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Ankle Biomechanics and Muscle Activation During Repeated Single-Leg Lateral Hopping: Effects of Speed and Distance

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Abstract

Objective: This study investigated the effects of hopping speed and distance on ankle joint moments, power, and muscle activation during repeated lateral hopping tasks. **Methods:** Twelve trained male athletes involved in team sports performed repeated lateral hop tasks at varying speeds (slow and fast) and hop widths (30 and 45 cm). Kinematics, kinetics, and muscle activation were recorded using ten optoelectronic cameras, two force platforms, and four channels of electromyography at the peroneus longus, tibialis anterior, lateral gastrocnemius, and soleus muscles. A two-way repeated-measures ANOVA assessed the effects of speed and distance, and secondary outcomes were adjusted for multiple comparison using the false discover rate procedure. **Result:** Hopping speed significantly influenced ankle joint moment ($p < 0.05$) and produced selective altered joint power and muscle activation ($p < 0.05$), while distance affected ankle joint moment and range of motion in the frontal plane ($p < 0.05$), but not muscle activation. Several of these effects remained significant after correction for multiple comparisons. A significant speed × distance interaction was observed for frontal plane peak ankle moment ($p < 0.05$). **Conclusions:** Hopping speed modulates ankle joint moment most consistently, with more variable changes in joint power and relative muscle activation, while distance manipulation selectively affects mechanical loading without altering muscle activation. Faster hopping increased ankle joint loading and altered relative muscle activation, while wider hopping distances could progressively load frontal plane structures. The results may inform progressive loading strategies for multidirectional sport training in healthy athletes, though clinical applicability remains to be confirmed.

Keywords: ankle biomechanics; muscle activation; lateral hop; progressive load

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1. Introduction

Lateral movements such as cutting, side-stepping, and rapid directional changes are essential fundamental movements that reflect sports performance in field and court sports, with 300–700 directional changes per game in football [1,2]. These movements place substantial mechanical and neuromuscular demands on the lower extremity, especially the ankle complex, and are frequently associated with injury during rapid changes in direction or landing tasks [3]. In particular, dorsiflexion range of motion directly influences force absorption and postural control during landing, with restricted mobility leading to altered landing kinematics and reduced landing stability [4]. The functional side hop test was developed to evaluate lateral movement skills by performing repeated medial and lateral hops over a set distance [5], which requires rapid acceleration, deceleration, and changes in direction that are typical of multidirectional sports movements. This test is used in athletic contexts to assess sport-specific movement quality [6], reflecting its relevance to performance in field and court sports. Given the importance of lateral movement proficiency for both performance and injury prevention, developing training strategies that address the explosive and plyometric nature of these movements is essential.

Plyometric training improves dynamic movement control and athletic performance by exploiting the stretch-shortening cycle of muscles [7]. During this cycle, muscles rapidly stretch and then contract to produce greater force and power. Exercises such as jumping, bounding, and hopping utilise this mechanism effectively [8]. The lateral hop stabilisation drill, or lateral hop exercise, is a fundamental plyometric movement often incorporated into training and rehabilitation programmes to enhance neuromuscular control, movement strategy, and mechanical loading capacity during multidirectional sports movements in both injured and healthy populations [9–11]. This exercise involves repeated single-leg hops in medial and lateral directions, with emphasis on controlled force transitions and progressive balance training that gradually advances in intensity, contrasting with the explosive effort required in the functional side-hop test. Importantly, the lateral specificity is critical, as Asadi et al. (2016) revealed that vertical plyometric exercises do not improve change in direction ability, whereas lateral-specific exercises (lateral bounds, side hops) effectively enhance multidirectional performance [12]. For example, McCormick et al. (2015) reported that basketball players demonstrated faster lateral movements after participating in a plyometric program that included the repeated lateral hop exercise [10]. Similarly, participants showed improvements in repeated counter-movement jump performance, characterised by shorter landing times and greater knee flexion range during landing, following a program that incorporated related repeated lateral hops in healthy individuals [11]. Moreover, comparable benefits have also been observed in injured populations, with greater ankle dorsiflexion and ankle muscle activation during landing after completing hop-stabilisation programmes [9,13]. These findings highlight the role of plyometric exercises in enhancing neuromuscular coordination and lower limb control. However, the effectiveness of plyometric training depends on the design and progression of an appropriate program. Selecting the right intensity and progression strategy is crucial for driving adaptation while avoiding excessive loading. Understanding progressive overload principles is, therefore, essential for optimising training outcomes.

Progressive loading is a key principle in rehabilitation and performance training, increasing the physical demands through intensity, frequency, or duration modification to facilitate adaptation and restore function [14,15]. Progression strategies are tailored to specific goals such as strength, stability, or movement coordination [16]. For rapid movement performance, speed-based progressions, such as reducing ground contact time, enhance reactive strength and dynamic movement control during the task [17]. McCormick et al. (2015) progressively increased training intensity by reducing ground contact time, leading to faster lateral shuffle and longer lateral hop distance in healthy basketball players [10]. While focused on rehabilitation, Ardakani et al. (2019) demonstrated similar benefits of movement speed progression in athletes with chronic ankle instability, advancing from self-selected speed to 120 ground contacts per minute (2 Hz) during their exercise programme, resulting in improved landing strategy [9]. However, these studies focus on functional performance outcomes such as movement speed and landing strategy without quantifying the underlying changes in joint kinetics, kinematics, or muscle activation. As a result, it remains unclear how speed progression alters lower limb mechanical loading during lateral movement tasks. In addition to speed, distance is another key used to progress exercise intensity, which requires greater force production, stability, and neuromuscular coordination. Previous studies have employed incremental distance progression, such as drop and depth jump, typically starting at 20–30 cm and progressing by a similar magnitude, particularly in vertical plyometric tasks such as drop- and depth-jump variations [18,19]. Progressing to greater landing height increases loading mechanics as reflected by higher peak lower extremity joint angle and joint moment in the sagittal plane during landing [20,21], together with increased tibialis anterior and lateral gastrocnemius activation [22]. However, distance progression in horizontal or lateral direction has not been examined, and it remains unclear whether similar mechanical consequences occur when distance increases laterally. Moreover, lateral tasks have different mechanical demands that are not fully captured by vertical landing assessment, due to greater medio-lateral control and landing stabilisation requirements. While both speed and distance are used as progression tools in practice, no study has yet examined their biomechanical effects within the same lateral hopping protocol, making it unclear whether they produce distinct or overlapping mechanical demands. This gap highlights the need to examine how speed- and distance-based progressions independently influence the biomechanical demands of lateral movement tasks in order to inform more effective training strategies.

The analysis of biomechanical jump landing tasks is inherently complex; however, it plays a critical role in enhancing sports performance, mitigating injury risks, and informing and supporting rehabilitation strategies [23,24]. Among various landing strategies, lateral landings are particularly relevant in multidirectional sports and have been extensively studied in terms of both kinematics and kinetics, joint moments and joint power [24,25]. In lateral landing tasks, ankle joint moments measure the rotational demands on the joint, while joint power indicates the musculature's ability to absorb and produce force, both crucial for dynamic stability and movement efficiency [25]. The ankle complex was selected as the primary focus given the large and rapid impulse loads it experiences during unilateral landing tasks, making it the joint most susceptible to mechanical demands during lateral movements [26]. Although several studies have investigated lateral hop tasks in athletic populations using functional side hop tests to analyse joint kinematics, joint moments, and joint power under maximal effort conditions [5,25], none have examined how systematic manipulation of speed or distance influences ankle biomechanics during these tasks. Existing evidence based on loading progression in plyometric tasks is largely based on vertical jumps and drop jumps, which may not translate directly to lateral movement. To our knowledge, no studies to date have examined such movements in the context of training or rehabilitation, where the manipulation of speed or distance is used to increase loading intensity progressively. Thus, this study aimed to investigate how hopping speed and distance affect ankle joint moments, power, and muscle activation during repeated lateral hopping tasks. We hypothesised that increased task demands through greater hopping speed and wider distance would result in progressively higher ankle joint kinetics (moments and power), particularly in the frontal plane, and ankle muscle activation during the lateral landing and take-off phase. No specific interaction between speed and distance was hypothesised a priori.

2. Materials and Methods

2.1. Participants

Twelve male sportspeople (age 24 ± 4 yrs, weight 78.3 ± 13.0 kg, height 1.82 ± 0.09 m) volunteered for this study. The sample size was calculated using G*Power analysis (v.3.1.9.7, Heinrich-Heine-Universität Düsseldorf, Germany), based on an effect size of 0.65 (η_p^2 = 0.3) derived from a two-way interaction for peak ankle joint moment during a lateral cutting task [27], as no prior study had reported a speed x distance interaction for repeated lateral hopping. With an alpha level (α) of 0.05 and a power of $1 - \beta$ set at 0.95, the minimum required sample was 7; twelve were recruited to allow a margin for dropout. Only male participants were recruited to minimise variability attributable to sex-related differences in hormonal status, joint laxity, and neuromuscular control, which may confound biomechanical comparisons across participants [28]. All participants were classified as trained (Tier 2) according to the Participant Classification Framework [29], based on their regular involvement in organised sport training and competition. Participants were required to actively engage in team sports (e.g., soccer, basketball, rugby) for at least three sessions per week, with each session lasting a minimum of one hour. Additionally, participants needed to be free from any musculoskeletal lower-limb injuries for six months before participation. Participants received complete information regarding the study's objectives, processes, and possible risks prior to giving their written informed consent to take part. All procedures were approved by the University Research Ethics Committee of Liverpool John Moores University (No. 24/SPS/051, date of approval: 19 September 2024) and the Mahidol University Institutional Review Board (COA No. MU-MOU 2025/098.3103, date of approval: 31 March 2025) and conducted in accordance with the Declaration of Helsinki.

2.2. Experiment Protocol

The Liverpool John Moore University lower limb trunk model utilised 44 retroreflective markers placed on the participant's anatomical landmarks [30]. This 6DOF model features functional hip joints and knee axes. The model consisted of trunk, pelvis, thigh, shank and foot segments. Furthermore, wireless surface EMG electrodes were placed on the dominant leg, targeting the tibialis anterior, peroneus longus, gastrocnemius lateralis, and soleus, in accordance with SENIAM recommendations [31].

Kinematic and kinetic data were collected using the Qualisys Track Manager (QTM, v2024.1, Qualisys, Gothenburg, Sweden), which synchronised ten motion capture cameras (Arqus A12, Qualisys, Sweden), two force platforms (Kistler Instrument AG, Winterthur, Switzerland), and a four-channel surface electromyography (EMG) system (Noraxon DTS, Noraxon, Scottsdale, AZ, USA). Kinematic data were captured at a sampling rate of 250 Hz, while kinetic and EMG data were recorded at 1500 Hz. The force platforms were used to identify ground contact and toe-off during the landing phase.

Upon arriving at the biomechanics laboratory, participants engaged in a ten-minute warm-up routine that consisted of mobility exercises, dynamic stretching, and dynamic movement activities targeting the lower extremities. Following this, participants completed three trials of four tasks, each involving ten repeated unilateral lateral hops (20 hops total, alternating medial-to-lateral and lateral-to-medial), in a random order: (1) hopping at a speed of 60 beats per minute (bpm) with a distance of 30 cm between the hopping areas, (2) hopping at a speed of 120 bpm with a distance of 30 cm between the areas, (3) hopping at a speed of 60 bpm with a distance of 45 cm between the areas, and (4) hopping at a speed of 120 bpm with a distance of 45 cm between the areas. The faster hopping speed of 120 bpm was adopted from a hop-stabilisation training protocol used in athletes with chronic ankle instability [9], while 60 bpm represented a 50% reduction to establish a controlled slow condition. The 30 cm distance was selected based on its use in the functional side hop test [25,32] and drop jump training protocol [33], with 45 cm (an increase of 15 cm, or 50% above the baseline) selected to represent a progressive increase in hopping distance.

All participants received guidance on the necessary repeated unilateral side hops and were subsequently allowed to practice and familiarise themselves with the movements, including maintaining pace with the metronome cadence for each condition. They were instructed to stand on a force platform and then jump from side to side for a lateral hop between two parallel lines. Each participant was required to perform ten repeated single-leg lateral hops using their dominant leg (preferred leg for kicking a ball) [25] (Figure 1), with each hop initiated immediately upon the metronome beat. Each metronome beat served as a signal to hop, alternating between a medial-to-lateral hop and a lateral-to-medial hop on consecutive beats, with participants landing and stabilising before hopping back on the subsequent beat, such that two consecutive beats (one medial-to-lateral and one lateral-to-medial hop) represented one complete cycle (20 hops total per trial). Participants completed three successful trials per condition, with one minute of rest between trials to minimise fatigue. Successful trials were recorded if participants could achieve the 10-jump repetitions without their hopping leg stepping on the line or their untested leg touching the floor. However, if there were any unsuccessful trials, the tests would be discarded, and the participant would repeat the task.

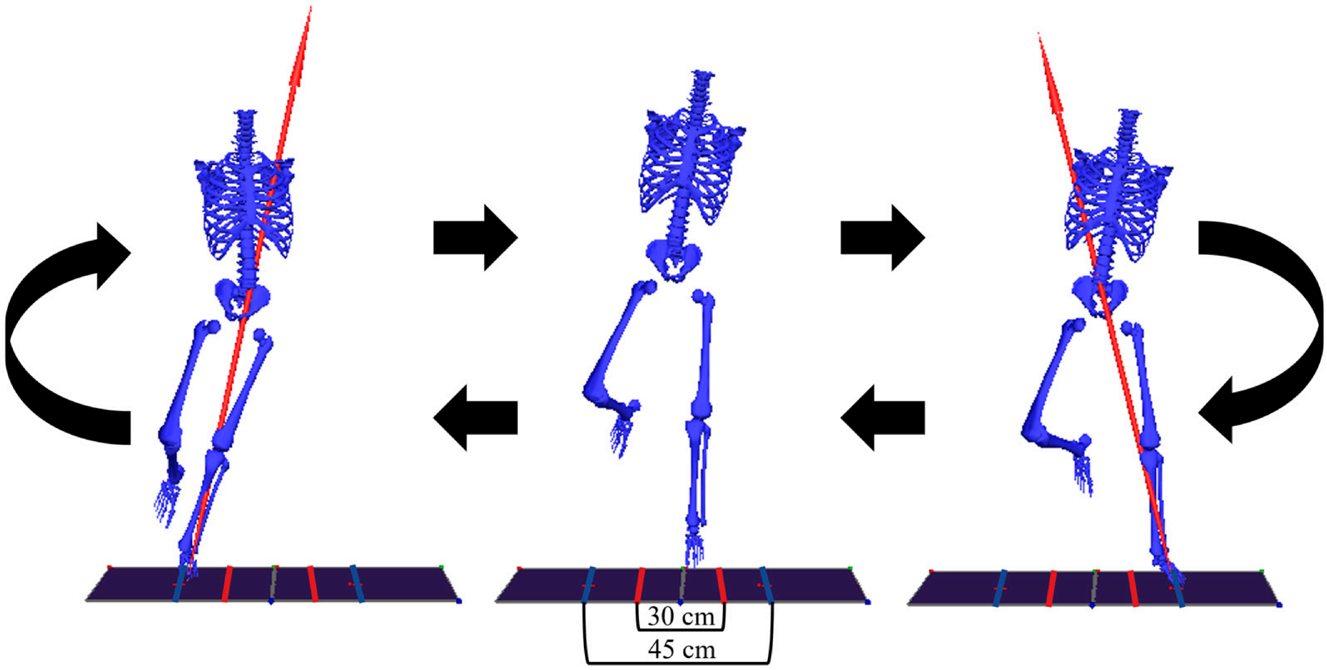

Figure 1. Sequence of repeated single-leg lateral hops. The lateral hop distances between landing locations were set at 30 cm (red lines) and 45 cm (blue lines), defined relative to a central reference line. Red arrows represent the visualised vertical ground reaction force.

2.3. Data Processing

All data were processed using Visual 3D (v2024.10.11, HAS-Motion Inc., Kingston, ON, Canada), and only lateral contacts were analysed, as lateral landings are more frequently associated with ankle injuries and impose greater neuromuscular and mechanical demand on the ankle complex [3,5]. The initial contact and toe-off for each task were defined as the moments when the vertical ground reaction force exceeded 20 N and subsequently fell below that threshold [25]. Maximum knee flexion was used to delineate the end of the landing phase and the beginning of the take-off phase. Landing contact time was calculated from initial contact to the maximum knee flexion, expressed in seconds (s).

The three-dimensional marker trajectories and three components of ground reaction force were filtered with a 20 Hz low-pass Butterworth filter. Ankle joint kinematics and kinetics were calculated in sagittal and frontal planes, representing dorsiflexion–plantarflexion and inversion–eversion, respectively, based on the joint coordinate system

(Figure 2). Positive values represent dorsiflexion and eversion, while negative values represent plantarflexion and inversion. Ankle range of motion and peak joint moment during the landing phase, and peak negative and positive ankle joint power during the landing and take-off phases, respectively, were extracted.

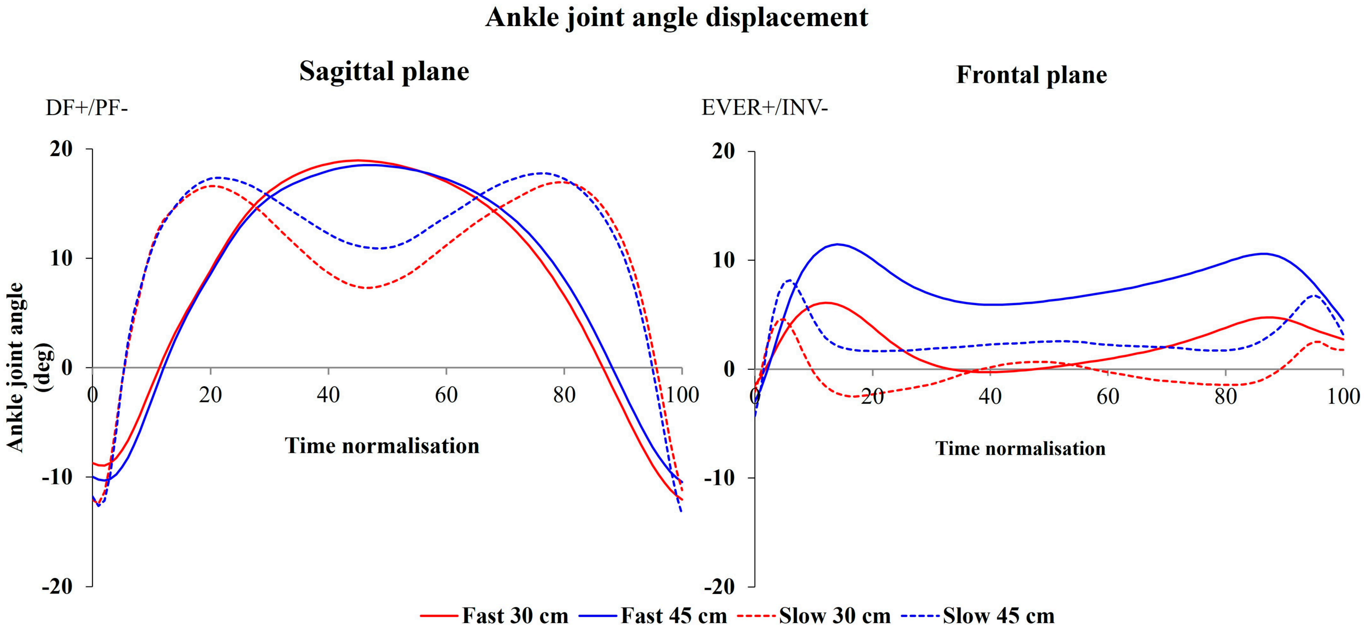

Figure 2. Ankle joint angle displacement in the sagittal (**left**) and frontal (**right**) planes during the lateral hop contact phase from one representative participant. Red = 30 cm; blue = 45 cm; dashed = slow; solid = fast. Positive values represent dorsiflexion/eversion; negative values represent plantarflexion/inversion.

The raw sEMG signals were band-pass filtered within a range of 15–500 Hz and processed using the root-mean-square method with a 150 ms smoothing window. For each participant and muscle, signals were normalised to the peak amplitude of each muscle across all speed x distance conditions [34], expressed as a percentage of this individual, muscle-specific peak (%peak task activation, distinct from %MVC). Peak muscle activation was then analysed during both the landing and take-off phases.

Each trial consisted of 10 complete cycles (20 hops: 10 medial-to-lateral and 10 lateral-to-medial). We analysed the lateral direction hop from 6 of 10 cycles, excluding the first two and last two cycles to minimise variability from movement initiation and fatigue, which may not accurately reflect steady-state hopping performance [5,25]. For each participant, data from these 6 cycles were averaged within each trial, and then trial-level means were averaged across three trials for statistical comparison.

2.4. Statistical Analysis

All statistical analyses were conducted using SPSS version 30.0 (IBM Inc., Armonk, NY, USA), and all data were presented as the mean and standard deviation (SD). Shapiro–Wilk tests were used to determine normality. Although some conditions showed mild deviation from normality, repeated-measures ANOVA is considered robust to mild violations of this assumption, particularly in balanced within-subject designs [35]. A two-way repeated measure ANOVA was used to evaluate the main effects of speed (slow vs. fast) and distance (30 cm vs. 45 cm) on joint kinematics, moment, and power, as well as muscle activation during the repeated lateral hop tasks. Peak ankle joint moment (eversion) in the frontal plane and peak ankle joint moment (dorsiflexion) in the sagittal plane were designated a priori as primary outcomes, reflecting ankle joint loading in the two planes primarily involved in lateral hopping. All remaining kinetic and kinematic variables were treated as secondary/exploratory outcomes, and *p*-values across these outcomes were adjusted for multiple comparisons using the Benjamini–Hochberg false discovery rate procedure [36]. When a significant interaction was observed, Bonferroni post hoc comparisons were conducted to identify the interaction effect, and simple main effects with 95% confidence intervals were estimated. Where post hoc comparisons alone were insufficient to characterise the interaction, a difference-in-differences contrast was calculated from individual difference scores. Effect sizes for the ANOVA were expressed as partial eta squared (η_p^2) to facilitate cross-study comparisons, interpreted as small (η_p^2 = 0.01), medium (η_p^2 = 0.06), and large (η_p^2 = 0.14) [37]. The level of statistical significance was set at $p < 0.05$ for primary outcomes, and Benjamini–Hochberg-adjusted significance was used for secondary outcomes.

3. Results

3.1. Kinetic and Kinematics

Table 1 presents the mean, standard deviation, and 95% confidence intervals for the kinetics and kinematics variables across different hopping speed and distance tasks. A significant interaction effect was observed only for the peak ankle joint moment in the frontal plane ($F_{1,11}$ = 8.64, $p < 0.05$, η_p^2 = 0.440 (Table 1). Post hoc comparisons with Bonferroni correction revealed that peak ankle joint moment (eversion) was significantly greater at 45 cm compared to 30 cm under both slow (mean difference = 0.102 Nm/kg, 95% CI [0.065, 0.139], $p < 0.001$) and fast (mean difference = 0.150 Nm/kg, 95% CI [0.090, 0.211], $p < 0.001$) hopping speeds. To directly test whether this simple effect of distance differed in magnitude between speeds, an interaction contrast compared the two simple effects. The distance effect was greater at fast speed than at slow speed, difference-in-differences = 0.048 Nm/kg, 95% CI [0.012, 0.084]. When examining speed effects, the faster hopping condition resulted in notably shorter ground contact times ($F_{1,11}$ = 83.28, $p < 0.05$, η_p^2 = 0.883). Participants also exhibited increased peak ankle moments during fast hops across both sagittal ($F_{1,11}$ = 84.03, $p < 0.05$, η_p^2 = 0.884) and frontal planes ($F_{1,11}$ = 6.38, $p < 0.05$, η_p^2 = 0.367). This speed-related increase also extended to joint power production, though findings were more variable across measures. In the sagittal plane, peak positive power increased with speed ($F_{1,11}$ = 25.63, $p < 0.05$, η_p^2 = 0.700) and remained significant after correction for multiple comparisons, whereas the increase in peak negative power ($F_{1,11}$ = 5.77, $p < 0.05$, η_p^2 = 0.344) did not remain significant after correction for multiple comparisons. In the frontal plane, increases in both peak negative ($F_{1,11}$ = 7.72, $p < 0.05$, η_p^2 = 0.412) and positive phases ($F_{1,11}$ = 7.39, $p < 0.05$, η_p^2 = 0.402) also did not remain significant after correction for multiple comparisons. Furthermore, distance manipulation between conditions (30 versus 45 cm) produced distinct biomechanical adaptations. Wider hops demanded increased ankle mobility in the frontal ($F_{1,11}$ = 43.11, $p < 0.05$, η_p^2 = 0.797) and sagittal planes ($F_{1,11}$ = 14.69, $p < 0.05$, η_p^2 = 0.572), alongside elevated peak moments in corresponding to sagittal ($F_{1,11}$ = 5.08, $p < 0.05$, η_p^2 = 0.316) and frontal ($F_{1,11}$ = 35.03, $p < 0.05$, η_p^2 = 0.761). The 45 cm condition was also associated with greater negative joint power generation in the frontal plane ($F_{1,11}$ = 6.15, $p < 0.05$, η_p^2 = 0.359), although this effect did not remain significant after correction for multiple comparisons.

3.2. Electromyography

Table 2 shows the mean, standard deviation, and 95% confidence interval for ankle muscle activation across different hopping-speed and distance tasks. There was no significant main effect of distance and no interaction effect between hopping speed and distance for any muscle activation (Table 2). However, hopping speed had a significant effect on activations during the landing and take-off phases. During landing, fast hopping resulted in greater activation of the soleus muscles ($F_{1,11}$ = 21.26, $p < 0.05$, η_p^2 = 0.659), while lower activation in the tibialis anterior ($F_{1,11}$ = 15.63, $p < 0.05$, η_p^2 = 0.587) compared to slower hopping during the landing phase, both of which remained significant after correction for multiple comparisons. Peroneus longus activation also appeared greater during fasting hopping ($F_{1,11}$ = 7.32, $p < 0.05$, η_p^2 = 0.399), but this did not remain significant after correction. During take-off, slower hopping was associated with greater tibialis anterior ($F_{1,11}$ = 6.36, $p < 0.05$, η_p^2 = 0.366), lateral gastrocnemius ($F_{1,11}$ = 6.88, $p < 0.05$, η_p^2 = 0.385), and soleus ($F_{1,11}$ = 6.93, $p < 0.05$, η_p^2 = 0.387), although none of these effects remain significant after correction for multiple comparisons.

Table 1. Landing contact time, ankle range of motion, peak ankle joint moment, and peak negative joint power, measured during the landing phase; and positive joint power, measured during the take-off phase; across different hopping speed and distance conditions. Data are presented as mean ± SD with 95% confidence intervals in parentheses.

Parameters	**Slow Speed**		**Fast Speed**		**Speed**	**Distance**	**Interaction**
	30 cm	**45 cm**	**30 cm**	**45 cm**	***p*-Value** ***p*-Valueadj** **(η_p^2)**	***p*-Value** ***p*-Valueadj** **(η_p^2)**	***p*-Value** ***p*-Valueadj** **(η_p^2)**
Landing contact time (s)	0.20 ± 0.02 (0.19, 0.22)	0.19 ± 0.02 (0.18, 0.21)	0.16 ± 0.01 (0.16, 0.17)	0.16 ± 0.01 (0.16, 0.17)	<0.001 * <0.001 * (0.883)	0.306 0.520 (0.095)	0.618 0.828 (0.023)
Frontal plane							
Ankle range of motion (deg)	7.46 ± 2.61 (5.81, 9.12)	12.62 ± 3.45 (10.42, 14.81)	8.06 ± 2.42 (6.52, 9.59)	14.16 ± 3.76 (11.770, 16.55)	0.069 0.207 (0.270)	<0.001 * <0.001 * (0.797)	0.239 0.467 (0.123)
Peak ankle joint moment (Nm/kg)	0.07 ± 0.10 (0.01, 0.13)	0.17 ± 0.10 (0.11, 0.23)	0.12 ± 0.18 (0.00, 0.23)	0.27 ± 0.21 (0.13, 0.40)	0.028 * - (0.367)	<0.001 * - (0.761)	0.013 - (0.440)
Peak positive joint power (W/kg)	0.18 ± 0.14 (0.09, 0.26)	0.15 ± 0.09 (0.09, 0.21)	0.24 ± 0.16 (0.14, 0.33)	0.23 ± 0.13 (0.15, 0.31)	0.020 * 0.098 (0.402)	0.716 0.107 (0.013)	0.368 0.591 (0.074)
Peak negative joint power (W/kg)	−0.12 ± 0.06 (−0.16, −0.08)	−0.17 ± 0.06 (−0.21, −0.13)	−0.16 ± 0.08 (−0.21, −0.11)	−0.21 ± 0.08 (−0.26, −0.16)	0.018 * 0.098 (0.412)	0.031 * 0.107 (0.359)	0.970 0.982 (0.000)
Sagittal plane							
Ankle range of motion (deg)	29.34 ± 5.74 (25.70, 33.00)	33.41 ± 4.93 (30.30, 36.50)	32.26 ± 5.18 (29.00, 35.60)	33.71 ± 6.92 (29.30, 38.10)	0.085 0.228 (0.246)	0.003 * 0.023 * (0.572)	0.086 0.228 (0.245)
Peak ankle joint moment (Nm/kg)	−2.28 ± 0.40 (−2.53, −2.02)	−2.03 ± 0.35 (−2.25, −1.81)	−2.97 ± 0.58 (−3.34, −2.61)	−2.82 ± 0.39 (−3.06, −2.57)	<0.01 * - (0.884)	0.046 - (0.316)	0.487 - (0.045)
Peak positive joint power (W/kg)	5.23 ± 1.04 (4.57, 5.90)	5.07 ± 1.16 (4.33, 5.80)	6.21 ± 1.29 (5.38, 7.03)	5.96 ± 1.25 (5.16, 6.75)	<0.001 * 0.005 * (0.700)	0.312 0.520 (0.093)	0.756 0.895 (0.009)
Peak negative joint power (W/kg)	−5.94 ± 1.35 (−6.80, −5.09)	−6.28 ± 0.96 (−6.89, −5.67)	−6.56 ± 1.75 (−7.67, −5.45)	−6.92 ± 1.3 (−7.74, −6.09)	0.035 * 0.113 (0.344)	0.182 0.389 (0.156)	0.972 0.982 (0.000)

Note: positive values of ankle joint moment represent ankle eversion and ankle dorsiflexion moment in the frontal and sagittal planes, and negative values represent ankle inversion and plantarflexion moment in the frontal and sagittal planes. *p*-valueadj = (Benjamini–Hochberg correction for false discovery rate), reported for secondary outcomes only; peak ankle joint moment (frontal and sagittal planes) was designated a primary outcome and evaluated at $p < 0.05$ without adjustment, indicated by "-". * Statistically significant for both raw and Benjamini–Hochberg-adjusted *p*-values ($p < 0.05$). (Table S1 in Supplementary File).

Table 2. Muscle activation (% peak task activation) across different hopping speed and distance conditions during landing and take-off phases. Data are presented as mean ± SD with 95% confidence intervals in parentheses.

Parameters	Slow Speed		Fast Speed		Speed	Distance	Interaction
	30 cm	45 cm	30 cm	45 cm	*p*-Value *p*-Valueadj (η_p^2)	*p*-Value *p*-Valueadj (η_p^2)	*p*-Value *p*-Valueadj (η_p^2)
Landing							
Peroneus longus	45.50 ± 9.33 (39.57, 51.43)	46.58 ± 8.99 (40.87, 52.29)	55.47 ± 15.41 (45.68, 65.25)	57.46 ± 14.93 (47.97, 66.94)	0.020 * 0.098 (0.399)	0.464 0.720 (0.050)	0.825 0.928 (0.005)
Tibialis anterior	42.50 ± 10.66 (35.70, 49.30)	43.60 ± 13.66 (34.90, 52.30)	29.23 ± 7.63 (24.40, 34.10)	30.50 ± 8.47 (25.10, 35.90)	0.002 * 0.018 * (0.587)	0.586 0.828 (0.028)	0.964 0.982 (0.000)
Lateral gastrocnemius	43.08 ± 16.28 (32.70, 53.40)	43.92 ± 14.36 (34.80, 53.00)	51.84 ± 15.06 (42.30, 61.40)	52.97 ± 13.00 (44.70, 61.20)	0.190 0.389 (0.151)	0.564 0.819 (0.031)	0.947 0.982 (0.000)
Soleus	37.39 ± 13.05 (29.10, 45.70)	34.46 ± 13.71 (25.70, 43.20)	55.21 ± 12.03 (47.60, 62.90)	52.16 ± 10.92 (45.20, 59.10)	<0.001 * 0.008 * (0.659)	0.121 0.289 (0.205)	0.982 0.982 (0.000)
Take-off							
Peroneus longus	62.64 ± 14.36 (53.5, 71.80)	62.23 ± 11.90 (54.70, 69.80)	55.89 ± 15.75 (45.90, 65.90)	53.44 ± 15.93 (43.30, 63.60)	0.122 0.289 (0.203)	0.483 0.725 (0.046)	0.646 0.828 (0.020)
Tibialis anterior	42.53 ± 11.11 (35.50, 49.60)	40.66 ± 15.25 (31.00, 50.40)	31.59 ± 10.55 (24.90, 38.30)	34.22 ± 9.79 (28.00, 40.40)	0.028 * 0.105 (0.366)	0.807 0.928 (0.006)	0.308 0.520 (0.094)
Lateral gastrocnemius	52.06 ± 16.08 (41.80, 62.30)	51.37 ± 16.73 (40.70, 62.00)	47.62 ± 17.56 (36.50, 58.80)	44.02 ± 15.67 (34.10, 54.00)	0.024 * 0.098 (0.385)	0.263 0.493 (0.112)	0.640 0.828 (0.021)
Soleus	59.48 ± 9.38 (53.50, 65.40)	56.73 ± 12.94 (48.50, 64.90)	52.66 ± 12.58 (44.70, 60.60)	46.93 ± 12.53 (39.00, 54.90)	0.023 * 0.098 (0.387)	0.145 0.326 (0.183)	0.662 0.828 (0.018)

Note: *p*-valueadj = (Benjamini–Hochberg correction for false discovery rate), reported for all outcomes. * Statistically significant for both raw and Benjamini–Hochberg-adjusted *p*-values ($p < 0.05$). (Table S2 in Supplementary File).

4. Discussion

This study investigated the effects of varying hopping speed and distance on ankle joint mechanics and muscle activation during repeated lateral hopping tasks, focusing on the lateral landing and subsequent take-off phase. We hypothesised that increased task intensity by greater hopping speed and distance would result in higher ankle joint kinetics and muscle activation. Our findings partially supported a priori hypothesis, as variations in hopping speed and distance were associated with significant changes in ankle joint moment, as well as for some measures, joint power and muscle activation. Specifically, faster hopping generally increased the joint moments, whereas effects on joint power and muscle activation depended on the specific variable and task condition. Moreover, wide-distance hopping increased the ankle joint moment and range of motion.

The observed changes in peak ankle joint moments, joint power, and muscle activation across different hopping speeds reflect a coordinated adaptation of the ankle kinetic output and neuromuscular responses to varying mechanical demands. Speed manipulation significantly increased ankle joint moment, with particularly pronounced effects in the frontal plane than in the sagittal plane. Sagittal plane peak positive power also increased with speed and remained significant after correction for multiple comparisons, while frontal plane power did not remain significant after correction. Although research examining speed effects during lateral movements remains limited, this pattern is consistent with forward locomotion studies, where increasing running speed similarly elevates ankle joint moments and power during landing and propulsion phases [38–40]. de David et al. (2014) demonstrated that faster running speeds produced greater ankle joint moments in both sagittal and frontal planes [40]. Similarly, Jin et al. (2024) reported progressive increases in ankle moments and peak joint power during the transition from walking to running speeds [39]. This relationship between speed and joint kinetics appears to be a general feature of dynamic lower limb tasks, our findings suggest it also applies to lateral movements. Although absolute kinetic values were greater in the sagittal plane, the proportional increase from slow to fast speed was consistently larger in the frontal plane. This suggests that faster lateral hops primarily increase demand on frontal plane stabilisation to control rapid medio-lateral redirection, while contributions in the sagittal plane also rise, but to a lesser extent. Moreover, the observed increases in ankle joint moments across both planes with faster hopping speeds may reflect a progressive overload mechanism similar to that seen in resistance training. This pattern is consistent with the findings of Kipp et al. (2020), who investigated alterations in ankle net joint moments during squatting at progressively increasing intensities and demonstrated that ankle net joint moments increased in response to greater external loads, supporting the notion that mechanical demands on the ankle joint scale with task intensity [41]. Taken together, these findings suggest that speed-based progression may increase mechanical loading on the ankle joint without adding external load. Whether this approach could be applied safely or effectively for injury prevention purposes remains to be tested in future prospective training studies.

During the landing phase, Soleus activation increased and tibialis anterior activation decreased at fast compared to slow speed, both remaining significant after correction for multiple comparisons. The greater soleus muscle activation under fast hopping conditions relative to slow speed suggests higher stabilisation demands during landing, likely reflecting the increased eccentric loading demands at faster speed. The soleus provides sagittal plane ankle control and contributes to shock absorption [42], though whether its elevated activation also serves a preparatory role for the subsequent take-off cannot be determined from peak amplitude measures alone. In contrast, tibialis anterior activation decreased during faster hopping tasks. This finding aligns with prior research, which shows greater tibialis anterior activity during slower locomotor tasks (jogging) compared to faster movements (running) in the early half of the contact phase [43]. This elevated activation may reflect longer ground contact time at slower speeds, which requires sustained tibialis anterior activation to maintain ankle stability and control foot positioning before initiating the next take-off phase. Surprisingly, peak ankle joint moment (eversion) increased with speed but peroneus longus activation did not change accordingly. Despite being the primary muscle responsible for mediolateral stabilisation at the ankle [44], its activation showed no significant difference between speeds after correction for multiple comparisons. During the take-off phase, none of the muscles examined showed significant differences in activation between speeds after correction for multiple comparisons, despite an apparent reduction in tibialis anterior, soleus, and gastrocnemius activation at faster speeds prior to correction. One possible explanation is more effective use of the stretching-shortening cycle at faster speeds. Shorter contact times tend to mean more elastic energy return and less reliance on active muscle force generation [17]. However, these effects did not remain significant after correction for multiple comparisons. This interpretation remains speculative.

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Landing contact time was shorter at fast repeated lateral hopping speeds compared to the slower condition, reflecting the naturally shorter ground contact during faster hops. This finding demonstrates consistency with previous research, where participants exhibited ground contact times ranging from 0.17 to 0.26 s when completing side hop tests within 5–8 s [25,45]. The observed reduction in contact time at higher speeds reflects a fundamental characteristic of efficient athletic movement, as shorter ground contact times are associated with enhanced movement efficiency in sports performance, particularly during tasks requiring rapid force production and quick directional changes [45]. The reduced contact duration corresponds to observed increases in ankle joint moment and activation changes in key stabilising muscles, indicating coordinated mechanical and neuromuscular adaptations during faster movements.

The coordinated mechanical, neuromuscular, and temporal adaptation to increased hopping speed may be relevant to the demands of multidirectional sport performance, although this was not directly assessed. Faster lateral hopping increases eccentric loading, neuromuscular adaptation and challenges associated with shorter ground contact time, features that are conceptually associated with the movement demands of rapid cutting, side stepping, and changing direction. The extent to which these acute biomechanical changes translate into improved sport-specific capacity would require testing in longitudinal training studies. These findings may help inform future progression designs, where slower speeds provide more controlled movement demands before advancing to faster sport-specific conditions, a logic broadly consistent with hop-stabilisation programmes shown elsewhere to reduce ground contact time in individuals with ankle instability [9,13]. However, further research is needed to confirm the applicability of speed-based progression across different movement patterns, such as cutting or multidirectional tasks.

Beyond speed modification, distance manipulation offers a distinctly different progression strategy through directionally selective loading. Wider hop distances substantially increased frontal plane joint moment and kinematics demands compared to minimal changes in the sagittal plane. The significant interaction between speed and distance for peak ankle eversion moments suggests that wider hop distances imposed greater eversion demands under both speed conditions, with a more pronounced increase in eversion moment magnitude at fast compared to slow hopping speeds, indicating that the two parameters combined to amplify frontal plane loading. Notably, despite these mechanical increases, muscle activation and ground contact time remained unchanged, suggesting that increased hop distance elevated mechanical joint loading without altering the activation of the measured ankle muscles, though the response of proximal muscles remains unknown. Also, the frontal plane alteration observed reflects the lateral trajectory of the hopping task. During lateral hops, the main movement direction results in greater medio-lateral forces and demands improved control of frontal plane stability, with sagittal-plane requirements also rising, but to a lesser degree. This frontal plane emphasis is consistent with previous research showing that lateral hops produce substantially greater frontal plane kinetics compared to vertical jumps [45], demonstrating the directional specificity of these movement patterns. Limited evidence exists regarding how distance manipulation affects ankle biomechanics during lateral hopping. While direct comparisons are limited, distance effects have been observed in the vertical distance manipulation, as greater drop height and step-down distance increase angular displacement during forward landing tasks [46,47]. Notably, greater vertical distance showed larger increases in sagittal plane ROM compared to the frontal plane [47], indicating that the primary plane of movement dictates the plane most sensitive to distance manipulation. Our findings expand this principle to lateral horizontal movement, showing that larger changes in range of motion and joint moment occur in the frontal plane compared to the sagittal plane. Forward and vertical tasks emphasise sagittal plane responses, while lateral tasks emphasise frontal plane responses [47]. This movement direction-specific pattern has important implications for selecting tasks to target specific planes of motion in training, though its application to rehabilitation settings requires further investigation.

Distance manipulation facilitates loading progression, particularly in the frontal plane, which is relevant for multidirectional sports. A wider distance hopping increases frontal plane joint loading without a corresponding increase in the peak activation of the ankle muscles, suggesting that mechanical and distal muscle activation demands do not increase in parallel with distance manipulation, though the response of proximal muscles remains unknown. This may complement speed-based progressions by offering an alternative means of increasing joint loading demand. Greater ankle dorsiflexion ROM has been associated with jump performance in other contexts [48], though whether this extends to wider hop distances is untested. The greater frontal plane seen with distance progression may be relevant to CAI rehabilitation, as individuals with CAI often show impaired frontal plane control during landing. However, we did not assess clinical populations or rehabilitation outcomes, so this application remains speculative and would need to be tested directly in the CAI population.

Several limitations should be acknowledged. A key limitation is that the study included only trained young men, which limits the generalizability of the findings to women, elite athletes, or individuals with ankle pathologies such as CAI. Although the achieved sample (n = 12) exceeded the a priori minimum requirement (n = 7), the power calculation relied on peak ankle plantarflexion moment (η_p^2 = 0.347) and peak subtalar inversion moment (η_p^2 = 0.291) during a lateral cutting task [27], as no prior study has reported a speed x distance interaction specifically for repeated lateral hopping. As the present study examined ankle kinetics in both sagittal and frontal planes, an intermediate value (η_p^2 = 0.30) was adopted. Given that the interacting factors and task differ from those in the present study, the true interaction effect for speed x distance may differ from this a priori estimate, and the resulting power for some non-significant findings should therefore be interpreted with caution. Primary outcomes were designated a priori and secondary outcomes were adjusted for multiple comparisons using the Benjamini–Hochberg false discovery rate procedure; however, given the task mismatch underlying our a priori power calculation (which was based on a lateral cutting task, as no prior study had reported a speed × distance interaction for repeated lateral hopping) and the relatively small sample size, both non-significant findings and those that did not remain significant after correction should be interpreted with appropriate caution, as the study may have been underpowered to detect some true effects. Only two hopping speeds and distances were tested without incorporating vertical displacement (e.g., jump or landing height), which may not fully capture the range of mechanical and muscle activation relevant to progressive training design. Additionally, fixed hop distances were used rather than distances normalised to leg length or body height, which may have resulted in varying relative task difficulty across participants. This approach was chosen to maintain a consistent 50% increment between distance conditions, mirroring the progression used for speed, but future studies could consider normalised distances to better account for individual anthropometric differences. Furthermore, intra-session reliability was not formally assessed. Future studies should report trial-to-trial reliability measures such as coefficients of variation to strengthen the confidence of the reported findings. EMG signals were normalised to peak amplitude across tasks rather than to maximum voluntary contraction (MVC), which may limit comparisons with studies using MVC-based normalisation, though this approach was appropriate for within-session between-condition comparisons. Only lateral contacts were analysed. As repeated side-to-side hopping involves both medial and lateral contacts, focusing on lateral landing alone may not fully capture the demands of the complete lateral movement task. Finally, the assessment of only distal muscles limits the comprehensive understanding of neuromuscular responses across conditions. Future research should include proximal muscle measurements to fully understand these movement strategies in lateral movement tasks. Furthermore, integrating electromyography with a musculoskeletal modelling approach may provide deeper insight into muscle coordination and synergy during multidirectional movements.

5. Conclusions

This study provides preliminary biomechanical evidence that speed and distance manipulations produce distinct effects on ankle joint mechanics during repeated lateral hop tasks in healthy male trained athletes. Speed manipulation increased mechanical demands and reduced ground contact time, while muscle activation responses varied across movement phases. Distance manipulation indicated selective mechanical loading without altering the activation of the measured distal ankle muscles. These findings represent different mechanical stimuli, with potential implications for progression in multidirectional sport training, although their application to rehabilitation populations, such as individuals with CAI, requires further investigation.

Supplementary Materials: The following supporting information can be downloaded at https://www.mdpi.com/article/10.3390/biomechanics6030076/s1, Table S1: Raw and FDR-adjusted *p*-values corresponding to Table 1 (landing contact time, ankle range of motion, and peak negative and positive joint power) across different hopping speed and distance conditions during landing and take-off phases. Data are presented as mean ± SD with 95% confidence intervals in parentheses; Table S2: Raw and FDR-adjusted *p*-values corresponding to Table 2 (muscle activation, % peak task activation) across different hopping speed and distance conditions during landing and take-off phases. Data are presented as mean ± SD with 95% confidence intervals in parentheses.

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

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