1   **Title:** Elastic ankle exoskeletons reduce soleus muscle force but not work in human hopping

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8   **Running Header:** Ankle exoskeletons reduce soleus muscle force but not work

9   **Abstract**

10   Inspired by elastic energy storage and return in tendons of human leg muscle-tendon units

11   (MTU), exoskeletons often place a spring in parallel with an MTU to assist the MTU.

12   However, this might perturb the normally efficient MTU mechanics and actually increase

13   active muscle mechanical work. This study tested the effects of elastic parallel assistance on

14   MTU mechanics. Participants hopped with and without spring-loaded ankle exoskeletons

15   that assisted plantar-flexion. An inverse dynamics analysis combined with in vivo ultrasound

16   imaging of soleus fascicles and surface electromyography was used to determine muscle-

17   tendon mechanics and activations. Whole-body net metabolic power was obtained from

18   indirect calorimetry. When hopping with spring-loaded exoskeletons, soleus activation was

19   reduced (30 - 70%) and so was the magnitude of soleus force (peak force reduced by 30%)

20   and the average rate of soleus force generation (by 50%). Although forces were lower,

21   average positive fascicle power remained unchanged owing to increased fascicle excursion

22   (+ 4-5 mm). Net metabolic power was reduced with exoskeleton assistance (19%). These

23   findings highlighted that parallel assistance to a muscle with appreciable series elasticity

24   may have some negative consequences and that the metabolic cost associated with

25   generating force may be more pronounced than the cost of doing work for these muscles.

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27   **Keywords-** Ultrasound, fascicle, tendon, metabolic power, plantar-flexors.

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Introduction

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32   Assistive exoskeletons or wearable robots have the potential to restore locomotor function

33   in individuals with musculo-skeletal disorders and augment locomotor function for healthy

34   persons. The desired outcome of wearing an exoskeleton is typically to reduce the demands

35   placed on the musculo-skeletal system during locomotion (17). This might be with the

36   intention of (i) lowering the metabolic cost of transport (7, 15, 25, 38); (ii) reducing musculo-

37   skeletal injury risk; and/or (iii) providing mechanical power output that the biological tissues

38   cannot (3).

39   One of the main challenges of designing exoskeletons is to minimise their mass but still have

40   them be capable of powering locomotion. One possible solution for this is to remove

41   powered actuators and replace them with passive springs that are lightweight but can store

42   and return energy to help power locomotion (7, 15, 17, 25, 40). This approach takes

43   advantage of the natural spring-like mechanics of the human leg during locomotion (35). A

44   simple spring-mass model can be used to replicate the motion of the body centre of mass

45   during locomotion (5), highlighting the potential for storage and return of energy within

46   elastic tissues in the legs during stance. In particular, elastic tissues in series with muscles

47   (e.g. tendons) can be used to store energy from, and return energy to, the centre of mass (2,

48   9).

49   Taking inspiration from this biological mechanism, passive exoskeletons using springs in

50   parallel with the muscles of the legs have been developed (7, 11, 15, 17, 25, 40). Grabowski

51   and Herr (25) showed that an exoskeleton with springs spanning all three joints (ankle, knee

52   and hip) of the leg could be used to reduce the metabolic cost of two-legged hopping in

53   place. These authors demonstrated that when hopping in the exoskeletons, individuals

54 reduced their biological (muscular) contribution to leg stiffness to maintain normal overall

55 stiffness (biological plus exoskeleton) and centre of mass mechanics. Similar effects have

56 been observed specifically at the ankle joint for humans hopping in ankle-foot orthoses that

57 were spring-loaded to assist plantar-flexion (11, 15, 17). In these studies, plantar-flexor

58 electromyographic activity was reduced when hopping with the device. This was shown to

59 reduce the biological contribution to ankle stiffness (17) and, at certain hopping

60 frequencies, net metabolic power during hopping (15).

61 Based on the aforementioned studies of joint and centre of mass level mechanics, one

62 might conclude that these spring-loaded exoskeletons are successful in achieving their goals

63 of reducing mechanical and metabolic demands on the musculo-skeletal system. However,

64 to date nobody has studied the effects of providing parallel assistance to a muscle-tendon

65 unit (MTU) on the mechanics of the MTU itself. This may be of particular importance for

66 MTU's such as those comprising the ankle plantar-flexors that have relatively short, pennate

67 fascicles in series with a longer, compliant series elastic element (SEE) composed of

68 aponeurosis and external tendon (21). This morphology allows length changes of the muscle

69 fascicles to be decoupled from ankle joint rotation because angular excursion at the joint

70 can be provided by stretch and recoil of the SEE (19, 37). One of the benefits of this

71 decoupling of muscle length change from joint excursion is that muscle fibres are potentially

72 able to produce force with minimal changes in length and at relatively slow velocities (37).

73 This should reduce the required muscle activation and metabolic energy consumption for a

74 given level of force production (16).

75 Ultrasound imaging studies of human plantar-flexor MTU mechanics have actually shown

76 that during the stance phase of walking, running and hopping, muscle fascicles contract

77 relatively slowly and length changes are primarily occurring in the SEE (14, 20, 22, 29, 30).

78 This allows the SEE to store and return energy, minimising the work that must be done by

79 active muscle that has the primary function of producing force isometrically (or with

80 minimal length change) to stretch the SEE. Therefore, it seems that, in a healthy individual,

81 muscle-tendon interaction within MTU's that have a compliant SEE is well tuned to provide

82 work output at the joint level with high efficiency (31, 32).

83 However, assuming that the SEE has a reasonably constant stiffness and that ankle joint

84 kinematics remain constant, the tuned interaction of muscle and tendon must require a

85 particular force profile to be applied to the SEE by the muscle. As stated above, assistive

86 ankle exoskeletons reduce plantar-flexor muscle activation (15, 17) leading to reduced

87 muscular contributions to joint stiffness (17). Presumably this indicates that the muscles are

88 producing lesser forces and thus, may not be able to stretch the SEE to the same extent as

89 when unassisted. Therefore, it could be that providing parallel assistance to a MTU with a

90 compliant SEE interferes with the MTU's efficient mechanics. This might mean that there are

91 some negative effects of providing such assistance as well as the previously stated benefits.

92 The aim of this study was to test, in vivo, whether providing exoskeletal assistance in

93 parallel to a MTU with a compliant SEE during a cyclic movement interferes with the

94 normally efficient muscle-tendon interaction that occurs. It was hypothesised that when

95 assistance is provided in parallel to the human plantar-flexors, soleus would reduce its

96 activation and force production levels, resulting in decreased stretch of the SEE and a

97 compensatory increase in length change of the muscle fascicles. Furthermore, it was

98 expected that the predicted increase in soleus fascicle length change would increase soleus

99 fascicle mechanical work despite the decreased load on the muscle.

Methods

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(a) Participants

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

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gave written informed consent to participate in this study. All participants were in good

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health and had no recent history of lower limb musculo-skeletal injury. All procedures were

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approved by an institutional review board and complied with the guidelines for research

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involving human participants as set out in the Declaration of Helsinki.

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(b) Experimental Protocol

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Bilateral hopping is a bouncing gait with similar spring-mass mechanics to running but

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simpler kinematics. The plantar-flexors undergo a stretch-shortening motion, meaning a

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simple spring-loaded ankle exoskeleton can be used to provide parallel assistance to the

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plantar-flexors. Therefore, each participant was required to perform bilateral hopping in

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place in time with the beat of a metronome operating at 2.5 Hz. This frequency was chosen

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because a previous study using the same movement found that this was the frequency

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around which metabolic cost was minimised when hopping in spring-loaded exoskeletons

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(25). The hopping task was performed for 4 minutes to allow the participants to reach a

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metabolic steady state and was completed under 3 experimental conditions: (1) With no

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exoskeleton (NE); (2) With bilateral ankle exoskeletons but no spring (NS); (3) With spring-

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loaded exoskeletons to assist plantar-flexion (S).

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(c) The Exoskeletons

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The devices have been previously described elsewhere (15) and a sketch of the exoskeletons

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used is shown in Figure 1. The exoskeleton consisted of a carbon fibre cuff around the upper

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122   shank which was connected to a carbon fibre foot section via two aluminium bars which had

123   a freely rotating joint aligned with the participants' malleoli. The foot section was

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

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

126   the foot segment via a number of metal links. The number of links was adjusted for each

127   participant such that the resting length of the spring coincided with an ankle angle of 127°

128   which has been determined as the typical angle at ground contact in hopping (17). This

129   same approach was used by Ferris et al. (17) for a similar exoskeleton. A compression load

130   cell (Omegadyne Inc., OH,USA) was placed on the inferior side of the bolt at the heel of the

131   foot segment and attached to the links in series with the spring. This was used to measure

132   the forces in the spring. The stiffness of the spring in tension was 5 kNm^{-1} and its moment

133   arm about the joints was 0.135 m. This gave a rotational stiffness of 1.59 Nm/° (91 $Nm·rad^{-1}$)

134   which is approximately 40% of ankle stiffness during unassisted hopping at preferred

135   frequency (17).

136   *(d) External Kinematics and Kinetics*

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

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

139   marker positions were filtered using a second order low-pass butterworth filter with a cut-

140   off of 10 Hz. A static standing trial was captured and the positions of markers on segment

141   end points were used to calibrate a four segment (pelvis, thigh, shank and foot) model for

142   each subject using established inertial parameters (12). Clusters of three or four markers on

143   rigid plates were attached to the pelvis, thigh and shank segments to track segment motion

144   during hopping. For the foot, a cluster of three markers was attached directly to the shoe.

145   Joint angles for the hip knee and ankle were computed in three dimensions as the

146   orientation of the distal segment with reference to the proximal segment.

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

148   computed during vertical hopping using a split belt instrumented treadmill (Bertec, OH,

149   USA) with the belts turned off. Participants hopped such that each foot was on a separate

150   half of the treadmill and thus, the two 3D force vectors could be attributed separately to the

151   left and right legs. Raw analogue force platform signals were filtered using a low-pass

152   Butterworth digital filter with the cut-off set to 35 Hz. Inverse dynamic analyses (41) were

153   then used to compute net joint moments at the hip, knee and ankle. Kinematics and kinetics

154   were calculated for the right leg only and it was assumed that the left leg behaved

155   symmetrically. Inverse dynamics procedures were performed with Visual 3D software (C-

156   motion Inc., Germantown, MD, USA). For the S condition, the contribution of the

157   exoskeleton to the net ankle joint moment had to be determined. The force in the spring

158   was computed from the load cell output voltage (from its factory calibration data) and

159   multiplied by the moment arm of the spring about the ankle joint. This gave the plantar-

160   flexion moment provided by the exoskeleton and this was subtracted from the total ankle

161   moment to give the moment provided by biological tissues.

162   *(e) Determination of Soleus Muscle Parameters*

163   Soleus (SO) muscle fascicle length during hopping was measured from B-mode ultrasound

164   images (27) (Figure 1). A linear ultrasound transducer (LV7.5/60/96Z, Telemed, Lithuania)

165   operating at 8.0 MHz was placed over the mid-belly of the SO and aligned so that SO

166   fascicles could be visualised from deep to superficial aponeuroses (Figure 1). The reliability

167   and accuracy of ultrasound measurements of fascicle length are reported elsewhere (1, 18,

168   36). Images were sampled at 50 Hz and a pulse from the ultrasound system that was high (3-

169   5 V) during recording and low (0 V) before and after was used to trigger collection of all

170   other data synchronously. To obtain fascicle length from each image, a custom MATLAB^{TM}

171   (The Mathworks Inc., Natick, MA) program was used to digitize the points of attachment of

172   a fascicle on the superficial and deep aponeuroses and the length was calculated as the

173   distance between these two points. Pennation angle was defined as the angle between the

174   digitized fascicle and the deep aponeurosis (Figure 1). The instantaneous length of the

175   whole SO MTU was calculated from ankle joint flexion-extension angle using the equations

176   of Hawkins and Hull (26). To obtain a value for the length of the SEE, the length of the

177   fascicle was multiplied by the cosine of pennation angle and subtracted from the MTU

178   length (Fukunaga et al., 1997; Figure 1). Initial fascicle length (L_i) was taken as the length of

179   the fascicle at landing. Following landing, fascicles lengthened then shortened. Fascicle

180   lengthening (ΔL_{FAS}^{+}) was calculated relative to L_i by subtracting L_i from the peak length

181   during stance. Fascicle shortening (ΔL_{FAS}^{-}) was calculated as the length at take-off minus the

182   peak length during stance. Overall length change (ΔL) was the sum of the absolute values of

183   ΔL_{FAS}^{+} and ΔL_{FAS}^{-}.

184   *(f) Soleus Muscle Kinetics*

185   Procedures for determining soleus kinetic data were similar to those employed by Farris and

186   Sawicki (14) previously for the gastrocnemius. Direct measurement of muscle force was not

187   possible and so it was estimated from inverse dynamics and SO muscle parameters. Forces

188   transmitted by the Achilles tendon (AT) to the calcaneus were calculated as the biologically

189   generated ankle moment divided by the moment arm of the AT about the ankle joint (14,

190   30). This moment arm was calculated as the first derivative of SO MTU length with respect

191   to ankle angle (6, 32). To reduce this force solely to that contributed by SO it was multiplied

192   by the relative physiological cross-sectional area (PCSA) of SO within the plantar flexors

193   (0.54 from ref. (23)). Next, SO force was divided by the cosine of SO pennation angle to

194   calculate the force generated along the line of the fascicle (F_{SO}). The average rate of SO

195   force production ($\bar{\dot{F}}_{SO}$) was calculated by differentiating SO force with respect to time,

196   integrating the period when the derivative was positive during a hop, and dividing the

197   integral by the time taken for an entire hop. This value was calculated for multiple hops (8-

198   10) and averaged.

199   The velocities of the SO fascicles, MTU and SEE were calculated as the first derivative of

200   their lengths with respect to time. The power output of the fascicles, SEE and MTU were

201   then calculated as the product of their respective forces and velocities. Positive work done

202   by fascicles, SEE and SO was estimated by integration of positive portions of each

203   component's power curve. Periods of positive power during each trial were integrated by

204   the trapezium method and summed, then divided by the number of hops taken in that trial

205   to calculate average positive work done per hop. These values were divided by hop cycle

206   time to convert to average positive powers for fascicle ($\overline{P}_{FAS}^{+}$), SEE ($\overline{P}_{SEE}^{+}$) and MTU ($\overline{P}_{MTU}^{+}$).

207   These average positive powers were considered indicative of the fascicle and tendon

208   interaction. For example, the overall MTU output would be most efficient if $\overline{P}_{FAS}^{+}$ was zero

209   (i.e. the fascicle is always isometric) and all of $\overline{P}_{MTU}^{+}$ was supplied by $\overline{P}_{SEE}^{+}$ (i.e. from recoil of

210   the SEE).

211   *(g) Electromyography*

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

213 Gastrocnemius (MG), Lateral Gastrocnemius (LG), Soleus (SO) and Tibialis Anterior (TA). All

214 four channels were recorded using wired electrodes (Biometrics Ltd, UK) that were carefully

215 placed over muscle bellies after the skin surface was prepared by light abrasion and cleaned

216 with an alcohol swab. D.C. offsets were removed from raw signals which were then band-

217 pass filtered (20-300 Hz). The data were then smoothed by calculating the root-mean-

218 squared (RMS) value of the signals over a rolling window of 20 ms. The RMS of each

219 muscle's signal was also calculated over the period of ground contact and the aerial phase

220 of each hop as a metric of total activity over these two phases of the hop. Processed EMG

221 signals for each muscle were normalised to the average of the local (within each hop)

222 maxima of the signal recorded in that muscle in the NS condition.

223 *(h) Metabolic Power*

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

225 recorded using a portable metabolic system (Oxycon Mobile, Viasys Healthcare, CA, USA).

226 Prior to hopping, measurements were made during five minutes of quiet standing and

227 values from the last two minutes were averaged and used to calculate rates of metabolic

228 energy consumption whilst standing. For the hopping trials, data from the last two of the

229 four minutes were averaged for the calculation of metabolic rate. Visual inspection of rates

230 of oxygen consumption with time (averaged over 30 s intervals) confirmed that participants

231 were at steady-state during this period and the respiratory exchange ratio was never

232 greater than one. Rates of oxygen consumption and carbon dioxide production were

233 converted to metabolic powers using standard equations detailed by Brockway (8). Net

234 metabolic powers during hopping were calculated by subtracting metabolic power during

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

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

237   unless otherwise stated.

238   *(i) Statistical Analyses*

239   All time-series data for individual participants were reduced to the mean of at least 10 hops

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

241   are the mean ± standard error for the whole participant group. To test for statistical

242   differences in dependent variables between conditions a one-way ANOVA with repeated

243   measures was employed using SPSS software (IBM, USA). The independent variable for the

244   ANOVA was spring condition (3 levels – NE, NS, S). F-ratios for main effects were considered

245   significant for $P < 0.05$. If a significant main effect was found, paired t-tests were used to

246   make pair-wise comparisons between spring conditions.

247   **Results**

248   Hop heights and duty factors (proportion of a hop cycle spent in contact with the ground)

249   were not significantly different between conditions (Table 1) indicating that the overall

250   mechanical demand of the hopping task on the lower-limbs limbs was not different between

251   conditions (15). RMS EMG for SO during the aerial phase ($SO_{RMSaerial}$) and ground contact

252   ($SO_{RMSground}$), peak F_{SO} and $\bar{F}_{SO}$ were all significantly ($P < 0.01$) less for the S condition than for

253   NS and NE conditions (Table 1, Figure 2 a,b,c). These reductions occurred concurrently with

254   a significant increase in both SO fascicle total excursion (lengthening + shortening) and

255   fascicle shortening for S compared to NS ($P = 0.01$) and NE ($P = 0.048$) during the stance

256   phase (Table 1, Figure 3a). There was no difference in length changes (relative to 0% hop

257   time) of the SEE or MTU between conditions (Figure 3 b,c). However, both the MTU and the

258   SEE were at significantly (P < 0.00) shorter lengths on average throughout the hop cycle in

259   the S condition (Table 1). This was associated with the ankle joint being more plantar-flexed

260   on average over a hop cycle for S (Table 1). A full description of joint kinematics and kinetics

261   has been previously published (15).

262   Fascicle length change increased and F_{SO} decreased in S (Figure 3a, 2b). This trade-off meant

263   that $\overline{P}_{FAS}^{+}$ was unchanged between conditions (Figure 4). However, both $\overline{P}_{SEE}^{+}$ and $\overline{P}_{MTU}^{+}$

264   were significantly less for S than for NS and NE (Figure 4). The net result of these findings

265   was that the ratio of $\overline{P}_{FAS}^{+}$ to $\overline{P}_{SEE}^{+}$ went down in S, indicating that a smaller proportion of

266   $\overline{P}_{MTU}^{+}$ was being provided by the return of elastic energy from the SEE. Whole body net

267   metabolic power was significantly less for S than NS (-19%, P = 0.010) and NE (-13%, P =

268   0.016).

269   **Discussion**

270   This study aimed to assess, in vivo, the effects on muscle-tendon mechanics of providing

271   parallel assistance to a MTU that has a compliant SEE. As predicted, soleus fascicle length

272   change increased when assistance was provided although, this did not lead to an increase in

273   $\overline{P}_{FAS}^{+}$. This was due to reductions in soleus activity and the resulting force production as

274   discussed below. It should be noted that more common locomotor tasks (i.e. walking and

275   running) utilise more complex kinematics than hopping and thus, care should be taken in

276   extrapolating these findings to such tasks. However, plantar-flexor stretch-shortening cycles

277   are important in walking and running and therefore, it is relevant to these tasks to study the

278   effects of a parallel spring on soleus stretch-shortening mechanics in a simpler motion.

279   *(a) Muscle activation and force production*

280   Based on previous studies using similar devices, the first hypothesis was that soleus

281   activation levels would be reduced when exoskeleton assistance was provided. As can be

282   seen from Figure 2a and Table 1, this hypothesis was supported because there were

283   reductions in $SO_{RMSaerial}$ (60-70%) and $SO_{RMSground}$ (≈30%). Most notably, reductions occurred

284   late in the aerial phase and early in stance (Figure 2a). These timings of reductions in SO

285   activation agree well with data from hopping in a similar device (17).

286   Also as predicted, there was a concurrent reduction in F_{SO} during the stance phase when

287   hopping with assistance compared to both other conditions (Figure 2b, Table 1). In fact, the

288   onset of force production during stance was later and the peak F_{SO} was less (≈30%) when

289   hopping with assistance. The later onset of force production may have been facilitated by

290   the lesser pre-activation of SO at the end of the aerial phase and lower activation in early

291   stance (Figure 2a). It seems that parallel assistance in the form of a spring-loaded ankle

292   exoskeleton was able to effectively reduce the activation and loading of SO during hopping.

293   One potential benefit of reduced force and activation is that the metabolic costs associated

294   with producing muscular force could be reduced. These costs have been considered to

295   account for ≈50% of the metabolic cost of transport in humans (24). Biewener (4) proposed

296   that two factors influence the energy cost of producing muscular force in mammalian

297   locomotion: (1) the magnitude of force generated per unit time and (2) the rate of force

298   development and frequency of activation. As indicated by Figure 2b, the magnitude of F_{SO}

299   was always less when assistance was provided. Hop cycle time was constant and thus the

300   magnitude of force per unit time was less for hopping with assistance. Furthermore, Figure

301   2c shows that $\bar{F}_{SO}$ was significantly less (≈50%) when the spring-loaded exoskeletons were

302   being used. Because the frequency of the cyclic hopping task was controlled, the frequency

303   of SO activation should not have changed. Overall, the metabolic cost of producing force in

304   SO should have been reduced when assistance was provided. Consistent with this, whole

305   body net metabolic power was significantly reduced with assistance (Figure 4). However,

306   studies of whole-body mechanics have suggested that the cost of force production accounts

307   for ≈50% of net metabolic costs (24) and so there may be other factors to consider at the

308   muscular level.

309   *(b) Muscle-tendon mechanics and energetics*

310   In addition to muscle force, mechanical work done by muscle has been cited as the other

311   major factor in determining metabolic energy consumption during gait (10, 13, 24). The

312   mechanical work done by muscle is determined by the length change of contractile

313   elements and the force they produce during this length change. It was predicted that

314   reduced force production by SO would result in a smaller stretch of the SEE and that this

315   would have to be compensated for by increased length changes of muscle fascicles,

316   resulting in greater average fascicle rate of work ($\overline{P}_{FAS}^{+}$).

317   The changes in the length of the SEE were not different when assistance was added (Table 1,

318   Figure 3b). Initially this seems improbable given the elastic nature of the SEE and the noted

319   reductions in F_{SO}. However, the fact that the average length of the SEE and the whole MTU

320   were both significantly reduced when assistance was used (Table 1) may provide a clue.

321   Tendons typically exhibit a non-linear 'toe region' in their force-elongation relationship at

322   shorter lengths (2). This region has been specifically observed for the SEE of the human

323   plantar-flexors (34). In this region the SEE will lengthen more for a given increase in force

324   than in the stiffer linear region that exists at longer lengths. Without individual force-length

325   plots for each of the study participants it is hard to conclusively say when their SEE was and

326   was not in the toe region. However, this would help to explain why, despite the smaller

327   increase in force observed in the S condition, the SEE undergoes similar changes in length to

328   NE and NS but at shorter absolute lengths. It could even be that in the S condition the SEE

329   was shorter than its slack length early in stance. Again, without individual force-length data

330   this is hard to confirm but, in the S condition SO was contributing no force early (and very

331   late) in stance. At these times, the entire plantar-flexion moment was due to forces in the

332   spring of the exoskeleton (Figure 2b). This was despite small amounts of SO activation and

333   could be explained by a slack SEE.

334   With the SEE being at shorter lengths with assistance, one might have expected the fascicles

335   to have to have compensated and operated at longer lengths and over greater excursions.

336   However, the ankle kinematics changed with the addition of the spring and the ankle joint

337   was significantly more plantar-flexed on average (Table 1). This resulted in a shorter average

338   length of the whole MTU and SEE, but there was no change in average fascicle length (Table

339   1). That said, there were differences in fascicle length changes for S compared to NE and

340   NS.

341   As expected, the total excursion of SO fascicles was increased by 4-5 mm ($\approx$ 2-3 mm of

342   increased lengthening and shortening) compared to the unassisted conditions (Figure 3,

343   Table 1). This was as hypothesised but, contrary to predictions, the increased excursion did

344   not lead to any significant changes in $\overline{P}^{+}_{FAS}$ (Figure 4). This is because the work done by a

345   fascicle is dependent upon its length change and its force production during that length

346   change. In the present study, although with assistance there was increased fascicle

347   shortening (shortening results in positive work) there was also a concurrent reduction in

348   force production. The result of this trade-off was no change in $\overline{P}^+_{FAS}$. Thus, the hypothesis

349   that parallel assistance would have some negative impact on metabolic cost by increasing

350   fascicle work was not supported. This also meant that a reduction in overall net metabolic

351   cost was achieved without altering work done by the SO muscle. Thus, the metabolic costs

352   associated with doing work may not be as important as those associated with producing

353   force for SO during bouncing gaits.

354   Whilst $\overline{P}^+_{FAS}$ was not increased by assistance, the ratio of $\overline{P}^+_{FAS}$ to $\overline{P}^+_{SEE}$ was altered. $\overline{P}^+_{SEE}$

355   accounted for 74-78% of $\overline{P}^+_{MTU}$ without assistance and only 63% with assistance (Figure 4).

356   This should not increase metabolic cost but it may affect the apparent efficiency of SO

357   mechanical work. Apparent efficiency is high for MTU's when most of the positive work is

358   provided by energy returned from stretch of the tendon and minimal work is done by active

359   muscle (38). By reducing the energy stored and returned in the SEE, apparent efficiency of

360    $\overline{P}^+_{MTU}$ may have been reduced in the assisted condition.

361   *(c) Contributions from other muscles*

362   Several of the central discussion points of this paper have attempted to relate SO mechanics

363   to whole body net metabolic power. Clearly there are other muscles driving this cost, both

364   at the ankle and at other joints. Based on PCSA, SO is the largest of the plantar-flexors

365   accounting for ≈54% of the summed PCSA of this muscle group (23). SO was chosen because

366   of this and it was anticipated that it would make the largest contribution to ankle plantar-

367   flexion moments. However, it should be noted that a significant portion of plantar-flexor

368   force and work may come from the gastrocnemius and this may influence metabolic energy

369   consumption. However, the fact that MG and LG activations during ground contact did not

370   change with assistance may indicate that much of the change in ankle mechanics is due to

371   the change in soleus activation.

372   On a similar theme, hopping was chosen partly because the overall power production of the

373   task can be controlled, and the ankle joint is the primary power source, making it ideal for

374   studying muscle-tendon mechanics and energetics of the plantar-flexors. As such, we note

375   that hop heights and duty factors were not significantly different between experimental

376   conditions (Table 1). This combined with the controlled hopping frequency indicated that

377   the total external power requirements of the hopping task were consistent across

378   conditions. Although the ankle dominates overall power output of the task (≈60% at 2.5 Hz),

379   the muscles acting at the knee (37%) and hip (3%) also contribute to mechanical power

380   output and because of their lesser series compliance may do so less efficiently (39). This

381   would not affect the changes in metabolic cost observed between conditions if knee and hip

382   contributions were constant across conditions. However, there was a reduction in the

383   contribution at the knee from 37% to 29% in the assisted condition. This probably

384   contributed to some of the observed reduction in metabolic cost. The reduction in average

385   positive power at the knee (-0.2 $W \cdot kg^{-1}$) was much less than at the ankle (-1.0 $W \cdot kg^{-1}$). If a

386   typical muscle efficiency of 0.25 were assumed for the muscles acting at the knee (39), this

387   reduction would account for 0.8 $W \cdot kg^{-1}$ of the 1.5 $W \cdot kg^{-1}$ reduction in whole-body net

388   metabolic power that was observed (Figure 4). Despite some metabolic reduction coming

389   from reduced knee mechanical power, a significant portion (~0.7 $W \cdot kg^{-1}$) of the total

390   reduction in metabolic power can still be attributed to reductions in mechanical power

391   output at the ankle joint, most of which is due to reductions in force, but not work, of the

392   soleus muscle fascicles.

393   *(d) Force sharing considerations*

394   The force sharing between SO, MG and LG was determined purely on their relative PCSA's.

395   This approach assumes that their relative activations (relative to their own maximum) are

396   similar and fluctuate similarly. Maximums were not tested but, if different muscles' relative

397   activation were different within an experimental condition, it would have introduced a

398   systematic error that would not change the main results of this study. Of greater concern

399   with the current study design is whether or not their relative activations remained similar

400   when assistance was added. From Table 1 it can be seen that SO activation during stance

401   was reduced with assistance but MG and LG's was not. This would imply that relative

402   activation changed for SO but not for LG and MG. However, because of the direction of this

403   change (i.e. soleus is at a lower percentage of its maximum activation in the S condition),

404   the current result would only be strengthened because accounting for this change would

405   further reduce F_{SO}.

406   Further to force sharing among plantar-flexors, it must be considered that the net joint

407   moment includes an antagonistic contribution from TA that was ignored. This would only

408   cause a systematic underestimate of F_{SO} if TA activation were constant across conditions.

409   Unfortunately, this was not the case as introducing the spring increased TA activation (Table

410   1). However, this increase was small for the stance phase, which is when the key dependant

411   variables were evaluated, and TA was minimally active. The small increase in stance phase

412   activity in TA would not explain the large reductions in F_{SO} that were observed. A detailed

413   discussion of why TA's activation increased in the S condition has been previously published

414   (15).

415   *(e) Exoskeletons and injury Prevention*

416   Aside from reducing metabolic energy consumption, exoskeletons could be used to reduce

417   musculo-skeletal injury risk by unloading musculoskeletal structures. For example, chronic

418   joint or tendon conditions might be due to high volumes of repetitive loading such as

419   experienced during prolonged walking and running on a daily basis. Alternatively, acute

420   injuries may occur due to instances of excessive loading of tissues. Intuitively, one might

421   consider the reduced loading observed here to indicate a reduction in risk level for such

422   injuries. This may be true for certain stress-related injuries but, muscular injuries have been

423   more closely linked to excessive muscle strain (33). It was shown here that unloading SO

424   actually increased fascicle lengthening by 2-3 mm (Figure 3a) and fascicles reached a

425   maximum length of 43 mm. Average fascicle lengths during hopping were 38-40 mm and

426   this is similar to resting SO fascicle lengths previously reported (28). Taking 38 mm as a

427   resting length, the exoskeleton increased fascicle strain by 5-8% to a maximum of 8%. This is

428   still well below the 25% strain that has been reported as a damaging strain level (33).

429   Furthermore, the fact that individuals adjusted their ankle kinematics to reduce MTU

430   lengths for S, may have been a strategy to reduce passive muscle stretch early in stance and

431   help prevent muscle damage. The MTU was up to 19 mm longer without assistance. Had the

432   fascicle been required to provide that lengthening, it would have reached damaging strain

433   levels. This highlights the point that reducing musculoskeletal loading may not lead to

434   reduced muscle strain, especially when significant series elastic tissues are present.

435   *(f) Conclusions*

436   This study tested the effects of providing passive exoskeletal assistance to the human ankle

437   joint on soleus MTU mechanics and whole body net metabolic power during bilateral

438   hopping. The passive elastic assistance reduced net metabolic power. It also reduced soleus

439   force production (and rate of force) but, increased soleus fascicle excursion. This trade-off

440   maintained fascicle average positive mechanical power output despite the reduction in

441   force. These results highlighted that the metabolic cost of producing force may be more

442   important than cost of work for MTU's with compliant SEE's and short fascicles, during cyclic

443   contractions.

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447

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544

545

Figure Captions

547   Figure 1. (a) A sketch of the spring-loaded ankle exoskeleton design. LC is the load cell and

548   US the ultrasound transducer which was held in position with elastic bandaging. (b) A

549   sample ultrasound image with a schematic of the transducer placement (inset). The

550   transducer surface is at the top edge of the image. A soleus fascicle is highlighted (dashed

551   line) and pennation angle is marked (φ).

552   Figure 2. (a) Group mean normalised and smoothed soleus (SO) EMG signals and (b) SO

553   force, for the NE, NS and S conditions. The grey curve in (b) represents the force in the

554   spring of the exoskeleton for the S condition (S_{spring}) The shaded areas indicate the aerial

555   phases. The start of the aerial phase for S, NS and NE are indicated by each of the

556   progressively darker shades of grey, respectively. (c) Shows the group mean (± sem) average

557   rate of force production by SO for NE (dots), NS (hatched) and S (solid black). Data in a and b

558   are plotted over 101 points, normalised to time over an entire hop.

559   Figure 3. Group mean length changes of the soleus (a) fascicle, (b) SEE, and (c) MTU. The

560   shaded areas indicate the aerial phases. The start of the aerial phase for S, NS and NE are

561   indicated by each of the progressively darker shades of grey, respectively. All length changes

562   were relative to the length at the beginning of ground contact. Data are normalised to 101

563   points over a hop cycle.

564   Figure 4. Group mean (± sem) average positive powers for soleus (SO) fascicles (FAS) and

565   series elastic element (SEE) are shown individually and stacked (left vertical axis). The entire

566   stacked bars indicate total MTU average positive power and the percentage contribution of

567   the SEE is noted. Hatched bars are the group mean (± sem) whole body net metabolic

568   powers for NE, NS and S (right vertical axis). *indicates significant differences in $\overline{P}^+_{mech}$ and

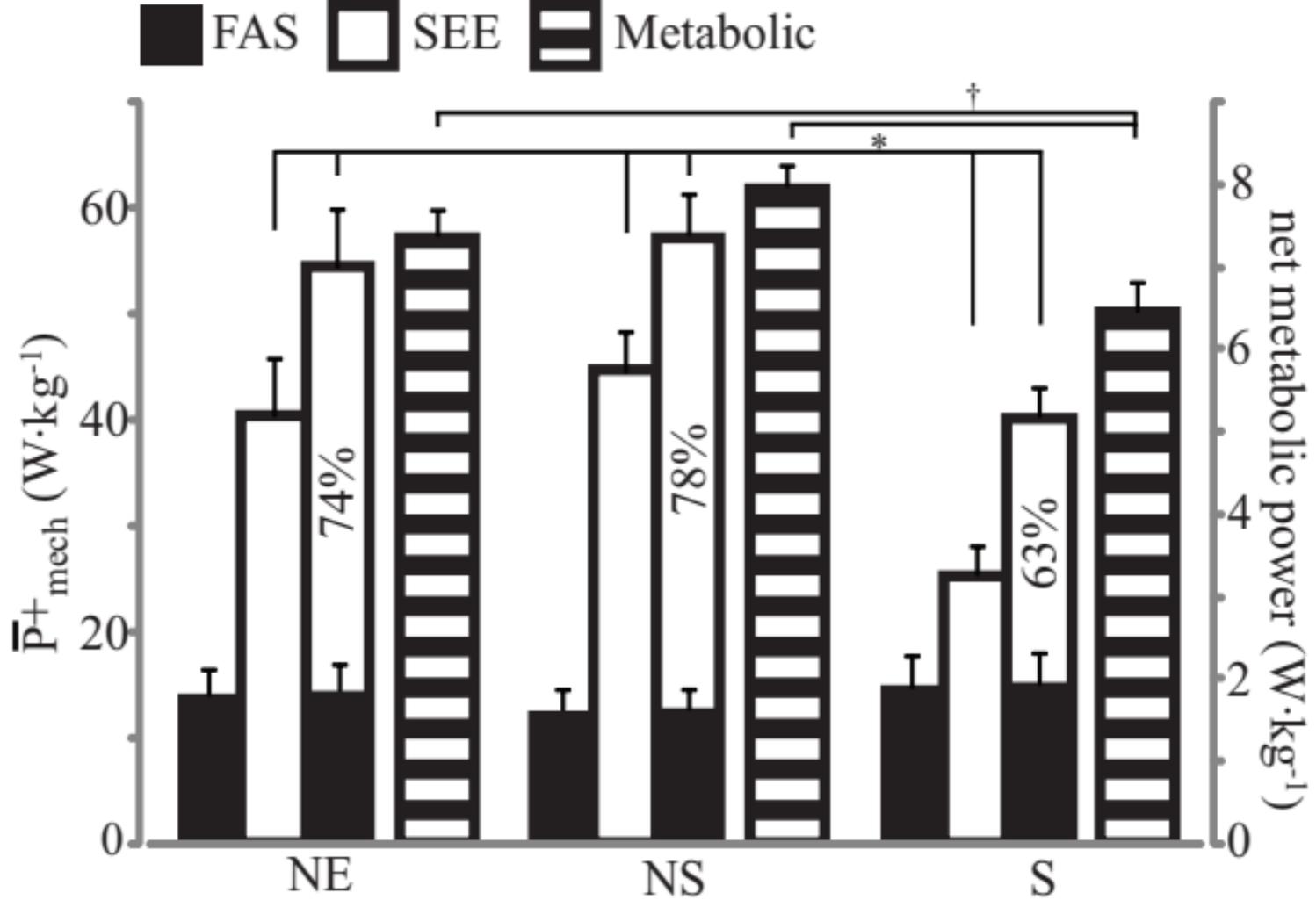569   † indicates significant differences in whole body net metabolic power.

a

b

Soleus

US

LC

φ
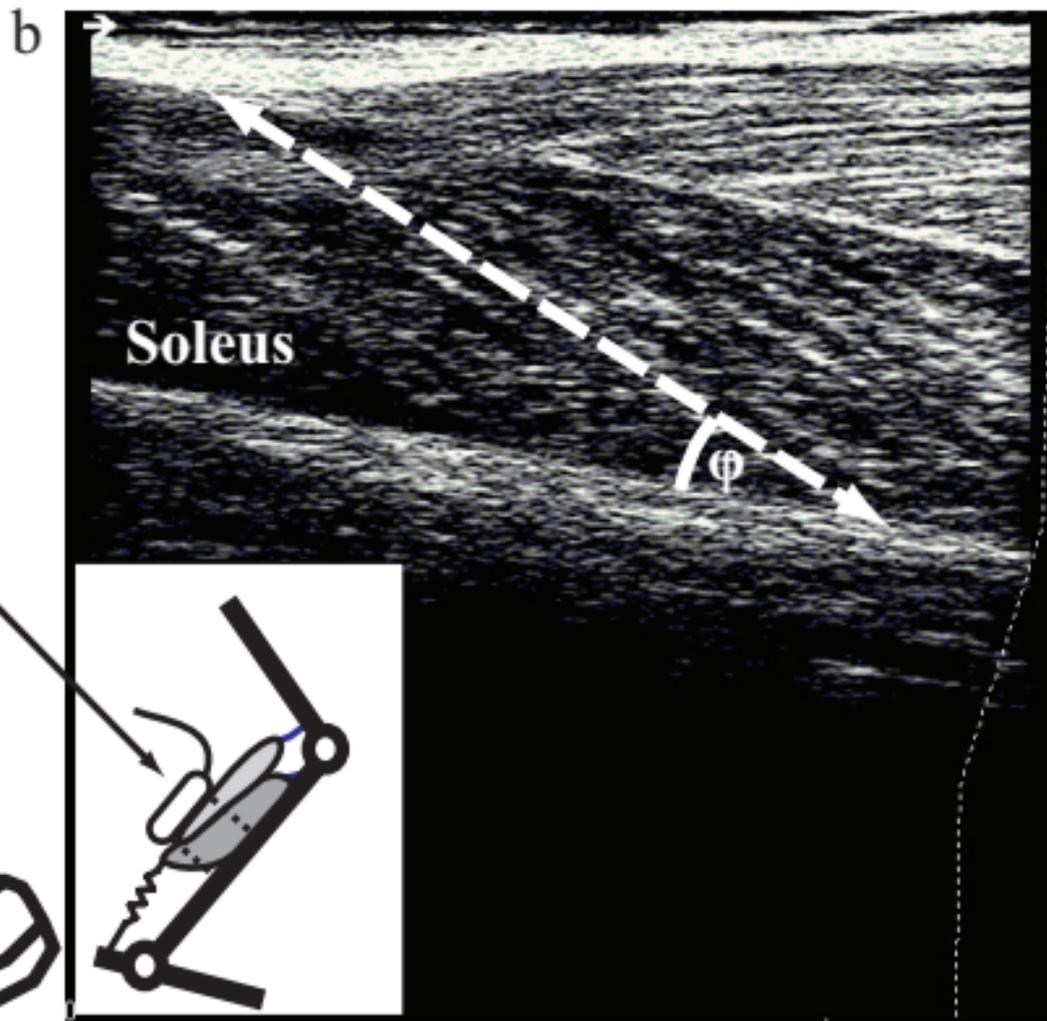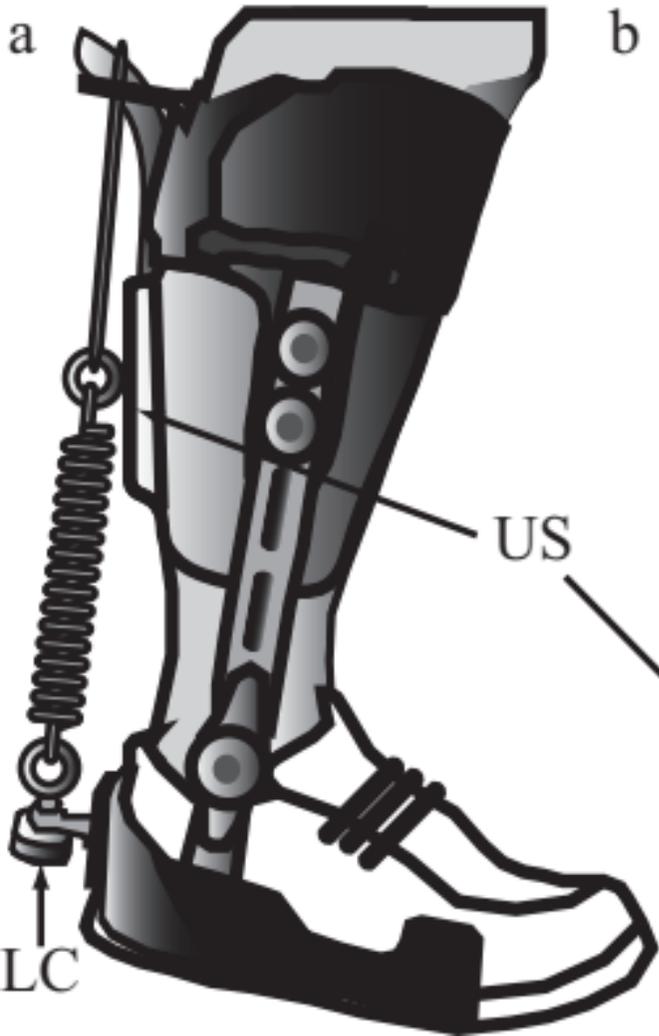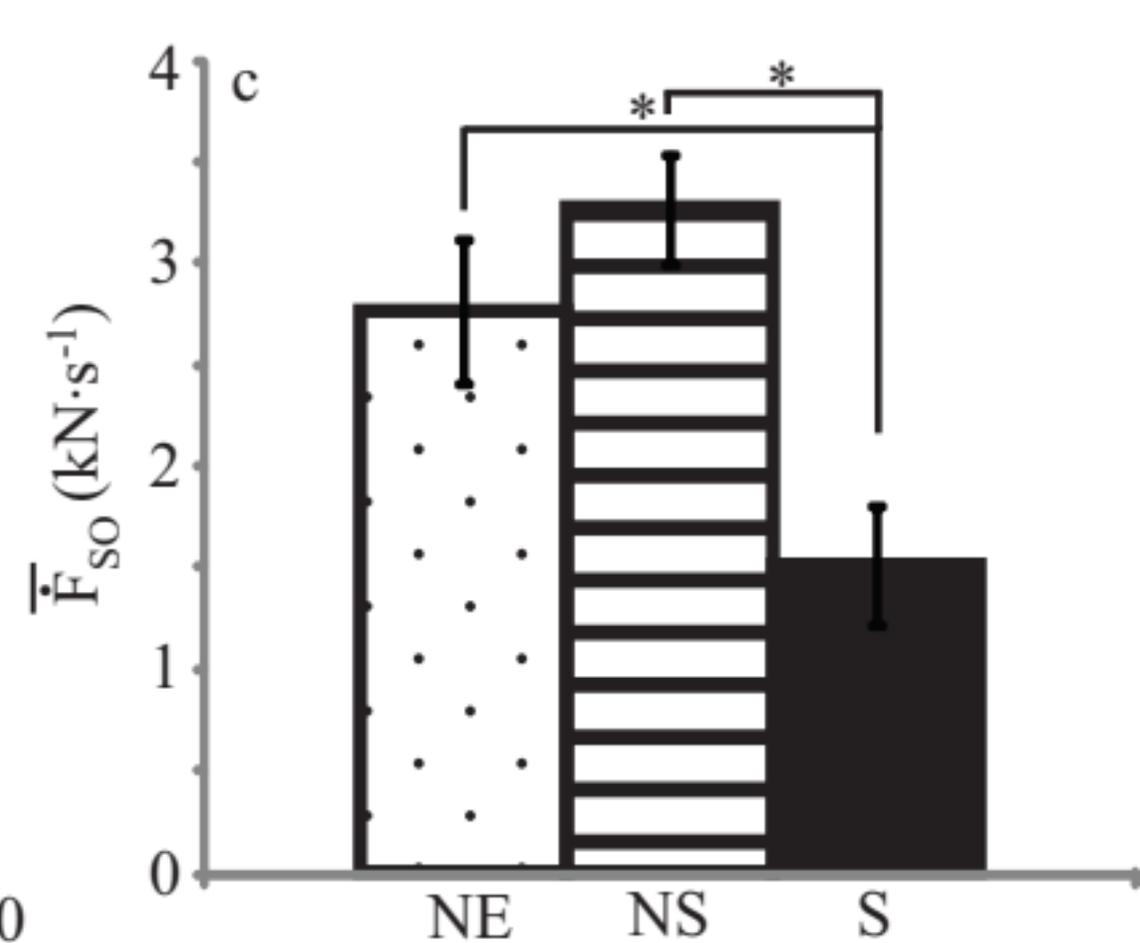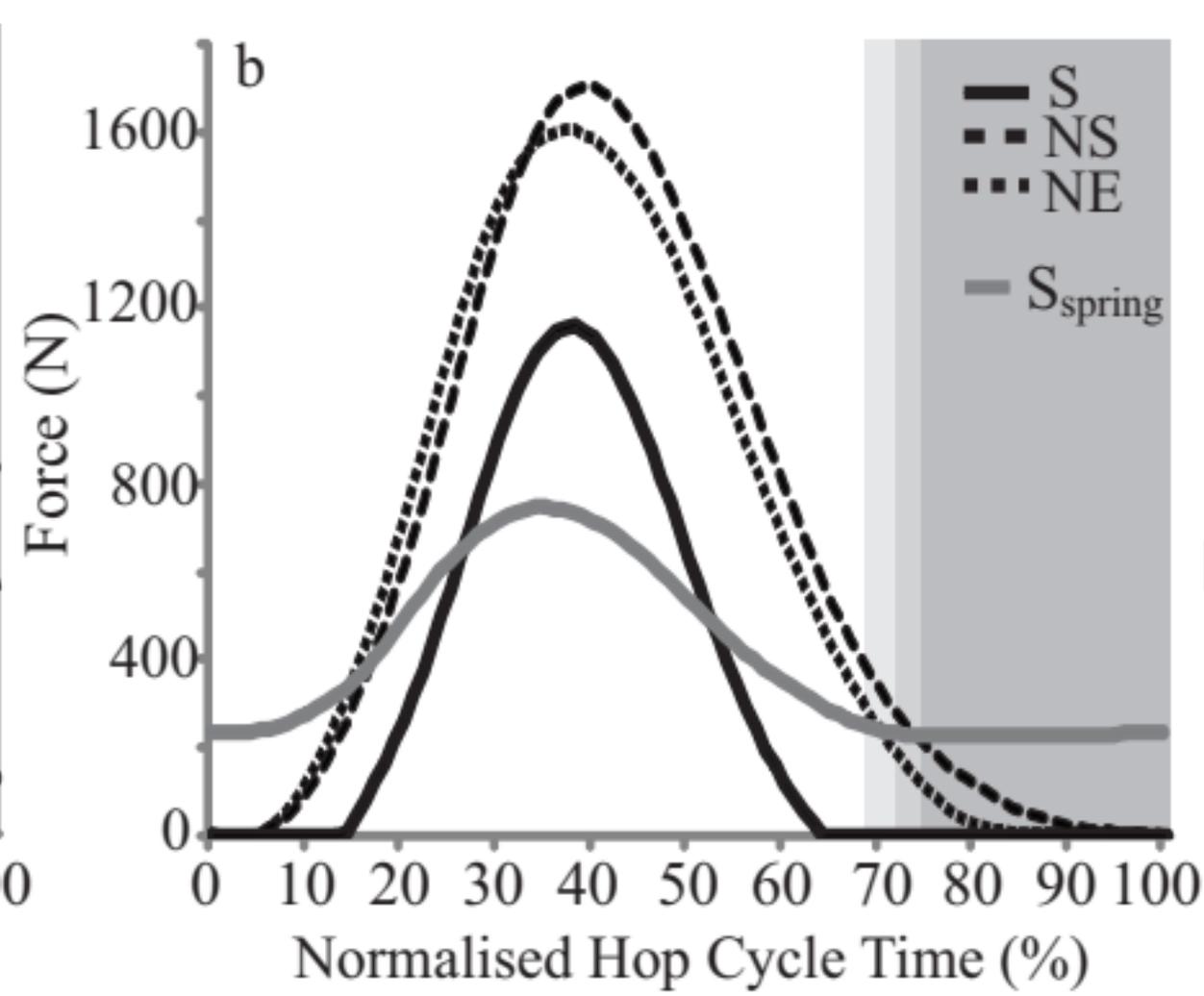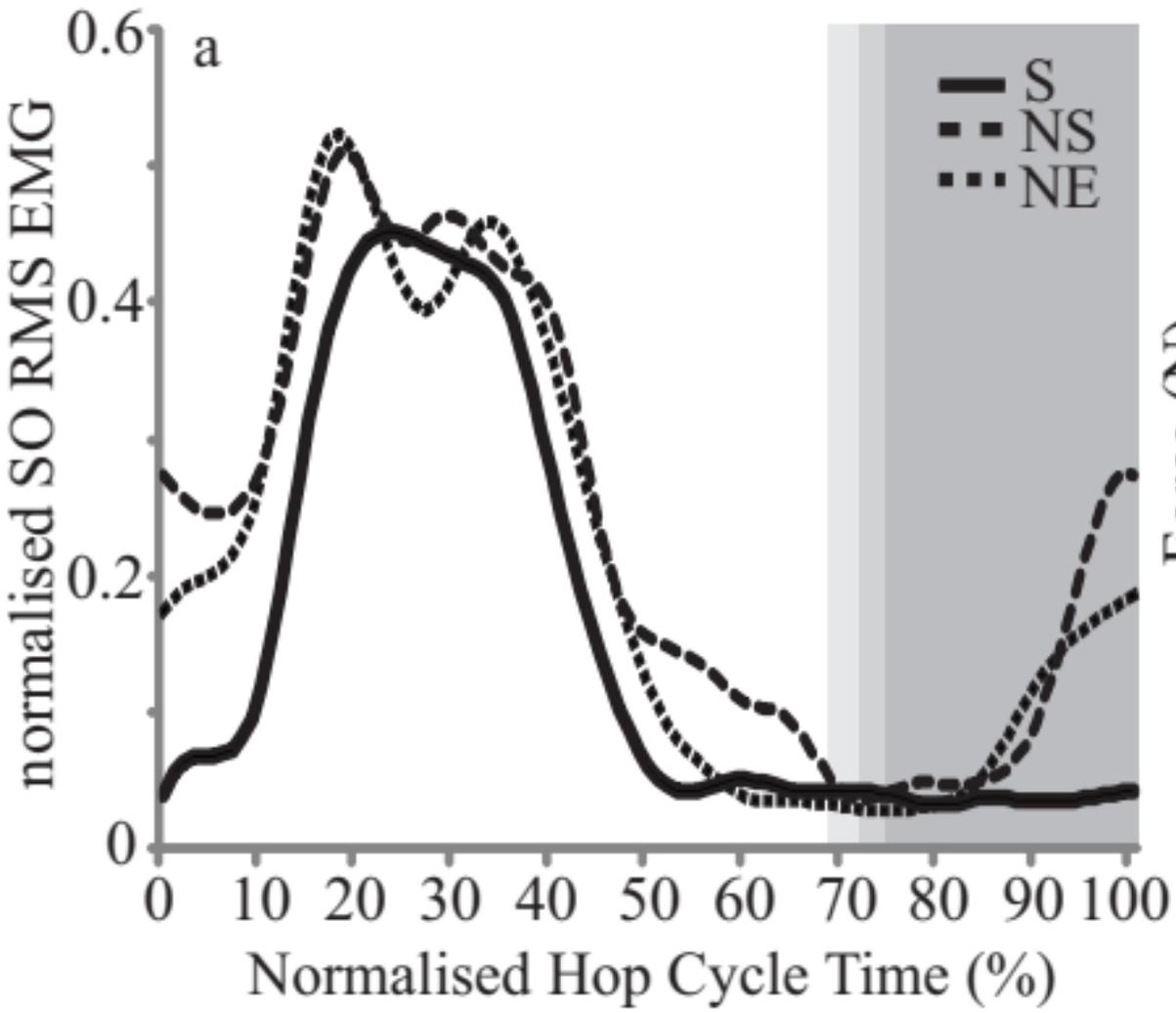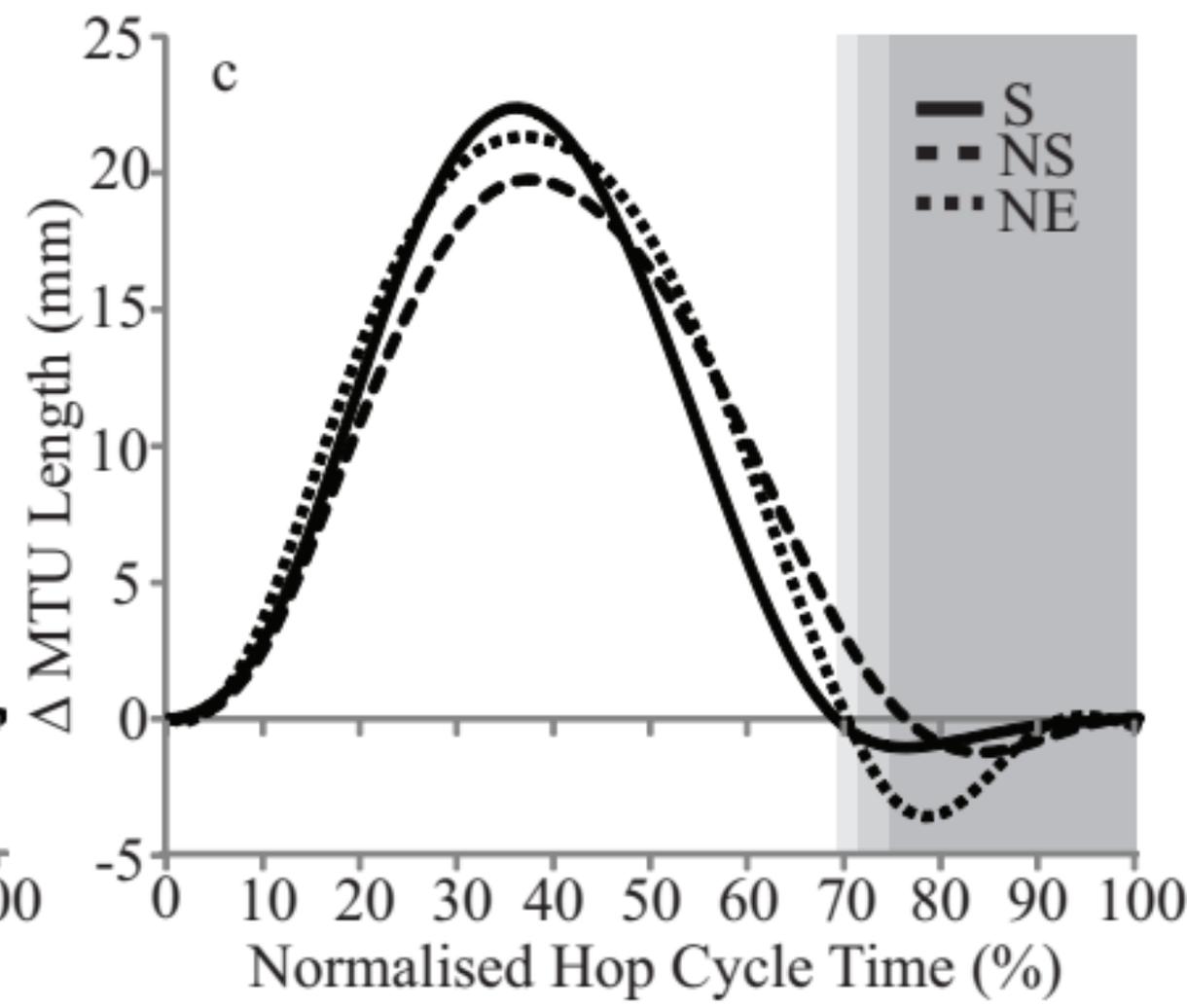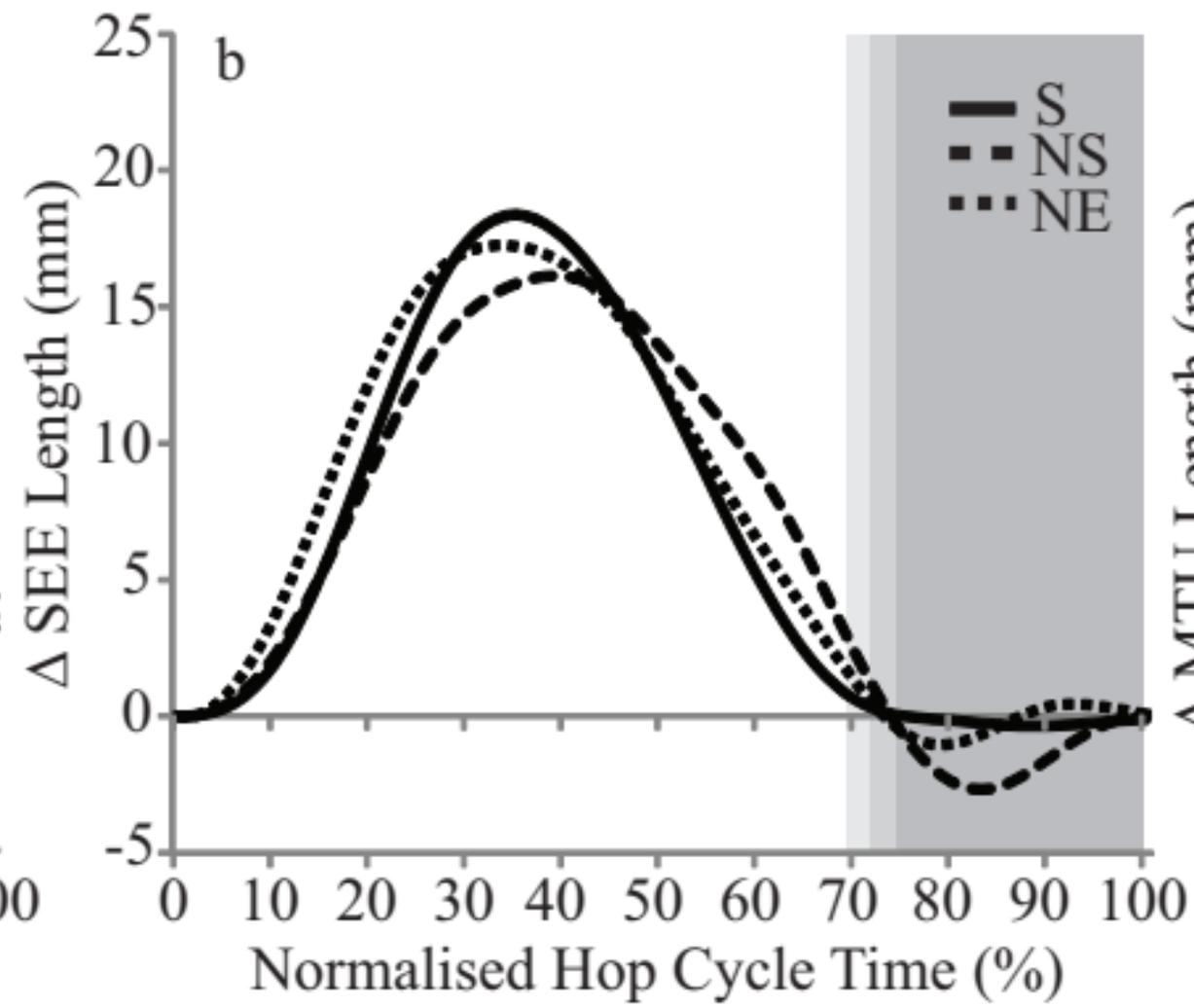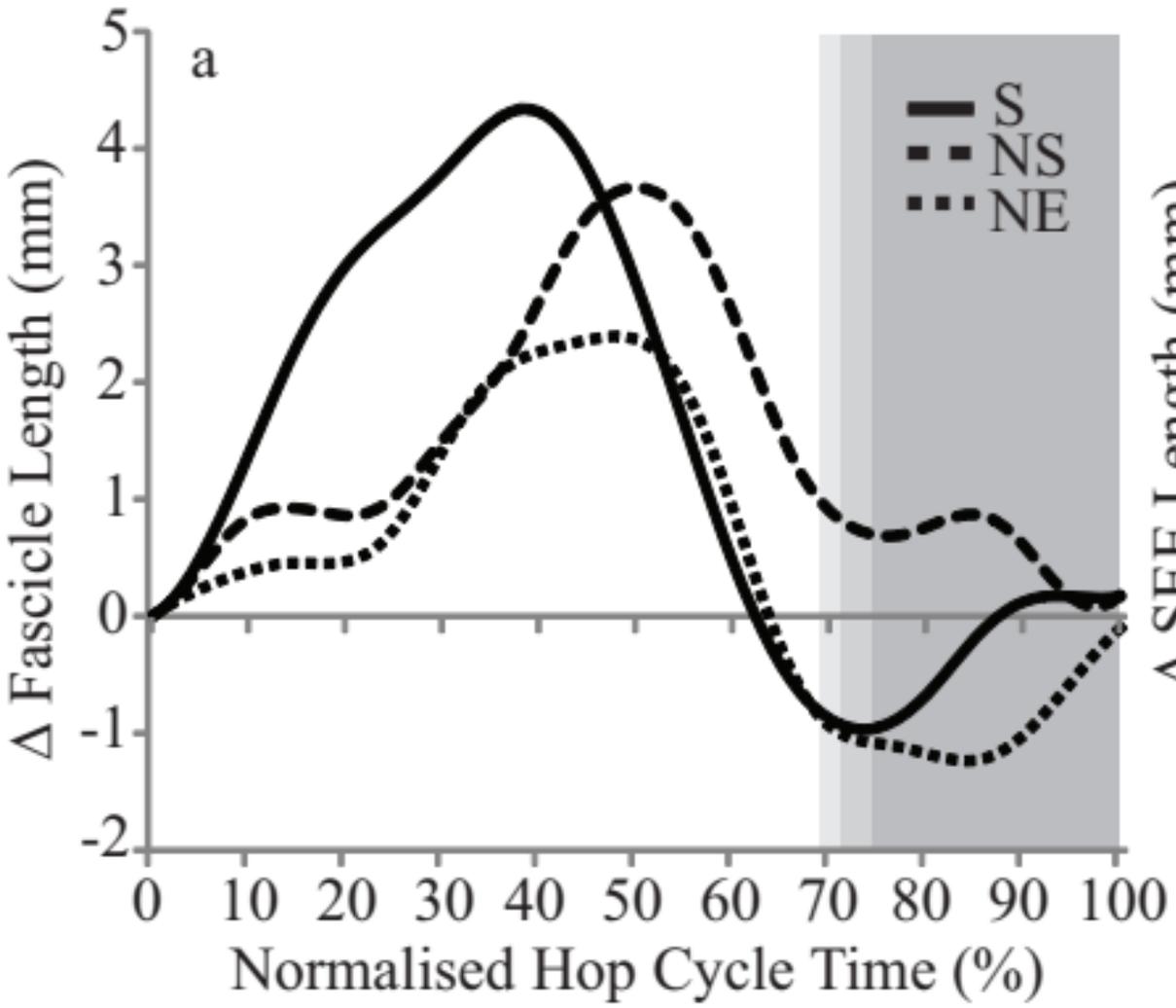
Table 1. Group mean (± sem) MTU metrics

	NE	NS	S
H (mm)	20 ± 8	13 ± 6	15 ± 9
Duty factor	0.72 ± 0.07	0.75 ± 0.06	0.69 ± 0.07
L_{SEE} (mm)	254 ± 9*	246 ± 8*	237 ± 7
L_{FAS} (mm)	39 ± 3	41 ± 4	38 ± 4
L_{MTU} (mm)	291 ± 8*	285 ± 8*	273 ± 6
ΔL_{FAS}^{+} (mm)	2 ± 1	4 ± 1	4 ± 1
ΔL_{FAS}^{-} (mm)	3 ± 1*	3 ± 1*	5 ± 1
ΔL_{TOT} (mm)	5 ± 1*	7 ± 1*	9 ± 1
Peak F_{SO} (N)	1623 ± 187*	1667 ± 119*	1166 ± 194
$\bar{\theta}_{ank}$ (°)	114 ± 1*	113 ± 2*	128 ± 2
TA_{rms} ground	1.07 ± 0.18*	1.00 ± 0.00*	1.42 ± 0.13
TA_{rms} aerial	1.07 ± 0.20*	1.00 ± 0.00*	2.84 ± 0.67
SO_{rms} ground	1.00 ± 0.06*	1.00 ± 0.00*	0.73 ± 0.06
SO_{rms} aerial	0.72 ± 0.10*	1.00 ± 0.00*	0.29 ± 0.10
MG_{rms} ground	1.05 ± 0.04	1.00 ± 0.00	1.05 ± 0.06
MG_{rms} aerial	1.02 ± 0.08*	1.00 ± 0.00*	0.40 ± 0.08
LG_{rms} ground	1.09 ± 0.02	1.00 ± 0.00	0.95 ± 0.20
LG_{rms} aerial	1.22 ± 0.20*	1.00 ± 0.00*	0.49 ± 0.20

H – Hop height. L_{SEE}, L_{FAS}, L_{MTU}- average lengths of the SEE, fascicle & MTU. ΔL_{FAS}^{+}, ΔL_{FAS}^{-}, ΔL_{TOT} - lengthening & shortening of fascicle during ground contact. $\bar{\theta}_{ank}$ – mean ankle angle during ground contact. $_{rms}$ground and $_{rms}$aerial are the root mean square EMG values from the stance and aerial phases for each muscle. *denotes statistically significant (P<0.05) difference from the S condition.