1 BIOMECHANICAL DEMANDS OF THE 2-STEP TRANSITIONAL GAIT CYCLES LINKING
2 LEVEL GAIT AND STAIR DESCENT GAIT IN OLDER WOMEN

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4 **ABSTRACT**

5 Stair descent is an inherently complex form of locomotion posing a high falls risk for older

6 adults, specifically when negotiating the transitional gait cycles linking level gait and descent.

7 The aim of this study was to enhance our understanding of the biomechanical demands by

8 comparing the demands of these transitions. Lower limb kinematics and kinetics of the 2-

9 step transitions linking level and descent gait at the top (level-to-descent) and the bottom

10 (descent-to-level) of the staircase were quantified in 36 older women with no falls history.

11 Despite undergoing the same vertical displacement (2-steps), the following significant

12 (p<.05) differences were observed during the top transition compared to the bottom

13 transition: reduced step velocity; reduced hip extension and increased ankle dorsiflexion

14 (late stance/pre-swing); reduced ground reaction forces, larger knee extensor moments and

15 powers (absorption; mid-stance); reduced ankle plantarflexor moments (early and late

16 stance) and increased ankle powers (mid-stance). Top transition biomechanics were similar

17 to those reported previously for continuous descent. Kinetic differences at the knee and

18 ankle signify the contrasting and prominent functions of controlled lowering during the top

19 transition and forward continuance during the bottom transition. The varying musculoskeletal

20 demands encountered during each functional sub-task should be addressed in falls

21 prevention programmes with elderly populations where the greatest clinical impact may be

22 achieved. Knee extensor eccentric power through flexion exercises would facilitate a smooth

23 transition at the top and improving ankle plantarflexion strength during single and double

24 limb stance activities would ease the transition into level gait following continuous descent.

25

INTRODUCTION

Descending stairs is a common task that permits functional ambulation between different levels. The knee extensors and ankle plantarflexors play an important role in stair descent biomechanics (McFadyen and Winter, 1988; Samuel et al., 2011) by dissipating mechanical energy and enabling forward progression, respectively (Cluff and Robertson, 2011). Considerable eccentric control of the knee and ankle musculature is required to resist the downward influence of gravity as the body undergoes repetitive free fall from one step to the next. Stair locomotion presents a considerable falls risk with early work indicating that 14% of all falls occur on stairs (Cohen et al., 1986) and 75% of all stair-related falls occur during descent compared to ascent in older adults (Masud and Morris, 2001). An important element in designing effective falls prevention programmes requires a comprehensive biomechanical understanding of task demand.

Studies have frequently analysed gait cycles that are initiated and terminated on independent steps while participants negotiate the stairs using a step-over-step, reciprocal gait pattern representative of continuous descent (McFadyen and Winter, 1988; Christina and Cavanagh, 2002; Hamel et al., 2005; Sheehan and Gottschall, 2011). During continuous descent, older adults operate within a higher proportion of their maximal dynamometer-derived capacity for both knee moments (old *vs.* young; 42% *vs.* 30%) and ankle dorsiflexion angle (107% *vs.* 91%) (Reeves et al., 2008). Further work has confirmed that mechanical demands at the knee are greater than at the hip with older adults using on average 100%, and in some cases 150% of available capacity (Samuel et al., 2011). Functional demands at the hip were on average ~20% of available isometric hip strength for both the flexor and extensor muscles (Samuel et al., 2011). Demands exceeding 100% of capacity may reflect the age-related differences in voluntary drive to activate muscles during selected testing protocols and variation in the protocols utilised (i.e., contraction type, chosen angular

52 position/ velocity) which makes direct comparisons challenging. Whilst it is well known that

53 continuous descent poses heightened mechanical demands for older adults, the kinematic

54 and kinetic demands of the transitions linking level and continuous descent gait are less well

55 understood.

56

57 One study investigating the influence of step location (comparison between continuous

58 descent in the top and mid-stair region) upon ground reaction forces (GRF) during descent

59 found altered GRF in both young and old (Christina and Cavanagh, 2002). Interestingly, an

60 interaction effect was observed (step location*age) such that loading rates were larger as

61 participants progressed down the staircase and this was more apparent for older adults. In

62 support of this, Lee & Chou, (2007) showed that both young and older adults completed the

63 bottom transition more quickly compared to continuous descent. Moreover, the same study

64 indicated that unlike the young, older adults were unable to reduce their centre of mass

65 (COM) sway angles from continuous descent to the bottom transition which the authors

66 suggested may represent a reduced ability to stabilise during this transition (Lee and Chou,

67 2007). Given the likely increased severity of injury that would result from a fall from the top

68 compared to the bottom of the staircase, and the progressive change in demands thought to

69 occur throughout descent, analysis of lower limb mechanics during both transitions is vital to

70 provide a thorough understanding of task demand and falls risk.

71

72 To the best of the authors' knowledge, only one early study directly compared the top and

73 bottom transitions in young adults. This work revealed that whilst lower limb joints operate

74 within a similar range of motion (ROM) during both transitions, differing kinematic profiles

75 were observed (Andriacchi et al., 1980). Moreover, increased external hip and knee flexor

76 moments and earlier onset of knee extensor muscle activity were noted for the top transition,

77 albeit these differences were not evaluated statistically (Andriacchi et al., 1980) and require

78 confirmation. Redirecting the COM from one level to another requires a prescribed change in

79 lower limb mechanics modulated by changes in both step height and depth in response to

80 staircases of varying design. These movement alterations require a superior level of postural

81 and motor control facilitating appropriate multi-segment co-ordination. The biomechanical

82 requirements to complete both transitional phases are likely to differ from one another as

83 has been demonstrated for stair ascent (Alcock et al., 2014a) and when comparing 1-step

84 transitions with continuous stair gait (Sheehan and Gottschall, 2011). Identifying the

85 biomechanical demands of these transitions would guide evidence-based recommendations

86 for targeted exercises, especially in high-falls risk groups, and encourage safer stair

87 locomotion. This could have greatest impact for older women due to their increased falls

88 occurrence and amplified falls risk associated with stair locomotion (Blake et al., 1988;

89 Campbell et al., 1989; Gine-Garriga et al., 2009).

90 Therefore, the aim of this study was to compare the lower limb mechanics involved in the 2-

91 step transition from the top and bottom of the staircase in older women with no falls history.

92 It was hypothesised that functional differences would exist between the transitions

93 particularly during stance, with the top transition necessitating greater controlled lowering

94 and presenting demands similar to that of continuous descent (i.e., greater eccentric control

95 of the knee extensors in terminal stance) and the bottom transition stance phase closely

96 representing level gait (i.e., greater concentric knee power generation mid-stance, and larger

97 ankle plantarflexor moments).

98

99 **METHODS**

100 **PARTICIPANTS**

101 Thirty-six female participants gave written informed consent to take part in this study which

102 received National Health Service ethical approval (08-H1305-91). Participants were recruited

103 through the local community and were pre-screened to exclude cardiovascular,

104 musculoskeletal or neurological complaints, visual or cognitive deficits, polypharmacy or a

105 history of falls. Group mean[SD] characteristics were: age 71.7[7]years, range 61-83 years;

106 height 162.8[6.6]cm; mass 70.7[12.7]kg. This study was embedded within a larger project

107 that quantified biomechanical profiles of older women completing daily activities (Alcock et

108 al., 2013; 2014a; 2014b)

109

110 **PROTOCOL**

111 3D kinematics of the 2-step transition from the top and the bottom of the stairs were

112 recorded using 14 ProReflex infrared cameras sampling at 100Hz (Qualisys, Sweden).

113 Spherical reflective markers (14mm) were placed upon the participants' lower limbs

114 bilaterally according to a six degrees-of-freedom marker system (Cappozzo et al., 1995). A

115 custom-built staircase was utilised (step height: 20cm, depth: 25cm, width: 80cm, top landing

116 depth: 80cm) as described previously (Alcock et al., 2014a). Orthogonal GRFs were

117 measured using two 400x600mm piezoelectric force platforms (model 9286AA, Kistler,

118 Winterthur, Switzerland) sampling at 500Hz. One platform was mounted within the first step

119 and measured forces from the 2-step transition from the top of the staircase; while one

120 ground-mounted platform recorded forces from the 2-step transition at the bottom of the

121 staircase (Figure 1). Analogue data were converted through a 64-bit analogue-to-digital

122 board and recorded synchronously with kinematic data. Participants were instructed to begin

123 each trial from the back of the top landing and completed either one or two gait cycles on the

124 landing before descending the stairs completing a total of 8-10 descent trials. Participants

125 were asked to continue walking beyond the bottom of the staircase (approximately 4 metres)

126 at their self-selected pace.

127

128   All participants used a reciprocal stepping pattern naturally and without prompt, and no

129   participant used the handrails. During descent, and on the 3-step staircase used in this study

130   (Figure 1), the lead limb initially descended from the top landing to step 2 (1-step top

131   transition). The trail limb then descended two steps from the top landing to step 1 (2-step top

132   transition). The next step of the lead limb was from step 2 to the ground (2-step bottom

133   transition). The trail limb then descended from step 1 onto the ground (1-step bottom

134   transition). It is noteworthy that, depending on the number of steps within a given staircase,

135   the lead/trail limb functions will alter during the bottom transition. This study is specifically

136   focused on comparing the 2-step transitions from the top and bottom of the staircase rather

137   than the 1-step transitions due to the larger vertical displacement involved and consequently

138   larger ROM required.

139

140   **VARIABLES**

141   Extracted temporal-spatial variables included velocity (m/s), cycle time (s) and stance phase

142   duration (%). Peak lower limb joint angles and ROM were calculated during each of the 2-

143   step transitions and joint angles were time-normalised to 100% gait cycle. The gait cycle was

144   standardised to begin with toe-off, thus presenting the swing phase first followed by the

145   stance phase, to facilitate comparisons with previous studies' (McFadyen and Winter, 1988;

146   Karamanidis and Arampatzis, 2010; Sheehan and Gottschall, 2011). Foot contact and toe-off

147   events were identified from the vertical displacement of the forefoot markers relative to the

148   staircase structure and corroborated with GRF data when available. Peak medial (F_{x1}),

149   lateral (F_{x2}), posterior (F_{y1}) and anterior (F_{y2}) GRF values were quantified. In addition,

150   peak vertical forces during early (F_{z1}) and late stance (F_{z3}), the minimum force mid-stance

151   (F_{z2}), and load and decay rates were analysed. GRF data were normalised to body mass

152   and time-normalised to 100% stance. Inverse dynamics were used to calculate lower limb

153   sagittal joint moments and powers and were time-normalised to 100% gait cycle. Body

154   mass-normalised peak joint powers were determined according to the specific bursts defined

155   by McFadyen & Winter (1988). To ensure that kinetic differences observed were not

156   influenced by alternative force plate mounting structures, fast Fourier analysis was

157   performed on the force plate in each of the settings used (concrete pit and wooden inset in

158   the staircase structure). This analysis revealed that kinetic data were not confounded as a

159   result of force platform mounting structure (Chesters et al., 2013) and results are presented

160   in the *supplementary material*.

161

162   **DATA ANALYSIS**

163   A static calibration trial was collected prior to the movement trials to define segment lengths

164   and identify lower limb joint centres. The hip joint centres were derived from the CODA

165   pelvis which was constructed in Visual 3D (Bell et al., 1989; Bell et al., 1990). The knee and

166   ankle joint centres were defined as the midpoint between the markers defining the lateral

167   and medial aspects (i.e. femoral epicondyles and malleolus of the fibula and tibia,

168   respectively) of two articulating segments (i.e. thigh and shank, respectively). Marker

169   trajectories were identified and labelled in Qualisys Track Manager (v.2.7, Qualisys,

170   Sweden), then exported to Visual 3D (v.3.90.7, C-Motion, Germantown, MD, USA) for

171   subsequent analysis. Kinematic data were interpolated over a maximum gap of ten frames

172   using a cubic spline algorithm and an X-Y-Z Cardan sequence defined the order of rotations

173   according to the right hand rule about the segment coordinate axes (x: flexion/extension, y:

174   abduction/adduction and z: longitudinal rotation). Kinematic and kinetic data were filtered

175   using a low-pass Butterworth filter with cut-off frequencies of 6Hz and 25Hz, respectively

176   (Siegel et al., 1996) and all data were averaged across the completed trials.

177

178 **STATISTICAL ANALYSIS**

179 Paired samples t-tests were conducted to analyse the biomechanical differences between

180 the top *vs.* bottom transitional gait cycles. Paired comparisons were split into three groups:

181 temporal-spatial, kinematic, and kinetic indices. A family-wise Hommel correction was used

182 to manage the Type I error associated with multiple comparisons (Hommel, 1988; Falk,

183 1989). Two-tailed significance was reported as the direction of the group differences was not

184 known. Where statistical differences were found, effect sizes (Cohen's d) were calculated to

185 verify these differences. Significance was set at p≤.05.

186

187 **RESULTS**

188

189 A significantly faster velocity, shorter cycle time and stance phase duration were observed

190 for the bottom transition compared to the top (p≤.0018; d=4.6-10.2, Table 1).

191

192 Significant kinematic differences were observed between the two transitions at the hip and

193 ankle (Table 1 and Figure 2). The limb completing the top transition demonstrated increased

194 peak ankle dorsiflexion (late stance) and ankle ROM (p=.0064) compared to the limb

195 executing the bottom transition. Peak hip extension (late stance) and ankle plantarflexion

196 (late swing/ early stance), were significantly greater during the bottom transition compared to

197 the top (p=.0064, d=9.3 and 3.7, respectively).

198

199 Several GRF parameters (Fy1, Fz1, load and decay rates) were found to be statistically

200 greater for the bottom transition compared to the top transition (Table 2 and Figure 3). The

201 limb completing the bottom transition generated significantly greater ankle plantarflexor

202 moments during early and late stance compared to the top transition (p=.0095). All

203 statistically significant differences were confirmed by moderate Cohen's d effect sizes

204 ranging from 6.7-20.7. The largest difference was the peak knee extensor moment (late

205 stance) which was reduced during the bottom transition compared to the top transition

206 (d=20.7, p=.009).

207

208 The limb completing the bottom transition generated significantly greater knee power mid-

209 stance (p=.0095, Table 3 and Figure 4). Knee power absorption (late stance) and ankle

210 power absorption (mid-stance) were significantly reduced during the bottom transition

211 compared to the top. Differences in the peak ankle power generation (mid-stance) were

212 reduced during the bottom transition and were non-significant post-corrective procedures.

213

214 **DISCUSSION**

215 This study investigated the differences between the 2-step transitions from the top and

216 bottom of the staircase during stair descent. Despite both gait cycles undertaking a 2-step

217 cycle, distinct biomechanical differences and contrasting functional demands were observed.

218 In agreement with our hypothesis, the top transition was characterised by controlled lowering

219 (represented by a larger knee extensor moment and eccentric extensor control), similar to

220 continuous descent (McFadyen and Winter, 1988). In comparison, the bottom transition was

221 completed more quickly with larger GRFs and plantarflexor moments indicating a greater

222 requirement for forward propulsion into level gait.

223

224 **Demands of descent transitions compared to level gait**

225 Level gait mechanics for the same cohort have been reported previously (Alcock et al.,

226 2013). Both stair transitions were completed more slowly, with an increased cycle time and

227 reduced stance phase duration compared to level gait. Knee ROM was considerably greater

228 during both transitions (~90°) than during level gait (~60°) as was ankle ROM due to greater

229 dorsiflexion (~two-fold increase) and plantarflexion (~four-fold increase). Both the knee

230 extensor moment and knee power absorption burst were largest during the top transition

231 compared to level gait and the bottom transition. Increased ankle power generation was

232 observed during both transitions compared with level gait.

233

234 **Comparison between top and bottom transitions**

235 The two descent transitions were distinguished by peak hip extension angles during late

236 stance (top=9.2° flexion *vs.* bottom=2.3° extension, p=.0064) such that the hip never fully

237 extended during the top transition. Moreover, the participants in the current study

238 demonstrated more hip extension compared to the findings presented in Samuel et al.

239 (2011) (20° flexion). Similar magnitudes of hip flexion were noted for the top transition in the

240 present study and the continuous cycle reported in Reeves et al. (2008a). This suggests that

241 the stance phase of the top transition in the present study (which was completed on the

242 staircase) exhibited similar mechanics to that observed during continuous descent gait.

243 Variations in hip extension profiles during the top transition between the present study and

244 that of Samuel et al. (2011) may be attributed to varying staircase dimensions (height x

245 depth: 20x25cm vs. 18.5x28cm for the current *vs.* Samuel et al. (2011) study, respectively).

246 The large magnitude of hip extension observed during the bottom transition acts to facilitate

247 the increase in step length of the ipsilateral limb onto level ground thus conforming more

248 closely to the level gait mechanics of forward propulsion. This is in contrast to the top

249 transition, whereby step length is dictated by the proceeding staircase dimensions.

250 Therefore chosen step length beyond the staircase was not restricted by the impending step

251 depth and increasing step length beyond the staircase inherently necessitates increased hip

252 extension.

253

254 Kinematic differences further distinguishing between the two transitions included a

255 significantly reduced dorsiflexion angle (late stance) and greater plantarflexion angle (late

256 swing/early stance), which resulted in reduced ROM during the bottom transition. Greater

257 plantarflexion upon contact increases functional leg length and thus facilitates appropriate

258 foot placement whilst requiring less pelvic movement in the frontal plane (i.e., pelvic obliquity

259 - not analysed in the present study). The most marked difference between the two transitions

260 was the peak dorsiflexion angle which was largest during the top transition. Maximising

261 ankle dorsiflexion may strategically increase the base of support (BOS) during the top

262 transition, and thus dynamic stability, as it allows a larger area of the foot surface to remain

263 in contact with the ground for longer (Lark et al., 2003). This strategy was observed in the

264 current study during the top transition and may indicate an intention to maximise dynamic

265 stability when eccentric demands at the knee are high. Consequently, improving locomotor

266 stability when descending from the top of the stairs may be achieved by enhancing ankle

267 ROM particularly within the dorsiflexion range. Moreover, concurrent use of the handrails

268 would further enhance dynamic stability during this demanding task, thus helping to alleviate

269 fall risk concerns in high risk groups.

270

271 The limb executing the top transition displayed many GRF parameters of reduced magnitude

272 (Table 2) compared to the bottom transition. These alterations may be attributed to the

273 increased velocity observed during the bottom transition as demonstrated previously

274 following continuous, rhythmic descent (Lee and Chou, 2007). In addition, it is conceivable

275 that locomotor confidence may increase as a person descends, as the severity of potential

276 fall-related injuries may reduce closer to the bottom of the stairs. This effect may be even

277 more pronounced on a conventional staircase comprising a greater number of steps during

278 which online motor programmes may be fine-tuned and automated (Schmidt, 1975)

279 according to standardised staircase dimensions. Further work is required to determine

280 whether these discrete transitional forms of locomotion may impact falls risk due to temporal-

281   spatial disparities, varying staircase designs, and the presence of a fear of falling which

282   should be monitored in future studies.

283

284   A limitation of the current study was the use of only a 3-step staircase and the lack of

285   reciprocal, continuous descent gait cycles separating one transition from another. It may be

286   expected that a longer staircase comprising a greater number of consecutive steps (a

287   minimum of four steps is required to permit analysis of a single continuous cycle and top and

288   bottom transitions) would likely result in greater momentum generated at the bottom of the

289   stairs. However differences between transitions were still detected with the present 3-step

290   staircase and may be further amplified when ambulating at faster velocities and thus with

291   greater momentum. Future work may incorporate a longer top landing and explore the

292   chosen foot placement strategies adopted in the approach to stair descent, in conjunction

293   with both 1-step and 2-step transitional biomechanics, to provide greater detail about this

294   potentially hazardous transitional phase. Integrating COM and BOS calculations would help

295   to determine whether older adults strategically choose foot placement to optimise global

296   stability during transitional phases.

297

298   Participants self-selected their lead limb for each trial to represent their habitual descent

299   biomechanics/ patterns most accurately. Lower limb mechanics were considered

300   symmetrical during level gait for the same cohort and as such it was not expected that limb

301   preference due to asymmetry would have influenced the data presented (Alcock et al.,

302   2013). However, it would be interesting to understand whether participants with large

303   between-limb strength differences and asymmetry (i.e., due to disease, disuse or trauma)

304   elect to use the preferred limb for a particular transition given the varying demands exposed

305   in this study. Enhancing our understanding of transitional mechanics should be extended to

306   comparisons with young individuals, fallers and those with compromised balance to further

307 understand transitional demands. Moreover, it is important to consider the adjacent steps to

308 each of these transitions (i.e. 1-step transition or continuous stair gait of the contralateral

309 limb) given the influence on the bilateral coordination of temporal-spatial, kinematic and

310 kinetic indices. Finally it is noteworthy to highlight the variety of methods used to define a

311 continuous *vs.* transitional gait cycle and the gait event (foot contact/ toe-off) that is used to

312 define the beginning of the gait cycle (stance/ swing). It is critical that clear definitions and

313 consistent terminology are established for stair phase gait mechanics to facilitate appropriate

314 comparisons. We propose that a continuous gait cycle is defined as one that is initiated *and*

315 terminated on an independent step, not including that of the floor level and thus all other gait

316 cycles would be classified as transitional.

317

318 This study is the first to identify the functional biomechanical demands of transitions between

319 level and stair descent gait in older women. Some preliminary recommendations for stair

320 decent rehabilitation may be made for maintaining strength and joint ROM and evaluating

321 these parameters in exercise-based interventions with other older adult populations (fallers,

322 individuals with balance impairments etc.) should be the focus of future work. Exercise

323 recommendations may include incorporating the considerable eccentric control required from

324 the knee extensors (power absorption, late stance), concentric and eccentric control from the

325 plantarflexors (power absorption and generation mid-stance) and greater magnitudes of

326 ankle dorsiflexion and ROM required during the top transition. In contrast, increased

327 concentric knee power generation (mid-stance) and ankle plantarflexor moments (early and

328 late stance) were observed during the bottom transition and improving ankle plantarflexion

329 strength during single and double limb stance activities would ease the transition into level

330 gait following continuous descent. Finally, reduced hip and ankle joint mobility, particularly

331 for joint extension, may restrict the propulsion away from the stairs and consequently inhibit

332 initiation of level gait and limit step length beyond the stairs.
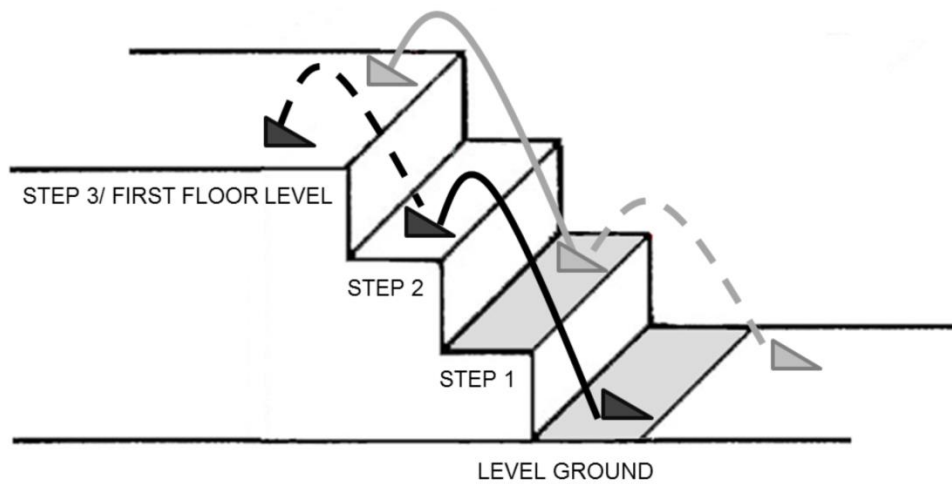
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340   **FIGURE 1** Schematic demonstrating the lead (black line) and trail (grey line) limb gait cycles

341   during stair descent

342   The dashed lines represent the 1-step transitional gait cycles of the lead and trail limbs, while the solid lines represent

343   the 2-step transitional gait cycles that were selected for further analysis. The grey shaded steps denote the positioning

344   of force plates for kinetic data acquisition of the lead (ground) and trail (step 1) limbs. Both gait cycles studied were

345   initiated and terminated by toe-off and data are presented firstly by swing, followed by stance.

346

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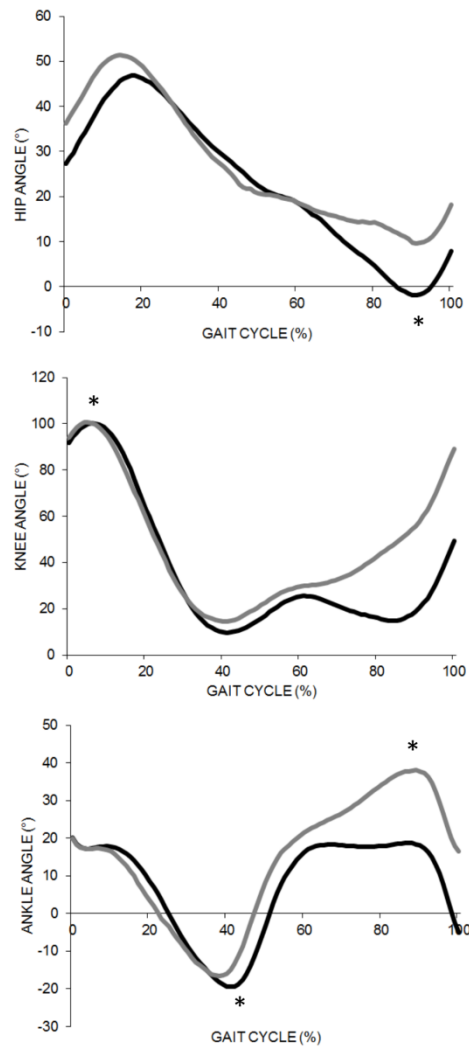
FIGURE 2 Ensemble average and time-normalised sagittal plane joint angle profiles (degrees) of the limb completing the level-to-descent gait transition (grey line, top floor level to step 1) and the limb completing the descent-to-level gait transition (black line, step 2 to level ground)

* indicates significant between-limb differences (p≤.05) post corrective procedures. Negative [+] values indicate extension and plantarflexion for the hip and ankle angles, respectively.
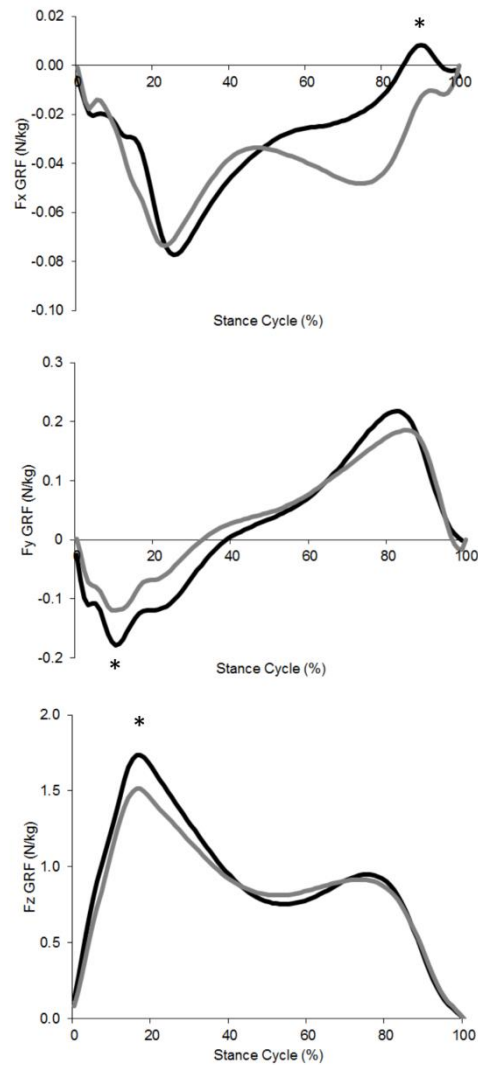
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FIGURE 3 Ensemble average and time-normalised orthogonal ground reaction forces (N/kg) of the limb completing the gait-to-descent transition (grey line, top floor level to step 1) and the limb completing the descent-to-gait transition (black line, step 2 to level ground)

* indicates significant between-limb differences (p≤.05) post corrective procedures. Negative [-] ground reaction force values indicate lateral (Fx2) and posterior (Fy2) force components.
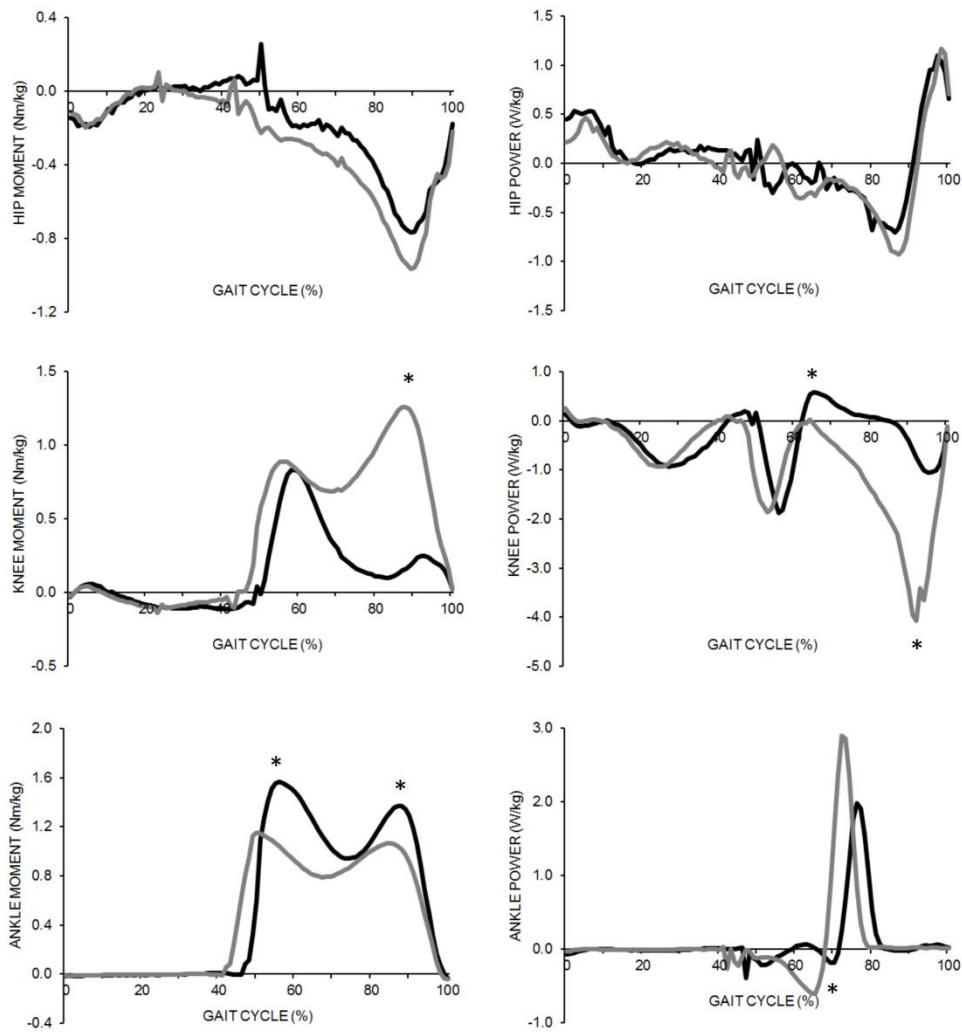
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FIGURE 4 Ensemble average and time-normalised sagittal hip, knee and ankle joint moments (Nm/kg) and joint power profiles (W/kg) of the limb completing the gait-to-descent transition (grey line, top floor level to step1) and the limb completing the descent-to-gait transition (black line, step 2 to level ground)

* indicates significant between-limb differences (p≤.05) post corrective procedures. At the hip and knee, a positive [+] value indicates an extensor moment; at the ankle, a positive [+] value indicates a plantarflexor moment. Positive [+] powers denote concentric power generation and negative [-] powers denote eccentric power absorption at the respective joints.

372 **TABLE 1** – Mean [SD] temporal-spatial and peak joint kinematics and ROM (degrees) parameters of the limb completing the top transition (top

373 floor level to step 1) and the limb completing the bottom transition (step 2 to level ground)

VARIABLE	TOP TRANSITION	BOTTOM TRANSITION	95% CONFIDENCE INTERVAL (Lower : Upper)	t	SIG.	CORRECTED SIG.	COHEN'S d
TEMPORAL-SPATIAL							
GAIT SPEED (m/s)	0.64 [0.1]	0.84 [0.2]	0.17 : 0.21	16.7	.001	.0018	9.4
CYCLE TIME (s)	1.36 [0.3]	1.22 [0.2]	-5.91 : -2.91	-6.0	.001	.0018	4.6
STANCE (%)	57.7 [3.6]	53.3 [3.6]	-0.18 : -0.11	-8.4	.001	.0018	10.2
JOINT KINEMATICS (degrees)							
HIP FLEXION (Early swing)	53.3 [7.8]	46.8 [10.0]	-8.73 : -0.52	-2.3	.029	.1128	
HIP EXTENSION (Late stance)	9.2 [11.4]	-2.3 [9.3]	-14.15 : -8.78	-8.7	.001	.0064	9.3
HIP ROM	44.4 [8.2]	50.1 [6.9]	1.02 : 8.48	2.6	.014	.0713	
KNEE FLEXION (Early swing)	103.1 [7.2]	100.5 [9.4]	-5.00 : -0.25	-2.2	.031	.1128	
KNEE ROM	91.0 [5.4]	92.3 [7.3]	-1.49 : 3.95	0.9	.362	1.000	
ANKLE DORSIFLEXION (Early swing)	18.8 [8.3]	20.7 [7.6]	-0.88 : 4.79	1.4	.170	.541	
ANKLE PLANTARFLEXION (Late swing/ Early stance)	-18.3 [5.8]	-21.0 [6.6]	-3.99 : -1.38	-4.2	.001	.0064	3.7
ANKLE DORSIFLEXION (Late stance)	39.4 [7.8]	22.6 [4.9]	-19.38 : -14.30	-13.5	.001	.0064	22.0
ANKLE ROM	57.7 [6.1]	45.1 [5.7]	-14.92 : 10.21	-10.8	.001	.0064	17.7

374

375 ROM denotes range of motion. Shaded areas indicate significant between-limb differences. At the hip and ankle joints, a negative value [-] indicates hyper[extension] and plantarflexion,
376 respectively.

377

378

379 **TABLE 2 -** Mean [SD] ground reaction forces (GRFs) and peak internal joint moments (Nm/kg) of the limb completing the top transition (top

380 floor level to step 1) and the limb completing the bottom transition (step 2 to level ground)

381

VARIABLE	TOP TRANSITION	BOTTOM TRANSITION	95% CONFIDENCE INTERVAL (Lower : Upper)	t	SIG.	CORRECTED SIG.	COHEN'S d
GROUND REACTION FORCES (N/Kg)							
MEDIAL FX1 GRF	0.01 [0.02]	0.02 [0.02]	0.00 : 0.02	2.474	.020	.1562	
LATERAL FX2 GRF	-0.08 [0.02]	-0.08 [0.03]	-0.01 :0.01	-0.294	.771	1.0000	
POSTERIOR FY1 GRF	-0.13 [0.03]	-0.18 [0.03]	-0.06 : -0.03	-5.836	.001	.0095	10.2
ANTERIOR FY2 GRF	0.21 [0.05]	0.21 [0.04]	-0.02 : -0.02	0.147	.884	1.0000	
VERTICAL FZ1 GRF	1.53 [0.19]	1.76 [0.22]	0.16 : 0.29	7.412	.001	.0095	9.1
VERTICAL FZ2 GRF	0.80 [0.09]	0.77 [0.10]	-0.08 : 0.00	-1.843	.076	.5021	
VERTICAL FZ3 GRF	0.94 [0.10]	0.97 [0.09]	-0.01 : 0.08	1.635	.114	.6994	
LOAD RATE [*N/kg/s]*	12.6 [3.9]	16.4 [4.5]	2.42 : 5.08	5.813	.001	.0095	7.5
DECAY RATE [*N/kg/s]*	4.6 [1.2]	6.2 [1.3]	0.98 : 2.11	5.600	.001	.0095	9.1
JOINT MOMENTS (Nm/Kg)							
HIP FLEXOR MOMENT (Late stance)	-1.05 [0.5]	-0.88 [0.3]	-0.38 : 0.16	1.420	.168	.9018	
KNEE EXTENSOR MOMENT (Early stance)	0.93 [0.5]	0.85 [0.4]	-0.22 : 0.06	-1.152	.259	1.0000	
KNEE EXTENSOR MOMENT (Late stance)	1.23 [0.5]	0.31 [0.1]	-1.11 : -0.73	-9.903	.001	.0095	20.7
ANKLE PLANTARFLEXOR MOMENT (Early stance)	1.21 [0.3]	1.50 [0.4]	-0.38 : -0.19	6.330	.001	.0095	6.7
ANKLE PLANTARFLEXOR MOMENT (Late stance)	1.13 [0.1]	1.36 [0.2]	-0.32 : 0.13	4.860	.001	.0095	9.1

382

383 Shaded areas indicate significant between-limb differences, Negative [-] ground reaction force values indicate lateral (Fx2) and posterior (Fy2) force

384 components. At the hip and knee, positive [+] values indicate extensor moments and at the ankle joint, positive [+] values indicate a plantarflexor moment

385

TABLE 3 - Mean [SD] peak joint powers (W/kg) of the limb completing the top transition (top floor level to step 1) and the limb completing the

bottom transition (step 2 to level ground)

388

VARIABLE	TOP TRANSITION	BOTTOM TRANSITION	95% CONFIDENCE INTERVAL (Lower : Upper)	t	SIG.	CORRECTED SIG.	COHEN'S d
JOINT POWERS [W/kg]							
HIP POWER GEN (Early swing)	0.59 [0.48]	0.67 [0.33]	0.03 : 1.14	0.865	.396	1.0000	
HIP POWER GEN (Late stance)	1.27 [0.61]	1.35 [0.88]	-0.29 : 0.44	0.428	.672	1.0000	
KNEE POWER ABS (Mid-swing)	-0.75 [0.29]	-0.75 [0.30]	-0.11 : 0.12	0.122	.904	1.0000	
KNEE POWER ABS (Early stance)	-2.10 [1.22]	-2.10 [1.48]	-0.59 : 0.60	0.029	.977	1.0000	
KNEE POWER GEN (Mid-stance)	0.34 [0.54]	0.75 [0.40]	0.13 : 0.70	2.955	.006	.0095	5.8
KNEE POWER ABS (Late stance)	-3.91 [1.41]	-1.24 [0.32]	2.12 : 3.23	9.911	.001	.0095	18.3
ANKLE POWER ABS (Early stance)	-1.02 [1.46]	-1.31 [2.33]	-2.81 : -1.29	-1.570	.126	.7215	
ANKLE POWER ABS (Mid-stance)	-0.78 [0.57]	-0.36 [0.48]	0.52 : 1.43	4.642	.001	.0095	6.1
ANKLE POWER GEN (Mid-stance)	3.52 [1.42]	3.09 [1.24]	0.42 : 0.89	-2.090	.046	.3292	

389

390   Shaded areas indicate significant between-limb differences. GEN denotes generation and ABS denotes absorption.

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462

464   **Introduction**

465   Musculoskeletal modelling of human movement requires the capture of accurate and valid kinetic

466   data. Instrumented staircases such as the one in the present study are often unique in design, but

467   permit kinetic data collection via force platforms embedded into metal or wooden staircases (Nadeau

468   et al., 2003; Mian et al., 2007; Reeves et al., 2008), independent step structures (McFadyen and

469   Winter, 1988), or concrete supports (Hamel et al., 2005), and those consisting of a structure placed

470   on top of existing floor-mounted platforms (Lark et al., 2003). However, staircase design may

471   introduce error when comparing stairway-derived forces with ground-mounted force platforms due to

472   the material properties of the mounting structure.

473   Studies utilising instrumented staircases composed of wooden steps supported by metal frames

474   (Chapdelaine et al., 2005; Della Croce and Bonato, 2007) have reported reductions in the natural

475   frequency from staircases placed upon existing ground-mounted platforms when compared to stair-

476   mounted platforms (Della Croce and Bonato, 2007). Conversely Chapdelaine *et al.* (2005) were

477   unable to detect a natural frequency in the vertical direction due to a small oscillation impulse

478   amplitude. Whilst alterations in the natural frequency have been shown to not impede upon the low

479   frequencies typically associated with foot contact during gait and stair locomotion (Antonsson and

480   Mann, 1985; Chapdelaine et al., 2005), it is not clear if the experimental set-up used in the present

481   study provides robust kinetic data. Many studies employing the use of instrumented stairways or

482   walkways have neglected to quantify the spectral power lost due to force plate mounting or define the

483   signal filter introduced. Custom built experimental staircases are often constructed from wood (Lark et

484   al., 2003; Nadeau et al., 2003; Vanicek et al., 2010; Alcock et al., 2014) conforming to building

485   regulation dimensions with three steps (Andriacchi et al., 1980; Lu and Lu, 2006; Mian et al., 2007;

486   Beaulieu et al., 2008; Vanicek et al., 2010; Alcock et al., 2014). Therefore, to validate such designs

487   this supplementary material presents an evaluation of the power lost and signal filter introduced by

488   the 3-step custom-built staircase utilised in the current study and others published previously (Vanicek

489   et al., 2010; Alcock et al., 2014).

Methods

Staircase design and kinetic data acquisition

490

491

492   Dimensions and structure of the custom-built wooden staircase and associated force plate mounting

493   are presented in Figure S.1. The 3-step staircase was comprised of two independent sections

494   allowing a platform to be embedded in the first step with a 10mm gap around the platform edge.

495   Vertical ground reaction forces (GRF) were collected from a piezoelectric force platform (model

496   9286AA, Kistler, Winterthur, Switzerland) sampling at 500Hz through a 64-bit analogue-to-digital

497   board. A 3kg medicine ball was released by hand from a 1-metre height (measured by a stadiometer)

498   onto the platform and allowed to bounce once, two such trials were completed. This process was

499   performed with the force platform embedded into: (1) a floor-mounted level concrete pit (FP_{GROUND});

500   and (2) the first step of a wooden 3-step stairway (FP_{STEP}).

Spectral analysis

501

502   Spectral analysis (SA) of the vertical GRF from each trial was performed by FFT between 0-250 Hz in

503   2048 bins at a resolution of 0.122 Hz using Matlab (R2008a, Mathworks, Natick, MA). Mean power

504   spectrums were produced. 50SA (median frequency) and 95SA, defined as the spectral frequency at

505   which 50% and 95% of the power fell below; and total energy (TE) of each spectrum were calculated.

506   Additionally, the transfer function for FP_{STEP} with respect to FP_{GROUND} was calculated between 0Hz

507   and 18Hz. SA performed on a previously recorded vertical GRF trace recorded during gait analysis

508   (Male, age=27yrs, height=1.84m, mass=78.1kg, gait speed=1.12m/s) defined this frequency range as

509   containing 99.95% of spectral power during foot strike. This transformation also allowed the volume of

510   spectral power lost (%) during gait due to the transfer function to be calculated.

Statistical analysis

511

512   Independent samples t-tests were performed on 50SA, 95SA, and TE for each condition using SPSS

513   (v18.0, SPSS Inc., Chicago, IL). Homogeneity of variance was assessed using Levene's statistic and

514   equal variances were assumed. Statistical significance was set at $p \leq 0.05$.

25

Results

516 Impulse from the ball-drop contained energy across a wide range of frequencies (see Figure S.2).

517 50SA was reported as 16.58 Hz for FP_{GROUND}, indicating most of this energy was at low frequencies.

518 Power spectrum for FP_{STEP} deviated from FP_{GROUND} at ~10 Hz. Significant differences were observed

519 for 50SA and TE. The mean 50SA (M = 16.58, SD = 0.92, CI = 15.44:17.71) and mean 95SA (M =

520 52.47, SD = 3.02, CI = 48.72:56.21) for FP_{STEP} were significantly different to the mean 50SA (M =

521 20.39, SD = 0.46, 95%CI = 19.81:20.96) and mean 95SA (M = 52.78, SD = 1.16,

522 95%CI = 51.34:54.23) for FP_{GROUND}. Similarly, when considering TE, FP_{STEP} (M = 13344.78,

523 SD = 872.42, 95%CI = 12261.52:14428.04), was significantly different to FP_{GROUND} (M=17107.46,

524 SD=578.83, 95%CI=15967.93:17939.99).

525 The calculated transfer function indicated that limited signal filtering occurred and only at the highest

526 frequencies for FP_{STEP} (Figure S.3). When considered with respect to the power spectrum during foot

527 contact, the total loss of power was found to be 2.2% for FP_{STEP} (Figure S.3).

528

Discussion

530 This analysis has shown that kinetic data sampled by force platforms embedded in instrumented

531 wooden stairways were altered at high frequencies. However, in the frequency range of interest to

532 gait (0-18 Hz) and in this study, the differences were considered minimal. Significant differences

533 observed in 95SA and TE reflected energy across the complete frequency spectrum. This suggests

534 modifications in high frequency platform response may have occurred when mounted in the wooden

535 step. 50SA was changed significantly; this may have indicated some alteration of the low frequency

536 force response relevant to gait in this condition. However, the analysis of transfer functions suggests

537 only a small portion of TE was lost when the platform was staircase-mounted. This loss was

538 considered negligible in comparison to other errors introduced during in motion capture (Chiari et al.,

539 2005).

540 Whilst other studies have investigated the acquisition of kinetic data from instrumented stairways,

541 those studies applied impulses of low magnitude (Della Croce and Bonato, 2007) (0.1kg from 1-metre

542   height), and only considered natural frequencies. The experiment presented in this supplementary

543   material considered a much larger impulse and is the first to quantify the energy lost due to staircase

544   design. It was noted the drop-ball procedure provided energy from a wide range of frequencies and

545   TE from ball-drop was ~2 times that produced by a foot strike during gait. Activities such as stair

546   climbing and faster walking produce greater energy, thus, the impulse selected was of suitable size to

547   assess force platform performance in a gait laboratory. As the largest component of the GRF vector,

548   the vertical GRF was analysed due to its considerable influence on kinetic computations.

549   Furthermore, the vertical GRF is thought to be the most consistent during gait, as the medio-lateral

550   and anterior-posterior forces can vary substantially, and was therefore appropriate to represent

551   analysis of force platform performance.

552   In conclusion, this analysis found that negligible power was lost when mounting a force plate into a 3-

553   step wooden staircase structure and may alleviate concerns that the kinetic differences highlighted

554   between the transitional steps at the top and bottom of the staircase may have been filtered

555   substantially as a result of staircase mounting. Moreover, this methodology may be repeated in gait

556   laboratories using custom-built staircases made of alternative materials and comprising of more steps

557   and force platforms.

558

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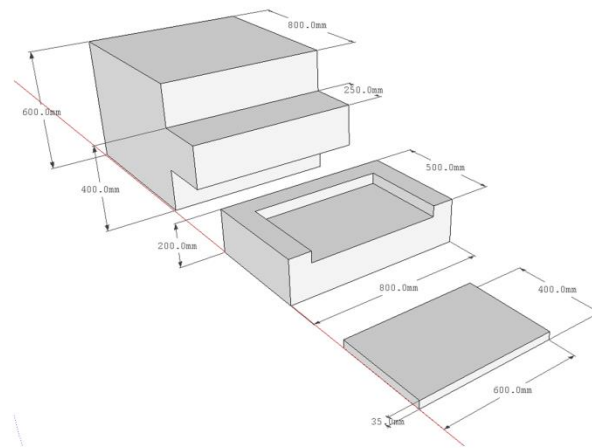
598

599   Figure S.1 – Geometric drawing of the 3-step custom built staircase components depicting the main

600   structure comprising steps 2 and 3 (far left), integrated first step housing the force plate (centre;

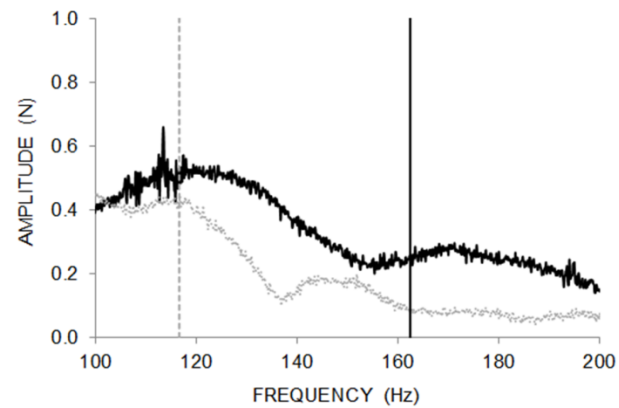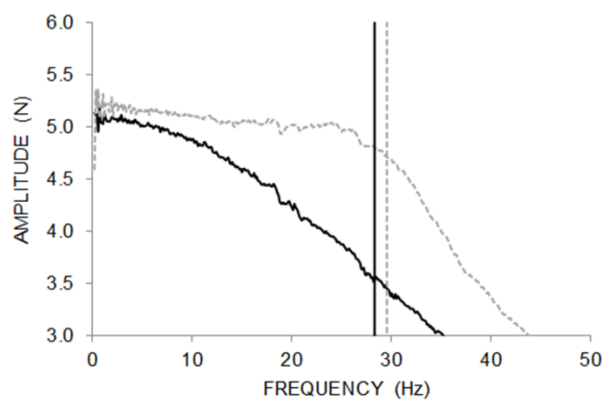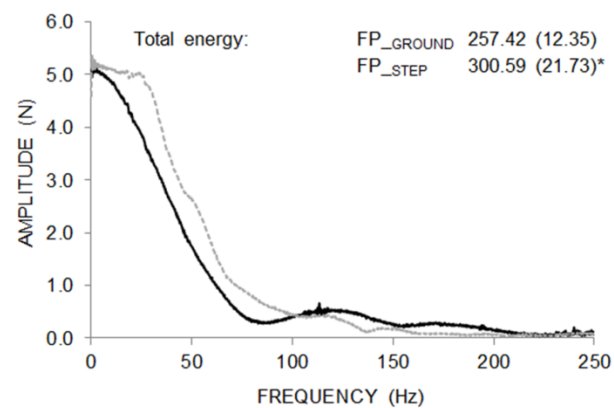601 FP_{STEP}) and the force plate (right; FP_{GROUND})
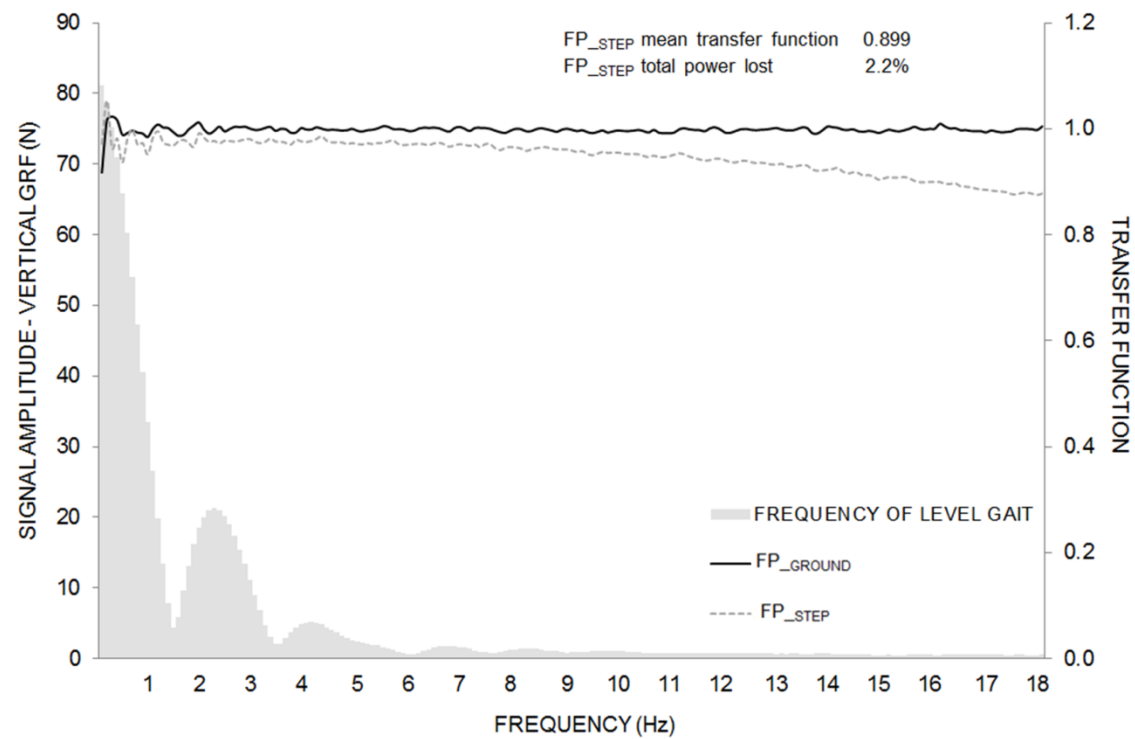
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603

604   Figure S.2 – Power spectrum of the ball drops displayed on a normal scale (top), and magnified scale depicting the mean 50SA (bottom left) and mean 95SA

605   (bottom right) for FP_{GROUND} (black solid line) and FP_{STEP} (grey dashed line)

606

607 Figure S.3 – Power spectrum of an example foot contact during level gait (shaded grey) up to 95% power, and transfer functions for FP_{GROUND} (black solid

608 line) and FP_{STEP} (blue dashed line)

609