pm 3.60^{b,c,d}$	45.21 ± 3.85^{a}	43.71 ± 4.72^{a}	44.35 ± 5.56^{a}	<0.001**	0.400
Ankle abduction	$2.82 \pm 1.72^{b,c,d}$	2.13 ± 1.60^{a}	2.03 ± 1.56^{a}	1.90 ± 1.74^{a}	<0.001**	0.385
Peak joint moments (N·m/BW)						
Knee flexion	2.21 ± 0.43	2.12 ± 0.44	2.29 ± 0.35	2.14 ± 0.29	0.312	0.080
Knee abduction	0.13 ± 0.07^{d}	0.16 ± 0.06^{d}	0.15 ± 0.13^{d}	$0.22 \pm 0.09^{a,b,c}$	0.001**	0.321
Ankle dorsiflexion	$1.19 \pm 0.38^{b,d}$	1.71 ± 0.59^{a}	1.46 ± 0.41	1.70 ± 0.55^{a}	<0.001**	0.357
Ankle abduction	0.22 ± 0.08^{d}	0.24 ± 0.10	0.25 ± 0.09	0.26 ± 0.11^{a}	0.046*	0.171
CCI before landing (%)						
RF-BF	$52.99 \pm 10.05^{b,c,d}$	60.27 ± 8.56^{a}	64.27 ± 14.90^{a}	63.61 ± 11.25^{a}	0.001**	0.748
LG-TA	$57.64 \pm 11.65^{b,c,d}$	68.67 ± 12.49^{a}	63.99 ± 14.86^{a}	68.43 ± 14.26^{a}	<0.001**	0.860
CCI after landing (%)						
RF-BF	72.94 ± 10.69^{d}	70.30 ± 10.25	73.97 ± 11.75^{d}	$65.50 \pm 7.46^{a,c}$	0.004	0.273
LG-TA	74.15 ± 13.10	74.69 ± 14.13	79.78 ± 13.73^{d}	70.46 ± 12.24^{c}	0.001**	0.755

*Mean ± standard deviation; CCI: co-contraction index; NC: no catching; BULC: bilateral upper limb catching; RULC: right upper limb catching; LULC: left upper limb catching; RF: rectus femoris; BF: biceps femoris; LG: lateral gastrocnemius; TA: tibialis anterior; BW: body weight; ANOVA: analysis of variance; ^{a}: differences from NC; ^{b}: differences from BULC; ^{c}: differences from RULC; ^{d}: differences from LULC; * is a significantly different ($p < 0.05$); ** is a very significantly different ($p < 0.01$).*

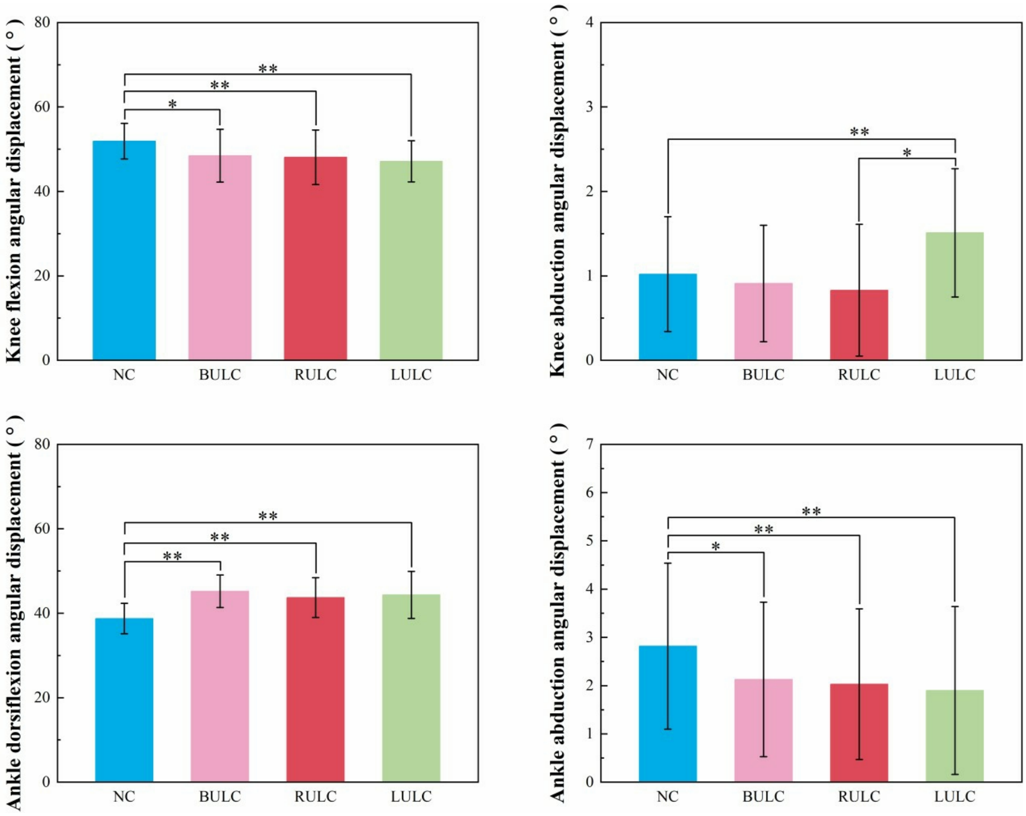

FIGURE 2. Knee and ankle joint angle displacement after landing. NC: no catching; BULC: bilateral upper limb catching; RULC: right upper limb catching; LULC: left upper limb catching; * is a significantly different ($p < 0.05$); ** is a very significantly different ($p < 0.01$).

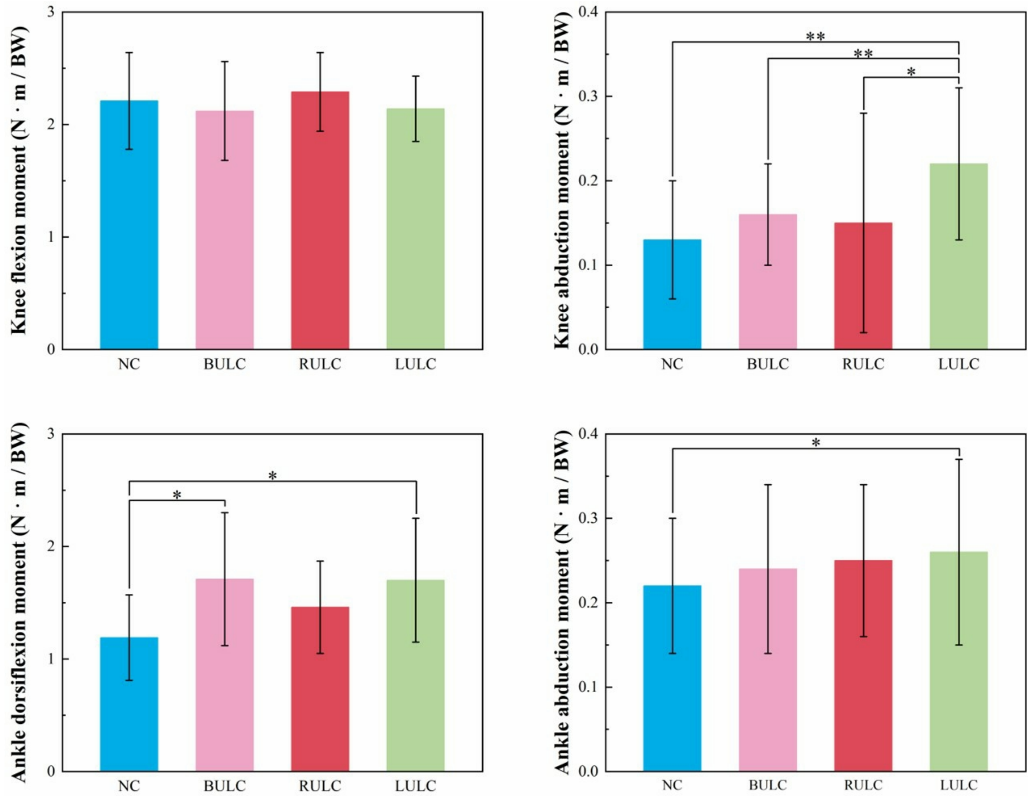

FIGURE 3. Knee and ankle joint moment after landing. NC: no catching; BULC: bilateral upper limb catching; RULC: right upper limb catching; LULC: left upper limb catching; BW: body weight; * is a significantly different ($p < 0.05$); ** is a very significantly different ($p < 0.01$).

movements compared to the aerial NC movement, the ankle abduction were significantly higher in the aerial LULC (p = 0.014, d = 0.96, CI (0.02, 0.08)) compared to the aerial NC movement (Fig. 3).

3.3 CCI of the knee and ankle joint muscles

Significant differences in CCI of the knee and ankle joint muscles were observed among the four aerial movements during the before landing (100 ms prior to initial contact) (RF and BF: $p = 0.001$, $\eta^2 = 0.748$; LG and TA: $p < 0.001$, $\eta^2 = 0.860$) (Table 1). *Post-hoc* analysis revealed that (Fig. 4) the CCI of the RF and BF was significantly higher in the aerial RULC (p = 0.018, d = 0.93, CI (0.35, 4.57)), LULC (p = 0.033, d = 0.85, CI (0.32, 4.28)) and BULC (p = 0.042, d = 0.81, CI (0.14, 3.60)) movements compared to aerial NC movement. Similarly, the CCI of the LG and TA was significantly higher in all aerial catching movements compared to aerial NC, including aerial RULC (p = 0.025, d = 0.88, CI (0.27, 3.99)), LULC (p = 0.004, d = 1.12, CI (0.77, 4.56)) and BULC (p = 0.015, d = 0.95, CI (0.58, 4.89)) movements (Fig. 4).

Significant differences were also found in the CCI of knee and ankle joint muscles among the four aerial movements during the after landing (100 ms following initial contact) (RF and BF: $p = 0.004$, $\eta^2 = 0.273$; LG and TA: $p < 0.001$, η^2 = 0.755). *Post-hoc* analysis indicated that the CCI of the RF and BF was significantly higher in the aerial RULC (p = 0.011, d = 0.99, CI (0.38, 4.23)) and NC (p = 0.037, d = 0.83, CI (0.23, 4.18)) movements compared to the aerial LULC movement. Additionally, the CCI of the LG and TA in the aerial RULC condition was significantly higher than in the aerial LULC (p < 0.001, d = 1.55, CI (0.65, 6.43)) movement.

4. Discussion

This study investigated the effects of various aerial catching movements performed following take-off on the CCI of the knee and ankle muscles, as well as on joint angular displacements and joint moments during landing. The results indicate that, compared to the aerial NC movement after takeoff, all aerial catching movements led to increased CCI of the RF and BF, and CCI of the LG and TA before landing. Additionally, there was a smaller knee flexion angular displacements and larger ankle dorsiflexion angular displacements, as well as increased flexion and abduction moments after landing. Furthermore, the aerial LULC movement led to a decrease in CCI of the RF and BF after landing, along with a greater knee abduction angular displacement and moment. These results validate our hypothesis.

The reduction in knee flexion angular displacement limits the ability to dissipate ground reaction forces through eccentric muscle action, thereby increasing the load on the knee joint and decreasing shock absorption capacity [24]. A stiffer landing strategy, characterized by limited knee flexion, has been identified as a risk factor for ACL injury [4, 6]. The insertion angle of the patellar tendon relative to the longitudinal axis of the tibia decreases with increasing knee flexion displacement [25]. The insertion angle of the patellar tendon has a large influence on the tibial shear force because the anterior component of the

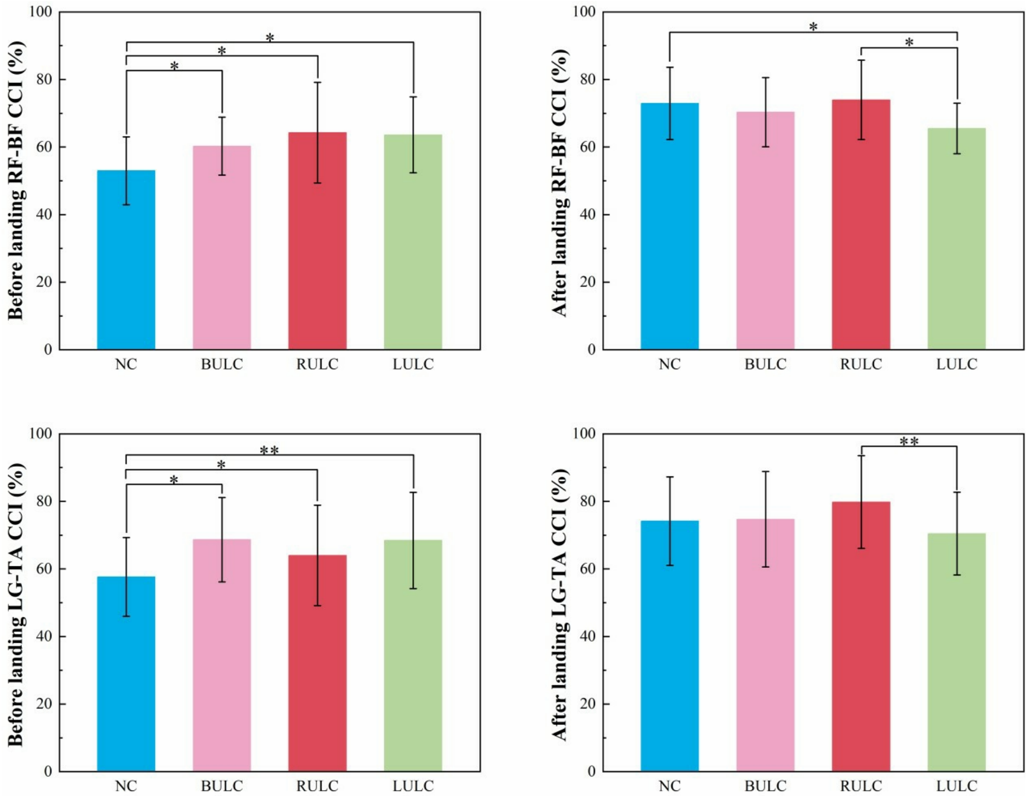

FIGURE 4. Knee and ankle joint muscles CCI. CCI: co-contraction index; NC: no catching; BULC: bilateral upper limb catching; RULC: right upper limb catching; LULC: left upper limb catching; RF: rectus femoris; BF: biceps femoris; LG: lateral gastrocnemius; TA: tibialis anterior; * is a significantly different ($p < 0.05$); ** is a very significantly different ($p < 0.01$).

quadriceps and patellar tendon forces is a multiple of the sine of the patellar tendon insertion angle [26]. During landing braking, a slight flexion of the knee, taking into account the forces of the quadriceps and patellar tendon, will increase the anterior tibial shear force, which in turn will increase the load on the cruciate ligaments and increase the risk of injury [24–26]. This study shows that all aerial catching movements led to a reduction in knee flexion angular displacements during landing compared to NC movement (as shown in Fig. 2), suggesting that all aerial catching movements reduce the ability of the knee joint to absorb shock during landing. We also observed an increase in CCI of the RF and BF (as shown in Fig. 4), before landing during all aerial catching movements. Previous studies have found that excessive co-contraction of the RF and BF during the flight phase reduces knee flexion angle during the landing phase [17]. This result is similar to ours, so the reduction in knee flexion angular displacements during landing in all aerial catching movements may be due to the increase in CCI of the RF and BF before landing. This may reflect a dual-task interference effect. Performing upper limb tasks during the aerial phase may increase cognitive and neuromotor demands, thereby reducing the attentional resources allocated to lower limb impact preparation. This redistribution of focus may impair proprioceptive acuity in the lower extremities prior to landing, potentially leading to an unconscious increase in muscle co-contraction as a compensatory strategy to prepare for ground contact. Moreover, decreased proprioception can increase the risk of injury [27].

Notably, the aerial LULC movement after takeoff led to a decrease in CCI of the RF and BF after landing (as shown in Fig. 4), with an increase in knee abduction angular displacements and moment (as shown in Fig. 3). The co-contraction between the quadriceps and hamstrings is related to knee joint stability [13]. Many studies on ACL injury risk assessment have shown that an increase in knee abduction angular displacements and moment is closely related to increased load on the ACL, leading to a higher risk of ACL injury. Therefore, the aerial LULC movement may not only reduce the sagittal plane mobility and knee shock absorption ability [24], but also increase knee joint motion in the frontal plane after landing, which may indicate a decrease in stability. The reduction in CCI of the RF and BF may further suggest that knee joint stability is compromised [13]. According to the logic of the CCI formula, the decrease in CCI of the RF and BF is due to the reduction in activation of the less active muscle (hamstrings). A greater hamstring co-contraction can limit anterior tibial translation and increase joint stability [12, 13]. Insufficient hamstring contraction is a primary mechanism for anterior tibial translation, leading to increased load on the ACL [28], which increases the risk of ACL injury, putting the knee in a vulnerable state during aerial LULC landing. Studies have shown that preventive strategies such as neuromuscular training and targeted hamstring strengthening particularly through eccentric exercises like the Nordic hamstring exercise—can enhance eccentric muscle strength and improve the hamstring-to-quadriceps strength ratio [29, 30]. Implementing these strategies may help mitigate the potential destabilizing effects observed under conditions such as the aerial LULC task.

Moreover, all aerial catching movements led to an increase in CCI of the LG and TA before landing (as shown in Fig. 4), with an increase in ankle dorsiflexion angular displacements and flexion moment after landing (as shown in Fig. 2), while ankle inversion angular displacements decreased and abduction moment increased (as shown in Fig. 3). This indicates that more ankle joint involvement was used to decelerate the landing [31]. According to the conservation of angular momentum, upper limb motion can influence the total body angular momentum [32], thereby affecting lower limb landing mechanics. Previous research has also shown that when proximal joint control is compromised, the neuromuscular system typically redistributes control to distal joints, particularly the ankle joint [33], to restore postural balance. This joint-specific response pattern may represent a neuromechanical compensation aimed at counteracting instability caused by upper body motion. However, the increase in ankle dorsiflexion moment may result in an impact on the knee through the tibia, and increased ankle dorsiflexion is also a related factor for ACL injury [27, 34]. Additionally, we observed that after performing the aerial LULC movement after takeoff, CCI of the LG and TA after landing was lower compared to the aerial RULC movement. This may be because all EMG electrodes in our experiment were placed on the right leg. The unilateral movements may lead to an imbalance in the landing mechanics of both lower limbs, which could be the cause of the CCI of the LG and TA difference between the aerial LULC and RULC movements after landing.

This study has several limitations. First, our sample consisted only of healthy young male athletes, which may limit the generalizability of the findings. Previous research has shown that, compared to males, female athletes are at greater risk of injury during athletic activities [35, 36], and may be more susceptible to the effects of aerial upper limb movements. However, male athletes typically achieve greater jump heights [4, 6], which may result in higher mechanical loads on the lower limbs during landing and potentially amplify the influence of upper limb movements in the air. Additionally, EMG data in this study were collected only from the dominant leg, and the potential asymmetric loading caused by unilateral catching movements was not considered. Second, although we controlled for fatigue and standardized movement instructions, individual differences in jumping strategies and upper limb coordination may still have introduced variability.

5. Conclusions

Aerial catching movements lead to higher co-contraction of the knee and ankle muscles in the lower limbs before landing, and a lower knee flexion angular displacement, which may increase the risk of knee injury during landing. Moreover, the increase in ankle dorsiflexion angular displacement after landing may serve as a compensation for the reduced knee joint flexion angular displacement, likely as a neuromuscular protective mechanism. However, the aerial LULC movement resulted in decreased knee joint muscle co-contraction after landing, along with increased frontal plane motion of the knee, indicating a diminished knee joint stability and potentially increasing the risk of knee injury. These findings may provide preliminary insights for training considerations. Coaches and practitioners might consider paying attention to the influence of upper limb movements during jump-landing tasks. Integrating upper limb coordination drills or asymmetric movement tasks particularly involving the non-dominant upper limb could potentially contribute to improving sensorimotor control and joint stability.

AVAILABILITY OF DATA AND MATERIALS

The data presented in this study are available on reasonable request from the corresponding author.

AUTHOR CONTRIBUTIONS

TW—Conceptualization; Investigation; Methodology; Validation; Formal analysis; Writing–Original Draft; Project administration. YK, SW and PLF—Investigation; Methodology. YK, PLF and SK—Conceptualization; Methodology; Resources; Supervision; Visualization; Writing–Original Draft; Writing–Review. All authors read and approved the final version of the manuscript.

ETHICS APPROVAL AND CONSENT TO PARTICIPATE

This study complies with the Declaration of Helsinki. The study protocol was explained to all participants, and they signed a formal informed consent form from. The protocol of this study was approved by the Jeonbuk National University Institutional Review Board and was performed in accordance with relevant guidelines/regulations (JBNU2022-04-008-002, 01 April 2022).

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CONFLICT OF INTEREST

The authors declare no conflict of interest. Sukwon Kim is serving as one of the Guest editors of this journal. We declare that Sukwon Kim had no involvement in the peer review of this article and has no access to information regarding its peer review. Full responsibility for the editorial process for this article was delegated to RFSO.

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