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4 **Mechanisms to attenuate load in the intact limb of transtibial amputees when**  
5 **performing a unilateral drop landing**

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19 **Abstract**

20 Individuals with unilateral transtibial amputations experience greater work demand and loading  
21 on the intact limb compared to the prosthetic limb, placing this limb at a greater risk of knee  
22 joint degenerative conditions. It is possible that increased loading on the intact side may occur  
23 due to strength deficits and joint absorption mechanics. This study investigated the intact limb  
24 mechanics utilised to attenuate load, independent of prosthetic limb contributions and  
25 requirements for forward progression, which could provide an indication of deficiencies in the  
26 intact limb. Amputee and healthy control participants completed three unilateral drop landings  
27 from a 30 cm drop height. Joint angles at touchdown, range of motion, coupling angles, peak  
28 powers, and negative work of the ankle, knee and hip were extracted together with isometric  
29 quadriceps strength measures. No significant differences were found in the load or movement  
30 mechanics ( $p \geq .312$ ,  $g \leq 0.42$ ), despite deficits in isometric maximum (20%) and explosive  
31 (25%) strength ( $p \leq .134$ ,  $g \geq 0.61$ ) in the intact limb. These results demonstrate that, when the  
32 influence from the prosthetic limb and task demand are absent, and despite deficits in strength,  
33 the intact limb adopts joint mechanics similar to able-bodied controls to attenuate limb loading.

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35 **Keywords:** quadriceps strength, knee osteoarthritis, joint loading, amputee

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37 **Word Count:** 4290/4000

## Introduction

39 Previous research on individuals with a transtibial amputation (ITTAs) has suggested  
40 that the mechanics of the prosthetic limb may influence the intact limb mechanics, and  
41 subsequently the magnitude and rate of load experienced in walking<sup>1,2</sup>, running<sup>3,4</sup>, and start-  
42 stop tasks.<sup>5</sup> This is postulated to result from the inability of the prosthesis to generate the  
43 propulsion required to continue forward progression<sup>1</sup> or, in bilateral jump landings, from  
44 inadequate absorption of high forces through the prosthesis.<sup>6</sup> These interactions between the  
45 prosthetic and intact limb mechanics may explain the altered shock absorption approach  
46 observed in the intact limb (i.e., reduced joint angles and powers) during the initial loading  
47 response phase of running, step/stair negotiation, and bilateral jump landing.<sup>4,6-8</sup> Thus, the  
48 intact limb must perform greater work to either continue forward progression or arrest the  
49 lowering of the centre of mass<sup>9</sup>, which results in high load compared to the prosthetic limb.<sup>10</sup>  
50 However, no research has assessed the shock absorption approach of the intact limb to attenuate  
51 load without the influence of the prosthetic limb and the requirement to continue forward  
52 progression. This could provide an indication of deficiencies in the intact limb following  
53 amputation, which may be useful for informing rehabilitation protocols.

54 A unilateral drop landing onto the intact limb can be used to examine joint mechanics  
55 and load attenuation in response to a consistent vertical momentum. Reducing vertical  
56 momentum is required in many movements such as walking, running, and jump landings, and  
57 occurs through joint flexion and eccentric work to efficiently absorb rapid impact forces.  
58 Deficiencies in muscle strength of the knee extensors may also play a role in load attenuation.  
59 Decreased maximum muscle strength has been identified as a key risk factor accompanying  
60 degenerative loading diseases<sup>11</sup> and has been suggested as an indication of increased limb  
61 loading.<sup>12,13</sup> Furthermore, frontal plane knee valgus motion can be increased 3-fold from  
62 decreased quadriceps muscle force,<sup>14,15</sup> which has been identified as a risk factor associated

63 with joint degeneration.<sup>16</sup> Increasing trunk flexion when landing has been found as a  
64 compensatory strategy to reduce the reliance on the eccentric contraction of the quadriceps.  
65 Greater trunk flexion is related to greater flexion at all lower-limb joints when landing from a  
66 jump which could aid in reducing knee joint loading.<sup>17,18</sup> Substantial deficits in quadriceps  
67 muscle strength of 30-39% have previously been reported in the intact limb of ITTAs compared  
68 to able-bodied individuals;<sup>13,19</sup> however, it is currently unknown how the intact limb may  
69 accommodate for decreased quadriceps strength.

70         When landing from a jump, the time to develop muscular force to control joint motion  
71 is limited. Generation of rapid muscle force has been shown to be important for re-stabilisation  
72 of the lower-limb joints following mechanical perturbations.<sup>20-22</sup> The inability to stabilise and  
73 prevent the rapid flexion of the knee joint during jump landings can lead to various acute and  
74 repetitive knee overloading injuries, e.g. osteoarthritis and non-specific knee pain.<sup>23</sup> Rapid  
75 muscle force production has not been examined in ITTAs yet could provide important  
76 information on the ability to initially stabilise the joints upon landing.

77         A study assessing bilateral jump landings<sup>6</sup> found that the intact limb of ITTAs  
78 underwent a smaller range of motion (ROM) at all lower-limb joints compared to the control  
79 population and experienced significantly greater peak vertical ground reaction force (vGRF).  
80 This suggests that ITTAs utilise a more extended landing strategy in the intact limb. However,  
81 the ITTA study assessed a bilateral landing, thus, the restricted mechanics from the prosthesis  
82 could have influenced the results. Reduced lower-limb joint flexion is possibly a compensation  
83 to limit the eccentric work required from the knee joint musculature<sup>24</sup>, yet this may lead to the  
84 impact forces being absorbed by the surrounding tissue structures.<sup>25</sup> Individuals who perform  
85 a more extended landing strategy also utilise a different joint absorption approach as measured  
86 by joint power and work.<sup>26,27</sup> While the knee joint is a consistent contributor to dissipating the  
87 kinetic energy, the percentage contribution of the ankle and hip joint work can be altered as the

88 degree of knee flexion during landing changes.<sup>26-28</sup> These studies suggest that specific  
89 coordination strategies of the lower-limb joints may be related to the load experienced. It is  
90 possible that without the influence from the prosthetic limb, the intact limb may be able to  
91 adopt a more flexed landing strategy thereby reducing the limb and joint load experienced.

92 In ITTAs, the intact limb is at a greater risk of experiencing knee pain, subsequent joint  
93 degeneration, and the development of comorbidities when compared to the prosthetic limb and  
94 the general population.<sup>29-31</sup> The pathogenesis of joint degeneration is thought to stem from  
95 repetitive overloading in a limb<sup>32</sup>, however, only one study has been conducted on landings in  
96 the ITTA population<sup>6</sup> where only the peak vGRF was assessed. Research assessing overloading  
97 injuries has examined various discrete features within the GRF<sup>33</sup>, knee joint moment<sup>34,35</sup>, and  
98 knee intersegmental force<sup>36,37</sup> waveforms. There is no clear consensus on the most appropriate  
99 reduction of these loading waveforms to assess overloading associated with joint degeneration.  
100 Statistical parametric mapping is an approach which analyses a waveform in its original  
101 temporal-spatial format<sup>38</sup> to remove the bias from an *a priori* approach when assessing limb or  
102 joint loading.

103 The purpose of this study was, therefore, to investigate limb loading in the intact limb  
104 of ITTAs compared to able-bodied controls during a unilateral drop landing, independent of  
105 prosthetic limb interactions and the requirement of forward progression; and assess the  
106 mechanisms underpinning any differences, including quadriceps maximal and rapid muscle  
107 force production and joint absorption mechanics. It is hypothesised that, compared to the  
108 control limb, the intact limb will 1) present with reduced quadriceps muscular strength and  
109 rapid muscle force production, 2) experience a greater magnitude of load throughout the  
110 absorption phase as assessed by examining the loading pattern using statistical parametric  
111 mapping, and 3) perform altered discrete joint mechanics in the sagittal plane for the ankle,  
112 knee, and hip joints and altered trunk flexion and knee joint valgus motion.

113

## Methods

114           Eight recreationally active ITTAs and twenty-one controls volunteered to participate in  
115 the study (Table 1). Ethical approval was obtained from the University of Roehampton's Ethic  
116 Committee (LSC 16/176) and the National Health Services Health Research Authority  
117 (17/NW/0566). All participants provided written informed consent prior to any assessment.  
118 Inclusion criteria required all participants to be physically active (i.e. requiring moderate or  
119 greater physical effort) a minimum of 2-3 days per week. Participants were excluded if they  
120 had sustained a musculoskeletal injury in the six months prior or were experiencing pain in  
121 their back or lower-extremities. ITTAs included in the study had a grading of K3/K4, as  
122 determined by their physicians, to ensure that the participants could perform high impact  
123 movements safely. A K3/K4 level is defined as an amputee that has the ability or potential to  
124 negotiate environmental barriers and for prosthetic ambulation that exhibits high impact, stress,  
125 or energy levels. ITTA participants had amputations due to traumatic incidents (e.g.,  
126 automobile accident) and were a minimum of 6-months post-amputation (mean  $\pm$  SD: 12.2  $\pm$   
127 11.5, range: 1.5-29 years) (Table 1).

128           All strength and biomechanical features were extracted from the intact limb of ITTAs  
129 ( $n = 8$ ) and the dominant control limb ( $n = 21$ ). Dominance was defined as the limb that was  
130 chosen first to complete a unilateral landing. Participants attended three data collection sessions  
131 each separated by 3-7 days: 1) familiarisation of strength measures, 2) strength data collection,  
132 and 3) biomechanical testing of drop landings. The data were collected in the order presented  
133 below for all participants.

134           Strength Data Collection: Quadriceps isometric strength data were collected using an  
135 isokinetic dynamometer (Humac Norm, Massachusetts, USA). The knee joint angle was set so  
136 that the angle during active maximal extension was 110° and the hip angle was set to 100° (full

137 extension = 180°). Adjustable straps across the pelvis and shoulders were tightened to ensure  
138 no extraneous movement. Arm placement during contractions was chosen by the participants  
139 and typically consisted of crossed arms over the chest or by their sides holding handles. The  
140 torque signal was sampled at 2000 Hz using an external A/D converter (16-bit signal recording  
141 resolution; Micro 1401, CED, Cambridge, UK) and interfaced with a PC using Spike 2 software  
142 (version 8; CED). All torque data were filtered using a fourth-order Butterworth filter with a  
143 cut-off frequency of 10 Hz, and were corrected for the weight of the limb by subtracting  
144 baseline resting torque.

145         Participants performed a series of warm-up contractions of increasing torque values for  
146 2-3 minutes. Following the warm-up, three maximal voluntary isometric contractions lasting  
147 ~3 s each were performed with a ~45 s rest in between each attempt. Additional attempts were  
148 required if peak force continued to increase with each subsequent effort. The only instruction  
149 provided was to ‘push as hard as possible’ and strong verbal encouragement was given  
150 throughout the contraction to encourage maximal effort. Real-time biofeedback of the torque-  
151 time curve and the peak torques achieved in each contraction were provided on a computer  
152 monitor in front of the participants. Maximum voluntary torque (MVT, considered a measure  
153 of maximum strength), was determined as the greatest peak torque recorded during any  
154 maximal or rapid muscle force contractions (see below), and normalised to body mass.

155         Rapid muscle force contractions were performed separate to the maximal  
156 contractions.<sup>39,40</sup> Participants completed 10 rapid muscle force isometric contractions each  
157 separated by ~20 s rest. Participants were instructed to ‘push as fast and as hard as possible’  
158 for ~1 s, with an emphasis on ‘fast’, and aimed to achieve a minimum of 80% of MVT as  
159 quickly as possible. Real-time biofeedback was again provided to denote the participant’s best  
160 performance; the peak rate of torque development (RTD) was highlighted from the slope of the  
161 torque-time curve (15 ms time-constant). Resting torque was additionally monitored to ensure

162 that no countermovement or pre-tension occurred before the contraction. Peak RTD was  
163 averaged from the three rapid muscle force voluntary contractions with the highest peak  
164 RTDs<sup>41</sup> and expressed relative to body mass.

165 Biomechanical Data Collection: Joint motion data were captured at 200 Hz using  
166 twelve Vicon Vantage V5 motion capture cameras (Vicon, Oxford, UK) and force data were  
167 sampled at 1000 Hz using Kistler force platforms (Type 9281c; Kistler, Hampshire, UK).  
168 Kinematic and kinetic data were filtered using a fourth-order low-pass Butterworth filter with  
169 cut-off frequencies of 15 Hz and 200 Hz, respectively, in Vicon Nexus 2.6.1.<sup>42</sup> Data extraction  
170 and analysis was performed using custom-made code in MATLAB (R2017a, The Mathworks  
171 Inc, Natick, MA).

172 Retroreflective markers (14 mm) were placed on the skin according to the Plug-In-Gait  
173 full-body marker set.<sup>43</sup> Drop landings were performed from a custom-made hanging frame that  
174 was vertically adjusted to ensure all participants landed from a drop height of 30 cm based on  
175 the distance of the heel of the shoe from the ground as measured by a ruler. Participants were  
176 given time to become comfortable with the required movement (typically 2-5 trials) to ensure  
177 stable recovery. Trials were excluded if the participants raised their centre of mass by pulling  
178 themselves up on the bar prior to dropping, did not land with their foot fully on the force  
179 platform, or were unable to recover from the drop as denoted by stepping with their  
180 contralateral limb. Data collection continued until three 'successful' trials were captured.<sup>44</sup>  
181 Data were averaged from the three trials to be used in further analysis.

182 All loading and movement features were extracted from the absorption phase of  
183 landing. This phase was defined from touchdown, based on a 20N threshold in the vGRF,  
184 through to maximal knee flexion. The duration of the absorption phase was calculated in  
185 seconds as a measure of the time taken to absorb the impact forces from landing. The loading



186 waveforms extracted for analysis including the GRF, knee moments, and intersegmental knee  
187 forces in all three planes of motion. Knee moments and intersegmental knee forces were  
188 derived from inverse dynamic calculations using the Plug-In-Gait model in Vicon. Loading  
189 waveforms were linearly time-normalised to 100% of the absorption phase based on the  
190 average length of the phase across all participants (40 frames) to avoid over-stretching or -  
191 shrinking of the data.<sup>45</sup>

192 Discrete movement features were extracted from the sagittal plane ankle, knee and hip  
193 joints including touchdown angles, ROM, coupling angles, peak absorption powers, and  
194 negative joint work. ROM was determined as the difference from minimal to maximal flexion  
195 during the absorption phase. Joint coordination coupling angles were derived from angle-angle  
196 plots (Figure 1A) and represents the angle of the vector between two adjacent points relative  
197 to the right horizontal (Figure 1B).<sup>46,47</sup> The calculated coupling angle can lie anywhere between  
198 0° and 360°, where 0°, 90°, 180°, and 270° represent single joint movement and 45°, 135°,  
199 225°, and 315° indicate equal motion between the two joints<sup>48</sup> (see Figure 3). The average  
200 coupling angles were calculated for the ankle-knee, knee-hip, and hip-ankle joint pairs from  
201 touchdown to peak vGRF to assess the initial loading coupling strategy.<sup>49</sup> Negative joint work  
202 was calculated as the area under the negative portion of the power-time curve using the  
203 trapezoidal rule. Trunk flexion angle at touchdown and ROM during the absorption phase were  
204 additionally extracted. This ROM was calculated based on angular change of the vertical axis  
205 and the vector defined by the shoulder and anterior superior iliac spine markers during the  
206 absorption phase. Lastly, in the frontal plane, knee joint touchdown angle and ROM were  
207 extracted.

208 All data were normally distributed as determined by the Shapiro-Wilk test of normality  
209 for the discrete features and D'Agostino-Pearson K2 normality tests in SPM for the loading  
210 waveforms. To assess differences between the intact and control limbs, independent *t*-tests

211 were performed for all strength, loading, and movement features. Loading waveforms were  
212 assessed using statistical parametric mapping.<sup>38</sup> Hedge's  $g$  was calculated to aid in the  
213 understanding of the results and was interpreted as a small (0.2), medium (0.5), or large effect  
214 (0.8).<sup>50</sup>

## 215 **Results**

216 Participant demographics were not significantly different between groups for age,  
217 height, or mass, although there was a moderate-to-large effect ( $g = 0.69$ ) for ITTAs to be older  
218 than controls (Table 1). Average drop landing height for both groups was  $30.7 \pm 3.4$  cm and  
219 was not significantly different between groups ( $p = .170$ ,  $g = 0.34$ ; ITTA:  $31.6 \pm 3.4$  cm,  
220 Control:  $30.4 \pm 3.4$  cm). The duration of the absorption phase was also similar between groups  
221 ( $p = .798$ ,  $g = 0.05$ ; ITTA:  $0.21 \pm 0.04$  s, Control:  $0.20 \pm 0.13$  s).

222 For the strength measures, there was a medium-to-large effect ( $g = 0.61$ ) for MVT to  
223 be lower in the intact limb although this difference was not statistically significant (Table 2).  
224 There was also a medium-to-large effect ( $g = 0.72$ ) for peak RTD to be lower in the intact limb  
225 compared to the control limb.

226 SPM results of the loading waveforms found no significant differences between the  
227 intact limb of ITTAs and control limbs for any loading waveform for the duration of the  
228 absorption phase (Figure 2 & Supplementary Figure 1).

229 Within the movement features, the intact and control limbs did not differ significantly  
230 at any lower-limb joint or at the trunk for the touchdown angles ( $p \geq .312$ ,  $g \leq 0.42$ ) or ROM  
231 ( $p \geq .339$ ,  $g \leq 0.39$ ) in the sagittal and frontal planes (Figure 3A). Joint coordination strategies  
232 were not significantly different between groups for any lower-limb joint pair ( $p \geq .385$ ,  $g \leq$   
233  $0.21$ ; Figure 3B). Peak negative absorption powers were also not significantly different  
234 completed (Figure 4A) was not significantly different between groups at the ankle ( $p = .950$ ,  $g$

235 = 0.03; Intact limb:  $-1.23 \pm 0.35$  J/kg, Control limb:  $-1.22 \pm 0.28$  J/kg), knee ( $p = .457$ ,  $g =$   
236  $0.29$ ; Intact limb:  $-0.57 \pm 0.28$  J/kg, Control limb:  $-0.67 \pm 0.36$  J/kg) or hip joints ( $p = .406$ ,  $g$   
237  $= 0.33$ ; Intact limb:  $-0.34 \pm 0.25$  J/kg, Control limb:  $-0.27 \pm 0.20$  J/kg). Both the intact and  
238 control limbs utilised the ankle joint as the primary joint to perform the negative work to reduce  
239 the momentum of the centre of mass (56-58%; Figure 4B). Small effect sizes were present for  
240 all movement feature comparisons.

## 241 **Discussion**

242 This study investigated intact limb loading and the mechanisms utilised to attenuate  
243 this load without the influence of the prosthetic limb or the requirement for forward  
244 progression. The main finding of this study was that there were no significant differences  
245 between groups for the strength features, the joint mechanics utilised to absorb the impact from  
246 landing or in the load experienced at the ground or at the knee joint. These results provide  
247 evidence to suggest that high load in the intact limb, compared to controls, that has been found  
248 in other studies and movements (e.g., walking, step negotiation) is due to either the influence  
249 of the mechanics from the prosthetic limb or the specific task demands. This suggests that the  
250 intact limb of ITTAs is not at a greater risk of injury in the intact limb when performing a  
251 unilateral landing from a drop height of 30 cm.

252 MVT was 20% and peak RTD was 25% lower in the intact limb of ITTAs compared to  
253 controls and medium-to-large effect sizes were apparent (0.61 and 0.72, respectively). The  
254 MVT deficits are smaller than those found in other ITTA studies that included amputees with  
255 a similar mean and range of ages as that of the current study. These studies have indicated that  
256 the intact limb produces 30-39% less maximum strength than an able-bodied control.<sup>13,19</sup>  
257 However, these earlier studies included individuals whose amputation occurred due to vascular  
258 diseases, thus, the greater deficiencies in muscular strength may be due to the effects of the

259 disease that are not present in traumatic amputations. Further, these studies did not present the  
260 activity level of their participants and the Pedrinelli et al.<sup>19</sup> study included participants who  
261 used walking aids (20% of total participants). It is additionally possible that the recreationally  
262 active nature of the participants in the current study may have attributed to the lower percentage  
263 deficits.

264 Past research has found negative correlations between quadriceps strength and peak  
265 vGRF in quadriceps inhibition<sup>51</sup> and anterior cruciate ligament injury jump landing studies<sup>52</sup>  
266 when landing from a height of 30 cm. Additionally, it is well known that quadriceps weakness  
267 is associated with joint degenerative diseases where strength deficits from 15-18% may be  
268 present prior to disease development.<sup>53,54</sup> Previous research has suggested that isometric MVT  
269 deficits in the quadriceps of greater than 15% can negatively impact the loading patterns and  
270 alter the joint mechanics when landing from a jump.<sup>55</sup> This can result in the absorption of  
271 impact forces by the tissue structures rather than by the bigger muscle groups, increasing the  
272 incidence of developing degenerative diseases.<sup>25,26</sup> However, the current study found no  
273 differences in loading patterns between groups suggesting that the strength deficits did not  
274 influence the magnitude or rate of load experienced. Maximal production of strength may not  
275 have been required for the movement performed in this study. Further research could assess  
276 the height about which compensations may occur in response to reduced quadriceps strength.

277 As far as the authors are aware, the current study is the first to assess RTD strength in  
278 the intact limb of ITTAs. Previous research has found that greater RTD can aid in dynamic  
279 balance recovery,<sup>56</sup> such as that seen in sporting movements, by rapid stabilisation of the lower-  
280 limb joints. Without stabilisation, the joints could move into injurious positions (e.g. reduced  
281 knee joint flexion) placing the load demand onto the cartilage.<sup>57</sup> However, in the current study,  
282 the intact limb did not exhibit significantly different lower-limb motion, coordination patterns,  
283 or a shift in the shock absorption approach as both groups completed the majority of energy

284 absorption in the ankle joint (56-58%). Additionally, small effect sizes were present in all  
285 movement features further confirming that no differences were present between the intact limb  
286 of ITTAs and controls. As the ITTA population in the current study did not experience greater  
287 limb or joint load, it is possible that both groups had sufficient quadriceps strength and were  
288 able to rapidly produce muscle force that allowed an adequate degree of joint flexion to  
289 attenuate the load during landing.<sup>51</sup>

290         Reduced quadriceps strength can be compensated for through a number of mechanisms  
291 including frontal plane knee valgus motion<sup>58</sup> and trunk flexion.<sup>59</sup> The current study, however,  
292 found no significant differences in the frontal plane knee motion or the sagittal plane trunk  
293 flexion, possibly as no significant differences were found in the strength measures. These  
294 results differ from previous research. Goerger et al.<sup>60</sup> suggested that when vGRF is similar,  
295 frontal plane motion may be altered as a possible compensation to absorb load when deficits in  
296 quadriceps strength are present. This was also reported by Palmieri-Smith et al.<sup>58</sup> who  
297 demonstrated that reduced quadriceps preparatory activation prior to touchdown was associated  
298 with increased peak knee valgus angles. Healthy participants, who landed with greater peak  
299 trunk flexion, had a reduced quadriceps activity and landing forces suggesting a reduced  
300 reliance on the eccentric contraction of the quadriceps to attenuate load.<sup>18</sup> Greater active trunk  
301 flexion during landing is also associated with a more flexed strategy at the knee and hip joints<sup>17</sup>  
302 potentially contributing to the reduced landing forces. That there were no significant  
303 differences between the intact limb ITTAs and control limbs in the current study, suggests that  
304 the 20% deficit in quadriceps maximal strength and 25% deficit in peak RTD did not elicit  
305 compensations in the landing mechanics. Additionally, these deficits did not impact the  
306 magnitude and rate of load experienced when landing from a drop height of 30 cm.

307         Both groups performed an ankle dominant joint absorption approach when landing  
308 (Figure 4B). Greater utilisation of the ankle joint to attenuate load has been found to be

309 associated with increases in peak vGRF, knee flexor moment, and anterior knee intersegmental  
310 force magnitudes.<sup>26,27,61</sup> Healthy individuals who performed a more extended landing strategy  
311 at all joints utilised the ankle joint to perform ~50% of the total joint work.<sup>26,27</sup> Rowley &  
312 Richards<sup>62</sup> determined that an optimal ankle plantarflexion angle at touchdown between 20-  
313 30° would limit the peak vGRF and vGRF loading rate when landing from a jump.  
314 Additionally, within this optimal plantarflexion range, the lower-limb joints' contribution  
315 relative to the support moment were found to be relatively equal (ankle, knee and hip joints  
316 between 30-40% of total). This suggests that in-phase joint flexion coordination could  
317 potentially reduce load at the ground and at the knee joint by absorbing the load equally at the  
318 lower-limb joints.<sup>49</sup> The ITTA and control participants in the current study landed with an  
319 'optimal' ankle plantarflexion angle. However, there was greater utilisation of the distal joints  
320 with 56-58% of the total joint work completed by the ankle. In comparison to unilateral drop  
321 landing research, the joint mechanics were similar to that in the current study.<sup>51,63</sup> It was  
322 suggested that a more extended landing strategy is performed in unilateral landings to maintain  
323 balance, despite the greater risk of injury when utilising this approach.<sup>63</sup> It is also possible that  
324 the extended landing strategy was performed by ITTAs in this study to limit the eccentric work  
325 required from the quadriceps. Thus, a unilateral landing did not elicit greater joint flexion in  
326 the intact limb when the prosthetic limb contribution was absent. Single-limb balance and  
327 quadriceps strength training may enable the intact limb to adopt a more flexed landing strategy  
328 which could be important in reducing load in many sporting manoeuvres.

329         Landing height has been shown to influence the landing joint mechanics as greater  
330 momentum is experienced as landing height increases.<sup>64-66</sup> Schoeman et al.<sup>6</sup> found greater  
331 vGRF was experienced in the intact limb compared to the control limb. However, the ITTA  
332 group landed from a significantly lower jump height than the controls. This could suggest that  
333 the vGRF should have been significantly greater when ITTAs landed from the same height as

334 the controls. However, the vGRF experienced in the intact limb in the current study was similar  
335 to that experienced in the intact limb of the Schoeman et al.<sup>6</sup> study. This occurred despite  
336 landing from almost double the height (15 cm vs 30 cm). One possible reason is that the intact  
337 limb in the current study performed greater joint ROM compared to the intact limb of the ITTAs  
338 who landed from half the height (15 cm). Further, the intact limb in the current study performed  
339 similar ROM at all lower-limb joints to the control group in the Schoeman et al.<sup>6</sup> study who  
340 landed from the same height (31 cm). This shock absorption adaptation has been seen in able-  
341 bodied individuals who increase joint flexion angles as the drop height increases thereby  
342 limiting the load experienced.<sup>66</sup> Therefore, the results from the current study suggest that  
343 ITTAs can adapt to the higher landing height and attenuate load without the influence from the  
344 prosthetic limb, by adopting shock absorption strategies similar to those of a control population.

345         The intact limb of ITTAs does not experience significantly different load and does not  
346 perform significantly different joint absorption mechanics compared to an able-bodied control,  
347 when landing on this limb from a drop height of 30 cm. This was despite deficits in the knee  
348 extensor MVT and peak RTD in the intact limb that were greater than deficits that have  
349 previously indicated altered joint mechanics and loading patterns. It is therefore plausible that  
350 without the influence from the prosthetic limb or the requirement for continued forward  
351 progression, the intact limb of ITTAs can attenuate load when landing from a jump up to 30  
352 cm in height similar to able-bodied controls. Further, the ITTA participants included in this  
353 study (otherwise healthy and recreationally active) would suggest that the risk for joint  
354 degeneration is potentially similar to those in uninjured persons. Utilisation of unilateral drop  
355 landings in rehabilitation and exercise programmes for less-active or non-established ITTAs  
356 could aid in the development of strength and coordination and increase participation in sport  
357 and exercise.

358

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**Tables**

533

Table 1 Participant demographics (mean  $\pm$  SD) for ITTAs and able-bodied controls

	ITTA	Control	<i>p</i> -value	Hedges <i>g</i>
Age (years)	40.0 $\pm$ 9.0	34.0 $\pm$ 6.5	.064	0.69
Mass (kg)	84.5 $\pm$ 18	83.4 $\pm$ 11	.769	0.08
Height (cm)	177 $\pm$ 7.4	179 $\pm$ 6.2	.400	0.30

534

535

Table 2 Maximal voluntary isometric torque (MVT) and peak rate of torque development (RTD) for the intact limb of ITTAs and dominant control limb, mean  $\pm$  SD

	Intact Limb	Control Limb	<i>p</i> -value	Hedges <i>g</i>
MVT (Nm/kg)	2.29 $\pm$ 1.2	2.79 $\pm$ 0.6	.134	0.61
Peak RTD (Nm/kg/s)	19.6 $\pm$ 9.9	25.3 $\pm$ 6.7	.084	0.72

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## Figure Captions

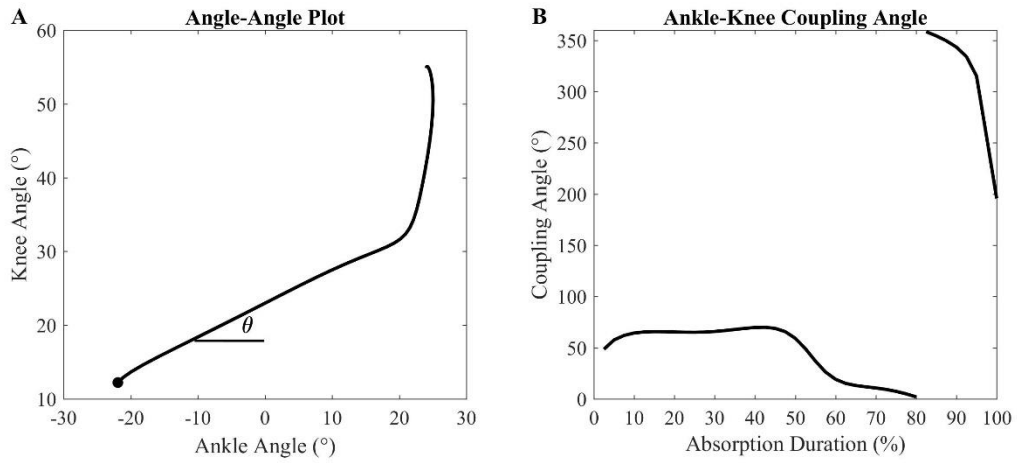
539 **Figure 1** – A) Angle-angle plot example of one healthy participant with touchdown denoted  
540 by the black circle and the coupling angle denoted by  $\theta$ . B) Coupling angle calculated from the  
541 angle-angle plot for the absorption phase.

542 **Figure 2** - Each row presents the 3-dimensional loading waveforms for the A) GRFs, B) knee  
543 moments, and C) intersegmental knee forces (KF) in the intact limb (IL; red dashed) and  
544 dominant control limb (DCL; black solid). Loading waveforms are presented for the duration  
545 of the absorption phase. Positive values are denoted first: GRFx = lateral-medial, GRFy =  
546 anterior-posterior, external knee adduction moment (KAM) = adduction-abduction, external  
547 knee flexion moment (KFM) = flexion-extension, external knee rotational moment (KMz) =  
548 internal-external, KFy = lateral-medial, KFx = anterior-posterior, and KFz = compression.

549 **Figure 3** - A) Joint angular position at touchdown (TD) and joint range of motion (ROM) in  
550 the sagittal and frontal planes, B) joint coordination coupling angle for the ankle-knee (AK),  
551 knee-hip (KH) and hip-ankle (HA), and C) peak joint absorption powers when landing at the  
552 ankle, knee, and hip joints in the intact limb (IL) and dominant control limb (DCL).

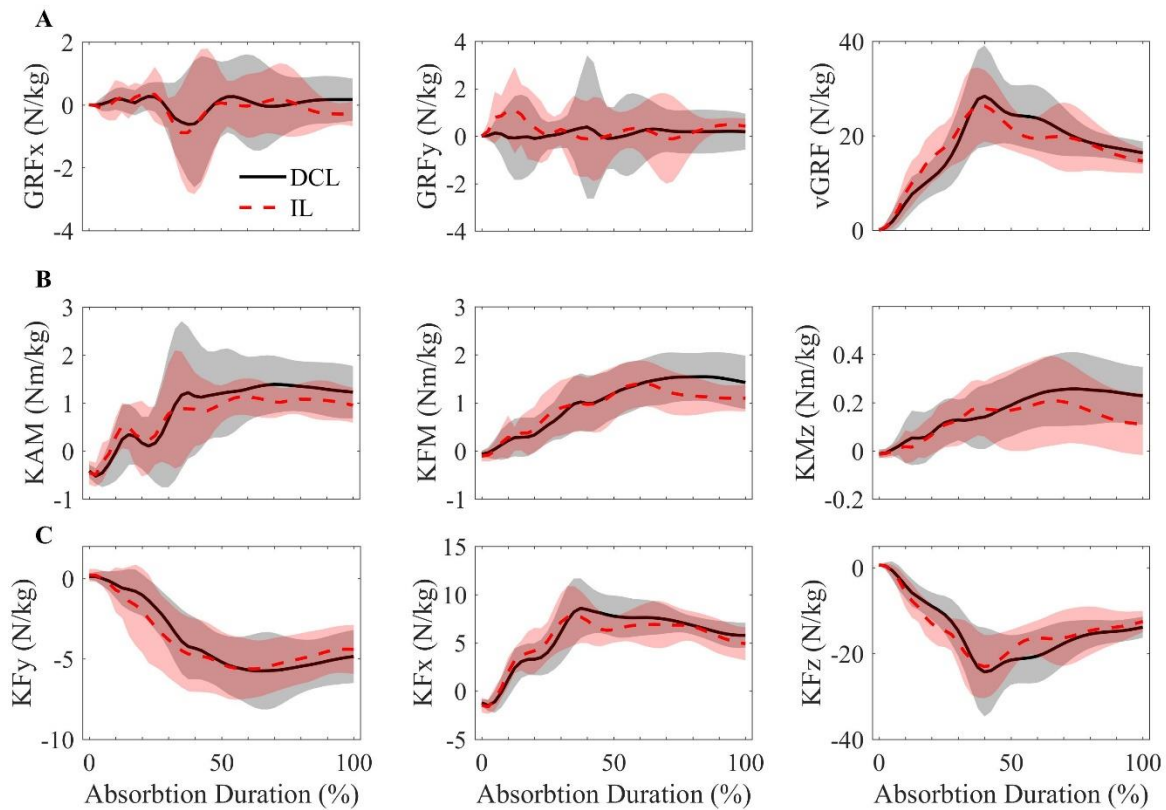
553 **Figure 4** - A) Individual joint work and B) joint percentage contribution of the total negative  
554 joint work performed at the ankle, knee, and hip joints for the intact limb (IL) and dominant  
555 control limb (DCL) during the absorption phase of landing.

556 **Supplementary Figure 1** – SPM {t}-statistic results for the loading waveform analysis  
557 (Figure 2). The horizontal red dashed lines represent the boundaries for statistical  
558 significance.



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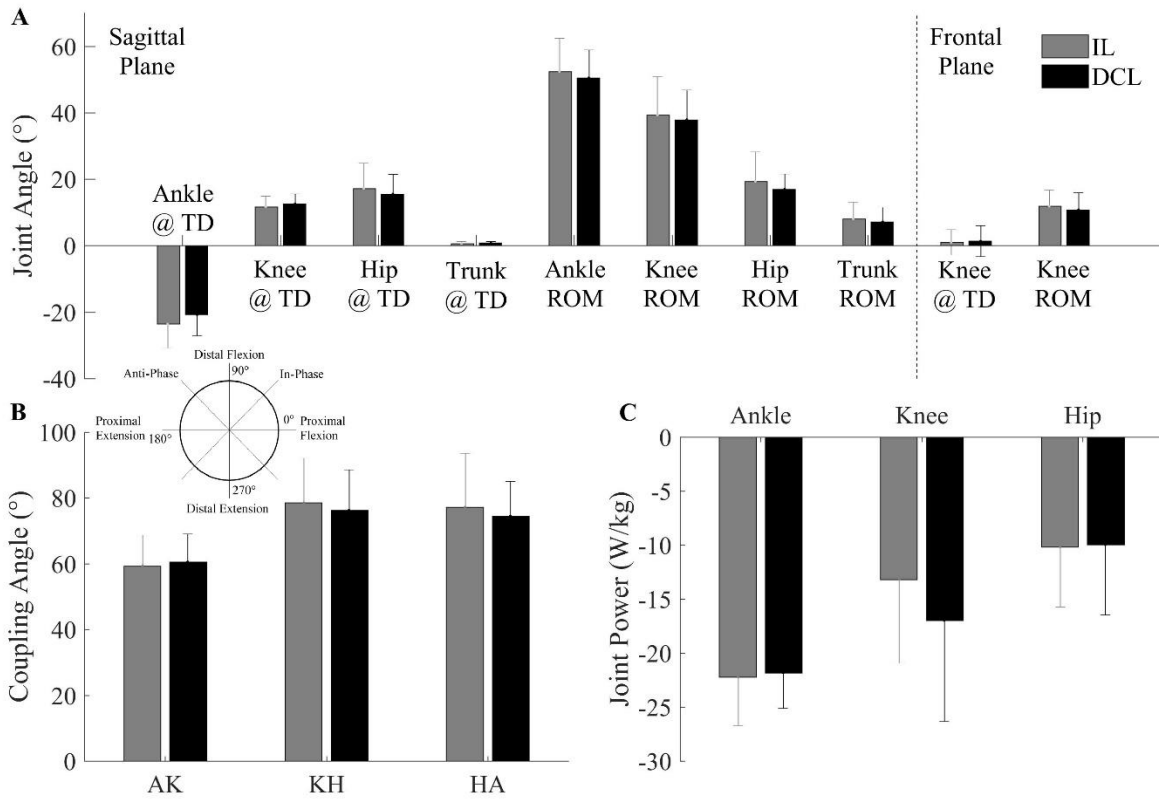
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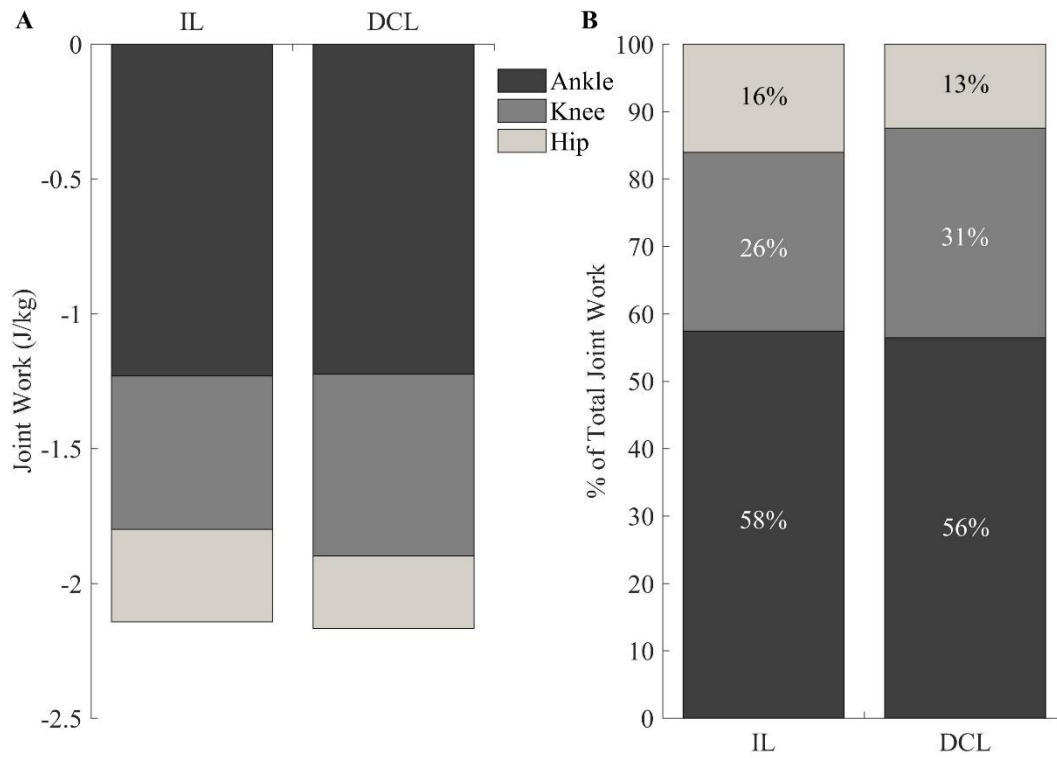
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573 **Figure 3 - A)** Joint angular position at touchdown (TD) and joint range of motion (ROM) in  
 574 the sagittal and frontal planes, B) joint coordination coupling angle for the ankle-knee (AK),  
 575 knee-hip (KH) and hip-ankle (HA), and C) peak joint absorption powers when landing at the  
 576 ankle, knee, and hip joints in the intact limb (IL) and dominant control limb (DCL).

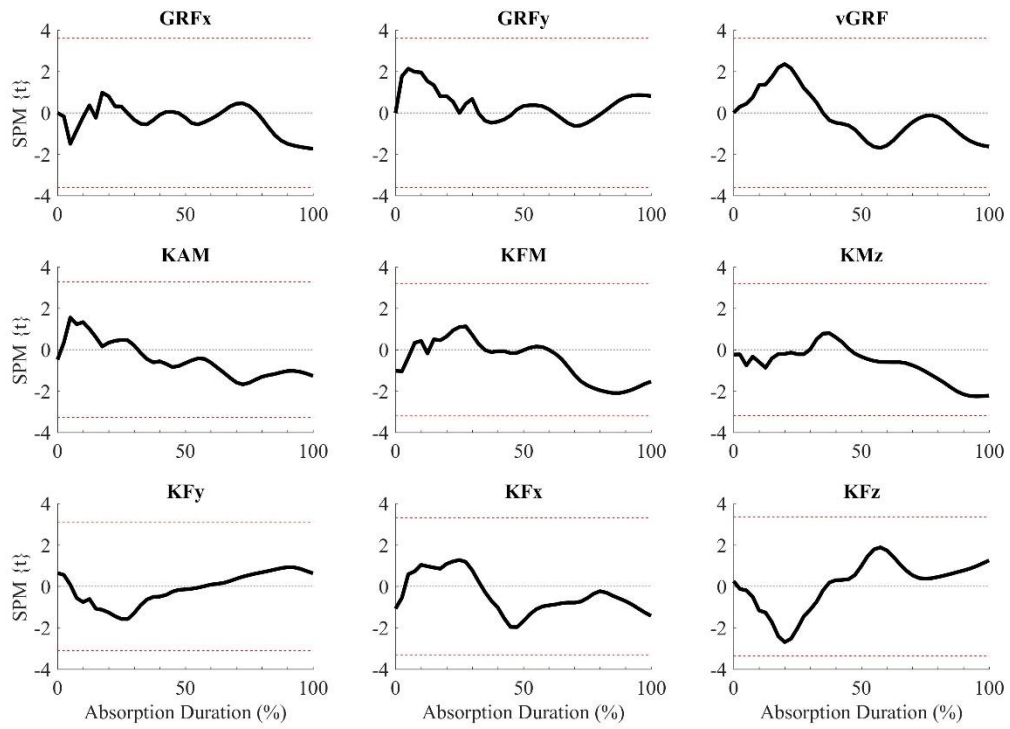
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579 **Figure 4** - A) Individual joint work and B) joint percentage contribution of the total negative  
 580 joint work performed at the ankle, knee, and hip joints for the intact limb (IL) and dominant  
 581 control limb (DCL) during the absorption phase of landing.

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583

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