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1 **Reliability of two-dimensional measures associated with bilateral drop-landing** 2 **performance**

3

4 **Abstract**

5 The aim of this study was to establish the within-session reliability for two-dimensional (2D)
6 video analysis of sagittal- and frontal-plane measures during bilateral drop-landing tasks.
7 Thirty-nine recreational athletes (22 men, 17 women, age = 22 ± 4 years, height = 1.74 ± 0.15
8 m, body mass 70.2 ± 15.1 kg) performed five bilateral drop-landings from 50%, 100% and
9 150% of maximum countermovement jump height, twice on the same day. Measures of
10 reliability for initial contact angle, peak flexion angle and joint displacement for the hip,
11 knee, and ankle joints, frontal-plane projection angles (FPPA), as well as inter-limb
12 asymmetries in joint displacement were assessed. No systematic bias was present between
13 trials ($p > 0.05$). All kinematic measurements showed relative reliability ranging from large
14 to near perfect (ICC = 0.52–0.96). Absolute reliability ranged between measures, with CV%
15 between 1.0–1.6% for initial contact angles, 1.9–7.9% for peak flexion angles, 5.3–22.4% for
16 joint displacement, and 1.6–2.3% for FPPA. Absolute reliability for inter-limb asymmetries
17 in joint displacement were highly variable, with minimal detectable change values ranging
18 from 6.0–13.2°. Therefore, 2D video analysis is a reliable tool for numerous measures related
19 to the performance of bilateral drop-landings.

20 **Key words:** within-session reliability, kinematics, landings

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26 **Fiabilité des mesures bidimensionnelles associées aux performances d'atterrissage en**
27 **chute bilatérale**

28

29 **Résumé**

30 Le but de cette étude était d'établir la fiabilité intra-session pour l'analyse vidéo
31 bidimensionnelle (2D) de mesures sur le plan sagittal et frontal lors de tâches d'atterrissage en
32 chute libre bilatérales. Trente-neuf sportifs sportifs (22 hommes et 17 femmes, âge = 22 ± 4
33 ans, taille = $1,74 \pm 0,15$ m, masse corporelle $70,2 \pm 15,1$ kg) ont effectué cinq atterrissages
34 bilatéraux à partir de 50%, 100% et 150% du maximum hauteur du saut en contre-
35 mouvement, deux fois le même jour. Mesures de fiabilité pour l'angle de contact initial,
36 l'angle de flexion maximal et le déplacement articulaire pour les articulations de la hanche, du
37 genou et de la cheville, les angles de projection dans le plan frontal (FPPA), ainsi que les
38 asymétries inter-membres dans le déplacement articulaire. Aucun biais systématique n'était
39 présent entre les essais ($p > 0,05$). Toutes les mesures cinématiques ont montré une fiabilité
40 relative allant de grande à quasi parfaite (ICC = 0,52–0,96). La fiabilité absolue variait d'une
41 mesure à l'autre, avec des CV% compris entre 1,0 et 1,6% pour les angles de contact initiaux,
42 entre 1,9 et 7,9% pour les angles de flexion maximaux, entre 5,3 et 22,4% pour les
43 déplacements articulaires et entre 1,6 et 2,3% pour les FPPA. La fiabilité absolue pour les
44 asymétries inter-membres dans le déplacement articulaire était très variable, avec des valeurs
45 de changement détectables minimales allant de 6.0 à 13.2°. Par conséquent, l'analyse vidéo
46 2D est un outil fiable pour de nombreuses mesures liées à la performance des atterrissages
47 bilatéraux.

48 **Mots clés:** Fiabilité intra-session, cinématique, atterrissages

49

50 **Introduction**

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

67 Although three-dimensional (3D) motion analysis is regarded as the gold standard in
68 exploring lower limb kinematics, in practice two-dimensional (2D) video analysis is more
69 accessible to practitioners (Munro, Herrington, & Carolan, 2012). However, before kinematic
70 measurements of bilateral landing tasks can be used for the purpose of screening, their
71 reliability must first be established. It is therefore important to quantify the noise (error) of
72 the proposed field-based measurements. For various landing tasks, 2D video analysis has
73 been shown to be a valid tool for measuring frontal-plane projection angle (FPPA),
74 significantly relating to measurements of knee abduction angle ($r = -0.38$) and external knee

75 abduction moment ($r = -0.59$) using 3D motion analysis. Furthermore, FPPA provides a
76 reliable representation of knee valgus/varus angle in the deepest landing position (Dingenen,
77 Malfait, Vanrenterghem, Verschueren, & Staes, 2014; McLean et al., 2005; Mizner,
78 Chmielewski, Toepke, & Tofte, 2012; Munro et al., 2012) and is a valid measure of frontal-
79 plane knee mechanics during landings when compared to 3D analysis. However, for joint
80 angle measurements in the sagittal-plane, only Dingenen et al. (2015) and King and Belyea
81 (2015) have investigated the reliability of 2D analysis for measurements of bilateral landing
82 activities. In all of these investigations, only peak angles for the hip, knee and ankle joints
83 were measured. At present, studies investigating the reliability of 2D analysis have not
84 considered other variables that may impact load dissipation during landings, such as initial
85 contact angles and joint displacement for the hip, knee, and ankle joints (Begalle et al., 2015;
86 Chappell et al., 2005; Rowley & Richards, 2015). Furthermore, there has been no
87 investigation of the reliability of 2D kinematic measures during a bilateral drop-landing, a
88 screening tool commonly used in practice (Bird & Markwick, 2016; Tran et al., 2015).

89 An additional consideration when analysing kinematic measures associated with bilateral
90 drop-landings is asymmetries in coordination. Asymmetry in landing strategies commonly
91 occur during bilateral landing tasks in uninjured (Schot et al., 1994) and injured populations
92 (Meyer, Gette, Mouton, Seil, & Theisen, 2018). Practitioners may attempt to determine
93 asymmetries in kinematic variables associated with landing performance, as individuals who
94 exhibit large asymmetries during bilateral landings may expose one leg to excessive loading
95 relative to the contralateral limb (Schot et al., 1994). However, the test re-test reliability for
96 2D video analysis to detect inter-limb asymmetries has not been established for kinematic
97 parameters of drop-landings.

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

100

101 **Methods**102 *Participants*

103 Thirty-nine recreational athletes volunteered for this study, consisting of 22 men (age = $23 \pm$
104 5 years; height = 1.80 ± 0.6 m; mass = 77.9 ± 14.0 kg) and 17 women (age = 20 ± 4 years;
105 height = 1.6 ± 0.9 m; mass = 60.3 ± 9.8 kg) with mean values for maximum
106 countermovement jump (CMJ) height of 0.34 ± 0.7 m and 0.24 ± 0.5 m, respectively.

107 Participants were excluded if they had a previous history of lower-extremity or spinal surgery
108 or had incurred a lower-extremity injury 6 months prior to testing. Participants were informed
109 of the risks associated with testing, completed a pre-exercise questionnaire and signed an
110 informed consent form before testing. Ethical approval was obtained by the Institutional
111 Research Ethics Panel of the lead author.

112

113 *Test procedures*

114 A within-session repeated measures design was used, with participants reporting to the
115 university laboratory for a single testing session. All test sessions were conducted between
116 10:00 am and 1:00 pm to control for circadian variation. All participants wore tight-fitting
117 shorts and vest so that key landmarks were recognisable by all cameras. Anthropometric data
118 was collected prior to completing a standardised warm-up routine consisting of a 5 min jog
119 and dynamic stretches including sumo squats, forward lunges, mountain climbers and leg
120 swings for 10 repetitions. Participants were then familiarised with performing a CMJ. For the
121 CMJ, participants stood bare feet with a hip-width stance with each foot placed on a separate
122 portable force platform recording at 1000 Hz (Pasco, Roseville, CA, USA). Each force
123 platform was positioned side-by-side, 0.05 m apart and embedded in custom-built wooden

124 mounts that were level with the force platforms and did not allow any extraneous movement
125 by the force platforms during the landing. Participants' hands were placed on their hips and
126 remained in this position throughout the jump to isolate the contribution from the lower-
127 extremity. Participants were then asked to rapidly descend prior to explosively jumping as
128 high as possible, with no control being placed on the depth or duration of the
129 countermovement (Benjanuvatra, Lay, Alderson, & Blanksby, 2013). For data collection,
130 three maximal effort CMJs were performed, with 60 s recovery between attempts. Following
131 the final CMJ, force-time data were analysed using the following equation (Moir, 2008) to
132 calculate jump height to the nearest cm:

133 *Equation: Time in the air jump height (cm) = $\frac{1}{2} g(t/2)^2$*

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

138 Following the performance of the CMJ, reflective markers were placed directly onto the
139 participants' skin by the same investigator using the anatomical locations for sagittal-plane
140 lower-extremity joint movements and frontal-plane projection angle (FPPA) outlined by
141 Dingenen et al. (2015) and Munro et al. (2012), respectively. For sagittal-plane views,
142 reflective markers were placed on both left and right acromioclavicular joints, greater
143 trochanters, lateral femoral condyles, lateral malleolus and 5th metatarsal heads (Dingenen et
144 al., 2015). Frontal-plane projection angle was calculated for the right knee joint only, with
145 reflective markers placed at the centre of the right knee joint (midpoint between the femoral
146 condyles), centre of the right ankle joint (midpoint between the malleoli) joint and on the
147 proximal right thigh (midpoint between the anterior superior iliac spine and the knee marker).

148 Midpoints for the knee and ankle were measured with a standard tape measure (Seca 201,
149 Seca, United Kingdom), as outlined by Munro et al. (2012).

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

173 calculate the mean based on previous investigations demonstrating a plateau in measures of
174 reliability for landing kinematics when >4 trials were used for data analysis (Ortiz et al.,
175 2007).

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

183

184 *Data analysis*

185 All video recordings were analysed with free downloadable software (Kinovea for Windows,
186 Version 0.8.15). For sagittal-plane joint movements, hip flexion, knee flexion and ankle
187 dorsiflexion angles were calculated at initial contact and the maximum flexion point deepest
188 landing position for both limbs. These angles were then used to calculate joint displacement
189 for each joint by subtracting the initial contact angle from the maximum flexion point. Initial
190 contact was defined as the frame prior to visual impact between the foot and the ground that
191 led to deformation of the foot complex. The maximum flexion point was identified visually
192 and defined as the frame where no more downward motion occurred at the hip, knee, or ankle
193 joints (Dingenen et al., 2015). Intra-rater reliability for identifying the moment of maximum
194 flexion as a reference point for peak joint angles during landing was performed using the first
195 trial from a drop height of 100% of CMJ height for 20 randomly selected participants (13
196 males and 7 females). Videos were examined twice by the same investigator, seven days

197 apart. Intra-class correlation coefficients (ICC) for time at the maximum flexion point were
198 0.99 and the standard error of measurement (SEM) were 0.01 s. Hip flexion angle was
199 calculated as the angle between a line formed between the acromioclavular joint and the
200 greater trochanter and a line between the greater trochanter and the femoral condyle. Knee
201 flexion angle was calculated as the angle between a line formed between the greater
202 trochanter and the femoral condyle and a line between the femoral condyle and the lateral
203 malleolus. Ankle dorsiflexion angle was calculated as the angle between a line formed
204 between the femoral condyle and the lateral malleolus and a line between the lateral
205 malleolus and the 5th metatarsal head. Frontal-plane projection angle was calculated for the
206 right limb at the deepest landing position, defined as the frame corresponding to maximum
207 knee flexion (Munro et al., 2012). This angle was calculated as the angle between the line
208 formed between the proximal thigh marker and the knee joint marker and a line between the
209 knee joint marker and the ankle joint marker (Munro et al., 2012). For initial contact and the
210 maximum flexion point, smaller values represented greater hip flexion, knee flexion and
211 ankle dorsiflexion for the hip, knee and ankle joints, respectively. For FPPA, values $< 180^\circ$
212 represented knee valgus and values $> 180^\circ$ represented knee varus.

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

217

218 *Statistical analysis*

219 Descriptive statistics (means \pm standard deviation) were calculated for initial contact angles,
220 peak flexion angle at the maximum flexion point and joint displacement for the right limb,

221 along with between-limb differences for joint displacement. The assumption of normality was
222 checked using the Shapiro-Wilk test. To account for heteroscedastic errors, the relationship
223 between the mean values between tests and the difference between repeat tests was evaluated
224 using Pearson's correlation coefficient. To establish systematic bias between test 1 and 2,
225 mean values for initial contact angle, peak flexion angles, joint displacement, FPPA, and
226 between-limb differences in sagittal-plane joint displacement was initially assessed using a
227 paired samples t-test (Atkinson & Nevill, 1998). The *a-priori* level of significance was set at
228 $p < 0.05$, with a Bonferroni correction applied post-hoc. Relative reliability was determined
229 using ICC as described by Hopkins (2018a) and reported with 95% confidence intervals, with
230 ICCs interpreted as follows: 0.01-0.3 poor, 0.3-0.5 moderate, 0.5-0.7 large, 0.7-0.9 very
231 large, and > 0.9 nearly perfect (Hopkins, 2018a). Absolute reliability was calculated using the
232 coefficient of variation, the 95% limits of agreement, SEM ($SD\sqrt{1-ICC}$) (Atkinson & Nevill,
233 1998) and minimal detectable change (MDC; $SEM*1.96*\sqrt{2}$) (Riemann & Lininger, 2018).
234 Due to the asymmetry in joint displacement being interval data, CV% was not determined.
235 ICC and CV% were calculated using a customised spreadsheet (Hopkins, 2018b). The CV%
236 was used as the primary measure of absolute reliability but we have reported a variety of
237 statistical interpretations to facilitate interpretation of the results by researchers and
238 practitioners. All statistical tests were performed using SPSS® statistical software package
239 (v.24; SPSS Inc., Chicago, IL, USA).

240

241 **Results**

242 There was no systematic bias found between test 1 and 2 for any variable for any drop height.
243 Relative reliability ranged from very large to near perfect (ICC = 0.87–0.93) and CV% for
244 initial contact variables ranged from 1.0–1.6% across all drop heights. For peak angles at the

245 maximum flexion point, relative reliability was near perfect (ICC = 0.92–0.95) and absolute
246 reliability ranged between 1.9–7.9% for CV% for the hip, knee and ankle joints, along with
247 FPPA for all drop heights. Relative reliability for joint displacement ranged from very large
248 to near perfect (ICC = 0.76–0.96). At drop heights of 50% CMJ height, greater absolute
249 variability was identified for joint displacement values (CV% =10.0–22.4%), but at a drop
250 height of 100% CMJ height, joint displacements values all possessed CV% < 10%. However,
251 at drop heights of 150% of CMJ height, joint displacement for the hip exceeded CV% > 10%.
252 Relative reliability for between-limb difference in sagittal-plane joint displacement ranged
253 from large to very large (ICC = 0.50–0.84) with MDC values ranging between 6.0–13.2°.

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266 **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	Test 2	Change in	95% LOA (°)	ICC (95% CI)	CV	SEM (°)	MDC
	Mean ± SD	Mean ± SD	mean (°)			(%)		(°)
	(°)	(°)						
<i>Initial contact angles</i>								
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
<i>Peak angles at maximum flexion point</i>								
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
<i>Joint displacement</i>								
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
<i>Asymmetries in joint displacement</i>								
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

268 **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	Test 2	Change in	95% LOA (°)	ICC (95% CI)	CV	SEM (°)	MDC
	Mean ± SD	Mean ± SD	mean (°)			(%)		(°)
	(°)	(°)						
<i>Initial contact angles</i>								
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
<i>Peak angles at maximum flexion point</i>								
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
<i>Joint displacement</i>								
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
<i>Asymmetries in joint displacement</i>								
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

270 **Table 3.** Within-session reliability for all dependant variables for bilateral drop-landing from a drop height equalling 150% of CMJ height ($n =$
271 39).

	Test 1	Test 2	Change in	95% LOA (°)	ICC (95% CI)	CV	SEM (°)	MDC
	Mean ± SD	Mean ± SD	mean (°)			(%)		(°)
	(°)	(°)						
<i>Initial contact angles</i>								
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
<i>Peak angles at maximum flexion point</i>								
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
<i>Joint displacement</i>								
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
<i>Asymmetries in joint displacement</i>								
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

273 **Discussion**

274 The primary aim of this investigation was to determine the within-session reliability of
275 kinematic variables using 2D video analysis during bilateral drop-landings from drop heights
276 equating to 50%, 100%, and 150% of an individual's maximum CMJ height. As part of our
277 investigation, we identified no systematic bias, indicating no evidence of a learning effect,
278 participant bias, or acute adaptations in movement strategies between tests using a within-
279 session design (Atkinson & Nevill, 1998). With *large to near perfect* ICC values and CV%
280 ranging between 1.0–22.4%, our findings suggest that 2D video analysis is sufficiently
281 reliable to determine typical changes in landing kinematics following training or therapeutic
282 interventions during bilateral drop-landings for most variables, although variability in error
283 will be influenced by the kinematic measurement analysed and the drop height. Previously,
284 2D video analysis has been validated against 3D motion analysis for both sagittal- and
285 frontal-plane lower extremity peak joint angles during landing tasks (Dingenen et al., 2014;
286 Dingenen et al., 2015; McClean et al., 2005; Mizner et al., 2012). In conjunction with the
287 findings of our investigation, 2D video analysis is therefore a viable tool for practitioners
288 when assessing bilateral drop-landing mechanics. However, the reliability values presented in
289 this study may not be directly applicable to all populations (i.e. elite athletes). As such,
290 practitioners should attempt to determine the reliability for these variables relative to the
291 population being assessed.

292 Our findings show that initial contact angles for both limbs can be reliably measured using
293 2D video analysis, with ICCs ranging from 0.87–0.93 and CV% between 1.0–1.6% across all
294 drop heights (Table 1–3). Previously, SEM values for establishing sagittal-plane knee and hip
295 angles at initial contact using 2D analysis during drop jumps have shown to range between
296 1.4–4.1° and 1.2–1.3°, respectively (King & Belyea, 2015). These values are similar to our
297 own findings (Table 1–3). To identify a preferred landing strategy, the initial contact angles

298 may provide valuable information regarding the athlete's efficiency for attenuating ground
299 reaction forces. Rowley and Richards (2015) showed that when participants consciously
300 increased their ankle plantarflexion angle from 10° to 30° at initial contact, vertical peak
301 ground reaction forces and loading rates significantly reduced during a bilateral drop-landing
302 from 100% of maximum CMJ height. Alongside landing with greater degrees of ankle
303 plantarflexion angle at initial contact, investigators also observed that participants landed
304 with increased hip and knee extension that was not actively encouraged as part of the study
305 design (Rowley & Richards, 2015). At 30° of ankle plantarflexion at initial contact, an even
306 contribution for shock absorption between the hip, knee and ankle joints occurred (Rowley &
307 Richards, 2015), which likely resulted from changes in joint angles at initial contact
308 increasing joint displacement following ground contact (Rowley & Richards, 2015). As
309 greater joint displacement reduces vertical leg stiffness during landings, peak vertical ground
310 reaction forces decrease as the centre of mass's vertical displacement increases (Ward et al.,
311 2018). These findings are supported by that of Kovács et al. (1999), who demonstrated that
312 bilateral landings with reduced ankle plantar flexion at initial contact led to greater force
313 dissipation via the knee and hip joint during the landing phase of a drop jump. Furthermore,
314 following ankle injury, Delahunt, Cusack, Wilson and Doherty (2013) showed that
315 individuals with chronic ankle instability landed with 3.0° less plantarflexion following ankle
316 mobilisation. Based on the absolute reliability values presented in Table 1–3, our
317 investigation indicates that regardless of box height, such subtle changes in hip, knee, and
318 ankle joint alignment at initial contact can be detected using 2D video analysis due to the
319 negligible error of this kinematic measure. Therefore, this test can be used to assess discrete
320 kinematic characteristics that may influence landing mechanics.

321 Peak joint angles for hip flexion, knee flexion, and ankle dorsiflexion demonstrated *nearly*
322 *perfect* relative reliability across all drop heights, with CV% ranging between 1.9-7.9%

323 (Table 1–3). Similar to our findings, Beardt et al. (2018) reported ICC values for measuring
324 peak hip and knee flexion angles using 2D analysis during bilateral drop jumps as 0.98 and
325 0.92, respectively. Likewise, King and Belyea (2015) reported comparable SEM values for
326 peak flexion angles for the hip ($SEM = 2.4^\circ$) and knee joint ($SEM = 3.1^\circ$) to that of our
327 investigation. During single-leg drop vertical jumps, peak hip angle is related to hip and knee
328 flexion moment, indicating that greater peak hip flexion as measured by 2D video analysis
329 results in greater hip flexion moments but reduced knee flexion moment (Dingenen et al.,
330 2015). Landing strategies that incorporate greater peak hip flexion have been shown to
331 produce less vertical ground reaction forces and reduced quadriceps muscle activity
332 (Blackburn & Padua, 2009). Furthermore, the increase hip flexion moment may potentially
333 increase the hip extensor muscle contribution to dissipate forces (Sigward, Pollard, &
334 Powers, 2012). As reduced hip extensor activation and elevated quadriceps activation during
335 landing tasks may be a risk factor for knee ligament injury (Withrow, Huston, Wojtys, &
336 Ashton-Mille, 2006), identifying landing strategies with reduced levels of peak hip flexion
337 has the potential to allow clinicians to identify athletes at greater risk of injury. Athletes with
338 limited sagittal-plane flexion strategies throughout the lower extremity have also been
339 suggested to lack the necessary shock absorption to attenuate forces during landing tasks
340 (Blackburn & Padua, 2009; Sigward et al., 2012; Zhang, Bates, & Dufek, 2000). Zhang et al.
341 (2000) showed that a 25.4° , 22.1° , and 5.9° reduction in peak hip flexion, knee flexion and
342 ankle dorsiflexion angles, respectively, between normal and ‘stiff’ landings, resulted in
343 significantly greater peak vertical ground reaction forces during bilateral drop-landings from
344 drop heights of 0.62 m. With greater peak forces during landing being associated with
345 increased lower-extremity injury risk (Hewett et al., 2005; Zadpoor & Nikooyan, 2011),
346 practitioners may wish to identify athletes using a stiff landing strategy and provide an
347 intervention to attenuate injury risk (Lopes et al., 2018). Based on CV% presented in Table

348 1–3, our findings indicate that changes in landing strategies for peak angles of hip flexion,
349 knee flexion, and ankle dorsiflexion, such as that shown by Zhang et al. (2000), may be
350 reliably identified using 2D video analysis. Our findings provide clinicians with practically-
351 relevant information that may guide the interpretation of bilateral landing tasks, with margins
352 for error in the test measures presented (Riemann & Lininger, 2018).

353 As a result of athletes displaying limited sagittal-plane contribution to attenuating load,
354 compensation may occur through excessive frontal- and/or transverse-plane lower-extremity
355 motion to lower their centre of mass for force dissipation (Sigward et al., 2012). The
356 development of compensation strategies most likely results in greater external knee valgus or
357 varus moments occurring (Kernozek, Torry, Van Hoof, Cowley, & Tanner, 2005). External
358 knee valgus moments and peak angles have previously been shown to recognise athletes at
359 greater risk for anterior cruciate ligament injury (Hewett et al., 2005). With peak FPPA
360 measured using 2D video analysis during landing tasks being shown to correlate with 3D
361 measures of knee valgus ($r = -0.38$) and knee abduction moment ($r = -0.59$) (Mizner et al.,
362 2012), our findings indicate that FPPA may be reliably measured during bilateral drop-
363 landings across various drop heights. SEM for FPPA across all drop heights ranged from 2.7–
364 4.3° for our investigation. These results are similar to the SEM values reported by Munro et
365 al. (2012) for FPPA during single-leg drop-landings (SEM = 2.7–2.9°) and bilateral drop
366 jumps (SEM = 3.0°) performed from a 0.28 m drop height. Therefore, using 2D video
367 analysis for identifying peak FPPA is a reliable means for assessing frontal-plane lower
368 extremity kinematics during bilateral drop-landings from heights ranging between 50–150%
369 of maximum CMJ height.

370 Joint displacement provides a general overview of the contribution from each joint towards
371 force attenuation during landing tasks (Decker, Torry, Wyland, Sterett, & Steadman, 2003,
372 2003). Our results indicate that measurements of joint displacement are reliable to detect

373 differences between- and within-participants in joint contribution from drop heights of 100%
374 and 150% of maximum CMJ height, with CV% ranging from 5.5–11.4%. Although a
375 threshold of 10% for CV% has been suggested to determine a measure as reliable (Stokes,
376 1985), the use of this arbitrary cut-off point has been contested on the basis that that it is not
377 based on a well-defined analytical goal (Atkinson & Nevill, 1998). As sagittal-plane joint
378 displacement has been shown to be >10% between populations and following an acute
379 intervention, we chose not to apply an arbitrary threshold for interpreting CV%. For example,
380 when investigating gender differences in joint displacement angles during bilateral drop-
381 landings from a 0.60 m drop height, mean differences between male and female participants
382 for the hip, knee, and ankle joints were 13.0%, 16.4% and 28.3%, respectively (Decker,
383 Torry, Wyland, Sterett, & Steadman, 2003). Similarly, with the application of a prophylactic
384 ankle brace to provide external support, Cordova, Takahashi, Kress, Brucker and Finch
385 (2010) found ankle joint displacement reduced by 19.5% during a drop-landing task. Based
386 on the absolute reliability established in our investigation (Table 1–3), such differences can
387 be detected using 2D video analysis from drop heights equating to 100% and 150% of an
388 individual's maximum CMJ height. However, absolute reliability for joint displacement
389 angles at the hip, knee and ankle were much greater in our investigation from drop heights of
390 50% of maximum CMJ height, with CV% ranging between 10.0-22.4%. It is possible that at
391 lower drop heights, the lower mechanical demand and thus relative ease of the task increases
392 degrees of movement freedom for participants, facilitating greater variability in joint
393 displacement angles for all segments (Nordin & Dufek, 2017). Our findings suggest that
394 greater change is required for joint displacement angles at the hip, knee, and ankle following
395 an intervention when lower relative drop heights are used for screening differences in
396 coordination strategies during bilateral drop-landings.

397 Between-limb differences in coordination strategies during bilateral drop-landing have been
398 identified in healthy (Pappas & Carpes, 2012) and previously injured populations (Meyer et
399 al., 2018). We determined the relative reliability for between-limb asymmetries in sagittal-
400 plane joint displacements to be large to very large as part of this investigation. However, the
401 absolute reliability values observed in this investigation indicated this measurement to be
402 highly variable. For example, the MDC values for between-limb asymmetries in ankle, knee
403 and hip joint displacement across each drop height ranged from 7.1–13.2°, 7.1–8.7° and 6.0–
404 9.3°, respectively (Table 1–3). Pappas and Carpes (2012) investigated gender differences for
405 between-limb joint kinematics during bilateral drop-landings from a 0.40 m drop height in
406 healthy recreational athletes. Between-limb differences for sagittal-plane joint displacement
407 at the ankle (male = 3.4°, females = 3.8°), knee (male = 3.6°, females = 3.8°) and hip joint
408 (male = 5.6°, females = 5.6°) would not exceed the MDC values presented in this
409 investigation. This is similar for between-limb differences observed in injured populations.
410 Using 3D analysis, Meyer et al. (2018) examined side-to-side differences in knee joint
411 alignment during a bilateral drop vertical jump from a 0.40 m drop height in 17 patients who
412 had undergone unilateral anterior cruciate ligament reconstructive surgery. For sagittal-plane
413 knee joint displacement, a 2.5° difference were found between the involved and uninvolved
414 limb (Meyer et al., 2018). Based on the findings of our investigation, it is likely that this
415 difference would not be detectable using 2D video analysis, irrespective of drop height.
416 Therefore, it is suggested that measurements of between-limb differences in sagittal-plane
417 joint displacement during bilateral drop-landings cannot be used to detect smaller, yet
418 clinically meaningful, changes.

419

420 **Conclusion**

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

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580 **Author contribution statement**

581 All authors contributed equally to the study design and writing of the manuscript. LH

582 collected and analysed all data.