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1 **Hip and knee joint contact loads in older adults during recovery from forward loss of 2**
3 **balance by stepping**

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21

22 **Abstract**

23 The purpose of this study was to use a musculoskeletal model to estimate hip and knee joint
24 contact loads in community dwelling older adults during maximal recovery from forward loss
25 of balance by stepping and to assess the association of contact loads with other measures of
26 recovery kinematics. Participants (n = 106) were released from a series of increasing static
27 forward lean angles until the maximum lean angle that each participant could recover from
28 with a single step was identified. Peak hip and knee joint contact loads following touchdown
29 of the stepping leg were computed using muscle force estimates obtained using static
30 optimisation. Peak contact loads ranged from 5.1-12.3 body weights for the hip and 3.2-10.7
31 body weights for the knee. Peak joint contact loads were significantly correlated with the
32 initial lean magnitude (Hip: $r = 0.55$; Knee: $r = 0.32$), as well as trunk flexion angle at foot
33 contact (Hip: $r = 0.30$; Knee: $r = 0.35$) and step length (Hip: $r = 0.54$; Knee: $r = 0.24$).
34 Overall findings indicated that older adults experience joint contact loads during maximal
35 balance recovery by stepping that are 3-4 times higher than those reported for normal gait,
36 and exceed hip contact loads previously reported to cause femoral fractures in individuals
37 with severe osteoporosis and suboptimal neuromuscular function. Improving trunk control
38 during recovery from forward loss of balance by stepping may decrease joint contact loads
39 and corresponding the risk of bone and/or joint injury.

40 Abstract length = 242 words

41 **Introduction**

42 Contact loads in the hip joint during normal walking are reported to be in the vicinity of 2-4
43 times body weight (Bergmann et al. 2001; Bergmann et al. 1993) and are considered unlikely
44 to cause spontaneous hip fracture because the mechanical failure load of cadaveric femurs

45 from older adults ranges from 5.5 to 14 body weights (Schileo et al. 2014). However
46 Viceconti et al. (2012) demonstrated via the use of a musculoskeletal modelling approach that
47 a combination of sub-optimal neuromuscular control and severe osteoporosis may make
48 spontaneous fracture during walking feasible, and thereby explain the small proportion of
49 femoral fractures that occur in the apparent absence of high-energy trauma typically
50 experienced due to a fall. It therefore follows that motor tasks where larger impulsive loads
51 than those associated with gait are applied, could produce hip loads that are in the range
52 associated with failure, perhaps even in the absence of degraded neuromuscular control and
53 severe osteoporosis. One such motor task where high joint contact loads are experienced is
54 the stumbling response used to recover balance from a trip perturbation. Bergmann et al
55 (1993) reported peak hip contact loads as high as 8.7 body weights in patients fitted with an
56 instrumented hip replacement during successful recovery from an unexpected trip
57 perturbation experienced during walking. At present however the magnitude of hip and knee
58 joint contact loads during maximal balance recovery by stepping, and the extent to which
59 these forces are affected by the balance perturbation intensity and motor control strategy used
60 during balance recovery by stepping remain unknown. Such information would inform efforts
61 to understand the mechanical risk factors associated with femoral fracture and implant
62 loosening and help identify ways by which hip and knee joint contact loads experienced
63 during balance recovery by stepping may be reduced.

64 The ability of older adults to recover from a large forward balance perturbation by stepping
65 significantly predicts the risk of real world falls in the following 12 months (Carty et al.
66 2015) and is largely determined by an ability to resist forward trunk flexion during the
67 stepping response (Barrett et al. 2012; Grabiner et al. 2008; Owings et al. 2001), take a
68 suitably long recovery step (Graham et al. 2015; Karamanidis et al. 2008; Schillings et al.

69 2005) and produce adequate hip and knee joint powers in the stepping limb (Carty et al.
70 2012b; Graham et al. 2015; Madigan 2006). Recovery step length, trunk angle at touchdown
71 of the stepping limb and lower limb joint moments and powers during recovery from forward
72 loss of balance are all reported to increase with balance perturbation intensity (Carty et al.
73 2012b; Madigan et al. 2005) and would therefore be expected to result in a corresponding
74 increase in lower extremity muscle force and hence joint contact loads for larger balance
75 perturbations. Poor trunk control in particular has been shown to result in more cocontraction
76 of spine, hip and knee muscles during balance recovery from an equivalent balance
77 perturbation (Graham et al. 2014) and might therefore be considered an example of
78 suboptimal motor control that adversely affects balance recovery and simultaneously increase
79 joint contact forces.

80 The purpose of this study was to use a musculoskeletal model to estimate hip and knee joint
81 contact loads in older adults during recovery from a forward loss of balance when released
82 from their maximum recoverable initial static forward lean angle. A secondary purpose was
83 to assess the association of contact loads with other measures of balance recovery kinematics.
84 We hypothesised that the magnitude of joint contact loads during balance recovery would be
85 positively correlated with initial lean magnitude as well as variables previously reported to
86 influence recovery performance, namely increased step length and increased trunk flexion
87 angle at foot contact of the stepping leg.

88

89 **Methods**

90 *Participants*

91 One hundred and six community dwelling older adults aged 65 to 80 years (age: 72.0 ± 4.8
92 years; height: 1.67 ± 0.09 m, mass: 75.4 ± 12.5 kg) were recruited at random from the local
93 electoral roll. Individuals previously diagnosed with neurological, metabolic,

94 cardiopulmonary, musculoskeletal and/or uncorrected visual impairment were excluded.
95 Ethics approval was obtained from the Institutional Human Research Ethics Committee and
96 all relevant ethics guidelines including provision of informed consent were followed.

97 *Experimental procedures*

98 The balance recovery protocol was undertaken as reported in Carty et al., (2011). Participants
99 stood barefoot with their feet shoulder-width apart in an upright posture and were
100 subsequently tilted forward, with their feet flat on the ground, until the required load in body
101 weight (BW) was recorded on a load cell (S1W1kN, XTRAN, Australia) placed in series with
102 an inextensible cable. One end of the cable was attached to a safety harness worn by the
103 participant at the level of their sacrum and the other end was attached to an electric winch on
104 a rigid metal frame located behind the participant. The length of the cable was adjusted until
105 the required force on the cable was achieved. Care was taken to ensure the cable was aligned
106 parallel with the ground and that participants kept their head, trunk and extremities aligned
107 prior to cable release. The cable was released at a random time interval (2-10 s) following
108 achievement of the prescribed posture and cable force ($\pm 1\%BW$), through the disengagement
109 of an electromagnet located in-series with the cable. Participants were instructed to relax their
110 muscles while leaning and to regain balance with a single step using the stepping lower limb
111 of their choice following cable release. The instruction to attempt to recover using a single
112 step was reiterated prior to every trial. A second cable, instrumented with a load cell
113 (S1W1kN, XTRAN, Australia), attached the safety harness to the ceiling, was used to prevent
114 participants from contacting the ground in the event of a failed recovery. Centre of pressure
115 location was displayed in real time on a computer monitor and was visually inspected by the
116 investigator to ensure anticipatory actions (e.g., antero-posterior and medio-lateral weight
117 shifting) were not evident in the period immediately prior to cable release. Following an
118 initial trial at a 15%BW lean angle the Maximal Recoverable Lean Angle (MRLA) was

119 determined by systematically increasing the lean magnitude by $\sim 1\%$ BW increments until the
120 participant could no longer recover with a single step. For each trial, participants were
121 classified as adopting either a single or a multiple step balance recovery strategy using
122 previously published criteria (Carty et al. 2011) where a multiple step was identified by a) a
123 second step of any kind by the stepping limb or progression of the non-stepping limb past the
124 stepping foot following the initial step, b) lateral deviation of the lateral malleolus marker on
125 the non-stepping foot by greater than 20% of body height from its position at cable release
126 and c) if a force of greater than 20% BW was detected in the load cell attached to the ceiling
127 restraint. For the purpose of this study only the MRLA trial was analysed. Trajectories of 51
128 reflective markers attached to each participant (Barrett et al. 2012) were recorded at 200 Hz
129 using a 10 camera, 3-dimensional motion capture system (Vicon Motion Systems, LA, USA).
130 Ground reaction forces were collected simultaneously at 1 kHz using two 900 mm x 600 mm
131 piezoelectric force platforms (Kistler Instruments, USA). One plate was located beneath the
132 two feet in the initial forward lean position and the second plate was located 800mm (center
133 of plate one to center of plate two) anterior to the first plate in order to record ground reaction
134 forces associated with touchdown of the stepping foot. Marker trajectory and ground reaction
135 force data were filtered using a 4th order, zero-lag, low-pass, Butterworth filter with a cut-off
136 frequency of 20Hz (Bisseling et al. 2006). Specific events during the stepping phase of
137 balance recovery were defined as follows: Cable release (CR) was identified from a 5 N drop
138 in force measured in the horizontal restraining cable, toe off (TO) was identified from the first
139 vertical motion greater than 2.5 mm of the great toe marker on the stepping foot (De Witt
140 2010) and foot contact (FC) from a force in excess of 5% of the participants body weight
141 recorded on the anterior force plate. For the purpose of this study the length of each trial was
142 the period from CR to 0.25 seconds after FC.

143 *Computation of hip and knee joint contact forces*

144 All data analysis were performed using OpenSim (version 3.2) (Delp et al. 2007) in
145 conjunction with custom Matlab scripts (Version 2014b, The Maths Works, USA). A scalable
146 anatomical model consisting of 17 bodies, 17 joints, 92 muscle actuators and 36 degrees-
147 offreedom (Hamner et al. 2010) was used as the initial generic model for analysis. A wrap
148 object was embedded in the generic model to maintain anatomically accurate erector spinae
149 muscles moment arms during trunk flexion (Graham et al. 2014). Model scaling and inverse
150 kinematic analysis (Lu et al. 1999) were performed by fitting the anatomical model to
151 measured 3D marker positions with a high weighting on virtual markers which defined the
152 joint centre of the hip, knee and ankle. Joint centres were estimated from experimental marker
153 trajectories: the regression equations of Harrington et al. (2007) were used for the hip joint (as
154 suggested by Kainz et al., 2015), while the knee and ankle joint centres were identified as the
155 midpoints of the femoral condyles and the medial and lateral malleoli respectively. Residual
156 Reduction Analysis (RRA) was subsequently performed to improve the dynamic consistency
157 between measured ground reaction forces and the mass-acceleration product of the model
158 (Delp et al. 2007).

159 The Static Optimisation tool in OpenSim was used to calculate muscle forces using a cost
160 function to minimise the sum of squared muscle activations within the force-length-velocity
161 constraints of each muscle. An evaluation of the simulations was conducted by comparing the
162 experimentally collected muscle EMG to the corresponding muscle activation from static
163 optimisation (see Supplementary Figure 1) in accordance with recommended best practice
164 (Hicks et al. 2015). Passive muscle forces were also checked for each simulation and found to
165 be negligible (i.e. muscles tended to operate on the ascending limb and plateau region of the
166 force-length relation). Joint contact loads were computed using the Joint Reaction analysis
167 available in OpenSim, which calculates contact loads through a recursive procedure

168 equivalent to resolving the free body diagrams of the rigid bodies included in the model,
169 starting from the most distal and moving proximally (a detailed description of the tool
170 implementation can be found in Steele et al., 2012). An example of contact loads calculated at
171 the hip and knee joint for a representative subject is presented in Figure 1, together with
172 ground reaction forces for both legs. The same OpenSim analysis was used to calculate joint
173 reactions by disabling the muscles and providing the joint moments necessary to equilibrate
174 the model through idealized torque actuators. The relative contribution of muscle forces to the
175 total joint contact load was obtained by subtracting the joint reaction load from the total joint
176 contact load.

177

178 *Statistical Analysis*

179 The Pearson Product Moment Correlation Coefficient was used to examine the relations
180 between joint contact loads and initial lean angle, step length normalised to participant leg
181 length (leg length was defined as the distance between the hip and ankle joint centres) and
182 trunk flexion angle at foot contact (relative to the vertical axis). Statistical analyses were
183 performed using SPSS (Version 22, IBM SPSS, USA). Significance was accepted for $P < 0.05$.

184 **Results**

185 Mean peak contact loads were approximately 8 and 6 times body weight for the hip and knee
186 respectively (Table 1). The largest peak joint contact loads experienced by an individual
187 were 12.3 BW for the hip joint and 10.7 BW for the knee joint. Muscle forces contributed
188 95% of the total hip joint contact force and 80% of the total knee joint contact force
189 respectively. Hip and knee joints joint contact loads were significantly correlated to maximal
190 recoverable lean angle (Figure 2) as well as trunk flexion angle at foot contact and step length
191 (Figure 3) ($p < 0.05$ for all correlations).

192 **Discussion**

193 This study confirmed that large peak contact loads are generated in maximal recovery from
194 forward loss of balance by stepping, with individual peak contact loads following touchdown
195 of the stepping leg ranging from 5.1-12.3 body weights for the hip and 3.2-10.7 body weights
196 for the knee. In support of our hypothesis, the magnitude of hip and knee joint contact load
197 was positively correlated with the intensity of the balance perturbation, and also with
198 variables previously demonstrated to be associated with recovery performance, namely trunk
199 flexion angle at touchdown of the stepping leg and recovery step length.

200 When released from the mean static forward lean angle of approximately 21 degrees, average
201 peak joint contact loads were approximately 8 BW for the hip and approximately 6 BW for
202 the knee. Average peak hip joint contact loads in the present study were therefore similar to
203 the peak value of 8.7 BW reported for stumbling by Bergmann et al (1993), approximately 4
204 times higher than previously reported for slow walking on level ground (Bergmann et al.
205 2001) and approximately 1.5 times larger than those reported for stair descent and running at
206 8 km/hr (Bergmann et al. 1993). Similarly, peak knee joint contact loads were approximately
207 3 times higher than those reported for walking (Fregly et al. 2012) and around 1.7 times
208 higher than those for stair descent (Kutzner et al. 2010). The relationship between joint

209 contact load and the intensity of the balance perturbation was strongest for the hip ($r = 0.55$),
210 where average peak hip loads increased by a factor of around 2 across the range of
211 perturbation intensities examined, compared with the knee ($r = 0.32$), which increased around
212 1.5 times over the same range of perturbation intensities. In agreement with previous reports
213 (Correa et al. 2010; Herzog et al. 2003; Winby et al. 2009) muscle force was the main
214 determinant of the joint contact force, which in the present study accounted for 95% of total
215 hip contact force and 80% of total knee contact force. The ability to generate large hip muscle
216 forces and sustain large hip joint contact loads therefore appears critical for successful
217 recovery from forward loss of balance by stepping which is consistent with the finding that
218 lower limb muscle weakness predicts the ability of older adults to recover from forward loss
219 of balance with a multiple compared to single steps (Carty et al. 2012a). The large peak hip
220 joint contact loads identified in the present study are also similar to the upper limit of around
221 9 BW reported by Martelli et al. (2011) to be feasible during walking in cases of severe
222 neuromotor degradation, and according to Viceconti et al. (2012), capable of producing
223 spontaneous hip fractures in the presence of severe osteoporosis of the hip and degraded
224 neuromuscular function. Balance recovery could therefore be a motor control task that
225 imposes risk of hip fracture in individuals, particularly following large balance perturbations
226 in individuals with suboptimal neuromuscular control and low bone mineral density.

227 Step length was correlated to the joint contact loads at the hip ($r = 0.54$) and knee ($r = 0.24$).
228 A long step is important for balance recovery because it places the base of support further in
229 front of the whole body centre of mass where GRF vector can more effectively reduce
230 forward and downward centre of mass progression. However a large step also comes at the
231 expense of larger joint contact forces, especially at the hip where a doubling of step length
232 corresponds to around 50% increase in hip joint contact force. It therefore follows that
233 participants that use short steps may do so to minimise joint loading, perhaps to maximise

234 joint stability (Bergmann et al. 2004) or minimise joint pain, even though balance recovery is
235 compromised.

236 The trunk flexion angle at foot contact was significantly correlated with hip ($r = 0.30$) and
237 knee ($r = 0.35$) joint contact force. Excessive trunk flexion represents suboptimal motor
238 control during balance recovery (Graham et al. 2014), and can also distinguish between older
239 adults that require multiple versus single steps to recover from a fixed initial lean magnitude
240 (Barrett et al. 2012; Grabiner et al. 2008; Owings et al. 2001). Given that the amount and rate
241 of trunk flexion during balance recovery can be improved through repeated exposure to the
242 task (Barrett et al. 2012; Carty et al. 2012c), training may be expected to reduce joint contact
243 loads during balance recovery through the combined effect of improved trunk control and
244 corresponding reduction in step length required to achieve dynamic stability.

245 A limitation of the present study was that, consistent with previous computational studies
246 aiming to estimate hip contact loads in daily living activities (Modenese et al. 2012;
247 Modenese et al. 2011), muscle forces were estimated using static optimisation with a cost
248 function that minimised muscle activation squared (Crowninshield et al. 1981). Joint contact
249 loads reported here are therefore unlikely to reflect suboptimal neuromuscular control
250 (Martelli et al. 2011; Modenese et al. 2013) including high levels of muscle co-contraction
251 which would be expected to result in even higher joint contact loads. Further Bergmann et al.
252 (2004) reported that unanticipated versus anticipated loss of balance resulted in higher versus
253 lower contact loads during balance recovery. As participants in the present study were aware
254 of their impending loss of balance, just not the exact timing, it is possible that the joint
255 contact loads reported here may be smaller than for a completely unanticipated fall. In future
256 it will be necessary to evaluate how the application of joint contact force vectors interact with
257 the geometry and material properties of the proximal femur to more accurately determine a
258 direct link to risk of femoral fracture.

259 **Conclusion**

260 Hip and knee joint contact loads in the stepping limb during recovery from forward loss of
261 balance in older adults are 2-3 times higher than those previously reported for normal gait.
262 Hip joint contact loads in particular were of similar magnitude to those previously reported to
263 cause femoral fracture in individuals with a combination of suboptimal control and severe
264 osteoporosis. Although a long recovery step is a feature of successful balance recovery, this
265 comes at the expense of increased hip and knee contact forces. Conversely, large trunk
266 flexion angles are a feature of poor balance recovery performance, and are also associated
267 with larger hip and knee contact forces. Balance training that improves trunk control may
268 therefore simultaneously improve balance recovery performance and decrease hip and knee
269 joint loading.

270 **Conflict of Interest**

271 The authors declare that they have no conflicting interests.

272

273 **Acknowledgement**

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Figure 1
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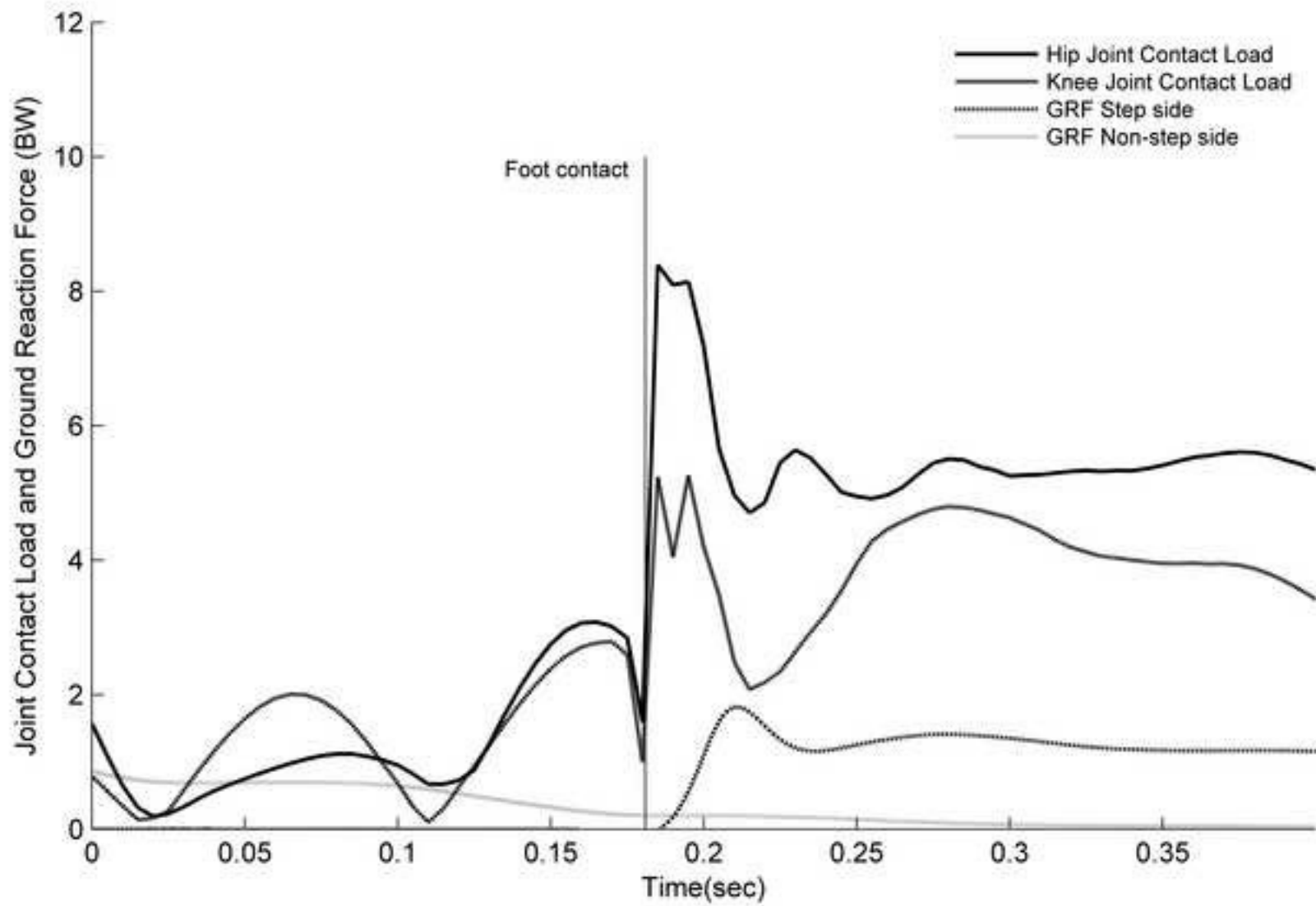


Figure 2
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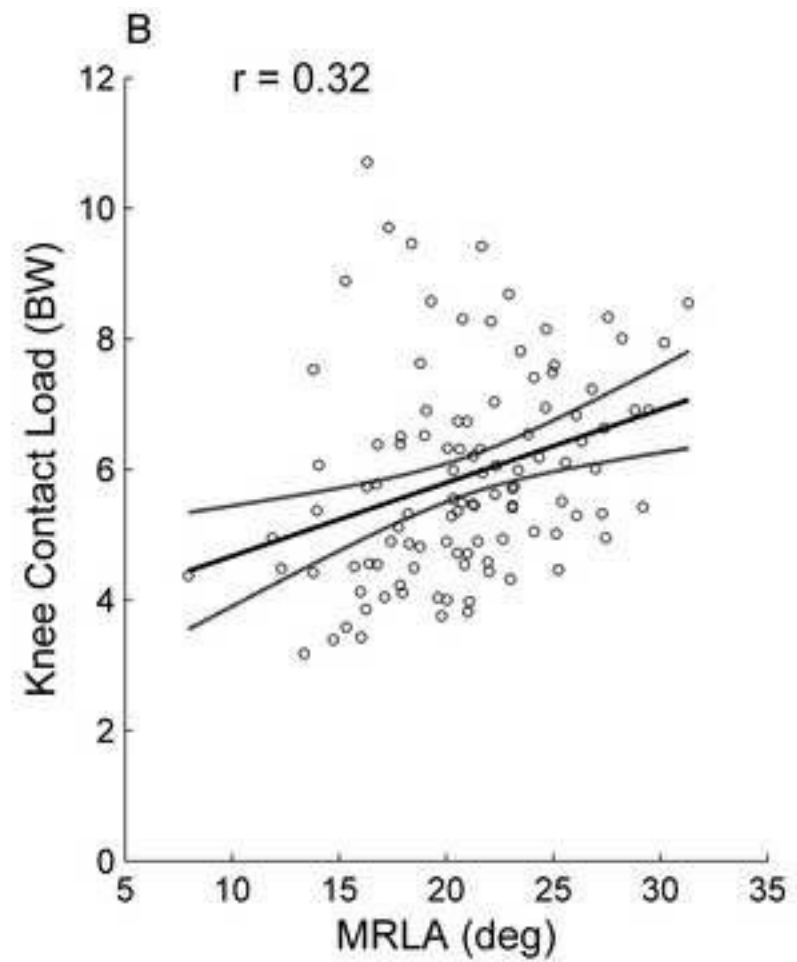
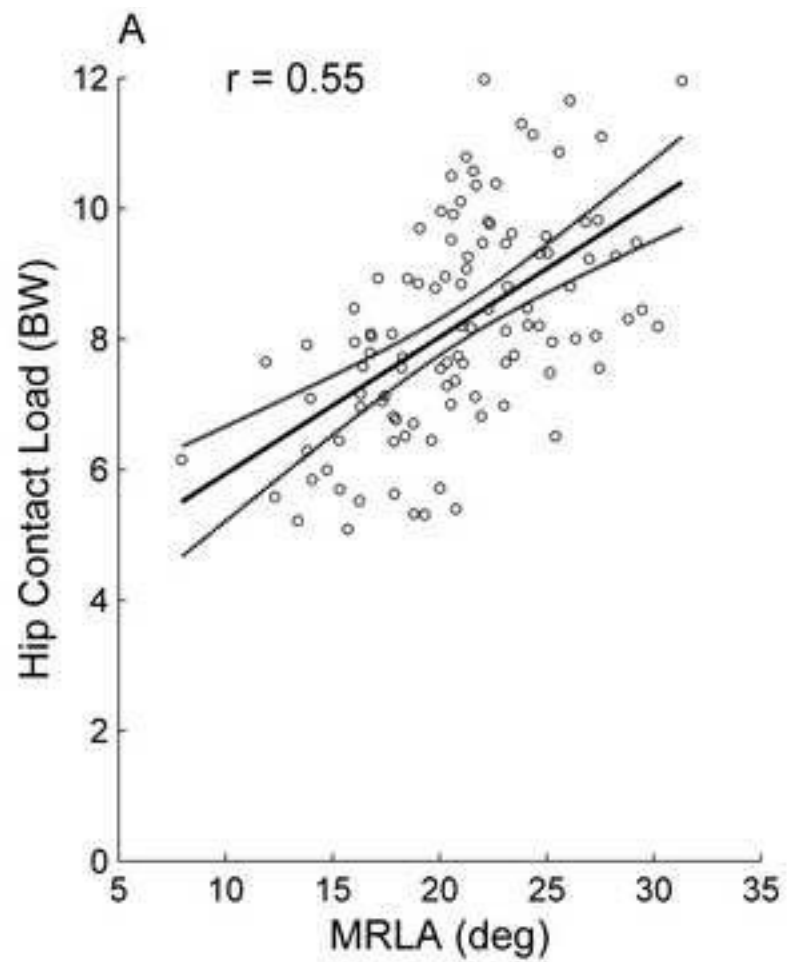


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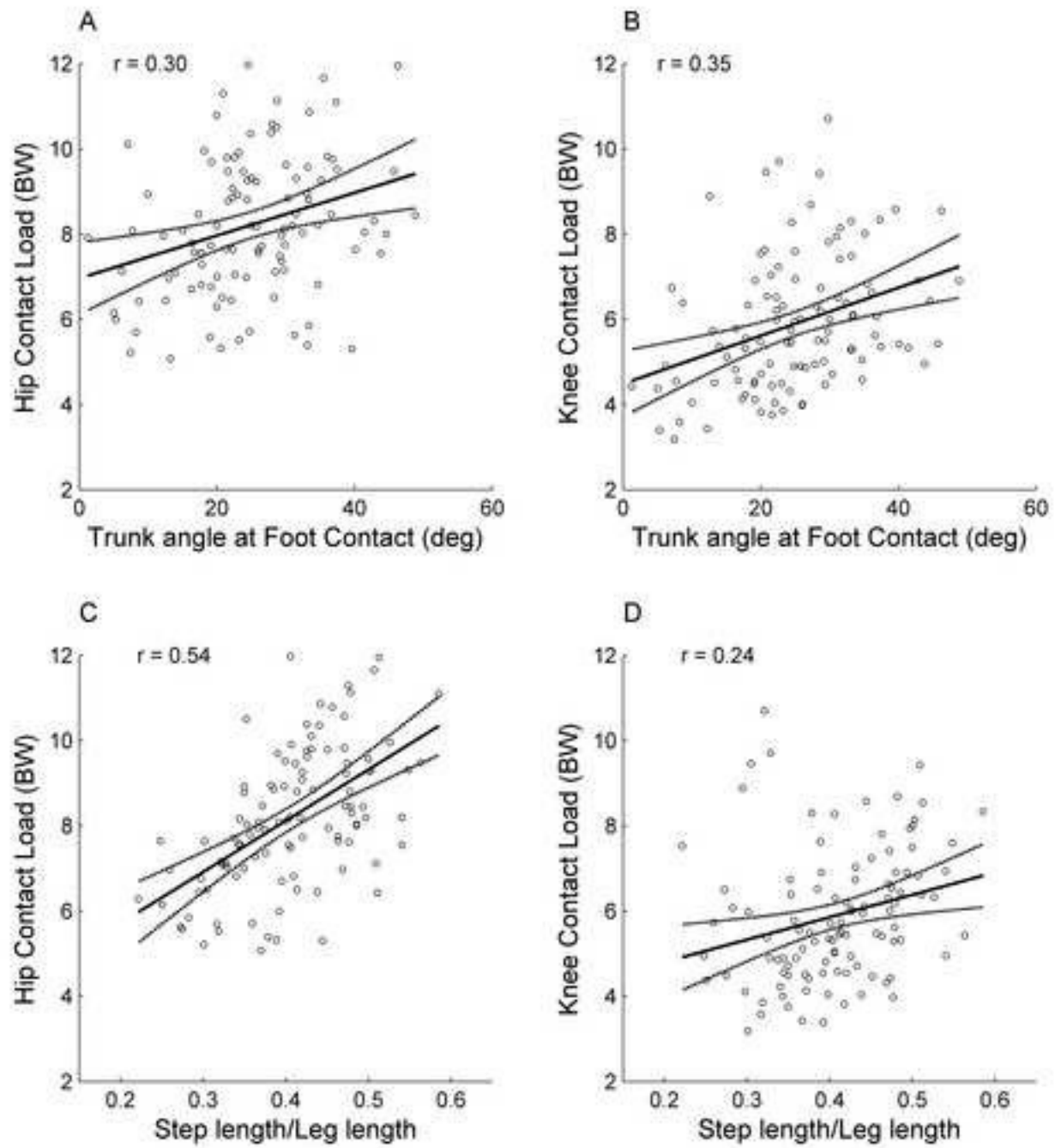


Table 1

Table 1. Summary data for balance recovery (n = 106).

| | Mean \pm SD | 95% CI | Range |
|--|-----------------|-----------|-----------|
| Maximum recoverable lean angle ($^{\circ}$) | 20.9 \pm 4.4 | 20.2-21.8 | 7.9-31.3 |
| Peak hip contact load (BW) | 8.22 \pm 1.68 | 7.94-8.55 | 5.1-12.3 |
| Peak knee contact load (BW) | 5.90 \pm 1.60 | 5.64-6.24 | 3.2-10.7 |
| Peak ground reaction force (BW) | 1.83 \pm 0.57 | 1.72-1.90 | 0.98-3.39 |
| Trunk flexion angle at foot contact ($^{\circ}$) | 25.0 \pm 9.9 | 23.3-26.9 | 11.3-48.8 |
| Step length/leg length | 0.41 \pm 0.08 | 0.40-0.42 | 0.22-0.59 |

ry Figure 1

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