

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

2 **Authors:** Dominic James Farris<sup>1</sup>, Benjamin D. Robertson<sup>1</sup> & Gregory S. Sawicki<sup>1</sup>

3 <sup>1</sup>Joint Department of Biomedical Engineering, University of North Carolina-Chapel Hill &  
4 North Carolina State University, EBIII, 911 Oval Drive, North Carolina State University,  
5 Raleigh, NC 27695-7115

6 **Corresponding author:** Dominic James Farris, School of Human Movement Studies, The  
7 University of Queensland, St Lucia, QLD 4072, Australia. Email: [d.farris@uq.edu.au](mailto:d.farris@uq.edu.au)

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.

26

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

28

29

30

## 31 **Introduction**

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.

## 100 **Methods**

### 101 *(a) Participants*

102 Seven male participants (mean  $\pm$  sd, age =  $28 \pm 7$ , height =  $1.8 \pm 0.06$  m, mass =  $80 \pm 10$  kg)  
103 gave written informed consent to participate in this study. All participants were in good  
104 health and had no recent history of lower limb musculo-skeletal injury. All procedures were  
105 approved by an institutional review board and complied with the guidelines for research  
106 involving human participants as set out in the Declaration of Helsinki.

### 107 *(b) Experimental Protocol*

108 Bilateral hopping is a bouncing gait with similar spring-mass mechanics to running but  
109 simpler kinematics. The plantar-flexors undergo a stretch-shortening motion, meaning a  
110 simple spring-loaded ankle exoskeleton can be used to provide parallel assistance to the  
111 plantar-flexors. Therefore, each participant was required to perform bilateral hopping in  
112 place in time with the beat of a metronome operating at 2.5 Hz. This frequency was chosen  
113 because a previous study using the same movement found that this was the frequency  
114 around which metabolic cost was minimised when hopping in spring-loaded exoskeletons  
115 (25). The hopping task was performed for 4 minutes to allow the participants to reach a  
116 metabolic steady state and was completed under 3 experimental conditions: (1) With no  
117 exoskeleton (NE); (2) With bilateral ankle exoskeletons but no spring (NS); (3) With spring-  
118 loaded exoskeletons to assist plantar-flexion (S).

### 119 *(c) The Exoskeletons*

120 The devices have been previously described elsewhere (15) and a sketch of the exoskeletons  
121 used is shown in Figure 1. The exoskeleton consisted of a carbon fibre cuff around the upper

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  $127^\circ$   
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  $5 \text{ kNm}^{-1}$  and its moment  
133 arm about the joints was 0.135 m. This gave a rotational stiffness of  $1.59 \text{ Nm}/^\circ$  ( $91 \text{ Nm}\cdot\text{rad}^{-1}$ )  
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™  
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 ( $\Delta L_{FAS}^+$ ) was calculated relative to  $L_i$  by subtracting  $L_i$  from the peak length  
181 during stance. Fascicle shortening ( $\Delta L_{FAS}^-$ ) was calculated as the length at take-off minus the  
182 peak length during stance. Overall length change ( $\Delta L$ ) was the sum of the absolute values of  
183  $\Delta L_{FAS}^+$  and  $\Delta L_{FAS}^-$ .

#### 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 ( $\bar{F}_{SO}$ ) 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 ( $\bar{P}_{FAS}^+$ ), SEE ( $\bar{P}_{SEE}^+$ ) and MTU ( $\bar{P}_{MTU}^+$ ).  
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  $\bar{P}_{FAS}^+$  was zero  
209 (i.e. the fascicle is always isometric) and all of  $\bar{P}_{MTU}^+$  was supplied by  $\bar{P}_{SEE}^+$  (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  $\pm$  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  $\bar{P}_{FAS}^+$  was unchanged between conditions (Figure 4). However, both  $\bar{P}_{SEE}^+$  and  $\bar{P}_{MTU}^+$   
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  $\bar{P}_{FAS}^+$  to  $\bar{P}_{SEE}^+$  went down in S, indicating that a smaller proportion of  
266  $\bar{P}_{MTU}^+$  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  $\bar{P}_{FAS}^+$ . 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  $SO_{RMSground}$  ( $\approx 30\%$ ). 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 ( $\approx 30\%$ ) 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  $\approx 50\%$  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 ( $\approx 50\%$ ) 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 ( $\bar{P}_{FAS}^+$ ).

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  $\approx 54\%$  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 ( $\approx 60\%$  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 ( $-0.2 \text{ W}\cdot\text{kg}^{-1}$ ) was much less than at the ankle ( $-1.0 \text{ W}\cdot\text{kg}^{-1}$ ). 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  $0.8 \text{ W}\cdot\text{kg}^{-1}$  of the  $1.5 \text{ W}\cdot\text{kg}^{-1}$  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 ( $\sim 0.7 \text{ W}\cdot\text{kg}^{-1}$ ) 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.

#### 444 **Acknowledgements**

445 The current affiliation for Dominic James Farris is – School of Human Movement Studies,  
446 The University of Queensland, St Lucia, QLD 4072, Australia.

447

#### 448 **Grants**

449 This study was in part funded by US Israel Binational Science Foundation Start up Grant  
450 #2011152 awarded to GSS

451 **References**

- 452 1. **Aggeloussis N, Giannakou E, Albracht K, and Arampatzis A.** Reproducibility of fascicle length  
453 and pennation angle of gastrocnemius medialis in human gait in vivo. *Gait & Posture* 31: 73-77,  
454 2010.
- 455 2. **Alexander R.** *Elastic mechanisms in animal movement.* Cambridge: Cambridge University  
456 Press, 1988.
- 457 3. **Bartonek A, Eriksson M, and Gutierrez-Farewik EM.** Effects of carbon fibre spring orthoses  
458 on gait in ambulatory children with motor disorders and plantarflexor weakness. *Dev Med Child*  
459 *Neurol* 49: 615-620, 2007.
- 460 4. **Biewener AA.** Biomechanics of mammalian terrestrial locomotion. *Science* 250: 1097-1103,  
461 1990.
- 462 5. **Blickhan R.** The spring mass model for running and hopping. *Journal of Biomechanics* 22:  
463 1217-1227, 1989.
- 464 6. **Bobbert MF, Huijing PA, and Schenau GJV.** A model of the human triceps surae muscle-  
465 tendon complex applied to jumping. *Journal of Biomechanics* 19: 887-898, 1986.
- 466 7. **Bregman DJ, Harlaar J, Meskers CGM, and de Groot V.** Spring-like ankle foot orthoses  
467 reduce the energy cost of walking by taking over ankle work. *Gait & Posture* 35: 148-153, 2012.
- 468 8. **Brockway JM.** Derivation of formulas used to calculate energy-expenditure in man. *Human*  
469 *Nutrition-Clinical Nutrition* 41C: 463-471, 1987.
- 470 9. **Cavagna GA.** Storage and utilization of elastic energy in skeletal muscle. *Exercise and Sport*  
471 *Sciences Reviews* 5: 89-129, 1977.
- 472 10. **Cavagna GA, and Kaneko M.** Mechanical work and efficiency in level walking and running.  
473 *Journal of Physiology* 268: 467-481, 1977.
- 474 11. **Chang YH, Roiz RA, and Auyang AG.** Intralimb compensation strategy depends on the nature  
475 of joint perturbation in human hopping. *Journal of Biomechanics* 41: 1832-1839, 2008.
- 476 12. **Dempster AD.** Space requirements of the seated operator Ohio: Wright-Patterson Air Force  
477 Base, 1955.
- 478 13. **Donelan JM, Kram R, and Kuo AD.** Mechanical work for step-to-step transitions is a major  
479 determinant of the metabolic cost of human walking. *Journal of Experimental Biology* 205: 3717-  
480 3727, 2002.
- 481 14. **Farris DJ, and Sawicki GS.** Human medial gastrocnemius force-velocity behavior shifts with  
482 locomotion speed and gait. *Proceedings of the National Academy of Sciences* 109: 977-982, 2012.
- 483 15. **Farris DJ, and Sawicki GS.** Linking the mechanics and energetics of hopping with elastic ankle  
484 exoskeletons. *J Appl Physiol* 113: 1862-1872, 2012.
- 485 16. **Fenn WO, and Marsh BS.** Muscular force at different speeds of shortening. *Journal of*  
486 *Physiology-London* 85: 277-297, 1935.
- 487 17. **Ferris DP, Bohra ZA, Lukos JR, and Kinnaird CR.** Neuromechanical adaptation to hopping  
488 with an elastic ankle-foot orthosis. *Journal of Applied Physiology* 100: 163-170, 2006.
- 489 18. **Finni T.** Structural and functional features of human muscle-tendon unit. *Scandinavian*  
490 *journal of medicine and science in sports* 2005.
- 491 19. **Fukunaga T, Ito M, Ichinose Y, Kuno S, Kawakami Y, and Fukashiro S.** Tendinous movement  
492 of a human muscle during voluntary contractions determined by real-time ultrasonography. *Journal*  
493 *of Applied Physiology* 81: 1430-1433, 1996.
- 494 20. **Fukunaga T, Kawakami Y, Kubo K, and Kanehisa H.** Muscle and tendon interaction during  
495 human movements. *Exercise and Sport Sciences Reviews* 30: 106-110, 2002.
- 496 21. **Fukunaga T, Kawakami Y, Kuno S, Funato K, and Fukashiro S.** Muscle architecture and  
497 function in humans. *Journal of Biomechanics* 30: 457-463, 1997.
- 498 22. **Fukunaga T, Kubo K, Kawakami Y, Fukashiro S, Kanehisa H, and Maganaris CN.** In vivo  
499 behaviour of human muscle tendon during walking. *Proceedings of the Royal Society of London*  
500 *Series B-Biological Sciences* 268: 229-233, 2001.

- 501 23. **Fukunaga T, Roy RR, Shellock FG, Hodgson JA, Day MK, Lee PL, Kwongfu H, and Edgerton**  
502 **VR.** Physiological cross-sectional area of human leg muscles based on magnetic-resonance-imaging.  
503 *Journal of Orthopaedic Research* 10: 926-934, 1992.
- 504 24. **Grabowski AM.** Metabolic and biomechanical effects of velocity and weight support using a  
505 lower-body positive pressure device during walking. *Archives of Physical Medicine and Rehabilitation*  
506 91: 951-957, 2010.
- 507 25. **Grabowski AM, and Herr HM.** Leg exoskeleton reduces the metabolic cost of human  
508 hopping. *Journal of Applied Physiology* 107: 670-678, 2009.
- 509 26. **Hawkins D, and Hull ML.** A method for determining lower extremity muscle tendon lengths  
510 during flexion/extension movements. *Journal of Biomechanics* 23: 487-494, 1990.
- 511 27. **Ishikawa M, Komi PV, Grey MJ, Lepola V, and Bruggemann G.** Muscle-tendon interaction  
512 and elastic energy usage in human walking. *Journal of Applied Physiology* 99: 603-608, 2005.
- 513 28. **Kawakami Y, Ichinose Y, and Fukunaga T.** Architectural and functional features of human  
514 triceps surae muscles during contraction. *Journal of Applied Physiology* 85: 398-404, 1998.
- 515 29. **Lichtwark GA, Bougoulias K, and Wilson AM.** Muscle fascicle and series elastic element  
516 length changes along the length of the human gastrocnemius during walking and running. *Journal of*  
517 *Biomechanics* 40: 157 - 164, 2007.
- 518 30. **Lichtwark GA, and Wilson AM.** In vivo mechanical properties of the human achilles tendon  
519 during one-legged hopping. *Journal of Experimental Biology* 208: 4715-4725, 2005.
- 520 31. **Lichtwark GA, and Wilson AM.** Is achilles tendon compliance optimised for maximal muscle  
521 efficiency during locomotion? *Journal of Biomechanics* 40: 1768-1775, 2007.
- 522 32. **Lichtwark GA, and Wilson AM.** Optimal muscle fascicle length and tendon stiffness for  
523 maximising gastrocnemius efficiency during human walking and running. *Journal of Theoretical*  
524 *Biology* 252: 662-673, 2008.
- 525 33. **Lieber RL, and Friden J.** Muscle damage is not a function of muscle force but active muscle  
526 strain. *Journal of Applied Physiology* 74: 520-526, 1993.
- 527 34. **Magnusson SP, Aagaard P, Rosager S, Dyhre-Poulsen P, and Kjaer M.** Load-displacement  
528 properties of the human triceps surae aponeurosis in vivo. *Journal of Physiology* 531: 277-288, 2001.
- 529 35. **McMahon TA, and Cheng GC.** The mechanics of running - how does stiffness couple with  
530 speed. *Journal of Biomechanics* 23: 65-78, 1990.
- 531 36. **Reeves ND, Maganaris CN, and Narici MV.** Ultrasonographic assessment of human skeletal  
532 muscle size. *European Journal of Applied Physiology* 91: 116-118, 2004.
- 533 37. **Roberts TJ.** The integrated function of muscles and tendons during locomotion. *Comparative*  
534 *Biochemistry and Physiology part A* 133: 1087-1099, 2002.
- 535 38. **Sawicki GS, and Ferris DP.** Mechanics and energetics of level walking with powered ankle  
536 exoskeletons. *Journal of Experimental Biology* 211: 1402-1413, 2008.
- 537 39. **Sawicki GS, Lewis CL, and Ferris DP.** It pays to have a spring in your step. *Exercise and Sport*  
538 *Sciences Reviews* 37: 130-138, 2009.
- 539 40. **Wiggin MB, Collins SH, and Sawicki GS.** An exoskeleton using controlled energy storage and  
540 release to aid ankle propulsion. In: *IEEE international conference on rehabilitation robotics*. Zurich,  
541 Switzerland: IEEE, 2011, p. 1-5.
- 542 41. **Winter DA.** Moments of force and mechanical power in jogging. *Journal of Biomechanics* 16:  
543 91-97, 1983.

544

545

546 **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 ( $\varphi$ ).

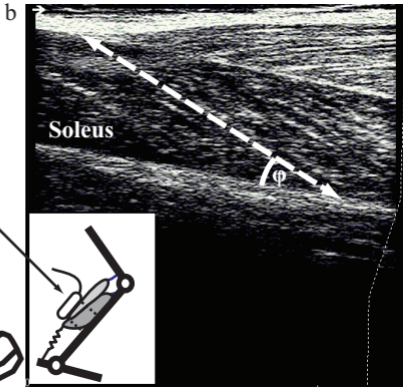
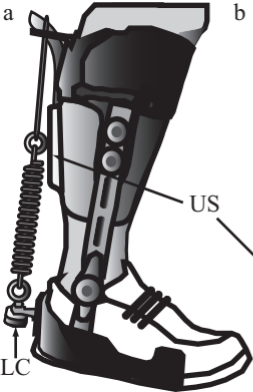
552 Figure 2. (a) Group mean normalised and smoothed soleus (SO) EMG signals and (b) SO  
553 force, for the NE, NS and S conditions. The grey curve in (b) represents the force in the  
554 spring of the exoskeleton for the S condition ( $S_{spring}$ ). The shaded areas indicate the aerial  
555 phases. 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 ( $\pm$  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.

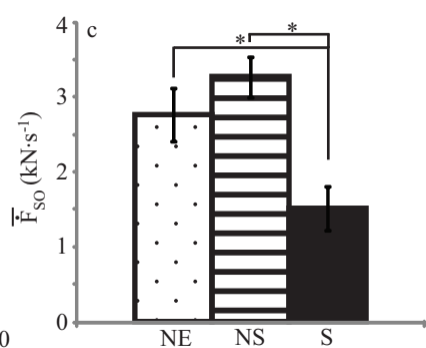
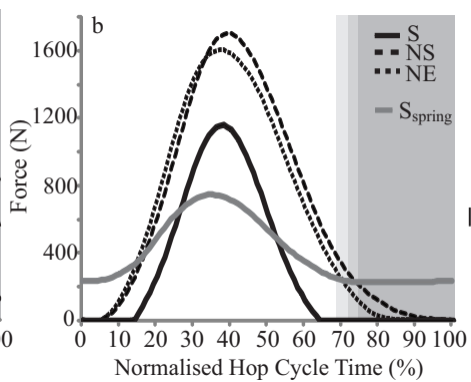
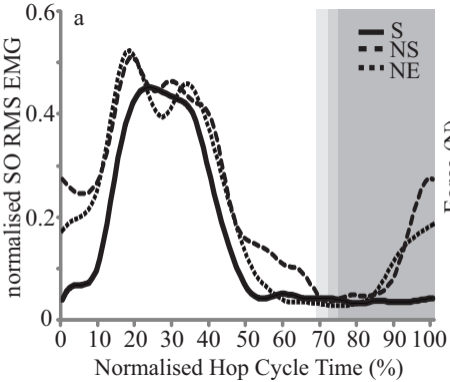
559 Figure 3. Group mean length changes of the soleus (a) fascicle, (b) SEE, and (c) MTU. The  
560 shaded areas indicate the aerial phases. The start of the aerial phase for S, NS and NE are  
561 indicated by each of the progressively darker shades of grey, respectively. All length changes  
562 were relative to the length at the beginning of ground contact. Data are normalised to 101  
563 points over a hop cycle.

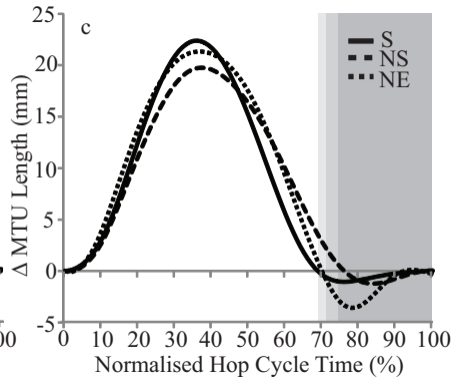
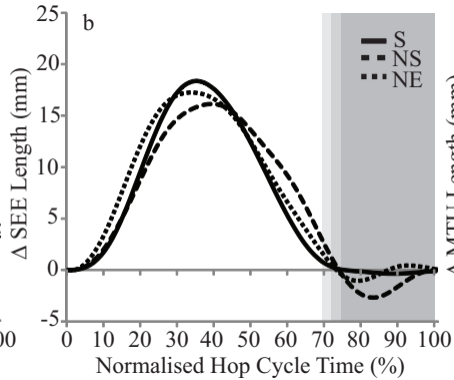
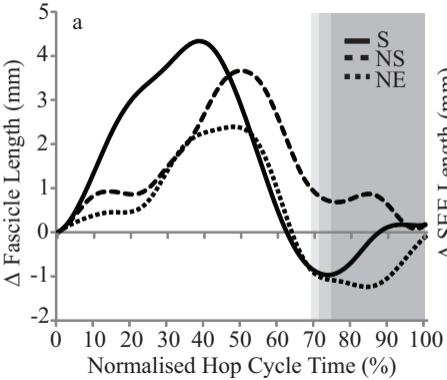
564 Figure 4. Group mean ( $\pm$  sem) average positive powers for soleus (SO) fascicles (FAS) and  
565 series elastic element (SEE) are shown individually and stacked (left vertical axis). The entire  
566 stacked bars indicate total MTU average positive power and the percentage contribution of  
567 the SEE is noted. Hatched bars are the group mean ( $\pm$  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.







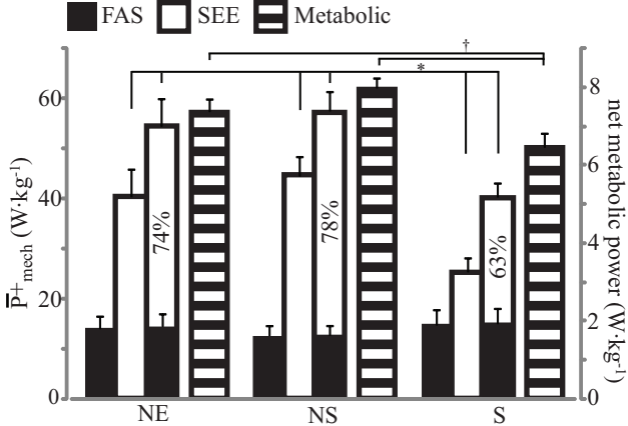


Table 1. Group mean ( $\pm$  sem) MTU metrics

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

H – Hop height. L $_{SEE}$ , L $_{FAS}$ , L $_{MTU}$  - average lengths of the SEE, fascicle & MTU.  $\Delta L_{FAS}^+$ ,  $\Delta L_{FAS}^-$ ,  $\Delta L_{TOT}$  - 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.