1   **Title –** The mechanics and energetics of hopping with elastic ankle exoskeletons

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7   **Running header -** mechanics and energetics of hopping with elastic ankle exoskeletons

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13   **Abstract**

14   The spring-like mechanics of the human leg during bouncing gaits has inspired the design of passive

15   assistive devices that use springs to aid locomotion. The purpose of this study was to test if a passive

16   spring-loaded ankle exoskeleton could reduce the mechanical and energetic demands of bilateral

17   hopping on the musculoskeletal system. Joint level kinematics and kinetics were collected with

18   electromyographic and metabolic energy consumption data for 7 participants hopping at 4

19   frequencies (2.2, 2.5, 2.8 & 3.2 Hz). Hopping was performed without an exoskeleton; with an spring-

20   less exoskeleton; and with a spring-loaded exoskeleton. Spring-loaded ankle exoskeletons reduced

21   plantar-flexor muscle activity and the biological contribution to ankle joint moment (15-25%) and

22   average positive power (20-40%). They also facilitated reductions in metabolic power (15-20%)

23   across frequencies from 2.2-2.8 Hz compared to hopping with a spring-less exoskeleton. Reductions

24   in metabolic power compared to hopping with no exoskeleton were restricted to hopping at 2.5 Hz

25   only (12%). These results highlighted the importance of reducing the rate of muscular force

26   production and work to achieve metabolic reductions. They also highlighted the importance of

27   assisting muscles acting at the knee joint. Exoskeleton designs may need to be tuned to optimise

28   exoskeleton mass, spring stiffness and spring slack length to achieve greater metabolic reductions.

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33   **Introduction**

34   When humans utilise bouncing gaits such as hopping and running, the supporting limb(s) during

35   ground contact can be considered to act like a compressive spring (3, 13, 29). This has led to such

36   motions commonly being modelled as a spring-mass system (12, 14, 19-20, 29). The spring-like

37   behaviour of the lower limb makes it possible for the muscle-tendon units acting around joints to

38   store and return energy in elastic structures (1, 6, 25, 28). Predominantly, energy is stored and

39   returned in tendons acting in series with the contractile elements of muscles (16, 31). The result of

40   this is a reduction in the mechanical work requirements of contractile elements and substantial

41   metabolic energy savings (2, 7). Some metabolic cost is still incurred owing to the contractile

42   elements having to produce force to stretch tendons (2, 31).

43   The way in which the muscle-tendon units of the leg utilise elastic energy cycling to efficiently power

44   bouncing gaits has inspired the design of orthoses and exoskeletons for the leg that seek to passively

45   store and return energy during movement (8, 18, 21). Such devices might be useful for rehabilitating,

46   restoring and/or augmenting human locomotor performance by reducing musculo-skeletal loading

47   and reducing metabolic energy consumption. Loading of the leg is expected to be reduced by

48   wearing assistive exoskeletons because the device bears some of the load normally taken by the

49   limb (18, 21). By using this loading to store and return energy in a spring, such devices may reduce

50   the mechanical demands on biological tissues which may, in turn, reduce metabolic energy

51   consumption (21).

52   Previous studies of the effects of passive lower-limb exoskeletons on human movement have

53   typically used hopping motions because of the simple biomechanical goals of the task (8) and its

54   similar spring-mass mechanics to running but, less complicated joint kinematics (18). Grabowski and

55   Herr (21) showed that a full leg exoskeleton crossing hip, knee and ankle joints significantly   reduced

56   the metabolic power requirement of bilateral hopping (up to 30%) over a range of frequencies from

57   2.0-2.6 Hz. Their participants reduced their biological contribution to leg stiffness to keep the

58   combined exoskeleton and biological stiffness similar to leg stiffness when hopping without an

59   exoskeleton. This helped maintain consistent centre of mass mechanics which appears to be a goal

60   of movement control for bouncing gaits (19-20).

61   A large proportion (≈ 45%) of the overall positive mechanical power output generated at leg joints

62   during human running comes from muscle-tendon units acting at the ankle joint (17). Also, leg

63   stiffness during hopping is predominantly determined by ankle joint stiffness (15). This highlights the

64   importance of the ankle joint in maintaining the mechanics of bouncing gaits. Therefore, providing

65   exoskeletal assistance at only the ankle joint could provide a significant proportion of the benefits

66   achieved with a full leg device, while simplifying its design.

67   Both Ferris et al. (18) and Chang et al. (8) investigated the effects of wearing an ankle-foot orthosis

68   that was spring-loaded to assist plantar-flexion on joint and limb stiffness during unilateral hopping.

69   Both studies showed that hoppers were able to reduce biological ankle stiffness to maintain overall

70   limb stiffness when wearing the devices. Ferris et al. (18) also showed a concurrent reduction in the

71   amplitude of electromyographic (EMG) signals from the plantar-flexor muscles. This suggested that

72   muscle activation was reduced to achieve lesser biological ankle joint stiffness when assistance was

73   being provided by the device. Whilst the studies of Ferris et al. (18) and Chang et al. (8) both suggest

74   that spring-loaded ankle exoskeletons might unload biological tissues, it was not the aim of these

75   studies to link the mechanics of hopping in ankle exoskeletons to the associated metabolic cost.

76   Therefore, they did not determine the mechanical demands and metabolic power requirements of

77   the task concurrently. Doing so would help to reveal if assisting only at the ankle joint could reduce

78   the mechanical demands on leg muscles sufficiently to reduce the metabolic cost of a bouncing gait.

79   The aim of this study was to investigate the effects of spring-loaded ankle exoskeletons on the

80   metabolic cost and mechanics of bilateral human hopping. It was hypothesised that when hopping

81   with the spring-loaded ankle exoskeletons, participants would reduce plantar-flexor muscle activity

82   and that this would reduce the mechanical power output by biological tissues (muscle-tendon units)

83   at the ankle. Because of reduced mechanical demand placed on muscles, it was also hypothesised

84   that hopping with spring-loaded exoskeletons would reduce metabolic power. Furthermore, it was

85   expected that the ankle joint would dominate overall mechanical input to the task and thus,

86   metabolic reductions would be proportionally less than those previously achieved with a full leg

87   exoskeleton (21) (i.e., if the ankle joint provides 60% of the mechanical input, the observed

88   reductions will be 60% of that with full leg assistance).

89   **Materials and Methods**

90   Seven male participants (mean ± sd, age = 28 ± 7, height = 1.8 ± 0.06 m, mass = 80 ± 10 kg) gave

91   written informed consent to participate in this study. All participants were in good health and had no

92   recent history of lower limb musculo-skeletal injury. All procedures were approved by an

93   institutional review board and complied with the guidelines for research involving human

94   participants as set out in the Declaration of Helsinki.

95   Because metabolic power required for hopping and the effect of exoskeletons is frequency

96   dependant (21), participants hopped in time with a metronome at four different frequencies (2.2

97   Hz, 2.5 Hz, 2.8 Hz, 3.2 Hz). There were three different conditions: (1) Without an exoskeleton (NE);

98   (2) With an exoskeleton but with no spring (NS); (3) With a spring-loaded exoskeleton to assist

99   plantar-flexion (S). Each trial lasted 4 minutes. Because of the physically challenging nature of the

100   task, each condition was performed on a separate day to minimise fatigue effects. The order of

101   conditions and frequencies was randomized. The intent of this experiment was to test the effects of

102   the exoskeletons on naïve users, so participants were given sufficient familiarisation with hopping in

103   the exoskeletons to feel comfortable but not for long enough to introduce any training effects.

104   An example of the exoskeletons used is shown in Figure 1. The exoskeleton consisted of a carbon

105   fibre cuff around the upper shank which was connected to a carbon fibre foot section via two

106   aluminium bars which had a freely rotating joint aligned with the participants' malleoli. The foot

107   section was embedded in a training shoe, through the sole and around the heel. An extension spring

108   could be attached to a bracket on the posterior aspect of the cuff and a bolt on the heel of the foot

109   segment via a number of metal links. The number of links was adjusted for each participant such

110   that the resting length of the spring coincided with an ankle angle of 127° which has been

111   determined as the typical angle at ground contact in hopping (18). This same approach was used by

112   Ferris et al. (18) for a similar exoskeleton. A compression load cell (Omegadyne Inc., OH,USA) was

113   placed on the inferior side of the bolt at the heel of the foot segment and attached to the links in

114   series with the spring. This was used to measure the forces in the spring. The stiffness of the spring

115   in tension was 5 kNm^{-1} and its moment arm about the joints was 0.135 m. This gave a rotational

116   stiffness of 1.59 Nm/° (91 Nm/rad) which is approximately 40% of ankle stiffness during unassisted

117   hopping at preferred frequency (18).

118   An eight camera motion analysis system (Vicon, Oxford, UK) was used to capture the three-

119   dimensional positions of 22 reflective markers attached to the pelvis and right leg. Raw marker

120   positions were filtered using a second order low-pass butterworth filter with a cut-off of 10 Hz. A

121   static standing trial was captured and the positions of markers on segment end points were used to

122   calibrate a four segment (pelvis, thigh, shank and foot) model for each subject using established

123   inertial parameters (10). Clusters of three or four markers on rigid plates were attached to the pelvis,

124   thigh and shank segments to track segment motion during hopping. For the foot, a cluster of three

125   markers was attached directly to the shoe. Joint angles for the hip knee and ankle were computed in

126   three dimensions as the orientation of the distal segment with reference to the proximal segment

127   and differentiated to calculate joint velocities.

128   Three-dimensional (3D) ground reaction forces applied to the left and right legs were computed

129   during hopping using a split belt instrumented treadmill (Bertec, OH, USA). Participants hopped such

130   that each foot was on a separate half of the treadmill and thus, the two 3D force vectors could be

131   attributed separately to the left and right legs. Raw analogue force platform signals were filtered

132   using a low-pass Butterworth digital filter with the cut-off set to 35 Hz. Inverse dynamic analyses

133   (Winter, 1990) were then used to compute net joint moments which were multiplied with joint

134   velocities to calculate joint powers at the hip, knee and ankle. Kinematics and kinetics were

135   calculated for the right leg only and it was assumed that the left leg behaved symmetrically. Inverse

136   dynamics procedures were performed with Visual 3D software (C-motion Inc., Germantown, MD,

137   USA).

138   The mechanical demands of hopping under different conditions was partly assessed by the average

139   positive power ($\bar{P}^+_{mech}$) of the task. More details of the computation of $\bar{P}^+_{mech}$ can be found

140   elsewhere (17). Briefly, it was calculated for each joint [ankle ($\bar{P}^+_{ank}$), knee ($\bar{P}^+_{knee}$) and hip ($\bar{P}^+_{hip}$)] as

141   the integral of all periods of instantaneous positive power for that joint during a hop, divided by the

142   time per hop. $\bar{P}^+_{ank}$, $\bar{P}^+_{knee}$ and $\bar{P}^+_{hip}$ were summed to compute total average positive power ($\bar{P}^+_{tot}$).

143   When the spring-loaded exoskeletons were being worn, only a portion of $\bar{P}^+_{ank}$ was provided by

144   biological tissues and the rest came from the exoskeleton. To separate out these contributions, the

145   power output of the exoskeleton was determined as follows. First, the force in the spring (measured

146   by the load cell, Figure 1) was multiplied by the spring's moment arm about the joint of the

147   exoskeleton, to give the moment provided by the spring. This moment was multiplied by ankle

148   angular velocity to compute the instantaneous power provided by the exoskeleton. The average

149   positive power of the exoskeleton ($\bar{P}^+_{exo}$) was calculate as it was for joints. The contribution of

150   biological tissues at the ankle ($\bar{P}^+_{bio}$) was calculated as $\bar{P}^+_{ank}$ minus $\bar{P}^+_{exo}$. All average powers were

151   normalised to body mass (W·kg^{-1}).

152   The rate at which muscles produce force may also be an important mechanical determinant of

153   metabolic cost (27). Therefore, a rate of force metric was also computed. This metric was the

154   average rate of joint moment production ($\bar{\dot{M}}^+$). Somewhat analogous to average positive power, this

155   value was calculated for each joint [ankle ($\bar{\dot{M}}^+_{ank}$), knee ($\bar{\dot{M}}^+_{knee}$) and hip ($\bar{\dot{M}}^+_{hip}$)] by integrating the

156   first derivative of each joint moment during periods of increasing moment production (i.e. when the

157   derivative was positive), summing each of the integral values during each hop and dividing the result

158   by the time taken for an entire hop cycle. As for average positive power, $\bar{M}_{ank}^+$ was reduced to

159   separate exoskeleton and biological contributions when a spring-loaded exoskeleton was being

160   used.

161   Surface electromyography (EMG) was used to record muscle activity from Medial Gastrocnemius

162   (MG), Lateral Gastrocnemius (LG), Soleus (SO) and Tibialis Anterior (TA). All four channels were

163   recorded using wired electrodes (Biometrics Ltd, UK) that were carefully placed over muscle bellies

164   after the skin surface was prepared by light abrasion and cleaned with an alcohol swab. D.C. offsets

165   were removed from raw signals which were then band-pass filtered (20-300 Hz). The data were then

166   smoothed by calculating the root-mean-squared (RMS) value of the signals over a rolling window of

167   20 ms. The RMS of each muscle's signal was also calculated over the period of ground contact and

168   the aerial phase of each hop as a metric of total activity over these two phases of the hop. For each

169   session of data collection, the participant performed a set of 20 hops at 2.2 Hz while wearing

170   exoskeletons without a spring. The maximum of the processed signals from each muscle for this

171   condition was used to normalise EMG signals from the experimental conditions on that day. This was

172   to avoid any effects of day-to-day variation in electrode placement on signal magnitudes.

173   Rates of oxygen consumption and carbon dioxide production during hopping trials were recorded

174   using a portable metabolic system (OXYCON MOBILE, Viasys Healthcare, CA, USA). Prior to hopping,

175   measurements were made during five minutes of quiet standing and values from the last two

176   minutes were averaged and used to calculate rates of metabolic energy consumption whilst

177   standing. For the hopping trials, data from the last two of the four minutes were averaged for the

178   calculation of metabolic rate. Visual inspection of rates of oxygen consumption with time (averaged

179   over 30 s intervals) confirmed that participants were at steady-state during this period and the

180   respiratory exchange ratio was never greater than one. Rates of oxygen consumption and carbon

181   dioxide production were converted to metabolic powers using standard equations detailed by

182   Brockway (5). Net metabolic powers during hopping were calculated by subtracting metabolic power

183   during standing from metabolic power during hopping and these values were normalized to

184   individual body mass ($W \cdot kg^{-1}$). Metabolic data were presented as the normalised net value, unless

185   otherwise stated.

186   All the kinematic and kinetic data for individual participants was reduced to the mean of at least 10

187   hops for each experimental condition. Unless otherwise stated, the values presented in this paper

188   are the mean ± standard error for the whole participant group. To test for statistical differences in

189   dependent variables between conditions a two-way ANOVA with repeated measures was employed

190   using SPSS software (IBM,USA). The two independent variables for the ANOVA were: 1. Spring

191   condition (3 levels – NE, NS, S) and 2. Hopping Frequency (4 levels – 2.2, 2.5, 2.8, 3.2 Hz). F-ratios for

192   main and interaction effects were considered significant for $P < 0.05$. If a significant main effect was

193   found, paired t-tests were used to make pair-wise comparisons between levels of independent

194   variables.

195   **Results**

196   The S condition was described by similar hip and knee joint kinematics to the other NS and NE

197   conditions but it did cause a shift in ankle angle to a generally more plantar-flexed position (Figure

198   2). This shift was systematic, allowing the ankle to move through a similar range of motion (15-30°,

199   dependant on frequency) in all conditions but with a more plantar-flexed touchdown angle for S.

200   Similarly, joint kinetics (moments and powers) were similar across the NE, NS and S conditions for

201   the hip and knee (Figures 3 and 4). Total ankle power had generally greater peaks in the S condition

202   (Figure 3) indicating that the sum of exoskeleton power and biological ankle power for S was greater

203   than the biological power in NE and NS. The ankle joint moments showed that the exoskeleton

204   provided between 30 and 50% of the total moment (Figure 3) and this significant contribution was

205   confirmed in the $\bar{P}^+_{mech}$ data which showed that $\bar{P}^+_{exo}$ accounted for 36-38% of $\bar{P}^+_{tot}$ while $\bar{P}^+_{bio}$

206   accounted for 23-38% (Figure 5).

207   Net metabolic power typically followed a U-shaped relationship with frequency for NE, NS and S

208   (Figure 6). This meant that net metabolic power requirements were least for the intermediate

209   frequencies (2.5 & 2.8 Hz). Notably, there was a relatively large increase in net metabolic power

210   from 2.8 Hz to 3.2 Hz in the S condition (Figure 6). This caused the mean value for net metabolic

211   power for S to be greater than for NE at 3.2 Hz when it had been consistently less than NE and NS at

212   all other frequencies. Net metabolic power was significantly less for S than NS at 2.2 (-16%, P=0.009),

213   2.5 (-19%, P = 0.006) and 2.8 Hz (-20%, P = 0.047) but only at 2.5 Hz (-13%, P = 0.016) compared to

214   NE (Figure 6).

215   The biological contribution to $\bar{P}_{tot}^+$ was significantly less for S than for NS (≈30-40%) and NE (≈20-

216   30%) at all frequencies of hopping (Figure 6b, ANOVA P < 0.001). As can be observed in Figure 6b,

217   this difference was greater at low frequencies than at high frequencies with the linear fits starting to

218   converge at high frequencies. Biological $\bar{P}_{mech}^+$ was greater at low frequencies than at high

219   frequencies for S, NS and NE. This was in contrast to the biological contribution to $\bar{M}_{ank}^+$ which

220   increased with hopping frequency. There was a significant difference in $\bar{M}_{ank}^+$ between the S than in

221   NS conditions at 2.2 (21%, P < 0.001), 2.5 (24%, P = 0.003) and 2.8 Hz (20%, P = 0.001) (Figure 6b).

222     $\bar{M}_{ank}^+$ was only significantly less for S than NS at 2.2 (18%, P = 0.007) and 2.5 Hz (19%, P = 0.05)

223   (Figure 6c).

224   Time histories of the smoothed and normalised EMG signals are shown in Figure 7. MG, LG and SO

225   all typically had lower activations toward the end of the aerial phase for S than for NE and NS. This

226   was apparent in the RMS values for these muscles during the aerial phase which were typically

227   significantly less for S than for NS and NE (Figure 8). During ground contact only SO had significantly

228   lower activation for S (Figure 8). Contrary to the plantar-flexor activations, TA exhibited significantly

229   increased activation during ground contact and the aerial phase for all frequencies in the S condition

230   compared to NE and NS (Figure 8).

231

232 **Discussion**

233   This study aimed to investigate the effects of spring-loaded ankle exoskeletons on the metabolic cost

234   and mechanics of bilateral human hopping. First, it was hypothesised that the use of bilateral

235   spring-loaded ankle exoskeletons during two-legged hopping would reduce ankle plantar-flexor

236   muscle activity. This prediction was supported for all the plantar-flexors tested during the aerial

237   phase but only soleus during ground contact. It was also hypothesised that the reduced activity

238   would reduce the contribution of plantar-flexors to mechanical power output. This was found to be

239   the case across all frequencies. The final prediction made was that using spring-loaded ankle

240   exoskeletons would lower the metabolic power requirement of hopping when compared to hopping

241   without an exoskeleton or without a spring. This prediction was supported for some conditions at

242   certain frequencies but not all. These findings raise interesting questions regarding the mechanisms

243   by which spring-loaded ankle exoskeletons may be able to lower metabolic cost and also highlights

244   some potential obstacles to this goal.

245   The results showed that the use of spring-loaded ankle exoskeletons resulted in significant

246   reductions in the $\bar{P}^{+}_{mech}$ production by biological tissues (muscle-tendon units) when hopping at all

247   frequencies compared to the NE and NS conditions (Figure 6). This was due to a reduction in the

248   biological plantar flexion moment at the ankle (Figure 3). This result was not surprising as it has been

249   previously shown that humans reduce biological ankle and limb stiffness to maintain overall system

250   stiffness when hopping with spring-loaded ankle or full leg exoskeletons (8, 18, 21). The reduction in

251   biological contribution to the plantar-flexion moment was primarily achieved by a reduction in

252   soleus muscle activity during the ground contact phase of hops (Figure 8). MG and LG RMS EMG

253   values were only less during the aerial phase and this was primarily due to a reduction in activity late

254   in this phase (a similar trend was apparent in soleus for the aerial phase). Therefore it seems that

255   pre-activation of plantar-flexors was also less when using spring-loaded exoskeletons.

256 The reduction in biological $\bar{P}^+_{mech}$ seen at all frequencies in the S condition was one indication that

257 the mechanical demands on biological tissues was less with assistance. However, significant

258 reductions in metabolic power only occurred at 2.5 Hz (-0.9 W·kg⁻¹) compared to the NE condition

259 and at 2.2 (-1.3 W·kg⁻¹), 2.5 (-1.5 W·kg⁻¹) and 2.8 Hz (-1.6 W·kg⁻¹) compared to the NS condition.

260 Furthermore, where there were reductions in metabolic power, they were not proportional with

261 reductions in biological $\bar{P}^+_{mech}$ (Figure 6). Therefore, whilst $\bar{P}^+_{mech}$ provides some indication of the

262 mechanical demands of the task, it alone cannot fully explain the trends in metabolic cost that were

263 observed. This is particularly notable at 3.2 Hz where the metabolic power requirement was greater

264 for the S condition than for the NE condition despite a reduction in $\bar{P}^+_{mech}$.

265 Another factor that may help to explain the observed trends in metabolic power is the metabolic

266 cost associated with producing muscular force. An increasing number of studies of human (and

267 animal) movement have indicated that a significant portion (up to 50%) of the metabolic cost of

268 locomotion is related to the metabolic energy used by muscles to produce force (22-23, 26-27, 32).

269 This cost has been proposed to increase proportionally with the rate at which force must be

270 produced (27). The cost of producing force may be particularly important in muscle groups such as

271 the plantar-flexors that rely on the series compliance of tendons to cycle energy while the muscle

272 itself contracts relatively isometrically during stance (25, 28). In such contractions, the contractile

273 elements of muscle do only small amounts of work but must produce forces sufficient to 'anchor'

274 the tendon at the myotendinous junction. In a study of human bouncing tasks, Dean and Kuo (9)

275 actually showed that the rate of force cycling in ankle plantar-flexors increased with bouncing

276 frequency and that this could explain why metabolic cost began to increase at frequencies above 3

277 Hz, despite muscular work decreasing. If the rate of force production becomes a significant factor in

278 determining metabolic cost at high frequencies (>3 Hz), it might explain why our spring-loaded

279 exoskeletons were unable to reduce metabolic power at 3.2 Hz despite reducing biological

280 mechanical power (assuming similarities between bouncing and hopping). In the present study $\bar{M}^+_{ank}$

281 was considered as a metric for rate of force production by the muscles acting at the ankle joint.

282 Figure 6 shows that $\bar{M}_{ank}^+$ increased with frequency lending some support to the notion that it

283 becomes more influential at higher frequencies. In fact, at the two highest frequencies in this study,

284 $\bar{M}_{ank}^+$ could explain the trends in metabolic power. At 3.2 Hz no significant differences in metabolic

285 power were observed and there were also no significant reductions in $\bar{M}_{ank}^+$ (Figure 6). For hopping

286 at 2.8 Hz there was a significant reduction in metabolic power for S compared to NS that coincided

287 with a reduction in $\bar{M}_{ank}^+$ but neither variable was reduced compared to NE (Figure 6). Therefore, at

288 higher frequencies metabolic power seems more dependent on $\bar{M}_{ank}^+$ than $\bar{P}_{mech}^+$ and at 3.2 Hz the

289 exoskeletons did not reduce $\bar{M}_{ank}^+$ sufficiently.

290 However, at lower frequencies this may not be the case. For both 2.2 Hz and 2.5 Hz there were

291 significant reductions in both $\bar{P}_{mech}^+$ and $\bar{M}_{ank}^+$ for S compared to NE and NS but metabolic cost was

292 not reduced at 2.2 Hz compared to NE (Figure 6). A possible explanation for this might be related to

293 the energy consumed by more proximal muscle groups. Although the muscles acting at the ankle

294 joint were responsible for the majority of power output, a non-trivial contribution was provided by

295 knee extensors (18-46%) and this was greatest at 2.2 Hz (Figure 5). Work provided by more proximal

296 muscle-tendon units has been postulated to be done less efficiently (33). This is owing to their lesser

297 ability to make use of series elastic compliance than muscle groups such as the plantar-flexors.

298 Therefore, although the contributions of more proximal muscle groups to power output was less

299 than at the ankle, they probably provided this power less efficiently. Thus, they may be responsible

300 for a larger proportion of the total energy consumed than mechanical power output suggests.

301 Because the muscles acting at the knee were providing a greater proportion of the total positive

302 power at 2.2 Hz, the effectiveness of the exoskeletons at reducing overall mechanical demands may

303 have been diminished. This may have contributed to the lack of significant reductions in metabolic

304 power compared to NE.

305　The influence of more proximal muscle groups was highlighted by the lesser reductions in metabolic

306　power that were achieved in this study than by a full leg exoskeleton in a previous study (21). With a

307　full leg exoskeleton, reductions were in the region of 2.0-2.5 W·kg^{-1} (≈30%) (21). By comparison, in

308　the present study the greatest reductions observed were approximately 1.5 W·kg^{-1} (≈20%).

309　Therefore, it seems that it would be beneficial to assist at more proximal joints as well as the ankle.

310　As can be seen from figure 5, the contribution of the hip to overall mechanical power output was

311　minimal. Therefore it might be that assisting at the ankle and knee joints would provide the desired

312　reductions in metabolic cost without requiring a device that assists at the hip too. However, this may

313　only apply to hopping in place which requires no forward swing of the leg.

314　Another further explanation for the lesser metabolic power reductions observed here may be

315　apparent in the RMS EMG data for TA. There was a significant increase in TA RMS EMG during both

316　the aerial and ground contact phases of hopping across all frequencies when hopping with the

317　spring-loaded exoskeletons. Particularly notable was the 2-3 fold increase in normalised EMG signal

318　magnitude during the aerial phase (Figures 7 and 8). In the NE and NS conditions there was apparent

319　pre-activation of the plantar-flexors prior to ground contact that was opposed by some co-activation

320　of TA (Figure 7). In this case, TA may be serving to stiffen the ankle joint to allow pre-activation of

321　the plantar-flexors without generating rotation at the ankle joint (there was <5° of rotation at the

322　ankle during the aerial phase despite considerable plantar-flexor activity). When hopping with

323　spring-loaded exoskeletons, the pre-activation of the plantar-flexors was less (Figures 7 and 8) and

324　this resulted in significantly smaller RMS EMG values for the plantar-flexors during the aerial phase

325　(Figure 8). This might lead one to think that TA should have exhibited lesser activation at the same

326　time to maintain equilibrium of the force couple at the ankle joint and consistent kinematics.

327　However, as shown in Figure 2, the spring was actually exerting force to generate a plantar-flexion

328　moment during the aerial phase. Therefore, TA had to be active to oppose this moment and was

329　effectively co-activating antagonistically with the spring. It was inferred from the raised activation of

330　TA in the spring-loaded condition that the moment generated by the spring during the aerial phase

331   was greater than that generated by the plantar-flexors in the NS and NE conditions. A consequence

332   of raised TA co-activation could be increased metabolic energy consumption that partially cancelled

333   out the benefits of reduced plantar-flexor activation. Elevated levels of antagonistic co-activation has

334   been cited as a factor contributing to increased energy consumption for locomotion in older

335   populations (24, 30). Therefore, the elevated activation of TA to work against the spring during the

336   aerial phase may have contributed to there being lesser metabolic power reductions than were

337   expected.

338   The apparent engagement of the spring during the aerial phase of hops likely represents a design

339   flaw of the exoskeletons. The spring was attached such that its slack length would be reached when

340   the ankle was at 127° of flexion. This angle was chosen from a previous study using similar devices as

341   the typical angle at touchdown (18). Thus, it was intended that the spring would remain slack (i.e.

342   exerting no force) until contact was made with the ground and it could be used to store energy

343   transferred from kinetic and potential energy of the body. However, Figure 2 shows that the

344   touchdown angle for the NS and NE conditions was less (more dorsi-flexed) than 127° (≈116-125°).

345   Extension of the spring would have been required to reach these angles when wearing the spring-

346   loaded exoskeletons. As can be seen in Figure 2, the touchdown angle of the ankle was greater

347   (more plantar-flexed) in the spring-loaded condition and at ≈132° should have placed the spring

348   below its slack length. However, the spring was still exerting force prior to touchdown and thus must

349   have been longer than its slack length. The results indicate that the angle at which the spring

350   engages is an important factor and should be optimised (possibly on an individual basis) to maximise

351   energy storage but minimise aerial co-activation of TA. Alternatively, a clutching mechanism that

352   selectively engages the spring after landing and disengages it after take-off might solve this issue.

353   Such mechanisms are useful for walking (34) and running (11) with spring-loaded assistive devices,

354   to prevent wearers from working against springs during the swing-phases of gait.

355 Other aspects of the exoskeleton that should be considered as 'tuneable' are its mass and spring

356 stiffness. Mass is a relatively simple concept in that it is a matter of making the exoskeletons as light

357 as possible. The exoskeletons used here were prototypes and thus were not fine-tuned to be as light

358 as would be desirable. The added mass of the exoskeletons adds mechanical demands and

359 associated metabolic cost to the task. In fact, it can be seen from the size of the pies in Figure 5 that

360 overall mechanical power output was greater in the NS and the S conditions than for NE. When

361 drawing comparisons between the S and NS conditions, significant reductions in metabolic cost were

362 observed at 2.2, 2.5 and 2.8 Hz. This indicated that the assistance provided by the spring was

363 typically sufficient to counteract the added mass of the exoskeleton, but only improved upon energy

364 consumption of normal hopping at 2.5 Hz. Reductions in metabolic cost relative to the NE condition

365 would be expected to reach similar values to reductions compared to the NS condition as the mass

366 of the device approaches zero.

367 With regard to spring stiffness, it may be necessary to design experiments to try and find an optimal

368 stiffness for metabolic reductions. The spring chosen for this study was selected as the stiffest spring

369 in which a pilot subject could complete the task. It was considered that this was the spring that

370 would provide maximal unloading of biological tissues. However, maximal unloading does not

371 necessarily correspond to the greatest reductions in metabolic cost (4). In fact, Grabowski and Herr

372 (21) showed with their full-leg exoskeleton that a stiffer spring can increase the net metabolic

373 power required for hopping compared to a less stiff spring. They proposed that this was because the

374 stiffer exoskeleton may have compromised stability and/or sensory feedback. As a proxy for stability

375 Grabowski and Herr (21) calculated the distance moved by the centre of pressure between

376 consecutive hops and showed that this distance was typically larger for the stiffer exoskeleton. In

377 this study the same metric was calculated (dHop) and was typically greater for the spring-loaded

378 condition, suggesting that participants were less stable in that condition. Therefore, although

379 intuition might suggest that a greater stiffness (up to the normal ankle stiffness) would be more

380 beneficial, it might actually impair the person's ability to hop. It seems plausible that there would be

381   some optimal stiffness that maximises energy storage and return without significantly impairing

382   stability and sensory feedback. Furthermore, this optimal stiffness might vary with hopping

383   frequency and between individuals depending on their body mass and stature. Careful studies

384   designed to explore these factors are still needed to fine-tune the exoskeleton design.

385   This study showed that spring-loaded ankle exoskeletons can be used to reduce the metabolic cost

386   of bilateral human hopping at certain frequencies. It has highlighted the potential for such devices to

387   assist with bouncing gaits and some of the important design problems that must be addressed.

388   Future studies should aim to investigate optimal spring stiffness and engagement angle. They should

389   also consider how the exoskeletons interact with the underlying muscle-tendon units and how they

390   might be re-designed to be used in running where the kinematics become more complex than for

391   hopping.

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395   **Disclosures**

396   The authors declare no conflicts of interest in relation to this study

397

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480   **Figure Captions**

481   Figure 1. Sketch of the spring-loaded exoskeleton and load-cell

482   Figure 2. Group mean ankle (A), knee (B) and hip (C) angles for each frequency, plotted over an

483   average hop (normalised over 101 points). The left column is the no exoskeleton (NE) condition,

484   middle is the no spring (NS) condition and right is the spring-loaded (S) condition. Vertical lines
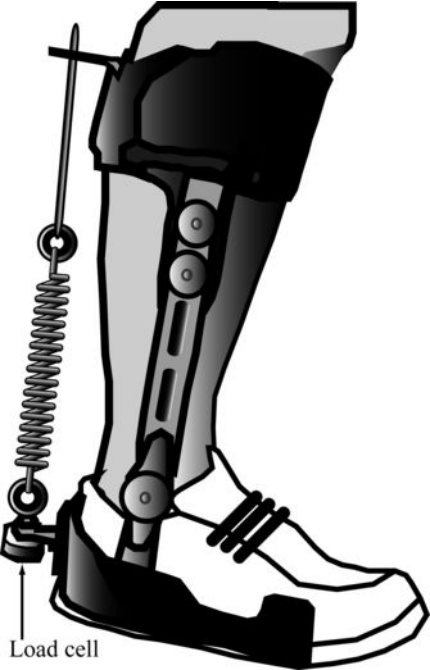
485   represent the range of normalised take-off times across frequencies.

486   Figure 3. Group mean ankle (A), knee (B) and hip (C) moments plotted over an average hop

487   (normalised over 101 points). The additional data series of smaller magnitude on the S ankle plot are

488   the exoskeleton moments for the corresponding frequencies and the hatched area highlights the

489   contribution from biological tissues to the total moment for 2.2 Hz. The left column is the No

490   Exoskeleton (NE) condition, middle is the no spring (NS) condition and right is for the spring-loaded

491   (S) condition. Vertical lines represent the range of normalised take-off times across frequencies.

492   Figure 4. Group mean ankle (A), knee (B) and hip (C) powers plotted over an average hop

493   (normalised over 101 points). The additional data series of smaller magnitude on the S ankle plot are

494   the exoskeleton powers for the corresponding frequencies and the hatched area highlights the

495   contribution from biological tissues to the total ankle power for 2.2 Hz. The left column is the no

496   exoskeleton (NE) condition, middle is the spring (S) condition and right is for the no spring (NS)

497   condition. Vertical lines represent the range of normalised take-off times across frequencies.

498   Figure 5. Pie charts showing the percentage of total average positive power contributed by the

499   biological tissues at the ankle (white), knee (thick hatching), hip (black) and by the exoskeleton

500   (narrow hatching) for the spring-loaded (A), no spring (B) and no exoskeleton (C) conditions. The

501   total area of the pies represents the total average positive power relative to all other pies.
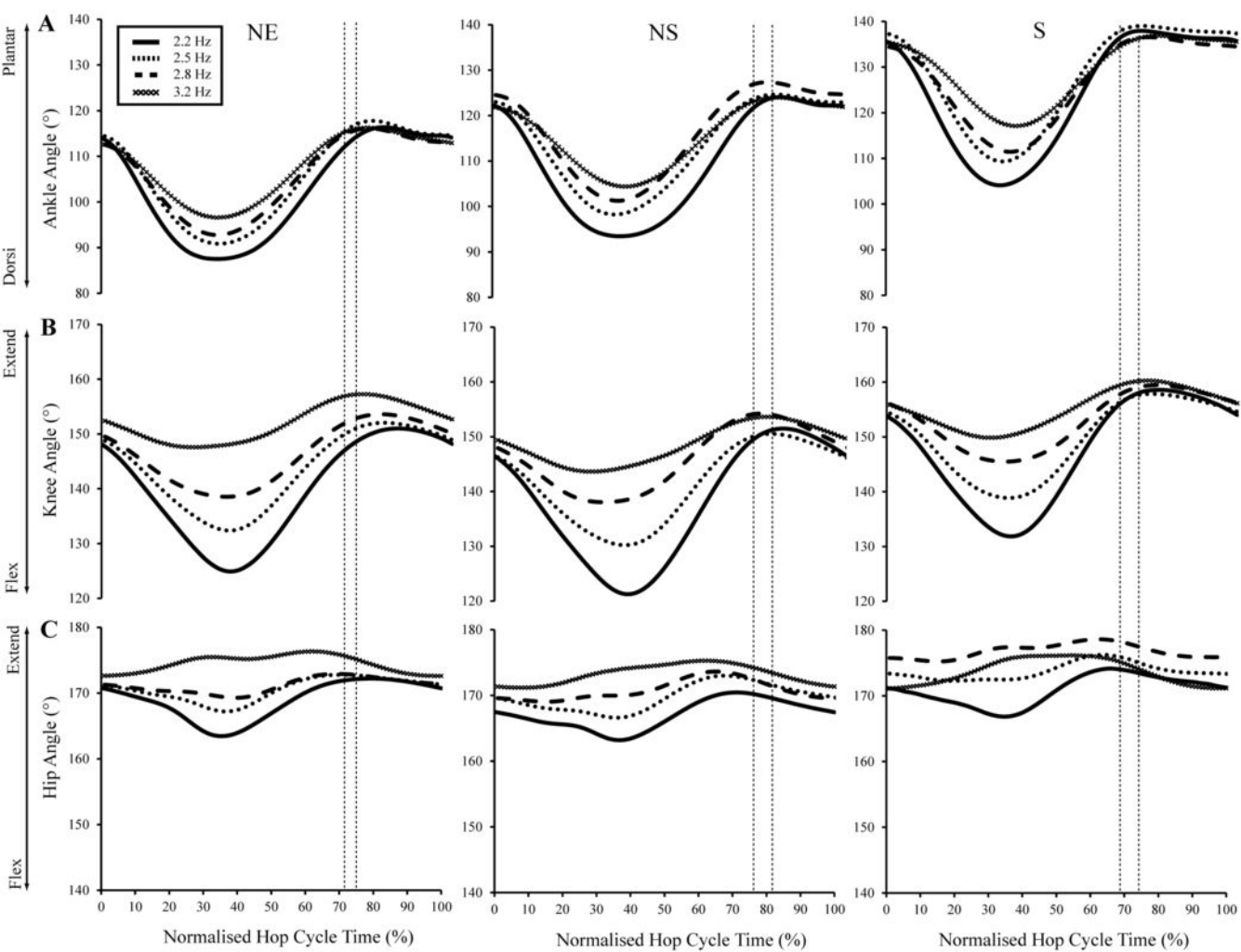
502   Figure 6. Metabolic power (A), biological contribution to $\bar{P}_{tot}^+$ (B) and Biological $\bar{\bar{M}}_{ank}^+$ (C) across

503   hopping frequency for S (filled diamonds), NS (filled circles) and NE (open triangles). * denotes

504   significant difference from S at that frequency and † denotes a significant main effect of frequency

505   Figure 7. Group mean normalised and smoothed time histories of EMG activity for MG (A), LG (B), SO

506   (C) and TA (D) at each frequency for NE, NS and S. Vertical lines indicate the range of normalised

507   take-off times across frequencies.

508   Figure 8. Group mean ± sem RMS EMG values for MG, LG, SO and TA during the ground contact (A)

509   and aerial (B) phases of hops. * denotes statistical significantly different from the S condition and †
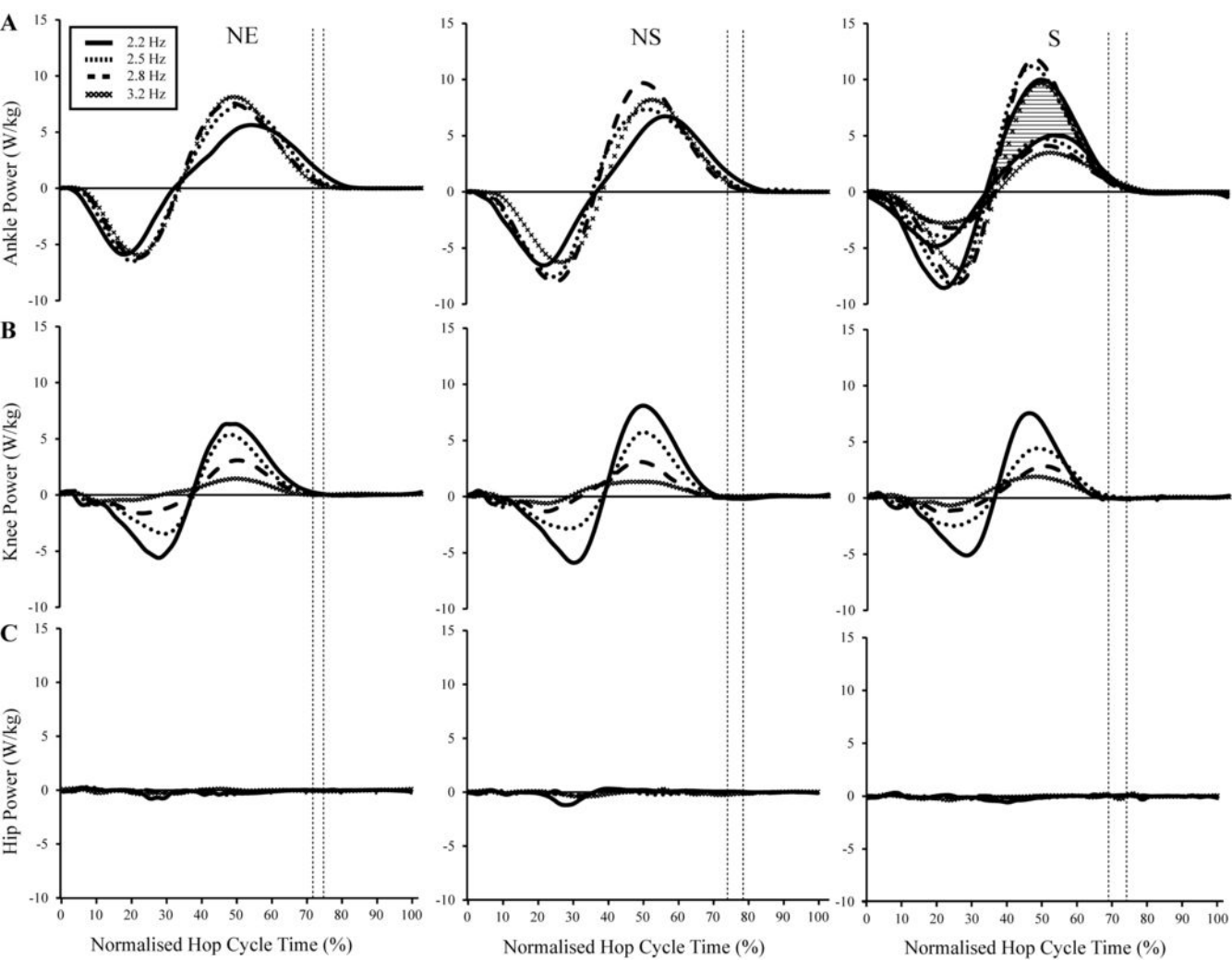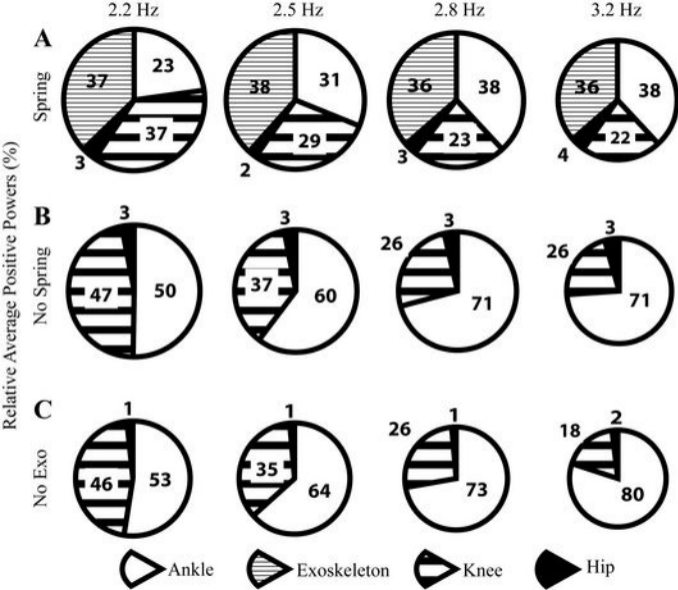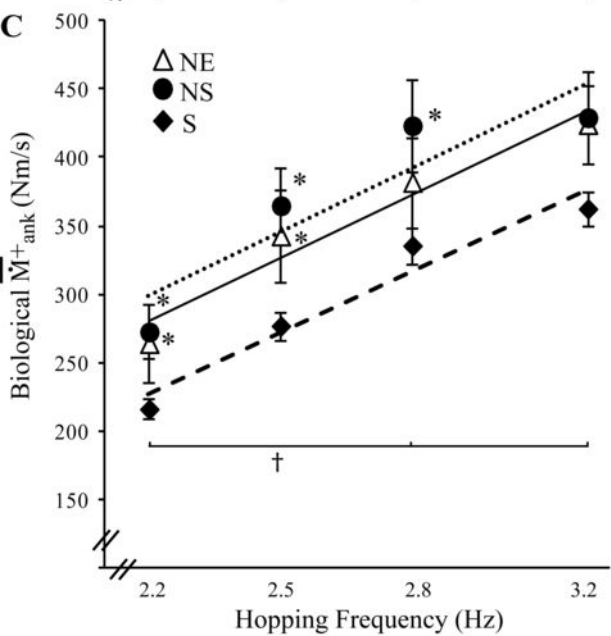
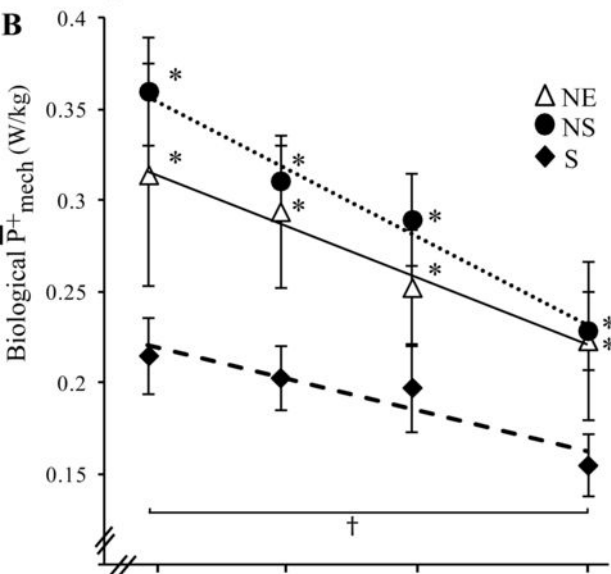510   denotes significant effect of frequency

Load cell

Figure legend:
- 2.2 Hz
- 2.5 Hz
- 2.8 Hz
- 3.2 Hz

A — NE / NS / S
Ankle Angle (°), Plantar ↑ / Dorsi ↓

B
Knee Angle (°), Extend ↑ / Flex ↓

C
Hip Angle (°), Extend ↑ / Flex ↓

Normalised Hop Cycle Time (%)

Relative Average Positive Powers (%)

	2.2 Hz	2.5 Hz	2.8 Hz	3.2 Hz

A Spring

B No Spring

C No Exo

▷ Ankle ▨ Exoskeleton ◁ Knee ◀ Hip
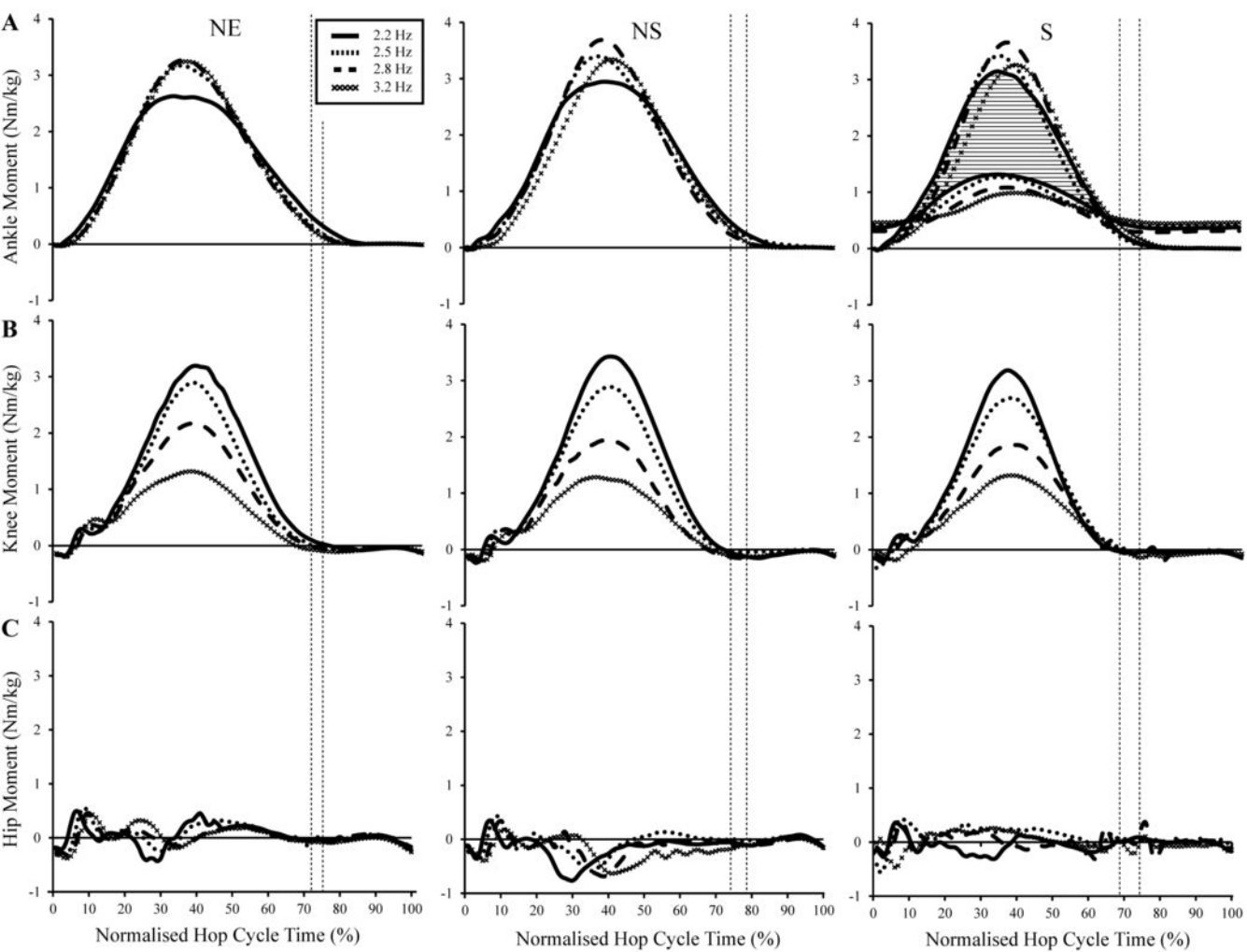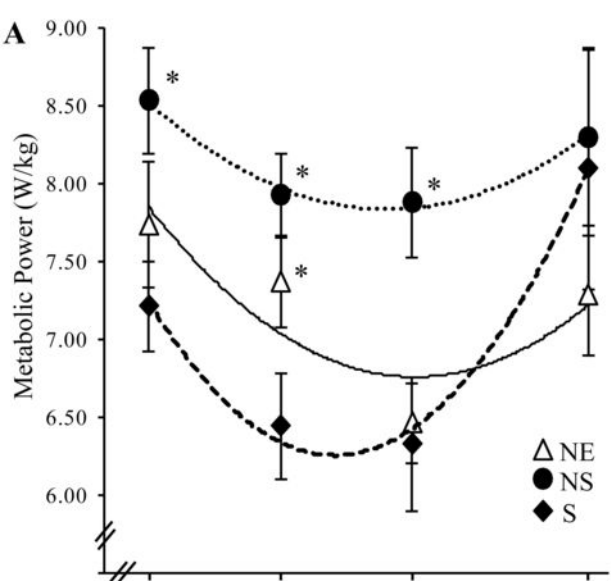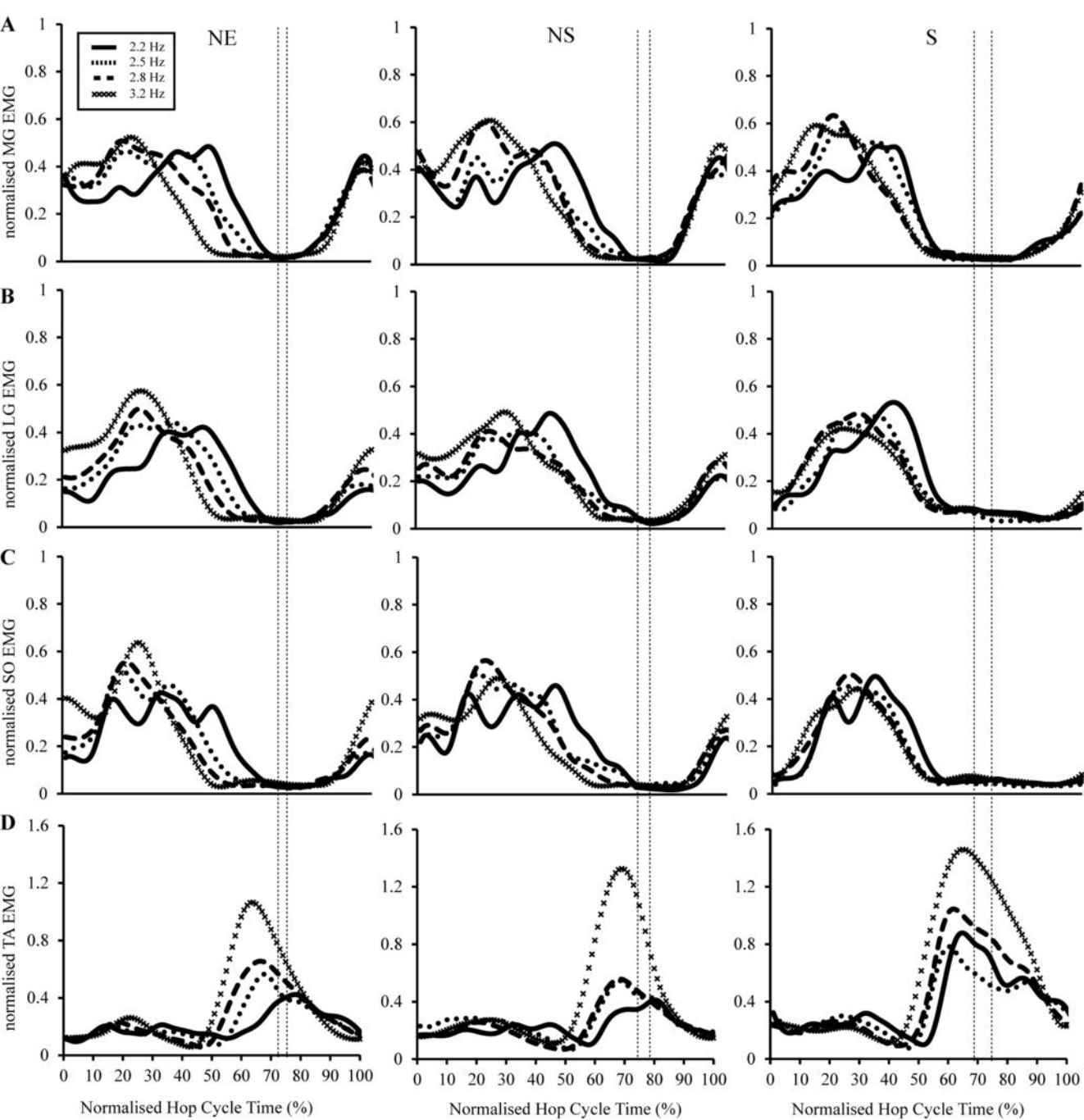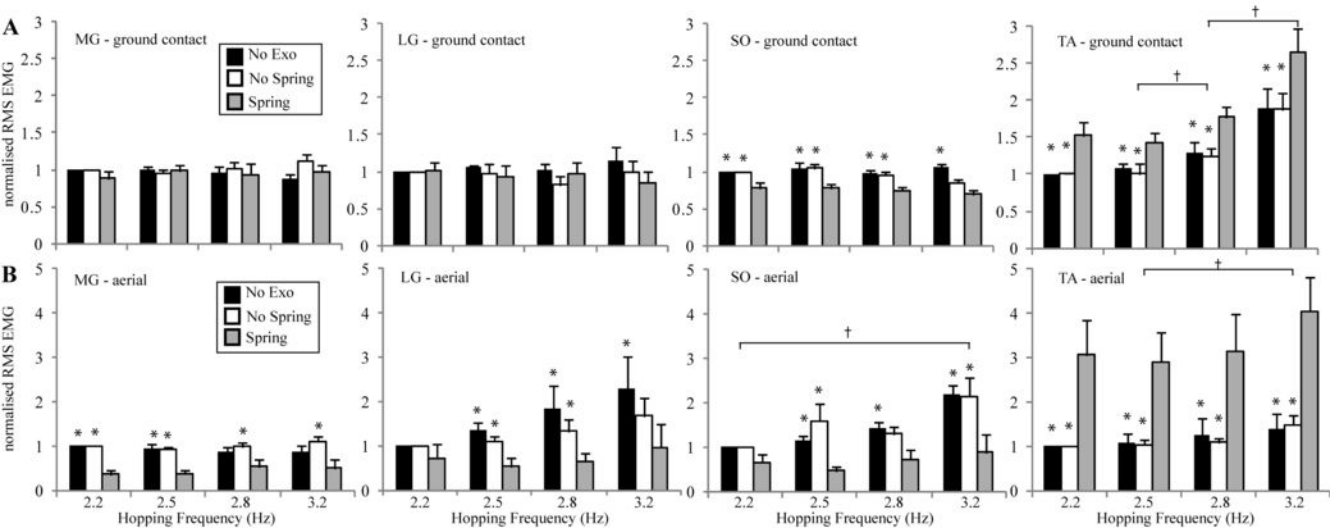
Table 1. Group mean ± sd hopping metrics at each frequency for each spring condition. ANOVA P values are for main effects of the spring condition (S) and frequency (ω) and the interaction effect between the two (S*ω).

	ω (Hz)	Hop Height (mm)	Ground Contact Time (ms)	Actual Frequency (Hz)	Duty Factor	dHop (mm)
No Exo	2.2	22 ± 8	337 ± 41	2.2 ± 0.1	0.75 ± 0.09	18 ± 5
	2.5	20 ± 8	285 ± 35	2.5 ± 0.1	0.72 ± 0.08	20 ± 4
	2.8	13 ± 5	256 ± 29	2.8 ± 0.0	0.72 ± 0.07	17 ± 5
	3.2	9 ± 4	225 ± 19	3.2 ± 0.1	0.73 ± 0.06	17 ± 5
No Spring	2.2	13 ± 6	352 ± 28	2.2 ± 0.1	0.78 ± 0.06	20 ± 5
	2.5	13 ± 6	300 ± 23	2.5 ± 0.0	0.75 ± 0.06	22 ± 10
	2.8	11 ± 4	264 ± 19	2.8 ± 0.0	0.74 ± 0.05	20 ± 7
	3.2	5 ± 2	264 ± 12	3.2 ± 0.1	0.79 ± 0.04	25 ± 7
Spring	2.2	17 ± 14	312 ± 35	2.3 ± 0.2	0.70 ± 0.05	25 ± 12
	2.5	15 ± 9	279 ± 27	2.5 ± 0.0	0.69 ± 0.07	22 ± 5
	2.8	13 ± 7	259 ± 23	2.8 ± 0.2	0.72 ± 0.04	28 ± 9
	3.2	9 ± 4	236 ± 22	3.1 ± 0.1	0.74 ± 0.05	29 ± 8
ANOVA P (S)		0.19	0.08	0.100	0.171	0.038*
ANOVA P (ω)		<0.000*	<0.000*	<0.000*	0.014	0.424
ANOVA P (S*ω)		0.259	0.003*	0.263	0.071	0.432

*Indicates statistically significant effect.