Research Square

Preprints are preliminary reports that have not undergone peer review.
They should not be considered conclusive, used to inform clinical practice,
or referenced by the media as validated information.

Impact kinetics, dynamic postural stability, and segmental acceleration during vertical and multiplanar single-leg drop landings : Insights for sports injury prevention

Indy Man Kit Ho

indyho@thei.edu.hk

Technological and Higher Education Institute of Hong Kong

Daniel Hung-Kay Chow

Education University of Hong Kong

Lana Ružić

University of Zagreb

Article

Keywords: Anterior cruciate ligament, Ground reaction force, Dynamic postural stability, Tibial shock, Drop landing, Inertial measurement unit

Posted Date: November 24th, 2026

DOI: https://doi.org/10.21203/rs.3.rs-11579957/v1

License: This work is licensed under a Creative Commons Attribution 4.0 International License.
Read Full License

Additional Declarations: No competing interests reported.

Impact kinetics, dynamic postural stability, and segmental acceleration during vertical and multiplanar single-leg drop landings : Insights for sports injury prevention

Indy Man Kit Ho[1,2], Daniel Hung-Kay Chow[3], Lana Ružić[2]

1 Department of Sport and Recreation, Technological and Higher Education Institute of Hong Kong, Hong Kong SAR, China

2 Faculty of Kinesiology, University of Zagreb, 10110 Zagreb, Croatia

3 Department of Health and Physical Education, The Education University of Hong Kong, Hong Kong SAR, China

Abstract

Multiplanar single-leg drop landings (MSLDL) may challenge impact absorption and postural control differently from vertical single-leg drop landings (SLDL). Forty-eight recreationally active collegiate athletes (24 males and 24 females) performed 30-cm vertical SLDL and six directional or rotational MSLDL tasks, with horizontal displacement standardized to 30% of body height. Ground reaction forces and segmental accelerations were recorded using a force platform and synchronized wireless inertial measurement units. Mixed-design analyses of variance and secondary comparisons restricted to MSLDL conditions were performed, with Holm–Bonferroni-adjusted post-hoc pairwise comparisons. Compared with baseline SLDL, pooled MSLDL values were higher for peak vertical ground reaction force (3.80 ± 0.76 versus 3.26 ± 0.63 body weights), vertical average loading rate (87.67 ± 32.39 versus 63.98 ± 25.93 body weights/s), and composite dynamic postural stability index (0.342 ± 0.046 versus 0.264 ± 0.040). Medial and lateral landings imposed the greatest mediolateral stabilization demands, whereas 90° internal rotation produced the highest peak vertical force and average loading rate. Acceleration attenuation and phase lag responses were task- and segment-specific. These findings support further evaluation of multiplanar landing assessments and future studies on injury prediction thresholds and prevention benefits require prospective validation on different target populations.

Keywords: Anterior cruciate ligament; Ground reaction force; Dynamic postural stability; Tibial shock; Drop landing; Inertial measurement unit.

1. Introduction

Non-contact lower-limb injuries, including anterior cruciate ligament (ACL) tears, knee pain, and lateral ankle sprains are important concerns in sport [1, 2]. A systematic review of team ball sports estimated that non-contact mechanisms accounted for 55% of ACL injuries [1]. Landing, cutting, and deceleration are key sports-specific skills highly relevant to ACL injuries [3]. Video-based reconstruction of ten ACL injury events suggested that injury occurred approximately 40 ms after initial contact (IC) [4]. Prospective and video-based studies have linked

knee loading and multiplanar movement patterns to ACL injury, although these associations do not establish an isolated risk factor, a universal injury mechanism or an individual risk threshold [5, 6, 7].

Single-leg drop landing (SLDL) is widely used to assess individual landing capabilities and dynamic postural control because participants need to rely on the robust and effective neuromuscular control to strategically attenuate and absorb landing forces, and stabilize over a small base of support (BOS) [8, 9, 10]. In stop-jump tasks, single-leg landings have produced greater peak vertical and posterior ground reaction forces (GRFs) than double-leg landings [11]. Given the unilateral weight bearing characteristics including the increased frontal and transverse stability challenge, SLDL assessments have also identified differences in landing kinetics and stabilization between athletes with and without chronic ankle instability [12].

Standardized vertical SLDL assessments do not capture all directional and rotational demands encountered in sports such as basketball, soccer, volleyball and netball. Multidirectional landing studies have demonstrated direction-dependent ankle mechanics and postural stability [13, 14, 15]. During mediolateral, oblique, or rotational landings, vertical (Fz) and horizontal (Fx and Fy) force components impose different braking and stabilization demands. Force platforms can quantify early impact kinetics using normalized peak vertical GRF (peak vGRF), vertical average loading rate (VALR), and vertical instantaneous loading rate (VILR). Dynamic postural control can be assessed using directional and composite dynamic postural stability indices (DPSI) and time to stabilization (TTS), although the analysis window and processing method affect these measures [10, 16].

Previous studies have compared forward, diagonal, and lateral landing tasks in healthy participants and athletes with chronic ankle instability [13, 14, 15]. Jolingana-Seoka et al. [17] also evaluated multidirectional unilateral landings in a pilot study of ten female netball players. Thus, directional landing comparisons are not new. Despite the extremely high vulnerability in the rotational movement on knee and ankle injuries, studies regarding rotational landing are lacking. Moreover, comparisons combining both vertical, directional and rotational SLDL conditions with force-platform and segmental accelerometry outcomes remain limited. Wearable inertial measurement units (IMUs) offer opportunities to collect movement data beyond laboratory settings, although much of the supporting evidence concerns running or task-specific wearable systems rather than validation of the present landing protocol [18, 19]. Tibial accelerometry provides information about segmental acceleration, but its interpretation depends on sensor placement, attachment, signal processing, and movement characteristics [20].

Several research questions therefore remain relevant: Do MSLDL tasks increase vertical loading rates or alter impact timing relative to vertical SLDL? Are changes in dynamic postural stability and TTS direction-specific? Do segmental acceleration attenuation and peak-timing measures differ between vertical SLDL and MSLDL tasks?

The purpose of this study was to compare landing kinetics, dynamic postural control, and wearable-derived acceleration measures between vertical SLDL and six MSLDL conditions (medial, lateral, anteromedial, anterolateral, 90° internal rotation, and 90° external rotation) in recreationally active athletes. We hypothesized that: (1) MSLDL tasks would produce higher vertical loading rates than baseline SLDL, with differences among MSLDL conditions; (2) MSLDL tasks would increase dynamic postural instability, with plane-specific changes in the directional subindices; (3) horizontal TTS (TTS_{Fx} and TTS_{Fy}) would be longer during MSLDL; (4) acceleration attenuation would be lower during MSLDL; and (5) phase lags would be shorter during MSLDL. The interpretation of attenuation and timing measures as indicators of landing stiffness was considered exploratory.

2. Methods

2.1 Participants and Ethical Approval

A total of 48 recreationally active collegiate athletes participated (24 males and 24 females; mean ± standard deviation: age, 22.4 ± 2.6 years; body mass, 64.8 ± 11.2 kg; height, 1.71 ± 0.09 m; body mass index, 22.0 ± 2.4 kg/m²). All participants had at least one year of jumping-related training experience in soccer, basketball, volleyball, handball, or track and field and had trained for at least 4 h per week during the preceding 12 months. Exclusion criteria were: (1) previous lower-limb surgery (e.g., ACL reconstruction or meniscal repair); (2) an acute lower-limb musculoskeletal injury within the preceding 6 months; (3) vestibular, visual, or neurological deficits; or (4) pain that altered natural landing execution. The experimental protocol was reviewed and formally approved by the Human Subjects Ethics Committee (HSEC) of Technological and Higher Education Institute of Hong Kong, as the Institutional Review Board (IRB), strictly adhering to the ethical principles outlined in the Declaration of Helsinki. Prior to experimental testing, all participants provided written informed consent.

2.2 Experimental Protocol and Landing Paradigms

Testing was conducted in a biomechanics laboratory after a standardized 5-min dynamic warm-up comprising 3 min of submaximal cycling, multidirectional lunges, body-weight squats, and submaximal pogo jumps. Participants performed SLDL from a 30-cm platform onto a floor-embedded AMTI force platform sampled at 1,000 Hz. The landing limb was the preferred take-off leg during unilateral jumping. Participants initiated each drop from the contralateral limb without jumping upward or stepping downward and landed on the center of the force platform. Arm movements were unrestricted to mimic natural sports-related balance reactions. Participants focused on an eye-level target 4 m in front of the force platform. After IC, they were instructed to stabilize as quickly as possible and maintain a motionless single-leg stance for 15 s.

Each participant completed two landing scenarios: (1) baseline vertical SLDL without horizontal displacement; and (2) six MSLDL conditions presented in randomized order: medial, lateral, anteromedial (45° oblique), anterolateral (45° oblique), 90° internal rotation, and 90° external rotation. Previous work has used single-leg drop-jump assessments [21] and multidirectional landing protocols [13], but neither reference establishes an identical setup to that used here. For MSLDL, the horizontal distance from the drop platform to the midpoint of the near edge of the force platform was standardized to 30% of each participant's height. This distance was adapted from the general multidirectional approach presented in Sonsukong et al. [13] rather than treated as an externally validated standard. For rotational landings, participants initially faced 90° relative to the target, rotated in mid-flight, and landed facing the target. A valid trial required a quiet single-leg stance for 15 s without contralateral foot touchdown, hopping, landing outside the force platform, or repositioning the landing foot. Unsuccessful trials were discarded and participants were required to repeat. Three valid trials were recorded per condition, with at least 60 s of rest between trials to eliminate fatigue.

2.3 Instrumentation, Signal Processing, and Biomechanical Endpoints

Triaxial GRFs and center of pressure (CoP) data were recorded at 1,000 Hz using an AMTI force platform (600 × 400 mm). Segmental acceleration and angular velocity were recorded simultaneously using wireless IMUs (Ultium Motion, Noraxon, Scottsdale, Arizona, USA) sampled at 200 Hz. Seven sensors were positioned according to the manufacturer's recommended anatomical locations: one over the flat surface of the sacrum; two over the anterior aspect of the distal half of each thigh, where muscle displacement was minimal; two over the flat anteromedial surface of each tibia, approximately 5 to 7.5 cm below the tibial tuberosity; and two over the dorsal midfoot, aligned with the second and third metatarsals. Participants performed all landing trials in bare feet. All IMUs were attached directly to the skin using custom double-sided tape to minimize tilting and movement artifacts. IMU and force-platform recordings were synchronized using a synchronization box, and IMU data were

upsampled to 1,000 Hz automatically via MyoResearch software provided by the Noraxon using spline interpolation method. Raw signals were filtered in two passes using a zero-phase, fourth-order Butterworth filter, with cutoff frequencies of 50 Hz for kinetic and DPSI measures, 20 Hz for accelerometry, and 15 Hz for TTS. IC (t = 0) was defined as the first instance at which vGRF exceeded 10 N.

The following outcomes were calculated: (1) peak vGRF, the maximum vertical force during the first 250 ms after IC, normalized to body weight (BW = body mass × 9.81 m/s²); (2) time to peak vGRF (TTP vGRF; s); (3) VALR (BW/s), the slope from IC to peak vGRF; (4) VILR (BW/s), the maximum instantaneous derivative of normalized Fz during the first 250 ms; (5) directional $DPSI_{AP}$ and $DPSI_{ML}$ and composite $DPSI_{Total}$, evaluated over the first 3 s after IC [10]; The isolated sagittal plane component of the DPSI formulation, reflecting shear instability was calculated by: $DPSI_{AP}$ = sqrt((sum(Fy²) / N_samples) / BW, where Fy represents anteroposterior component of ground reaction force at each time point.

$$\mathrm{DPSI_{AP}} = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N}\left(F_y[i]\right)^2}}{BW}$$

The isolated frontal plane component of the DPSI formulation, reflecting shear instability was calculated by: $DPSI_{ML}$ = sqrt((sum(Fx²) / N_samples) / BW, where Fx represents mediolateral component of ground reaction force at each time point.

$$\mathrm{DPSI_{ML}} = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N}(F_x[i])^2}}{BW}$$

Dynamic Postural Stability Index (DPSI Total labelled as "Target_DPSI_Total"): defined as the composite 3D dynamic postural stability index. It was evaluated over the first 3.0 seconds post-IC in accordance with the formulation of Wikstrom et al. [8]: DPSI Total = sqrt((sum(Fx²) + sum(Fy²) + sum((Fz - BW)²)) / N_samples) / BW, where Fx, Fy, and Fz represent mediolateral, anteroposterior, and vGRF at each time point, respectively.

$$\mathrm{DPSI_{Total}} = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N}\left[F_x[i]^2 + F_y[i]^2 + (F_z[i] - W)^2\right]}}{BW}$$

(6) horizontal TTS in the mediolateral (TTS_{Fx}) and anteroposterior (TTS_{Fy}) directions, calculated using an adapted Ross-range polynomial curve-fitting threshold method [8, 16]; The individualized baseline stability bandwidth was established via the quiet single-leg standing phase in the final 10 seconds. The peak-to-peak force range (max – min) was extracted across two consecutive blocks of 5-second windows (5.0-10.0 s and 10.0-15.0 s). The minimum peak-to-peak range was identified as the baseline tremor amplitude, and subsequently, the stabilization threshold was defined as 50% of this minimum range to match the amplitude of rectified deviations. After that, the primary impact shear peak was determined within the initial 500 ms post-IC to isolate the landing deceleration from subsequent postural sway, and the signal from peak shear onward was rectified around the mean baseline force of the final 5-second quiet stance period. A 3rd order polynomial curve was fitted to this rectified post-peak signal using least-squares regression. TTS was defined as the earliest time point at which the fitted polynomial trajectory decayed below the threshold boundary. Let $F_\tau(t)$ denote the low-pass filtered (15 Hz) horizontal shear force along axis $\tau \in \{x, y\}$:

$$\text{Range}_{\min,\tau} = \min(\text{ptp}(F_\tau[5.0\text{ s} \le t \le 10.0\text{ s}]),\ \text{ptp}(F_\tau[10.0\text{ s} \le t \le 15.0\text{ s}]))$$

$$T_\tau = \frac{\text{Range}_{\min,\tau}}{2}$$

$$F_{\text{base},\tau} = \frac{1}{N_{\text{quiet}}} \sum_{t=10.0\text{ s}}^{15.0\text{ s}} F_\tau(t)$$

$$y_\tau(t) = \left|F_\tau(t) - F_{\text{base},\tau}\right|, \quad \text{for } t \ge t_{\text{peak},\tau}$$

$$\hat{y}_\tau(t') = \beta_3(t')^3 + \beta_2(t')^2 + \beta_1(t') + \beta_0, \quad \text{where } t' = t - t_{\text{peak},\tau}$$

$$\text{TTS}_\tau = t_{\text{peak},\tau} + \min\{t' \ge 0 \mid \hat{y}_\tau(t') \le T_\tau\}$$

(7) segmental acceleration attenuation (dB), calculated as the logarithmic ratio of distal to proximal peak acceleration; and (8) phase lag, calculated as the difference between selected proximal and distal peak-acceleration times. These acceleration ratios and timing differences were not direct measures of energy absorption or tissue loading. The attenuation and phase lag equations were as follows:

$$\text{SA-ST} = 20\log_{10}\left(\frac{\max(a_{\text{shank}})}{\max(a_{\text{thigh}})}\right)$$

$$\text{SA-TP} = 20\log_{10}\left(\frac{\max(a_{\text{thigh}})}{\max(a_{\text{pelvis}})}\right)$$

$$\text{SA-FP} = 20\log_{10}\left(\frac{\max(a_{\text{foot}})}{\max(a_{\text{pelvis}})}\right)$$

$$\text{PL-FP} = \left(t_{\text{peak}}(\text{Pelvis}) - t_{\text{peak}}(\text{Foot})\right) \times 1000$$

$$\text{PL-SP} = \left(t_{\text{peak}}(\text{Pelvis}) - t_{\text{peak}}(\text{Shank})\right) \times 1000$$

Here, a denotes segmental acceleration, SA denotes acceleration attenuation, and PL denotes phase lag. ST indicates shank to thigh, TP indicates thigh to pelvis, FP indicates foot to pelvis, and SP indicates shank to pelvis. Positive attenuation values indicate a lower proximal than distal peak acceleration. Phase lag is expressed in milliseconds.

2.4 Statistical Analyses

Descriptive statistics are presented as mean ± standard deviation (SD). The distribution of standardized residuals was evaluated using Shapiro–Wilk tests, skewness, and kurtosis. Levene's test assessed homogeneity of variance between sexes. Two-way mixed-design analyses of variance (ANOVA) were performed for all 15 dependent variables, with sex (male or female) as the between-subject factor and landing condition (baseline SLDL and six MSLDL conditions) as the within-subject factor. Significant condition effects were followed by pairwise comparisons of each MSLDL condition with baseline SLDL. Step-down Holm–Bonferroni adjustments were applied to the pairwise comparisons. Statistical significance was set a priori at $\alpha = 0.05$. Effect sizes were expressed as partial eta squared (ηp^2; small: ≥ 0.01, medium: ≥ 0.06, large: ≥ 0.14) for omnibus effects and Hedges'

g (small: 0.20–0.49, medium: 0.50–0.79, large: ≥ 0.80) with 95% confidence intervals for pairwise comparisons. Relative and absolute test-retest reliability were assessed using intraclass correlation coefficients ($ICC_{3,1}$), standard error of measurement (SEM), minimal detectable change at 95% confidence (MDC_{95}), and coefficient of variation (CV%). Analyses were conducted in Python using SciPy v1.11 and statsmodels v0.14.

A secondary repeated-measures ANOVA was restricted to the six MSLDL conditions to examine directional and rotational differences without including the SLDL baseline, which lacked horizontal displacement. All MSLDL conditions used the same prescribed horizontal displacement normalized to body height. Significant condition effects were followed by all 15 pairwise comparisons among the six MSLDL conditions, with step-down Holm–Bonferroni adjustments and Hedges' g with 95% confidence intervals.

3. Results

Test-retest reliability was good for composite dynamic stability ($DPSI_{Total}$; $ICC_{3,1}$ = 0.77–0.87; CV ≤ 5.4%) and moderate to good for the primary impact measures (peak vGRF, TTP vGRF, VALR, and VILR; $ICC_{3,1}$ = 0.51–0.86; CV = 7.5–26.8%). $DPSI_{AP}$ showed poor to moderate repeatability ($ICC_{3,1}$ = 0.38–0.59; CV = 5.84–14.19%). $DPSI_{ML}$ generally showed moderate to good repeatability ($ICC_{3,1}$ = 0.61–0.84; CV = 4.85–14.96%), except during 90° external rotation (ICC = 0.235; CV = 17.58%). SA-TP showed poor to good repeatability across conditions ($ICC_{3,1}$ = 0.48–0.76; CV = 18.4–43.1%). Other attenuation and phase lag measures were less consistent, with poor reliability for majority outcomes. Both horizontal TTS measures showed poor reliability ($ICC_{3,1}$ = 0.129–0.477). Parametric F-tests were retained on the basis of the authors' assessment of distributional deviations and the balanced design (24 males and 24 females). Sphericity was assessed using Mauchly's test, and Greenhouse–Geisser corrections were applied when sphericity was violated ($p < 0.05$).

The mixed-design ANOVA showed no statistically significant sex × condition interaction ($p > 0.05$). Omnibus condition effects were significant for all variables ($p < 0.001$) except CoP length and shank-to-thigh attenuation (SA-ST; both $p > 0.05$). Table 1 presents baseline SLDL and pooled MSLDL descriptive values alongside the omnibus tests of seven conditions. Pooled MSLDL values were 16.5% higher for peak vGRF, 37.0% higher for VALR, and 28.6% higher for VILR; TTP vGRF was reported as 13.6% shorter. The corresponding omnibus partial η^2 values were 0.092, 0.219, 0.080, and 0.285, respectively. Pooled $DPSI_{AP}$, $DPSI_{ML}$, $DPSI_{Total}$, TTS_{Fx}, and TTS_{Fy} were also higher than baseline values. $DPSI_{ML}$ increased by 215.9% and TTS_{Fx} by 65.8%, with omnibus partial η^2 values of 0.926 and 0.560, respectively. Mean SA-TP increased from 5.68 to 5.88 dB, whereas SA-FP decreased from 8.62 to 8.11 dB. Mean PL-FP decreased from 42.32 to 37.15 ms, whereas PL-SP increased from 0.77 to 4.04 ms. The omnibus partial η^2 values for SA-TP, SA-FP, PL-FP, and PL-SP were 0.102, 0.087, 0.180, and 0.104, respectively. These descriptive pooled differences should be distinguished from the pairwise comparisons reported below.

Pairwise comparisons of peak vGRF and TTP vGRF between baseline SLDL and each MSLDL condition are shown in Fig. 1a,b.

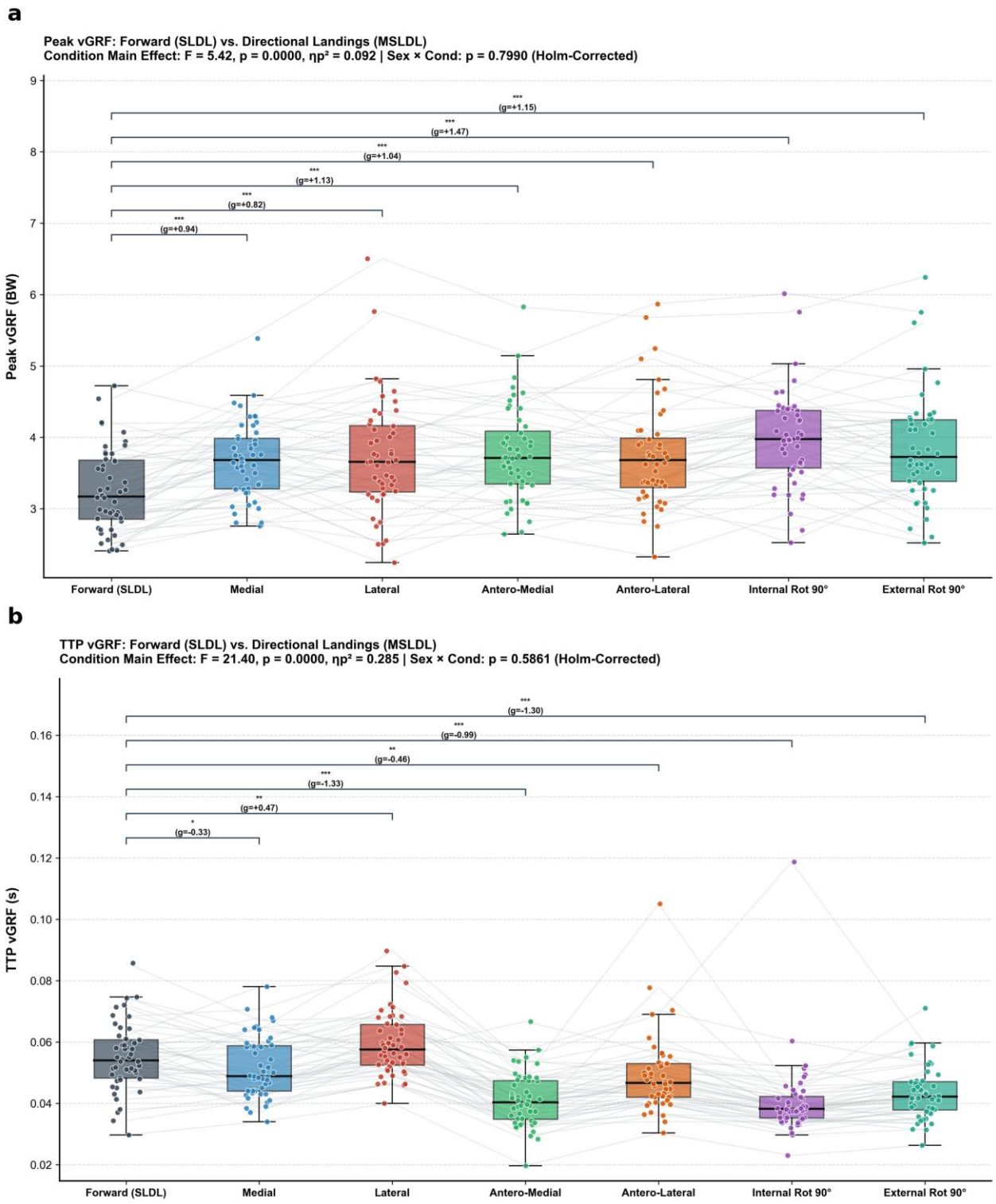

Figure 1. Peak vertical force and impact timing during vertical and multiplanar single-leg drop landings: (a) peak vertical ground reaction force (vGRF); (b) time to peak vGRF (TTP vGRF). SLDL, baseline vertical single-leg drop landing; MSLDL, multiplanar single-leg drop landing. Plots show comparisons of baseline SLDL with each MSLDL condition; comparisons among the six MSLDL conditions are not displayed. The cohort comprised 48 participants. Pairwise p-values were adjusted using the step-down Holm–Bonferroni procedure: *p < 0.05, **p < 0.01, ***p < 0.001. Values labelled g denote Hedges' g.

The corresponding pairwise comparisons of VALR and VILR are shown in Fig. 2a,b.

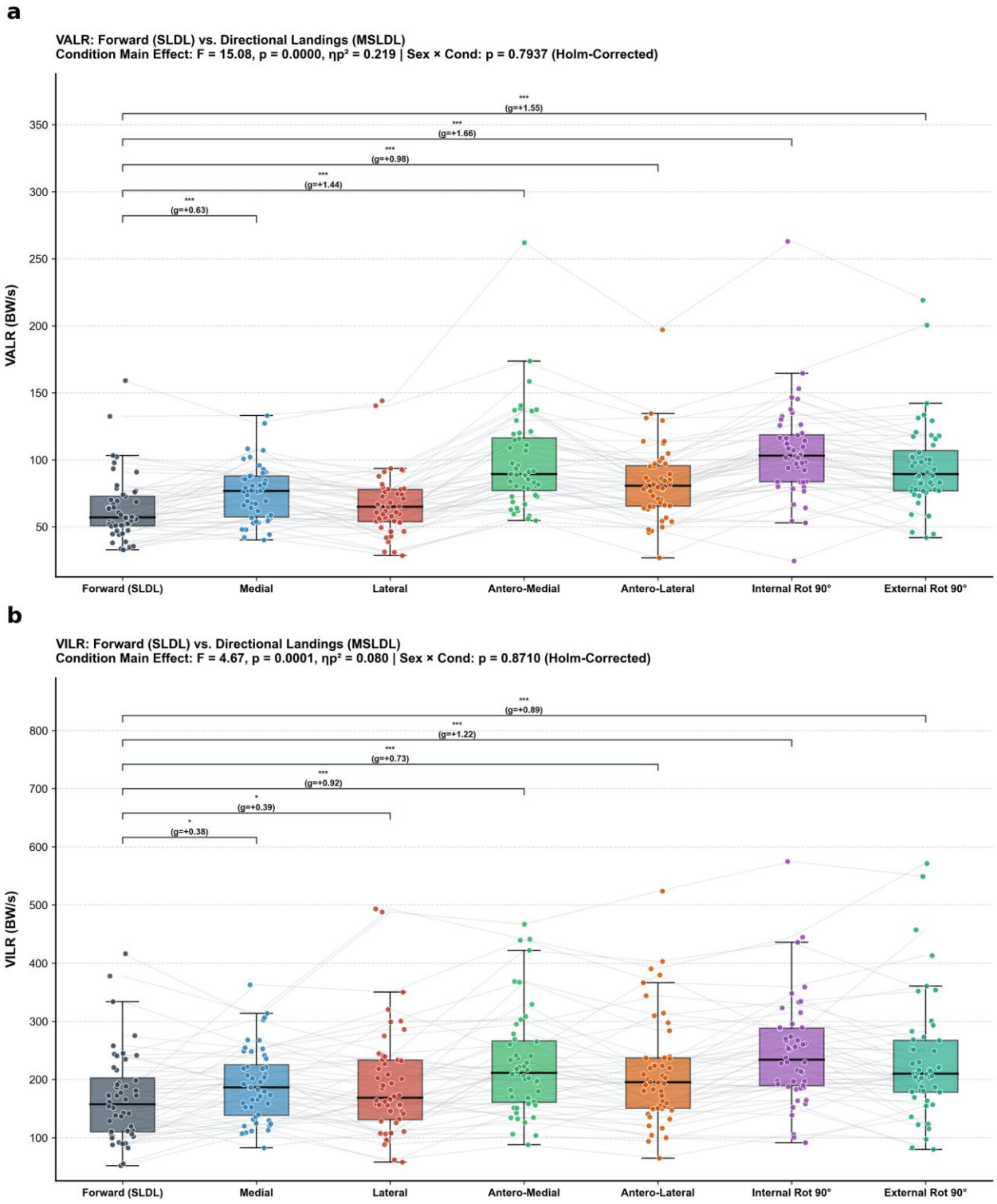

Figure 2. Vertical loading rates during vertical and multiplanar single-leg drop landings: (a) vertical average loading rate (VALR); (b) vertical instantaneous loading rate (VILR). SLDL, baseline vertical single-leg drop landing; MSLDL, multiplanar single-leg drop landing. Plots show comparisons of baseline SLDL with each MSLDL condition; comparisons among the six MSLDL conditions are not displayed. The cohort comprised 48 participants. Pairwise p-values were adjusted using the step-down Holm–Bonferroni procedure: *p < 0.05, **p < 0.01, ***p < 0.001. Values labelled g denote Hedges' g.

Among the six MSLDL conditions, the secondary repeated-measures ANOVA showed significant condition effects for peak vGRF ($p < 0.001$; $\eta p^2 = 0.098$), TTP vGRF ($p < 0.001$; $\eta p^2 = 0.542$), VALR ($p < 0.001$; $\eta p^2 = 0.523$), and VILR ($p < 0.001$; $\eta p^2 = 0.221$). Peak vGRF during 90° internal rotation (3.98 BW) was significantly

higher than during medial, lateral, anteromedial, and anterolateral landings ($p < 0.05$; $g = 0.43–0.61$). Anteromedial and 90° internal rotation landings had the shortest TTP vGRF (0.041 s). TTP vGRF during 90° internal rotation was significantly shorter than during medial ($p < 0.001$; $g = −1.07$), lateral ($p < 0.001$; $g = −1.58$), and anterolateral landings ($p < 0.001$; $g = −0.62$). TTP vGRF during anteromedial landing was also significantly shorter than during medial ($p < 0.001$; $g = −1.63$), lateral ($p < 0.001$; $g = −1.99$), and anterolateral landings ($p < 0.001$; $g = −0.70$). Medial landing produced a significantly shorter TTP vGRF than lateral landing ($p < 0.001$; $g = −0.89$).

VALR during 90° internal rotation was significantly higher than during medial ($p < 0.001$; $g = 1.23$), lateral ($p < 0.001$; $g = 1.66$), anterolateral ($p < 0.001$; $g = 1.33$), and 90° external rotation landings ($p < 0.001$; $g = 0.51$). In the same analysis, anteromedial landing also showed significantly higher loading-rate values than medial ($p < 0.001$; $g = 0.92$), lateral ($p < 0.001$; $g = 1.33$), and anterolateral landings ($p < 0.001$; $g = 0.75$). For VILR, 90° internal rotation values were significantly higher than medial ($p < 0.001$; $g = 0.87$), lateral ($p < 0.001$; $g = 0.82$), and anterolateral values ($p < 0.001$; $g = 0.72$). Anteromedial VILR was also significantly higher than medial ($p < 0.001$; $g = 0.57$) and lateral VILR ($p < 0.001$; $g = 0.54$).

$DPSI_{AP}$ was significantly higher during anteromedial, anterolateral, and 90° internal and external rotation landings than during baseline SLDL ($p < 0.001$; $g = 2.12–4.67$), but significantly lower during medial and lateral landings ($p < 0.001$; $g = −2.63$ to $−2.98$; Fig. 3a). $DPSI_{ML}$ was significantly higher in all MSLDL conditions, with the largest increases during lateral ($p < 0.001$; $g = 8.92$) and medial landings ($p < 0.001$; $g = 7.35$; Fig. 3b).

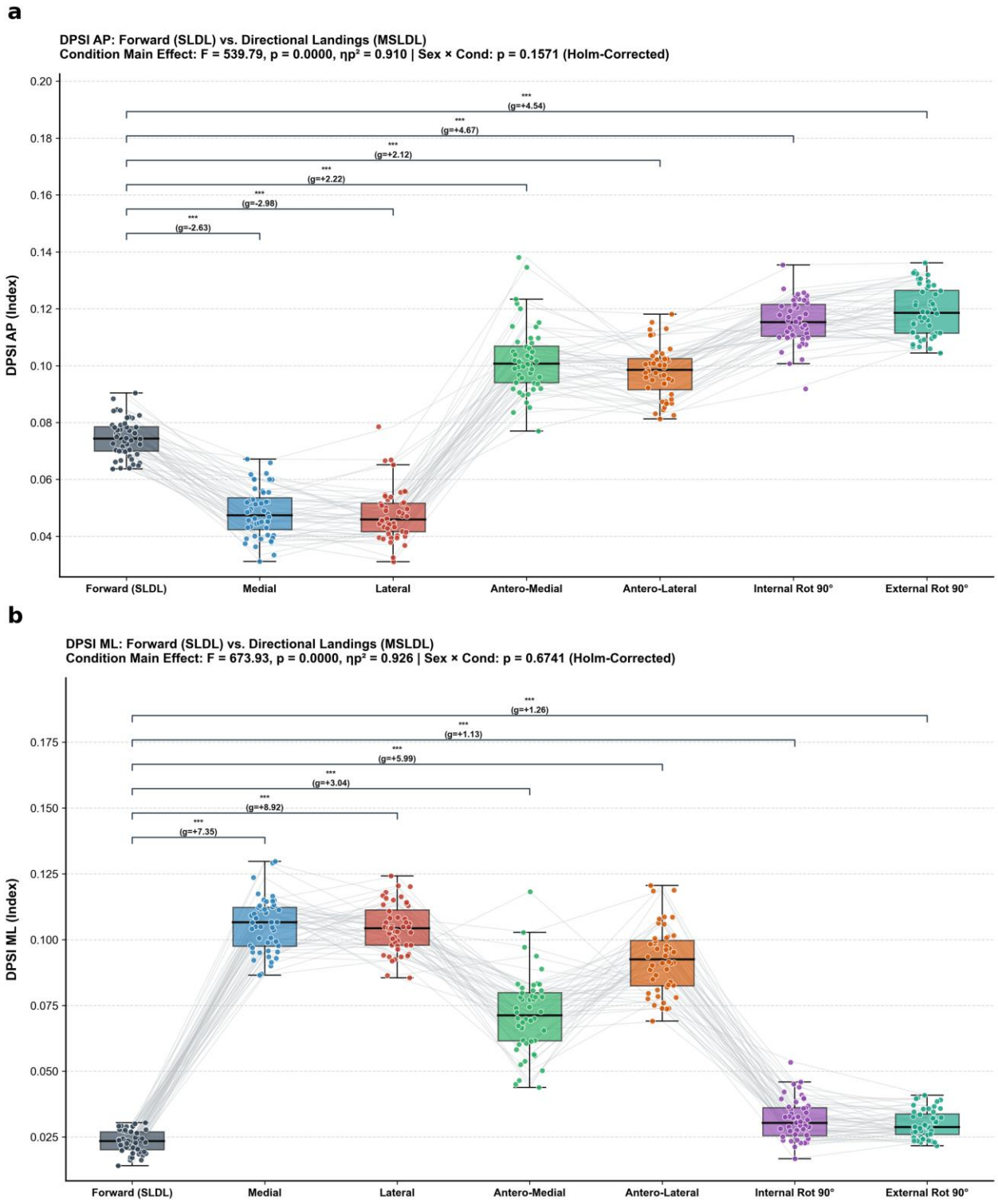

Figure 3. Directional dynamic postural stability during vertical and multiplanar single-leg drop landings: (a) anteroposterior index (DPSI AP); (b) mediolateral index (DPSI ML). SLDL, baseline vertical single-leg drop landing; MSLDL, multiplanar single-leg drop landing. Plots show comparisons of baseline SLDL with each MSLDL condition; comparisons among the six MSLDL conditions are not displayed. The cohort comprised 48 participants. Pairwise p-values were adjusted using the step-down Holm–Bonferroni procedure: *p < 0.05, **p < 0.01, ***p < 0.001. Values labelled g denote Hedges' g.

$DPSI_{Total}$ was significantly higher in all MSLDL conditions ($p < 0.001$; $g = 1.67–3.21$; Fig. 4).

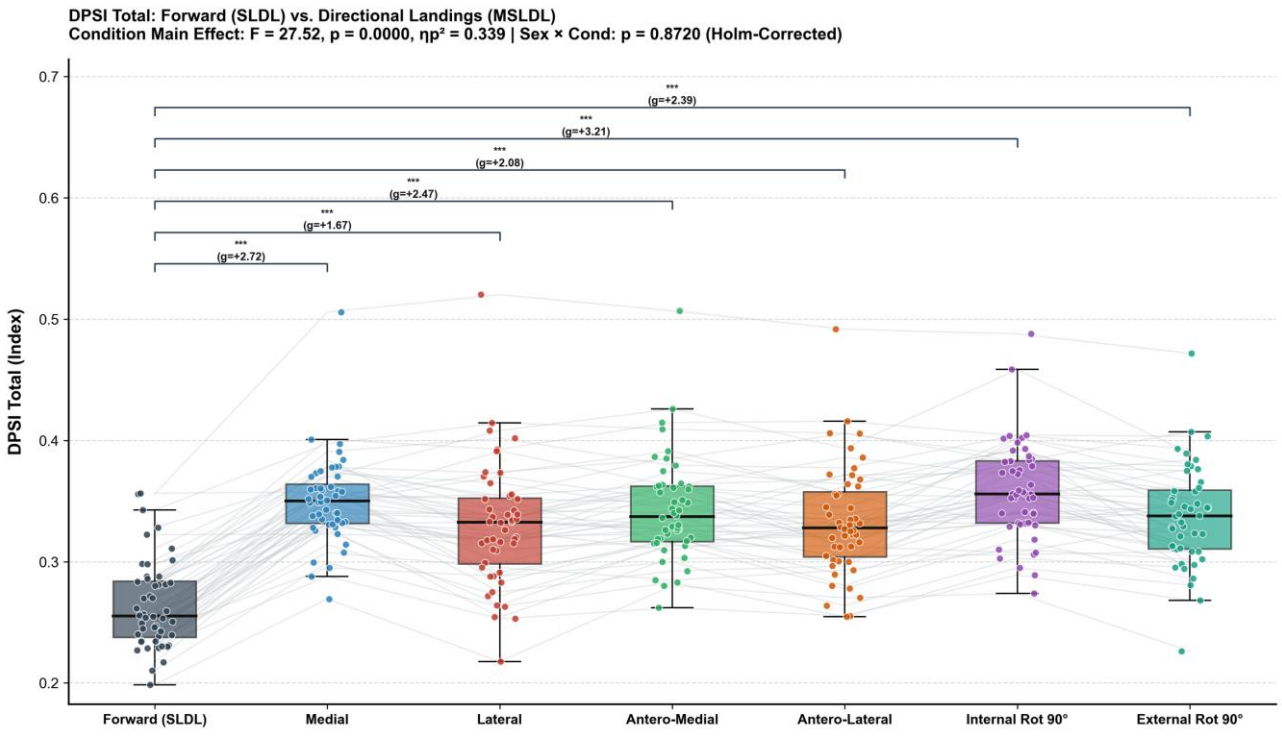

Figure 4. Composite dynamic postural stability index (DPSI Total) during vertical and multiplanar single-leg drop landings. SLDL, baseline vertical single-leg drop landing; MSLDL, multiplanar single-leg drop landing. Plots show comparisons of baseline SLDL with each MSLDL condition; comparisons among the six MSLDL conditions are not displayed. The cohort comprised 48 participants. Pairwise p-values were adjusted using the step-down Holm–Bonferroni procedure: *p < 0.05, **p < 0.01, ***p < 0.001. Values labelled g denote Hedges' g.

All MSLDL conditions produced significantly longer TTS_{Fx} than baseline SLDL ($p < 0.001$; $g = 0.82–2.72$; Fig. 5a). TTS_{Fy} was significantly longer during anteromedial, anterolateral, and 90° internal and external rotation landings ($p < 0.001$; $g = 1.07–1.50$), but significantly shorter during medial ($p < 0.01$; $g = −0.48$) and lateral landings ($p < 0.001$; $g = −0.61$; Fig. 5b).

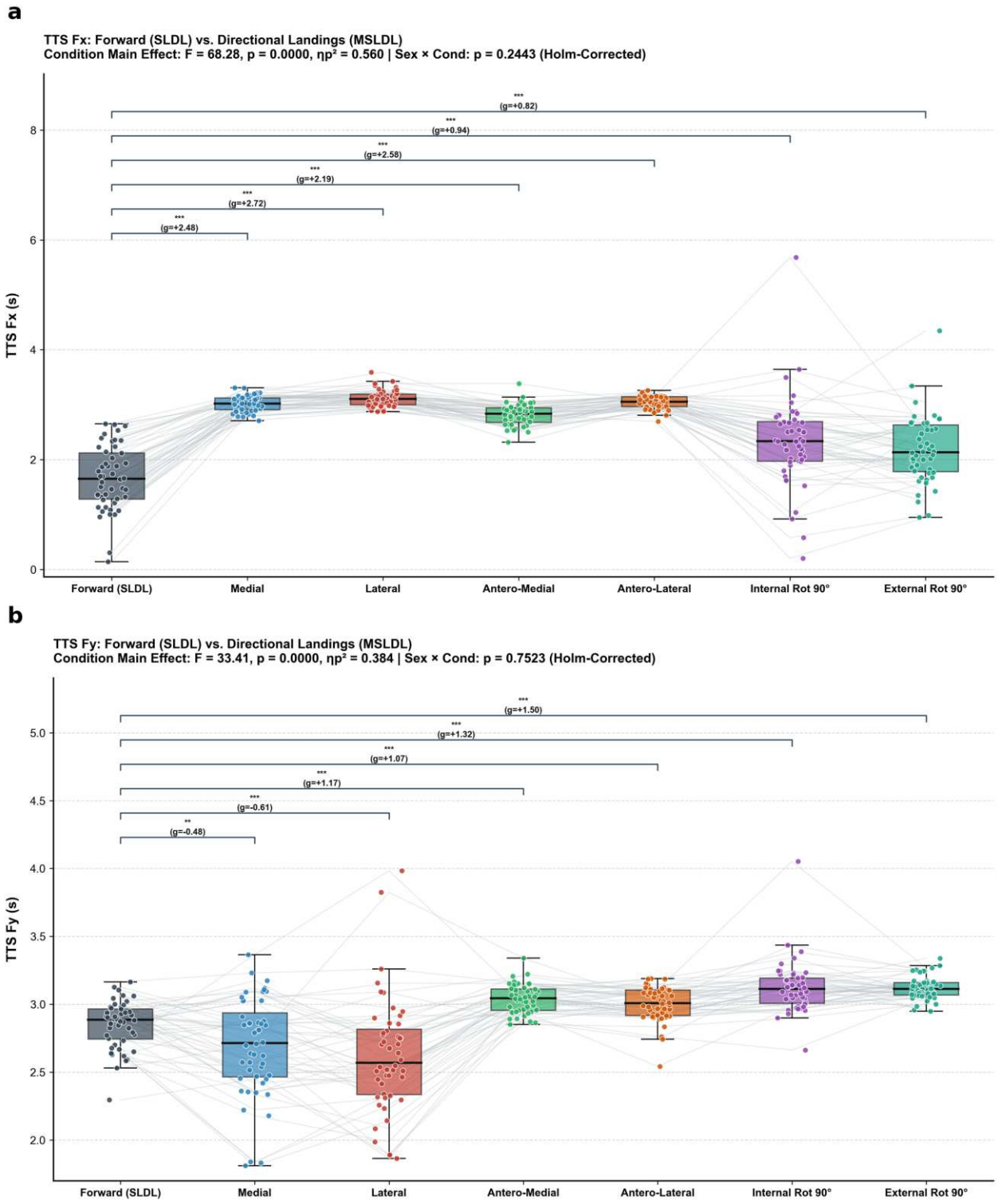

Figure 5. Horizontal time to stabilization during vertical and multiplanar single-leg drop landings: (a) mediolateral time to stabilization (TTS Fx); (b) anteroposterior time to stabilization (TTS Fy). SLDL, baseline vertical single-leg drop landing; MSLDL, multiplanar single-leg drop landing. Plots show comparisons of baseline SLDL with each MSLDL condition; comparisons among the six MSLDL conditions are not displayed. The cohort comprised 48 participants. Pairwise p-values were adjusted using the step-down Holm–Bonferroni procedure: *p < 0.05, **p < 0.01, ***p < 0.001. Values labelled g denote Hedges' g.

Among the six MSLDL conditions, significant condition effects were observed for $DPSI_{AP}$ ($p < 0.001$; $\eta p^2 = 0.942$), $DPSI_{ML}$ ($p < 0.001$; $\eta p^2 = 0.924$), $DPSI_{Total}$ ($p < 0.001$; $\eta p^2 = 0.255$), TTS_{Fx} ($p < 0.001$; $\eta p^2 = 0.532$), and TTS_{Fy} ($p < 0.001$; $\eta p^2 = 0.523$).

$DPSI_{AP}$ during 90° internal and external rotation was significantly higher than during medial ($p < 0.001$; $g = 5.76–5.90$), lateral ($p < 0.001$; $g = 5.25–5.70$), anteromedial ($p < 0.001$; $g = 1.11–1.35$), and anterolateral landings ($p < 0.001$; $g = 1.96$). Anteromedial and anterolateral $DPSI_{AP}$ values were also significantly higher than medial and lateral values ($p < 0.001$; $g = 3.92–4.36$). $DPSI_{ML}$ during medial and lateral landings was significantly higher than during anteromedial, anterolateral, and 90° internal and external rotation landings ($p < 0.001$; $g = 0.77–7.86$). Anteromedial and anterolateral $DPSI_{ML}$ values were significantly higher than the two rotational values ($p < 0.001$; $g = 2.65–4.64$). $DPSI_{Total}$ during medial and 90° internal rotation landings was significantly higher than during lateral, anterolateral, and 90° external rotation landings ($p < 0.05$; $g = 0.48–0.88$).

TTS_{Fx} during medial, lateral, and anterolateral landings was significantly longer than during anteromedial and 90° internal and external rotation landings ($p < 0.001$; $g = 0.89–1.58$). TTS_{Fy} during anteromedial, anterolateral, and both rotational landings was significantly longer than during medial and lateral landings ($p < 0.001$; $g = 0.90–1.29$). CoP path length was significantly greater during medial and lateral landings than during anteromedial landing ($p < 0.05$; $g = 0.46–0.76$); no other directional comparisons were statistically significant.

Compared with baseline SLDL, SA-TP was significantly lower during lateral landing ($p < 0.05$; $g = −0.40$) and significantly higher during anteromedial landing ($p < 0.05$; $g = 0.41$; Fig. 6a). SA-FP was significantly lower only during anterolateral landing ($p < 0.01$; $g = −0.57$; Fig. 6b).

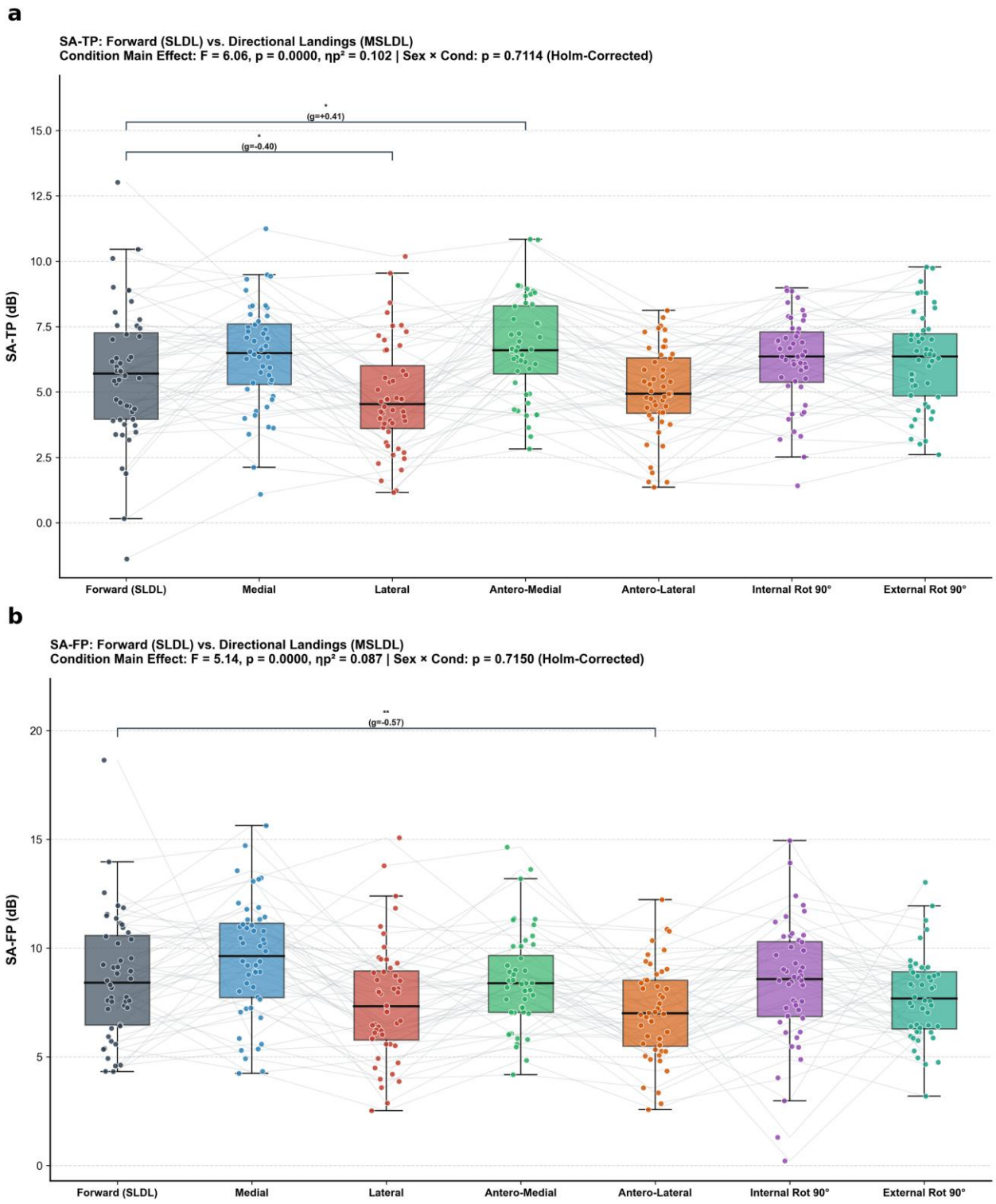

Figure 6. Segmental acceleration attenuation during vertical and multiplanar single-leg drop landings: (a) thigh-to-pelvis attenuation (SA-TP); (b) foot-to-pelvis attenuation (SA-FP). SLDL, baseline vertical single-leg drop landing; MSLDL, multiplanar single-leg drop landing. Plots show comparisons of baseline SLDL with each MSLDL condition; comparisons among the six MSLDL conditions are not displayed. The cohort comprised 48 participants. Pairwise p-values were adjusted using the step-down Holm–Bonferroni procedure: *p < 0.05, **p < 0.01, ***p < 0.001. Values labelled g denote Hedges' g.

The MSLDL-only analysis showed significant condition effects for SA-ST, SA-TP, and SA-FP ($p < 0.05$; $\eta p^2 = 0.06–0.274$). SA-FP during anterolateral landing was significantly lower than during medial, anteromedial, and 90° internal rotation landings ($p < 0.05$; reported $g = 0.48–0.84$). SA-TP during anterolateral landing was

significantly lower than during medial, anteromedial, and 90° internal and external rotation landings ($p < 0.001$; reported $g = 0.70–1.12$).

PL-SP was significantly longer during anteromedial, anterolateral, and 90° internal and external rotation landings than during baseline SLDL ($p < 0.05$; $g = 0.38–0.75$; Fig. 7a). PL-FP was significantly shorter in these conditions than during baseline SLDL ($p < 0.001$; $g = −0.65$ to $−1.44$; Fig. 7b).

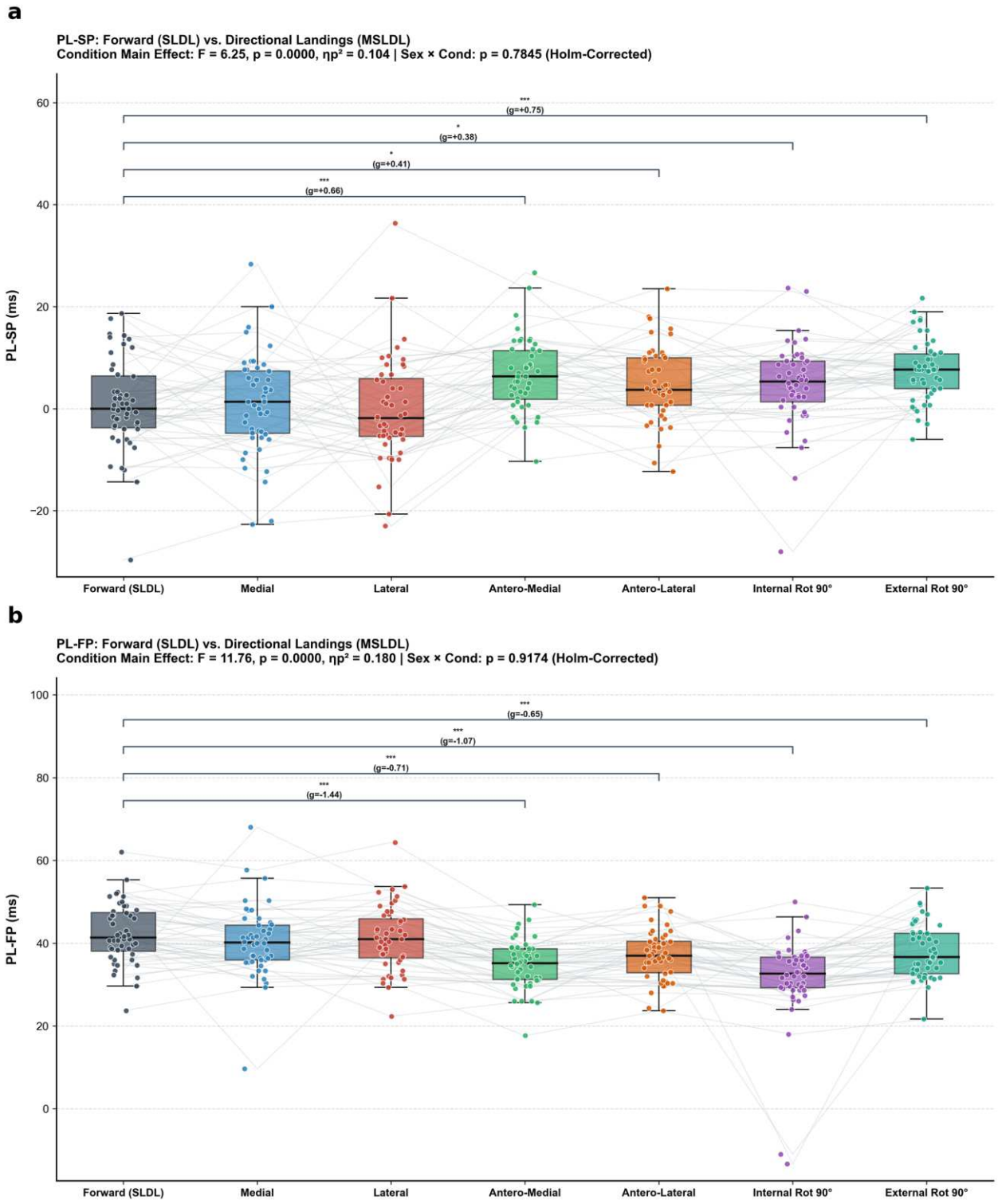

Figure 7. Distal-to-proximal peak-acceleration timing during vertical and multiplanar single-leg drop landings: (a) shank-to-pelvis phase lag (PL-SP); (b) foot-to-pelvis phase lag (PL-FP). SLDL, baseline vertical single-leg drop landing; MSLDL, multiplanar single-leg drop landing. Plots show comparisons of baseline SLDL with each MSLDL condition; comparisons among the six MSLDL conditions are not displayed. The cohort comprised 48 participants. Pairwise p-values were adjusted using the step-down Holm–Bonferroni procedure: *p < 0.05, **p < 0.01, ***p < 0.001. Values labelled g denote Hedges' g.

The MSLDL-only analysis showed significant condition effects for PL-FP and PL-SP ($p < 0.001$; $\eta p^2 = 0.184$–0.265). PL-FP during medial and lateral landings was significantly longer than during anteromedial, anterolateral,

and 90° internal rotation landings ($p < 0.05$; $g = 0.46–1.12$). PL-SP during 90° internal rotation was significantly longer than during medial, lateral, and anterolateral landings ($p < 0.05$; $g = 0.46–0.82$).

4. Discussion

This study compared impact kinetics, dynamic postural stability, and segmental acceleration measures between baseline vertical SLDL and six MSLDL conditions. A secondary analysis restricted to MSLDL examined directional and rotational differences under a common prescribed horizontal displacement. MSLDL tasks generally produced higher peak vGRF and loading rates, together with direction-dependent changes in DPSI and TTS. Changes in segmental acceleration attenuation were limited to selected comparisons, whereas peak-acceleration timing differed between segment pairs. The findings therefore supported the loading-rate and plane-specific stabilization hypotheses more consistently than the attenuation and timing hypotheses. No statistically significant sex × condition interaction was identified.

4.1 Impact Kinetics

Compared with baseline vertical SLDL, MSLDL tasks generally produced greater peak vGRF and vertical loading rates. Shorter TTP vGRF in selected conditions indicated that peak external vertical forces developed over a shorter interval. Because VALR depends on both force magnitude and time to peak, its increase should not be interpreted as an independent mechanical phenomenon. Importantly, baseline SLDL had no prescribed horizontal displacement, whereas MSLDL involved displacement equivalent to 30% of body height. Differences between these scenarios therefore cannot be attributed exclusively to landing direction.

The MSLDL-only comparisons help distinguish directional differences under a common prescribed horizontal displacement, although equal displacement does not guarantee equal landing velocity or momentum. The observed direction-dependent loading is consistent with evidence regarding directional increases in loading rates during unilateral cutting and oblique drop landing tasks [17, 22]. In our study, the 90° internal rotation SLDL produced the highest peak forces and loading rates magnitude than lateral, medial, and diagonal vectors. When the single-leg landing with whole body internal rotation were performed concurrently, ground friction caught and stopped foot rotation upon initial contact and landing phases while residual trunk momentum continued anteriorly and medially, imposing complex multi-axis braking demands before the effective buffering system by multiple lower limb joint flexion (hip, knee and ankle) as well as the neuromuscular control for “anti-rotation” could fully develop. Despite the plausible explanations from rotational braking and whole-body control, joint kinematics, muscle activity, and internal loading were not measured, so these mechanisms remain hypotheses. External GRFs should not be equated with ligament loading or tissue injury risk: running studies demonstrate that GRF metrics may be weakly related to internal tibial loading, while cadaveric landing studies show that muscle forces and knee configuration influence ACL strain [23, 24]. Further work combining kinetics and joint-level measurements is needed before the present external-loading differences can be interpreted as differences in knee injury risk. Nevertheless, strength coaches and clinical practitioners can still pay additional attention on the strength development especially the eccentric phase during particular MSLDL tasks, as well as the motor control to optimize the rapid attenuation and absorption of these observable increase on external load and loading rates.

4.2 Dynamic Postural Stability: Frontal and Transverse Plane Vulnerabilities

The directional DPSI results indicate task-specific stabilization demands rather than a uniform deterioration in balance. $DPSI_{Total}$ increased across MSLDL conditions, but medial and lateral landings increased $DPSI_{ML}$ while reducing $DPSI_{AP}$ relative to vertical SLDL. Conversely, anteroposterior instability was greater during diagonal and rotational tasks. TTS showed a similar directional pattern: medial and lateral landings required longer TTS_{Fx} but shorter TTS_{Fy} than baseline SLDL. These findings agree with earlier evidence that jump direction affects

directional stability indices [14, 15]. The netball pilot study did not identify significant directional differences in TTS [17]. Differences in participant characteristics (well-trained netball players vs. active healthy collegiate individuals), protocols (with shoes vs. bare feet; horizontal displacements: 70 cm vs. normalized with body height), and analysis methods (Different quiet stance threshold and TTS calculation methods) may contribute to these contrasting findings.

Direction-specific braking demands may require different combinations of ankle and proximal-limb control. Multidirectional landing studies have demonstrated changes in ankle kinematics and muscle activity across directions [13]. However, a study of ankle dorsiflexion stiffness and landing stability did not demonstrate significant associations with DPSI or TTS, underscoring the need to avoid attributing stability differences to a single anatomical characteristic [25]. In the present study, medial and lateral landings produced the largest $DPSI_{ML}$ values despite a common prescribed horizontal displacement. These findings support examining task-specific stability profiles, but not yet prescribing injury-prevention thresholds. Horizontal TTS had poor across-trial reliability and may be unsuitable as a standalone screening measure. $DPSI_{ML}$ reliability ranged from poor to good (ICC = 0.235–0.842; mean, 0.619; MDC_{95} = 0.0096–0.0202; mean, 0.0160; CV = 4.85–17.58%; mean, 9.99%). $DPSI_{Total}$ showed good overall reliability (ICC = 0.770–0.872; mean, 0.806; MDC_{95} = 0.0512–0.0596; mean, 0.0543; CV = 4.68–5.36%; mean, 4.98%). These indices may therefore warrant further study for repeated assessment, with condition-specific measurement error considered. Practitioners may consider including DPSI metrics especially those showing plausible across-trial reliability in the testing population, in their routine screening protocol for regular monitoring, biofeedback for landing motor control training or long-term evaluation purposes.

TTS_{Fx} exceeded 3 s during medial, lateral, and anterolateral MSLDL in this healthy cohort. These values do not establish a direction-specific injury-risk threshold. Associations reported for knee loading in prospective ACL studies should not be transferred directly to TTS or DPSI [5]. Comparisons with injured populations and prospective injury surveillance are needed to establish whether the present outcomes have prognostic value.

4.3 Shock Attenuation via Kinetic Chain

IMU-derived acceleration ratios showed task- and segment-specific changes rather than a uniform reduction in attenuation throughout the limb. SA-FP was lower during anterolateral landing, and SA-TP was lower during lateral landing relative to baseline SLDL. SA-ST showed no significant condition effect in the seven-condition analysis, although the MSLDL-only analysis identified a significant directional effect. These findings suggest that acceleration-ratio changes were localized and depended on the conditions included in the analysis.

The decibel measures used here are ratios of independently identified peak segmental acceleration amplitudes; they are not direct measurements of energy absorption, tissue stress, or joint damping. Earlier running studies have used acceleration and frequency-domain analyses to examine impact attenuation, but those approaches should not be treated as interchangeable with the present peak-ratio method [26, 27, 28]. Knee configuration can influence acceleration and force differently, further limiting simple mechanical interpretations [29]. Low across-trial reliability for attenuation measures (ICC = 0.150–0.758; mean, 0.470), especially SA-FP (ICC = 0.150–0.415; mean, 0.286), also requires caution. Sensor attachment and soft-tissue movement can influence accelerometry [20]. CoP path length showed no significant overall condition effect ($p = 0.206$), but this does not establish that CoP is insensitive to injury vulnerability or validate accelerometry as a replacement.

4.4 Distal-to-proximal peak-acceleration timing

Phase lags between distal and proximal acceleration peaks differed across tasks and segment pairs. PL-FP generally shortened during diagonal and rotational tasks, particularly 90° internal rotation and anteromedial landing. PL-SP, in contrast, increased from the small positive baseline value, with 90° internal rotation producing

longer delays than medial and lateral landing. These contrasting findings indicate changes in relative peak timing, not necessarily a uniform change in impact transmission along the limb. The IMUs were sampled at 200 Hz before upsampling, giving a native sampling interval of 5 ms. Interpolation to 1,000 Hz does not provide the temporal resolution of a native 1,000-Hz recording. Small shank-to-pelvis timing differences must therefore be interpreted cautiously. Without synchronized joint kinematics and electromyography, the observed differences cannot establish whether muscle co-contraction, joint stiffness, or independent local peak events were responsible.

4.5 Limitations and Future Directions

Several limitations should be considered. First, landings were performed from a standardized 30-cm platform in preplanned directions. Sport involves more complex interactions among movement, task, and environmental factors [3, 30]. Future studies should compare planned and reactive MSLDL tasks. Second, only healthy recreationally active collegiate athletes were included, limiting generalizability. Direction-dependent deficits reported in athletes with chronic ankle instability or ACL reconstruction support further comparisons with symptomatic cohorts [12, 15]. Third, joint angles, joint moments, and muscle activity were not measured. Combining force-platform data with synchronized motion capture and electromyography would help identify joint- and muscle-level strategies. Fourth, vertical SLDL and MSLDL differed in horizontal displacement, and velocity and momentum were not quantified. Finally, poor reliability for several outcomes and the native 200-Hz accelerometry sampling rate limit their use for individual-level interpretation and fine temporal comparisons.

5. Practical Applications for Sports Injury Prevention

MSLDL may complement vertical landing assessments by exposing participants to additional directional and rotational demands. The present results provide a rationale for further evaluation of medial, lateral, oblique, and rotational landing tasks relevant to an athlete's sport. Practitioners may consider including medial/lateral, oblique, and rotational single-leg landing tasks especially when selected landing directions are specific to the sports-specific demand of the athlete. They may also inform the design of progressive multidirectional landing and deceleration exercises. In particular, because MSLDL conditions impose additional stabilization and rapid force attenuation demands on the lower body, the evaluation, further development, and potential applications of these MSLDL tasks as new SL landing screening tools and as supplementary assessments after lower-limb injury are highly warranted. However, this study did not test a training intervention, injury prevention, or rehabilitation clearance. Validation in injured cohorts and prospective studies is needed before MSLDL outcomes are used to set injury-risk or return-to-sport thresholds. Measures with poor reliability, including TTS, require methodological refinement before individual changes can be interpreted confidently for providing meaningful clinical insights.

Our findings also shed the light that rotational SLDL especially the 90° internal rotational direction can induce additional vertical loading, impact and stabilization demand. Strength coaches and rehabilitation practitioners should therefore consider adopting this as a useful additional task in neuromuscular training programmes aiming at reducing non-contact knee injuries. The greater external loading observed during 90° internal rotation suggests that athletes should progressively develop the ability to control landing and deceleration while simultaneously managing rotational momentum. Such exercises may be particularly relevant in sports involving frequent changes of direction, pivoting, and unilateral landings.

Moreover, the pronounced mediolateral stabilization demands observed during frontal plane SLDL suggest that conventional vertical landing exercises may insufficiently challenge certain aspects of dynamic postural control relevant to both knee ACL tear and lateral ankle sprains. Athletes should consider progressively incorporating medial and lateral SLDL training activities when they are competent in managing the conventional vertical SLDL

training. These frontal-based SLDL may improve the specificity of neuromuscular adaptation, particularly in athletes participating in sports demanding on rapid lateral movements.

Last but not the least, the differing patterns of impact loading and directional postural stability indicate that these characteristics should not be considered interchangeable. Consequently, injury prevention programs may benefit from addressing rapid impact attenuation and multidirectional postural control as complementary neuromuscular abilities, rather than assuming that improvements in one automatically translate into improvements in the other.

6. Conclusion

Multiplanar single-leg drop landings produced task-specific differences in peak vertical force, loading rates, dynamic postural stability, and stabilization time relative to vertical SLDL. The MSLDL-only analysis also identified directional and rotational differences under a common prescribed horizontal displacement. Acceleration attenuation and peak-timing responses were segment-specific and less consistent. These findings support further development and validation of multiplanar landing assessments, but they do not establish injury-prediction thresholds, rehabilitation clearance criteria, or prevention benefits.

Data availability

The raw data and analysis code generated during the current study are available from the corresponding author upon reasonable academic request.

Declarations

Ethics Approval and Consent to Participate: The study protocol was approved by the Institutional Review Board and Human Subjects Ethics Committee of Technological and Higher Education Institution of Hong Kong (HSEC Ref. No:HE2025-01). Written informed consent was obtained from all individual participants prior to participation.

Consent for Publication

Not applicable.

Competing Interests

The authors declare that they have no competing financial or non-financial interests.

Funding

Nil.

Acknowledgement

The author specially thanks Ms Annie Chan for her efforts and assistance in the data collection processes.

Authors' Contributions

Conceptualization: [Indy M.K. Ho, Daniel H.K. Chow, Lana Ružić]; Methodology: [Indy M.K. Ho, Daniel H.K. Chow]; Writing – Original Draft Preparation: [Indy M.K. Ho]; Writing – Review & Editing: [Indy M.K. Ho,

Daniel H.K. Chow, Lana Ružić]; Supervision: [Indy M.K. Ho, Daniel H.K. Chow]. All authors read and approved the final manuscript.

Use of Generative AI and AI-Assisted Technologies: During the preparation of this work, the authors utilized Gemini (Google) to assist with Python scripting syntax (specifically implementing statistical analyses and visualization). During manuscript preparation, the authors used Perplexity AI to assist with English-language editing, condensation and restructuring of text, identification of potential citation inconsistencies, reference formatting, and figure arrangement. The authors reviewed and verified the resulting revisions and references and take full responsibility for the final manuscript.

References

1. Chia, L. et al. Non-contact anterior cruciate ligament injury epidemiology in team-ball sports: A systematic review with meta-analysis by sex, age, sport, participation level, and exposure type. *Sports Med.* **52,** 2447–2467 (2022). https://doi.org/10.1007/s40279-022-01697-w

2. Mineta, S., Koyama, T., Yamaguchi, S., Inagaki, H. & Sekine, Y. Epidemiology of lateral ankle sprain focusing on indirect contact mechanism in male and female soccer players: An 18.5-month cohort study. *Injury* **55,** 111699 (2024). https://doi.org/10.1016/j.injury.2024.111699

3. Boden, B. P., Sheehan, F. T., Torg, J. S. & Hewett, T. E. Noncontact anterior cruciate ligament injuries: Mechanisms and risk factors. *J. Am. Acad. Orthop. Surg.* **18,** 520–527 (2010). https://doi.org/10.5435/00124635-201009000-00003

4. Koga, H. et al. Mechanisms for noncontact anterior cruciate ligament injuries: Knee joint kinematics in 10 injury situations from female team handball and basketball. *Am. J. Sports Med.* **38,** 2218–2225 (2010). https://doi.org/10.1177/0363546510373570

5. Hewett, T. E. et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: A prospective study. *Am. J. Sports Med.* **33,** 492–501 (2005). https://doi.org/10.1177/0363546504269591

6. Quatman, C. E. & Hewett, T. E. The anterior cruciate ligament injury controversy: Is "valgus collapse" a sex-specific mechanism?. *Br. J. Sports Med.* **43,** 328–335 (2009). https://doi.org/10.1136/bjsm.2009.059139

7. Hewett, T. E., Torg, J. S. & Boden, B. P. Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: Lateral trunk and knee abduction motion are combined components of the injury mechanism. *Br. J. Sports Med.* **43,** 417–422 (2009). https://doi.org/10.1136/bjsm.2009.059162

8. Ross, S. E. & Guskiewicz, K. M. Examination of static and dynamic postural stability in individuals with functionally stable and unstable ankles. *Clin. J. Sport Med.* **14,** 332–338 (2004). https://doi.org/10.1097/00042752-200411000-00002

9. Sell, T. C. An examination, correlation, and comparison of static and dynamic measures of postural stability in healthy, physically active adults. *Phys. Ther. Sport* **13,** 80–86 (2012). https://doi.org/10.1016/j.ptsp.2011.06.006

10. Wikstrom, E. A., Tillman, M. D., Smith, A. N. & Borsa, P. A. A new force-plate technology measure of dynamic postural stability: The dynamic postural stability index. *J. Athl. Train.* **40,** 305–309 (2005).

11. Wang, L. I. The lower extremity biomechanics of single- and double-leg stop-jump tasks. *J. Sports Sci. Med.* **10,** 151–156 (2011).

12. Kawaguchi, K. et al. Dynamic postural stability is decreased during the single-leg drop landing task in male collegiate soccer players with chronic ankle instability. *Orthop. J. Sports Med.* **10,** 23259671221107343 (2022). https://doi.org/10.1177/23259671221107343

13. Sonsukong, A. et al. Ankle biomechanics during multidirectional landings in athletes with chronic ankle instability. *Foot Ankle Spec.* **17,** 249–258 (2024). https://doi.org/10.1177/19386400231208522

14. Wikstrom, E. A., Tillman, M. D., Schenker, S. M. & Borsa, P. A. Jump-landing direction influences dynamic postural stability scores. *J. Sci. Med. Sport* **11,** 106–111 (2008). https://doi.org/10.1016/j.jsams.2007.02.014

15. Zhang, X., Ruan, B. & Gao, Q. Dynamic postural stability indices in athletes: A case-control study on chronic ankle instability during multi-directional landing assessments. *BMC Musculoskelet. Disord.* **26,** 1093 (2025). https://doi.org/10.1186/s12891-025-09307-2

16. Fransz, D. P., Huurnink, A., de Boode, V. A., Kingma, I. & van Dieën, J. H. Time to stabilization in single leg drop jump landings: An examination of calculation methods and assessment of differences in sample rate, filter settings and trial length on outcome values. *Gait Posture* **41,** 63–69 (2015). https://doi.org/10.1016/j.gaitpost.2014.08.018

17. Jolingana-Seoka, T. T., Hammill, H. V., Willemse, Y. & Kramer, M. The relationship between ankle joint kinematics and impact forces during unilateral jump-landing tasks in university-level netball players: A pilot study. *Appl. Sci.* **13,** 9934 (2023). https://doi.org/10.3390/app13179934

18. Benson, L. C., Räisänen, A. M., Clermont, C. A. & Ferber, R. Is this the real life, or is this just laboratory? A scoping review of IMU-based running gait analysis. *Sensors* **22,** 1722 (2022). https://doi.org/10.3390/s22051722

19. Tedesco, S. et al. A wearable system for the estimation of performance-related metrics during running and jumping tasks. *Appl. Sci.* **11,** 5258 (2021). https://doi.org/10.3390/app11115258

20. Sheerin, K. R., Reid, D. & Besier, T. F. The measurement of tibial acceleration in runners—A review of the factors that can affect tibial acceleration during running and evidence-based guidelines for its use. *Gait Posture* **67,** 12–24 (2019). https://doi.org/10.1016/j.gaitpost.2018.09.017

21. Nardon, M. et al. Kinematics but not kinetics alterations to single-leg drop jump movements following a subject-tailored fatiguing protocol suggest an increased risk of ACL injury. *Front. Sports Act. Living* **6,** 1418598 (2024). https://doi.org/10.3389/fspor.2024.1418598

22. Dai, B. et al. The effects of 2 landing techniques on knee kinematics, kinetics, and performance during stop-jump and side-cutting tasks. *Am. J. Sports Med.* **43,** 466–474 (2015). https://doi.org/10.1177/0363546514555322

23. Matijevich, E. S., Branscombe, L. M., Scott, L. R. & Zelik, K. E. Ground reaction force metrics are not strongly correlated with tibial bone load when running across speeds and slopes: Implications for science, sport and wearable tech. *PLoS ONE* **14,** e0210000 (2019). https://doi.org/10.1371/journal.pone.0210000

24. Withrow, T. J., Huston, L. J., Wojtys, E. M. & Ashton-Miller, J. A. The relationship between quadriceps muscle force, knee flexion, and anterior cruciate ligament strain in an in vitro simulated jump landing. *Am. J. Sports Med.* **34,** 269–274 (2006). https://doi.org/10.1177/0363546505280906

25. Hatefi, M., Biglari, M. & Babakhani, F. Is there a correlation among landing stability, ankle dorsiflexion range of motion, and ankle stiffness during single-leg landing?. *BMC Sports Sci. Med. Rehabil.* **16,** 243 (2024). https://doi.org/10.1186/s13102-024-01035-9

26. Derrick, T. R., Hamill, J. & Caldwell, G. E. Energy absorption of impacts during running at various stride lengths. *Med. Sci. Sports Exerc.* **30,** 128–135 (1998). https://doi.org/10.1097/00005768-199801000-00018

27. Derrick, T. R., Dereu, D. & McLean, S. P. Impacts and kinematic adjustments during an exhaustive run. *Med. Sci. Sports Exerc.* **34,** 998–1002 (2002). https://doi.org/10.1097/00005768-200206000-00015

28. Gruber, A. H., Boyer, K. A., Derrick, T. R. & Hamill, J. Impact shock frequency components and attenuation in rearfoot and forefoot running. *J. Sport Health Sci.* **3,** 113–121 (2014). https://doi.org/10.1016/j.jshs.2014.03.004

29. Derrick, T. R. The effects of knee contact angle on impact forces and accelerations. *Med. Sci. Sports Exerc.* **36,** 832–837 (2004). https://doi.org/10.1249/01.mss.0000126779.65353.cb

30. Tsarbou, C. et al. ACL injury etiology in its context: A systems thinking, group model building approach. *J. Clin. Med.* **13,** 4928 (2024). https://doi.org/10.3390/jcm13164928

Table 1. Impact kinetics, dynamic postural stability, and segmental acceleration measures during baseline vertical SLDL and pooled MSLDL.

Values are mean ± SD. SLDL, baseline vertical single-leg drop landing; MSLDL, multiplanar single-leg drop landing, pooled across the six directional and rotational conditions. Δ and percentage differences describe pooled MSLDL relative to SLDL. F-values, p-values, and partial η^2 refer to the omnibus landing-condition effect across all seven conditions in the mixed-design ANOVA; they are not tests of the pooled MSLDL–SLDL contrast. No statistically significant sex × condition interaction was reported (all $p > 0.15$); descriptive values are pooled across 48 participants (24 males and 24 females). Partial η^2 thresholds were ≥ 0.01 (small), ≥ 0.06 (medium), and ≥ 0.14 (large). Percentage changes in dB measures or in phase lags close to zero should be interpreted cautiously. Numerical entries are retained as supplied pending author verification.

Biomechanical Variable	Domain	Vertical SLDL (Mean ± SD)	Pooled MSLDL (Mean ± SD)	Mean Diff (Δ)	% Diff	Condition F-value	p-value	Partial η^2
Peak vGRF (BW)	Impact Kinetics	3.26 ± 0.63	3.80 ± 0.76	+0.54	+16.5%	5.42	**< 0.001**	0.092
TTP vGRF (s)	Impact Kinetics	0.055 ± 0.012	0.048 ± 0.012	-0.007	-13.6%	21.40	**< 0.001**	0.285
VALR (BW/s)	Impact Kinetics	63.98 ± 25.93	87.67 ± 32.39	+23.68	+37.0%	15.08	**< 0.001**	0.219
VILR (BW/s)	Impact Kinetics	168.77 ± 85.56	217.05 ± 97.38	+48.27	+28.6%	4.67	**< 0.001**	0.080
$DPSI_{AP}$ (Index)	Dynamic Stability	0.074 ± 0.008	0.089 ± 0.011	+0.015	+19.6%	539.79	**< 0.001**	0.910
$DPSI_{ML}$ (Index)	Dynamic Stability	0.023 ± 0.005	0.073 ± 0.011	+0.050	+215.9%	673.93	**< 0.001**	0.926
$DPSI_{Total}$ (Index)	Dynamic Stability	0.264 ± 0.040	0.342 ± 0.046	+0.078	+29.5%	27.52	**< 0.001**	0.339
TTS_{Fx} (s)	Dynamic Stability	1.66 ± 0.76	2.75 ± 0.46	+1.09	+65.8%	68.28	**< 0.001**	0.560
TTS_{Fy} (s)	Dynamic Stability	2.85 ± 0.24	2.93 ± 0.30	+0.08	+2.9%	33.41	**< 0.001**	0.384
CoP Length (mm)	Postural Sway	786.89 ± 144.79	808.68 ± 156.82	+21.80	+2.8%	1.42	0.206	0.026
SA-ST (dB)	Kinematic Attenuation	0.08 ± 1.16	-0.15 ± 1.30	-0.23	-298.5%	1.42	0.206	0.026
SA-TP (dB)	Kinematic Attenuation	5.68 ± 2.85	5.88 ± 2.18	+0.20	+3.5%	6.06	**< 0.001**	0.102
SA-FP (dB)	Kinematic Attenuation	8.62 ± 3.75	8.11 ± 3.43	-0.51	-5.9%	5.14	**< 0.001**	0.087
PL-FP (ms)	Kinematic Phase Lag	42.32 ± 8.18	37.15 ± 10.27	-5.17	-12.2%	11.76	**< 0.001**	0.180
PL-SP (ms)	Kinematic Phase Lag	0.77 ± 11.58	4.04 ± 11.36	+3.27	+423.8%	6.25	**< 0.001**	0.104