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ARTICLE

Reliability of two-dimensional measures associated with bilateral drop-landing performance

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Abstract – The aim of this study was to establish the within-session reliability for two-dimensional (2D) video analysis of sagittal- and frontal-plane measures during bilateral drop-landing tasks. Thirty-nine recreational athletes (22 men, 17 women, age = 22 ± 4 years, height = 1.74 ± 0.15 m, body mass 70.2 ± 15.1 kg) performed five bilateral drop-landings from 50, 100 and 150% of maximum countermovement jump height, twice on the same day. Measures of reliability for initial contact angle, peak flexion angle and joint displacement for the hip, knee, and ankle joints, frontal-plane projection angles (FPPA), as well as inter-limb asymmetries in joint displacement were assessed. No systematic bias was present between trials ($P>0.05$). All kinematic measurements showed relative reliability ranging from large to near perfect (ICC = 0.52–0.96). Absolute reliability ranged between measures, with CV% between 1.0–1.6% for initial contact angles, 1.9–7.9% for peak flexion angles, 5.3–22.4% for joint displacement, and 1.6–2.3% for FPPA. Absolute reliability for inter-limb asymmetries in joint displacement were highly variable, with minimal detectable change values ranging from 6.0–13.2°. Therefore, 2D video analysis is a reliable tool for numerous measures related to the performance of bilateral drop-landings.

Keywords: within-session reliability, kinematics, landings

Résumé – Fiabilité des mesures bidimensionnelles associées aux performances d'atterrissage en chute bilatérale. Le but de cette étude était d'établir la fiabilité intra-session pour l'analyse vidéo bidimensionnelle (2D) de mesures sur le plan sagittal et frontal lors de tâches d'atterrissage en chute libre bilatérales. Trente-neuf sportifs sportifs (22 hommes et 17 femmes, âge = 22 ± 4 ans, taille = 1,74 ± 0,15 m, masse corporelle 70,2 ± 15,1 kg) ont effectué cinq atterrissages bilatéraux à partir de 50, 100 et 150 % du maximum hauteur du saut en contre-mouvement, deux fois le même jour. Mesures de fiabilité pour l'angle de contact initial, l'angle de flexion maximal et le déplacement articulaire pour les articulations de la hanche, du genou et de la cheville, les angles de projection dans le plan frontal (FPPA), ainsi que les asymétries inter-membres dans le déplacement articulaire. Aucun biais systématique n'était présent entre les essais ($p>0,05$). Toutes les mesures cinématiques ont montré une fiabilité relative allant de grande à quasi parfaite (ICC = 0,52–0,96). La fiabilité absolue variait d'une mesure à l'autre, avec des CV% compris entre 1,0 et 1,6 % pour les angles de contact initiaux, entre 1,9 et 7,9 % pour les angles de flexion maximaux, entre 5,3 et 22,4 % pour les déplacements articulaires et entre 1,6 et 2,3 % pour les FPPA. La fiabilité absolue pour les asymétries inter-membres dans le déplacement articulaire était très variable, avec des valeurs de changement détectables minimales allant de 6,0 à 13,2°. Par conséquent, l'analyse vidéo 2D est un outil fiable pour de nombreuses mesures liées à la performance des atterrissages bilatéraux.

Mots clés : fiabilité intra-session, cinématique, atterrissages

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1 Introduction

Jump landings expose athletes to large peak vertical forces that require attenuation during sporting activities (Chappell *et al.*, 2005). Landings have been identified as a mechanism for lower-extremity injuries in athletes during sport participation (Hewett, Myer, & Ford, 2006). Athletes at greater risk of injury during landing tasks tend to use less effective movement strategies to dissipate forces in multiple planes (Boling *et al.*, 2009; Hewett *et al.*, 2005; Padua *et al.*, 2009). For example, in the sagittal-plane, decreased knee flexion (Chappell *et al.*, 2005) and ankle plantarflexion angle at initial contact (Rowley & Richards, 2015), reduced hip (Blackburn & Padua, 2009) and knee flexion angle at the lowest point of the landing (Yu, Lin, & Garrett, 2006), and less ankle joint displacement following ground contact (Begalle *et al.*, 2015) have all been shown to increase mechanical loading throughout the lower extremity. In the frontal- and transverse-plane, greater peak knee valgus angle during landing tasks has also been shown to increase lower-extremity injury risk, secondary to higher knee abduction moments increasing the loading placed on passive structures at the tibiofemoral joint (Hewett *et al.*, 2005). Given their established relationship with risk of injury, it is common practice to pre-screen the movement strategies selected by athletes (Tran *et al.*, 2015).

Although three-dimensional (3D) motion analysis is regarded as the gold standard in exploring lower limb kinematics, in practice two-dimensional (2D) video analysis is more accessible to practitioners (Munro, Herrington, & Carolan, 2012). However, before kinematic measurements of bilateral landing tasks can be used for the purpose of screening, their reliability must first be established. It is therefore important to quantify the noise (error) of the proposed field-based measurements. For various landing tasks, 2D video analysis has been shown to be a valid tool for measuring frontal-plane projection angle (FPPA), significantly relating to measurements of knee abduction angle ($r = -0.38$) and external knee abduction moment ($r = -0.59$) using 3D motion analysis (Mizner, Chmielewski, Toepke, & Tofte, 2012). Furthermore, FPPA provides a reliable representation of knee valgus/varus angle in the deepest landing position (Dingenen, Malfait, Vanrenterghem, Verschueren, & Staes, 2014; McLean *et al.*, 2005; Mizner *et al.*, 2012; Munro *et al.*, 2012) and is a valid measure of frontal-plane knee mechanics during landings when compared to 3D analysis. However, for joint angle measurements in the sagittal-plane, only Dingenen *et al.* (2015) and King and Belyea (2015) have investigated the reliability of 2D analysis for measurements of bilateral landing activities. In all of these investigations, only peak angles for the hip, knee and ankle joints were measured. At present, studies investigating the reliability of 2D analysis have not considered other variables that may impact load dissipation during landings, such as initial contact angles and joint displacement for the hip, knee, and ankle joints (Begalle *et al.*, 2015; Chappell *et al.*, 2005; Rowley & Richards, 2015). Furthermore, there has been no investigation of the reliability of 2D kinematic measures during a bilateral drop-landing, a screening tool commonly used in practice (Bird & Markwick, 2016; Tran *et al.*, 2015).

An additional consideration when analysing kinematic measures associated with bilateral drop-landings is asymmetries in coordination. Asymmetry in landing strategies commonly occur during bilateral landing tasks in uninjured (Schot *et al.*, 1994) and injured populations (Meyer, Gette, Mouton, Seil, & Theisen, 2018). Practitioners may attempt to determine asymmetries in kinematic variables associated with landing performance, as individuals who exhibit large asymmetries during bilateral landings may expose one leg to excessive loading relative to the contralateral limb (Schot *et al.*, 1994). However, the test re-test reliability for 2D video analysis to detect inter-limb asymmetries has not been established for kinematic parameters of drop-landings.

The aim of this investigation, therefore, was to assess the reliability of kinematic measures using 2D video analysis during bilateral drop-landings across a range of heights.

2 Methods

2.1 Participants

Thirty-nine recreational athletes volunteered for this study, consisting of 22 men (age = 23 ± 5 years; height = 1.80 ± 0.6 m; mass = 77.9 ± 14.0 kg) and 17 women (age = 20 ± 4 years; height = 1.6 ± 0.9 m; mass = 60.3 ± 9.8 kg) with mean values for maximum countermovement jump (CMJ) height of 0.34 ± 0.07 m and 0.24 ± 0.05 m, respectively. Participants were excluded if they had a previous history of lower-extremity or spinal surgery or had incurred a lower-extremity injury 6 months prior to testing. Participants were informed of the risks associated with testing, completed a pre-exercise questionnaire and signed an informed consent form before testing. Ethical approval was obtained by the Institutional Research Ethics Panel of the lead author.

2.2 Test procedures

A within-session repeated measures design was used, with participants reporting to the university laboratory for a single testing session. All test sessions were conducted between 10:00 am and 1:00 pm to control for circadian variation. All participants wore tight-fitting shorts and vest so that key landmarks were recognisable by all cameras. Anthropometric data was collected prior to completing a standardised warm-up routine consisting of a 5-min jog and dynamic stretches including sumo squats, forward lunges, mountain climbers and leg swings for 10 repetitions. Participants were then familiarised with performing a CMJ. For the CMJ, participants stood bare feet with a hip-width stance with each foot placed on a separate portable force platform recording at 1000 Hz (Pasco, Roseville, CA, USA). Each force platform was positioned side-by-side, 0.05 m apart and embedded in custom-built wooden mounts that were level with the force platforms and did not allow any extraneous movement by the force platforms during the landing. Participants' hands were placed on their hips and remained in this position throughout the jump to isolate the contribution from the lower-extremity. Participants were then asked to rapidly descend prior to explosively jumping as high as possible, with no control being placed on the depth or duration of the countermovement (Benjanuvatra, Lay, Alderson, & Blanksby, 2013). For data collection, three maximal effort CMJs were performed, with 60 s recovery between attempts. Following the final CMJ, force-time data were analysed using the following equation (Moir, 2008) to calculate jump height to the nearest cm:

$$\text{Time in the air jump height(cm)} = {}^{1}\!\Big/_{2}\mathrm{g}(\mathrm{t}/2)^2,$$

where g represents the acceleration of gravity ($9.81\ \mathrm{m/s^2}$) and t represents the time in the air (s). Time in the air was determined as the period where force was less than 10 N. Using a custom-made Microsoft Excel spreadsheet, the maximum value of the three attempts was then used to calculate box height for the bilateral drop-landings.

Following the performance of the CMJ, reflective markers were placed directly onto the participants' skin by the same investigator using the anatomical locations for sagittal-plane lower-extremity joint movements and frontal-plane projection angle (FPPA) outlined by Dingenen *et al.* (2015) and Munro *et al.* (2012), respectively. For sagittal-plane views, reflective markers were placed on both left and right acromioclavicular joints, greater trochanters, lateral femoral condyles, lateral malleolus and 5th metatarsal heads (Dingenen *et al.*, 2015). Frontal-plane projection angle was calculated for the right knee joint only, with reflective markers placed at the centre of the right knee joint (midpoint between the femoral condyles), centre of the right ankle joint (midpoint between the malleoli) joint and on the proximal right thigh (midpoint between the anterior superior iliac spine and the knee marker). Midpoints for the knee and ankle were measured with a standard tape measure (Seca 201, Seca, United Kingdom), as outlined by Munro *et al.* (2012).

Participants then repeated the standardised warm-up before being familiarised with the bilateral drop-landings from drop heights of 50, 100, and 150% of maximum CMJ height. For familiarisation, participants performed bilateral drop-landings from each drop height. Familiarisation ceased once participants indicated they were comfortable with the technique and procedure. Bilateral drop-landings were performed with participants standing bare foot with their arms folded across their chest on a height-adjustable platform (to the nearest 0.01 m). Participants were then instructed to step off the platform, leading with the right leg, before immediately bringing the left leg off and alongside the right leg prior to impact with the ground. During this manoeuvre, participants were instructed to ensure that they did not modify the height of the centre of mass prior to dropping from the platform (James, Bates, & Dufek, 2003). To provide participants with a reference point for landing and to ensure landings were in full view of the video cameras, two force platforms were positioned 0.15 m away from the elevated platform (Munro *et al.*, 2012). Participants were instructed to "*land as softly as possible with both feet contacting the force platforms simultaneously and with equal weight distribution before returning to a standing position*". This instruction was provided to allow for focus of attention to be controlled between trials (Milner, Fairbrother, Srivatsan, & Zhang, 2012). No feedback on landing performance was provided at any point during testing. For each drop height, participants performed five landings for data collection, with 60 s recovery provided between landings. Following the performance of the initial five landings from each drop height (test 1), participants rested for 10 min prior to repeating the standardised warm-up and the bilateral drop-landing protocol (test 2). Drop height order was randomised using a counterbalanced design for both test 1 and 2. Mean values for all variables using all five trials were calculated for tests 1 and 2. Five trials were used to calculate the mean based on previous investigations demonstrating a plateau in measures of reliability for landing kinematics when >4 trails were used for data analysis (Ortiz *et al.*, 2007).

For 2D video analysis, sagittal- and frontal-plane joint movements were recorded using three standard digital video cameras sampling at 60 Hz (Panasonic HX-WA30). All cameras were set up using the procedures outlined by Payton (2007). For left and right sagittal-plane joint movements, cameras were positioned 3.5 m from the centre of either force platform (Dingenen *et al.*, 2015). To record frontal-plane kinematics, a camera was placed 3.5 m in front of the centre of the force platforms (Dingenen *et al.*, 2014). All cameras were placed on a tripod at a height of 0.60 m from the ground (Dingenen *et al.*, 2014, 2015).

2.3 Data analysis

All video recordings were analysed with free downloadable software (Kinovea for Windows, Version 0.8.15). For sagittal-plane joint movements, hip flexion, knee flexion and ankle dorsiflexion angles were calculated at initial contact and the maximum flexion point deepest landing position for both limbs. These angles were then used to calculate joint displacement for each joint by subtracting the initial contact angle from the maximum flexion point. Initial contact was defined as the frame prior to visual impact between the foot and the ground that led to deformation of the foot complex. The maximum flexion point was identified visually and defined as the frame where no more downward motion occurred at the hip, knee, or ankle joints (Dingenen *et al.*, 2015). Intra-rater reliability for identifying the moment of maximum flexion as a reference point for peak joint angles during landing was performed using the first trial from a drop height of 100% of CMJ height for 20 randomly selected participants (13 males and 7 females). Videos were examined twice by the same investigator, seven days apart. Intra-class correlation coefficients (ICC) for time at the maximum flexion point were 0.99 and the standard error of measurement (SEM) were 0.01 s. Hip flexion angle was calculated as the angle between a line formed between the acromioclavular joint and the greater trochanter and a line between the greater trochanter and the femoral condyle. Knee flexion angle was calculated as the angle between a line formed between the greater trochanter and the femoral condyle and a line between the femoral condyle and the lateral malleolus. Ankle dorsiflexion angle was calculated as the angle between a line formed between the femoral condyle and the lateral malleolus and a line between the lateral malleolus and the 5th metatarsal head. Frontal-plane projection angle was calculated for the right limb at the deepest landing position, defined as the frame corresponding to maximum knee flexion (Munro *et al.*, 2012). This angle was calculated as the angle between the line formed between the proximal thigh marker and the knee joint marker and a line between the knee joint marker and the ankle joint marker (Munro *et al.*, 2012). For initial contact and the maximum flexion point, smaller values represented greater hip flexion, knee flexion and ankle dorsiflexion for the hip, knee and ankle joints, respectively. For FPPA, values $< 180°$ represented knee valgus and values $> 180°$ represented knee varus.

Between-limb differences for sagittal-plane joint displacement was calculated by subtracting the left value from the right value for the ankle, knee and hip joints. A positive value indicated the right limb had greater joint displacement for the corresponding segment and *vice versa* for a negative value.

2.4 Statistical analysis

Descriptive statistics (means ± standard deviation) were calculated for initial contact angles, peak flexion angle at the maximum flexion point and joint displacement for the right limb, along with between-limb differences for joint displacement. The assumption of normality was checked using the Shapiro-Wilk test. To account for heteroscedastic errors, the relationship between the mean values between tests and the difference between repeat tests was evaluated using Pearson's correlation coefficient. To establish systematic bias between tests 1 and 2, mean values for initial contact angle, peak flexion angles, joint displacement, FPPA, and between-limb differences in sagittal-plane joint displacement was initially assessed using a paired samples *t*-test (Atkinson & Nevill, 1998). The α*-priori* level of significance was set at $P < 0.05$, with a Bonferroni correction applied post-hoc. Relative reliability was determined using ICC as described by Hopkins (2018a) and reported with 95% confidence intervals, with ICCs interpreted as follows: 0.01–0.3 poor, 0.3–0.5 moderate, 0.5–0.7 large, 0.7–0.9 very large, and > 0.9 nearly perfect (Hopkins, 2018a). Absolute reliability was calculated using the coefficient of variation (CV%), the 95% limits of agreement (LOA), SEM (SD$\sqrt{}$1-ICC) (Atkinson & Nevill, 1998) and minimal detectable change (MDC; SEM*1.96*$\sqrt{}$2) (Riemann & Lininger, 2018). Due to the asymmetry in joint displacement being interval data, CV% was not calculated. ICC and CV% were calculated using a customised spreadsheet (Hopkins, 2018b). The CV% was used as the primary measure of absolute reliability but we have reported a variety of statistical interpretations to facilitate interpretation of the results by researchers and practitioners. All statistical tests were performed using SPSS® statistical software package (v.24; SPSS Inc., Chicago, IL, USA).

3 Results

There was no systematic bias found between tests 1 and 2 for any variable for any drop height. Relative reliability ranged from very large to near perfect (ICC = 0.87–0.93) and CV% for initial contact variables ranged from 1.0–1.6% across all drop heights. For peak angles at the maximum flexion point, relative reliability was near perfect (ICC = 0.92–0.95) and absolute reliability ranged between 1.9–7.9% for CV% for the hip, knee and ankle joints, along with FPPA for all drop heights. Relative reliability for joint displacement ranged from very large to near perfect (ICC = 0.76–0.96). At drop heights of 50% CMJ height, greater absolute variability was identified for joint displacement values (CV% = 10.0–22.4%), but at a drop height of 100% CMJ height, joint displacements values all possessed CV% < 10%. However, at drop heights of 150% of CMJ height, joint displacement for the hip exceeded CV%>10%. Relative reliability for between-limb difference sin sagittal-plane joint displacement ranged from large to very large (ICC = 0.50–0.84) with MDC values ranging between 6.0–13.2°.

4 Discussion

The primary aim of this investigation was to determine the within-session reliability of kinematic variables using 2D video analysis during bilateral drop-landings from drop heights equating to 50, 100, and 150% of an individual's maximum CMJ height. As part of our investigation, we identified no systematic bias, indicating no evidence of a learning effect, participant bias, or acute adaptations in movement strategies between tests using a within-session design (Atkinson & Nevill, 1998). With large to near perfect ICC values and CV% ranging between 1.0–22.4%, our findings suggest that 2D video analysis is sufficiently reliable to determine typical changes in landing kinematics following training or therapeutic interventions during bilateral drop-landings for most variables, although variability in error will be influenced by the kinematic measurement analysed and the drop height. Previously, 2D video analysis has been validated against 3D motion analysis for both sagittal- and frontal-plane lower extremity peak joint angles during landing tasks (Dingenen *et al.*, 2014, 2015; McLean *et al.*, 2005; Mizner *et al.*, 2012). In conjunction with the findings of our investigation, 2D video analysis is therefore a viable tool for practitioners when assessing bilateral drop-landing mechanics. However, the reliability values presented in

Table 1. Within-session reliability for all dependant variables for bilateral drop-landing from a drop height equalling 50% CMJ height ($n = 39$).

	Test 1 Mean ± SD (°)	Test 2 Mean ± SD (°)	Change in mean (°)	95% LOA (°)	ICC (95% CI)	CV (%)	SEM (°)	MDC (°)
Initial contact angles								
Ankle plantarflexion	148.6 ± 6.9	147.6 ± 7.5	−0.9	0.9 ± 6.5	0.90 (0.82–0.95)	1.6	2.3	6.3
Knee flexion	169.4 ± 5.0	168.4 ± 5.6	−1.0	1.0 ± 4.6	0.91 (0.83–0.95)	1.0	1.6	4.5
Hip flexion	161.6 ± 7.0	161.0 ± 7.7	−0.6	0.6 ± 6.6	0.90 (0.82–0.95)	1.5	2.3	6.5
Peak angles at maximum flexion point								
Ankle dorsiflexion	105.5 ± 9.7	104.7 ± 8.9	−0.7	0.7 ± 6.7	0.94 (0.88–0.97)	2.3	2.3	6.5
Knee flexion	117.6 ± 17.3	117.0 ± 16.7	−0.6	0.6 ± 11.2	0.95 (0.90–0.97)	3.7	3.9	10.9
Hip flexion	127.1 ± 24.0	126.6 ± 24.6	−0.5	0.5 ± 18.5	0.93 (0.87–0.96)	5.6	6.5	18.0
Frontal plane projection angle	184.4 ± 10.7	184.2 ± 10.8	−0.1	0.1 ± 7.7	0.94 (0.88–0.97)	1.6	2.7	7.5
Joint displacement								
Ankle dorsiflexion	43.1 ± 7.5	42.2 ± 9.1	−1.0	1.0 ± 11.5	0.76 (0.59–0.87)	15.5	4.1	11.3
Knee flexion	51.8 ± 14.2	51.4 ± 14.1	−0.4	0.4 ± 11.6	0.92 (0.85–0.96)	10.0	4.1	11.3
Hip flexion	34.4 ± 19.6	34.3 ± 20.1	−0.1	0.1 ± 15.6	0.92 (0.86–0.96)	22.4	5.5	15.2
Asymmetries in joint displacement								
Ankle dorsiflexion displacement	11.7 ± 7.6	9.9 ± 10.1	−1.8	1.8 ± 13.4	0.72 (0.56–0.83)	N/A	4.8	13.2
Knee flexion displacement	10.3 ± 6.2	9.5 ± 7.0	−0.9	0.9 ± 8.8	0.78 (0.65–0.86)	N/A	3.1	8.7
Hip flexion displacement	6.2 ± 4.2	5.8 ± 5.3	−0.4	0.4 ± 6.1	0.80 (0.67–0.80)	N/A	2.1	6.0

this study may not be directly applicable to all populations (i.e. elite athletes). As such, practitioners should attempt to determine the reliability for these variables relative to the population being assessed.

Our findings show that initial contact angles for both limbs can be reliably measured using 2D video analysis, with ICCs ranging from 0.87–0.93 and CV% between 1.0–1.6% across all drop heights (Tables 1–3). Previously, SEM values for establishing sagittal-plane knee and hip angles at initial contact using 2D analysis during drop jumps have shown to range between 1.4–4.1° and 1.2–1.3°, respectively (King & Belyea, 2015). These values are similar to our own findings (Tables 1–3). To identify a preferred landing strategy, the initial contact angles may provide valuable information regarding the athlete's efficiency for attenuating ground reaction forces. Rowley and Richards (2015) showed that when participants consciously increased their ankle plantarflexion angle from 10° to 30° at initial contact, vertical peak ground reaction forces and loading rates significantly reduced during a bilateral drop-landing from 100% of maximum CMJ height. Alongside landing with greater degrees of ankle plantarflexion angle at initial contact, investigators also observed that participants landed with increased hip and knee extension that was not actively encouraged as part of the study design (Rowley & Richards, 2015). At 30° of ankle plantarflexion at initial contact, an even contribution for shock absorption between the hip, knee and ankle joints occurred, which likely resulted from changes in joint angles at initial contact increasing joint displacement following ground contact (Rowley & Richards, 2015). As greater joint displacement reduces vertical leg stiffness during landings, peak vertical ground reaction forces decrease as the centre of mass's vertical displacement increases (Ward *et al.*, 2019). These findings are supported by that of Kovács *et al.* (1999), who demonstrated that bilateral landings with reduced ankle plantar flexion at initial contact led to greater force dissipation via the knee and hip joint during the landing phase of a drop jump. Furthermore, following ankle injury, Delahunt, Cusack, Wilson and Doherty (2013) showed that individuals with chronic ankle instability landed with 3.0° less plantarflexion following ankle mobilisation. Based on the absolute reliability values presented in Tables 1–3, our investigation indicates that regardless of box height, such subtle changes in hip, knee, and ankle joint alignment at initial contact can be detected using 2D video analysis due to the negligible error of this kinematic measure. Therefore, this test can be used to assess discrete kinematic characteristics that may influence landing mechanics.

Peak joint angles for hip flexion, knee flexion, and ankle dorsiflexion demonstrated nearly perfect relative reliability across all drop heights, with CV% ranging between 1.9–7.9% (Tables 1–3). Similar to our findings, Beardt *et al.* (2018) reported ICC values for measuring peak hip and knee flexion angles using 2D analysis during bilateral drop jumps as 0.98 and 0.92, respectively. Likewise, King and Belyea (2015) reported comparable SEM values for peak flexion angles for the hip (SEM = 2.4°) and knee joint (SEM = 3.1°) to that of our investigation. During single-leg drop vertical jumps, peak hip angle is related to hip and knee flexion moment, indicating that greater peak hip flexion as measured by 2D video analysis results in greater hip flexion moments but reduced knee flexion moment (Dingenen *et al.*, 2015). Landing strategies that

Table 2. Within-session reliability for all dependant variables for bilateral drop-landing from a drop height equalling 100% CMJ height ($n = 39$).

	Test 1 Mean ± SD (°)	Test 2 Mean ± SD (°)	Change in mean (°)	95% LOA (°)	ICC (95% CI)	CV (%)	SEM (°)	MDC (°)
Initial contact angles								
Ankle plantarflexion	149.3 ± 7.6	148.5 ± 7.5	−0.7	0.7 ± 5.7	0.93 (0.87–0.96)	1.4	2.0	5.6
Knee flexion	167.6 ± 4.8	166.1 ± 5.3	−1.6	1.6 ± 5.1	0.87 (0.77–0.93)	1.1	1.6	5.0
Hip flexion	161.5 ± 6.9	160.2 ± 7.5	−1.3	1.3 ± 6.0	0.92 (0.85–0.95)	1.4	2.1	5.8
Peak angles at maximum flexion point								
Ankle dorsiflexion	104.7 ± 9.1	103.5 ± 8.7	−1.2	1.2 ± 5.5	0.95 (0.91–0.97)	1.9	2.0	5.5
Knee flexion	107.5 ± 17.6	105.1 ± 16.1	−2.4	2.4 ± 11.6	0.94 (0.89–0.97)	4.5	3.1	10.5
Hip flexion	114.4 ± 26.6	112.0 ± 25.6	−2.4	2.4 ± 11.6	0.96 (0.93–0.98)	6.0	5.0	13.8
Frontal plane projection angle	186.7 ± 14.0	187.8 ± 13.1	1.1	−1.1 ± 9.1	0.94 (0.90–0.97)	1.8	3.2	8.9
Joint displacement								
Ankle dorsiflexion	44.5 ± 7.1	45.0 ± 6.9	0.5	−0.5 ± 7.3	0.86 (0.76–0.93)	6.8	2.6	7.1
Knee flexion	60.1 ± 14.9	60.9 ± 13.0	0.9	−0.9 ± 10.7	0.93 (0.86–0.96)	6.6	3.8	10.5
Hip flexion	47.1 ± 22.2	48.2 ± 20.8	1.1	−1.1 ± 12.3	0.96 (0.92–0.98)	9.6	4.3	11.9
Asymmetries in joint displacement								
Ankle dorsiflexion displacement	4.3± 7.3	4.1 ± 6.7	−0.1	0.1 ± 8.8	0.81 (0.69–0.88)	N/A	3.1	8.6
Knee flexion displacement	6.4 ± 5.9	6.6 ± 6.0	0.2	−0.2 ± 8.8	0.73 (0.57–0.83)	N/A	3.1	8.7
Hip flexion displacement	3.9 ± 4.8	4.9 ± 4.7	1.0	−1.0 ± 8.1	0.63 (0.44–0.77)	N/A	2.9	8.0

Table 3. Within-session reliability for all dependant variables for bilateral drop-landing from a drop height equalling 150% of CMJ height ($n = 39$).

	Test 1 Mean ± SD (°)	Test 2 Mean ± SD (°)	Change in mean (°)	95% LOA (°)	ICC (95% CI)	CV (%)	SEM (°)	MDC (°)
Initial contact angles								
Ankle plantarflexion	149.6 ± 7.0	148.7 ± 7.4	−0.9	0.9 ± 5.2	0.93 (0.86–0.97)	1.3	1.8	5.1
Knee flexion	165.4 ± 4.5	164.3 ± 5.1	−1.1	1.1 ± 4.9	0.87 (0.77–0.93)	1.1	1.7	4.8
Hip flexion	160.4 ± 6.9	159.1 ± 7.1	−1.2	1.2 ± 6.2	0.90 (0.82–0.95)	1.4	2.2	6.0
Peak angles at maximum flexion point								
Ankle dorsiflexion	104.6 ± 8.4	103.9 ± 8.9	−0.8	0.8 ± 7.0	0.92 (0.85–0.96)	2.5	2.5	6.8
Knee flexion	101.7 ± 14.6	99.4 ± 15.2	−2.4	2.4 ± 11.1	0.93 (0.87–0.96)	4.6	3.9	10.8
Hip flexion	104.6 ± 26.4	102.1 ± 25.8	−2.6	2.6 ± 18.8	0.94 (0.88–0.97)	7.9	6.6	18.3
Frontal plane projection angle	187.5 ± 14.3	188.3 ± 15.5	0.9	−0.9 ± 12.3	0.92 (0.85–0.95)	2.3	4.3	12.0
Joint displacement								
Ankle dorsiflexion	45.0 ± 6.4	44.9 ± 6.2	−0.1	0.1 ± 6.1	0.88 (0.79–0.94)	5.3	2.2	6.0
Knee flexion	63.6 ± 12.5	64.9 ± 12.4	1.3	−1.3 ± 10.6	0.91 (0.83–0.95)	6.3	3.7	10.4
Hip flexion	55.7 ± 22.2	57.1 ± 21.6	1.3	−1.3 ± 16.9	0.93 (0.86–0.96)	11.4	6.0	16.5
Asymmetries in joint displacement								
Ankle dorsiflexion displacement	0.8 ± 6.5	1.2 ± 6.5	0.4	−0.4 ± 7.2	0.84 (0.75–0.91)	N/A	2.7	7.1
Knee flexion displacement	3.4 ± 5.3	4.9 ± 6.0	1.5	−1.5 ± 7.2	0.80 (0.67–0.88)	N/A	2.5	7.1
Hip flexion displacement	2.1 ± 4.8	3.6 ± 4.6	1.5	−1.5 ± 7.2	0.50 (0.27–0.67)	N/A	3.3	9.3

incorporate greater peak hip flexion have been shown to produce less vertical ground reaction forces and reduced quadriceps muscle activity (Blackburn & Padua, 2009). Furthermore, the increase hip flexion moment may potentially increase the hip extensor muscle contribution to dissipate forces (Sigward, Pollard, & Powers, 2012). As reduced hip extensor activation and elevated quadriceps activation during landing tasks may be a risk factor for knee ligament injury (Withrow, Huston, Wojtys, & Ashton-Mille, 2006), identifying landing strategies with reduced levels of peak hip flexion has the potential to allow clinicians to identify athletes at greater risk of injury. Athletes with limited sagittal-plane flexion strategies throughout the lower extremity have also been suggested to lack the necessary shock absorption to attenuate forces during landing tasks (Blackburn & Padua, 2009; Sigward *et al.*, 2012; Zhang, Bates, & Dufek, 2000). Zhang *et al.* (2000) showed that a 25.4°, 22.1°, and 5.9° reduction in peak hip flexion, knee flexion and ankle dorsiflexion angles, respectively, between normal and 'stiff' landings, resulted in significantly greater peak vertical ground reaction forces during bilateral drop-landings from drop heights of 0.62 m. With greater peak forces during landing being associated with increased lower-extremity injury risk (Hewett *et al.*, 2005; Zadpoor & Nikooyan, 2011), practitioners may wish to identify athletes using a stiff landing strategy and provide an intervention to attenuate injury risk (Lopes *et al.*, 2018). Based on CV% presented in Tables 1–3, our findings indicate that changes in landing strategies for peak angles of hip flexion, knee flexion, and ankle dorsiflexion, such as that shown by Zhang *et al.* (2000), may be reliably identified using 2D video analysis. Our findings provide clinicians with practically-relevant information that may guide the interpretation of bilateral landing tasks, with margins for error in the test measures presented (Riemann & Lininger, 2018).

As a result of athletes displaying limited sagittal-plane contribution to attenuating load, compensation may occur through excessive frontal- and/or transverse-plane lower-extremity motion to lower their centre of mass for force dissipation (Sigward *et al.*, 2012). The development of compensation strategies most likely results in greater external knee valgus or varus moments occurring (Kernozek, Torry, Van Hoof, Cowley, & Tanner, 2005). External knee valgus moments and peak angles have previously been shown to recognise athletes at greater risk for anterior cruciate ligament injury (Hewett *et al.*, 2005). With peak FPPA measured using 2D video analysis during landing tasks being shown to correlate with 3D measures of knee valgus ($r = -0.38$) and knee abduction moment ($r = -0.59$) (Mizner *et al.*, 2012), our findings indicate that FPPA may be reliably measured during bilateral drop-landings across various drop heights. SEM for FPPA across all drop heights ranged from 2.7–4.3° for our investigation. These results are similar to the SEM values reported by Munro *et al.* (2012) for FPPA during single-leg drop-landings (SEM = 2.7–2.9°) and bilateral drop jumps (SEM = 3.0°) performed from a 0.28-m drop height. Therefore, using 2D video analysis for identifying peak FPPA is a reliable means for assessing frontal-plane lower extremity kinematics during bilateral drop-landings from heights ranging between 50–150% of maximum CMJ height.

Joint displacement provides a general overview of the contribution from each joint towards force attenuation during landing tasks (Decker, Torry, Wyland, Sterett, & Steadman, 2003). Our results indicate that measurements of joint displacement are reliable to detect differences between- and within-participants in joint contribution from drop heights of 100 and 150% of maximum CMJ height, with CV% ranging from 5.5–11.4%. Although a threshold of 10% for CV% has been suggested to determine a measure as reliable (Stokes, 1985), the use of this arbitrary cut-off point has been contested on the basis that that it is not based on a well-defined analytical goal (Atkinson & Nevill, 1998). As sagittal-plane joint displacement has been shown to be > 10% between populations and following an acute intervention, we chose not to apply an arbitrary threshold for interpreting CV%. For example, when investigating gender differences in joint displacement angles during bilateral drop-landings from a 0.60-m drop height, mean differences between male and female participants for the hip, knee, and ankle joints were 13.0, 16.4 and 28.3%, respectively (Decker, Torry, Wyland, Sterett, & Steadman, 2003). Similarly, with the application of a prophylactic ankle brace to provide external support, Cordova, Takahashi, Kress, Brucker and Finch (2010) found ankle joint displacement reduced by 19.5% during a drop-landing task. Based on the absolute reliability established in our investigation (Tables 1–3), such differences can be detected using 2D video analysis from drop heights equating to 100 and 150% of an individual's maximum CMJ height. However, absolute reliability for joint displacement angles at the hip, knee and ankle were much greater in our investigation from drop heights of 50% of maximum CMJ height, with CV% ranging between 10.0–22.4%. It is possible that at lower drop heights, the lower mechanical demand and thus relative ease of the task increases degrees of movement freedom for participants, facilitating greater variability in joint displacement angles for all segments (Nordin & Dufek, 2017). Our findings suggest that greater change is required for joint displacement angles at the hip, knee, and ankle following an intervention when lower relative drop heights are used for screening differences in coordination strategies during bilateral drop-landings.

Between-limb differences in coordination strategies during bilateral drop-landing have been identified in healthy (Pappas & Carpes, 2012) and previously injured populations (Meyer *et al.*, 2018). We determined the relative reliability for between-limb asymmetries in sagittal-plane joint displacements to be large to very large as part of this investigation. However, the absolute reliability values observed in this investigation indicated this measurement to be highly variable. For example, the MDC values for between-limb asymmetries in ankle, knee and hip joint displacement across each drop height ranged from 7.1–13.2°, 7.1–8.7° and 6.0–9.3°, respectively (Tables 1–3). Pappas and Carpes (2012) investigated gender differences for between-limb joint kinematics during bilateral drop-landings from a 0.40-m drop height in healthy recreational athletes. Between-limb differences for sagittal-plane joint displacement at the ankle (male = 3.4°, females = 3.8°), knee (male = 3.6°, females = 3.8°) and hip joint (male = 5.6°, females = 5.6°) would not exceed the MDC values presented in this investigation. This is similar for between-limb differences observed in injured populations. Using 3D analysis, Meyer *et al.* (2018) examined side-to-side differences in knee joint alignment during a bilateral drop vertical jump from a 0.40-m drop height in 17 patients who had undergone unilateral anterior cruciate ligament reconstructive surgery. For sagittal-plane knee joint displacement, a 2.5°-difference was found between the involved and uninvolved limb (Meyer *et al.*, 2018). Based on the findings of our investigation, it is likely that this difference would not be detectable using 2D video analysis, irrespective of drop height. Therefore, it is suggested that measurements of between-limb differences in sagittal-plane joint displacement during bilateral drop-landings cannot be used to detect smaller, yet clinically meaningful, changes.

5 Conclusion

We have demonstrated that the use of 2D video analysis is a reliable tool for measuring kinematic variables associated with lower-extremity angles at initial contact and maximum flexion point during the bilateral drop-landings from a range of drop heights. With the absolute reliability values presented in this investigation, clinicians possess the tools to interpret an individual's coordination strategy, relative to inherent measurement error, during a bilateral drop-landing using 2D video analysis. However, the variability in asymmetry values found in this investigation indicates that inter-limb asymmetries in joint displacement during bilateral drop-landings may contain excessive amounts of error that impair the ability to interpret whether real change has occurred following intervention.

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