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Energy flow analysis of amputee walking shows a proximally-directed transfer of energy in intact limbs, compared to a distally-directed transfer in prosthetic limbs at push-off

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2 energy in intact limbs, compared to a distally-directed transfer in prosthetic limbs at push-off.

3

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22 Keywords: Power; gait; energy exchange; prosthesis

23

24

25 Abstract

26 Reduced capacity and increased metabolic cost of walking occurs in amputees, despite
27 advances in prosthetic componentry. Joint powers can quantify deficiencies in prosthetic gait,
28 but do not reveal how energy is exchanged between limb segments. This study aimed to
29 quantify these energy exchanges during amputee walking.

30 Optical motion and forceplate data collected during walking at a self-selected speed for
31 cohorts of 10 controls, 10 unilateral trans-tibial, 10 unilateral trans-femoral and 10 bilateral
32 trans-femoral amputees were used to determine the energy exchanges between lower limb
33 segments.

34 At push-off, consistent thigh and shank segment powers were observed between amputee
35 groups (1.12W/kg vs. 1.05W/kg for intact limbs and 0.97W/kg vs. 0.99W/kg for prosthetic
36 limbs), and reduced prosthetic ankle power, particularly in trans-femoral amputees
37 (3.12W/kg vs. 0.87W/kg). Proximally-directed energy exchange was observed in the intact
38 limbs of amputees and controls, while prosthetic limbs displayed distally-directed energy
39 exchanges at the knee and hip.

40 This study used energy flow analysis to show a reversal in the direction in which energy is
41 exchanged between prosthetic limb segments at push-off. This reversal was required to
42 provide sufficient energy to propel the limb segments and is likely a direct result of the lack
43 of push-off power at the prosthetic ankle, particularly in trans-femoral amputees, and leads to
44 their increased metabolic cost of walking.

45

46 Introduction¹

47 Despite advances in prosthetic lower limbs, amputees are still known to walk with
48 increased metabolic costs compared to able-bodied individuals, and with increasing
49 metabolic cost as the level of amputation becomes more proximal or when bilateral
50 amputation occurs [1-4]. To better understand why this may be the case, studies have
51 investigated the kinematics of lower limb amputees [5] and have consistently found reduced
52 knee flexion during weight-acceptance and reduced ankle plantar-flexion during late stance.
53 Recently however, studies have focussed on the kinetics and muscular activity of amputee
54 gait to provide a more complete picture of the biomechanics of the limbs and trunk during
55 amputee walking. Studies assessing the effect of different prosthetic components [6-8] and of
56 amputation level [9-14] during amputee gait have led to consistent findings of reduced peak
57 ankle plantar-flexion moment and power and increased peak hip power generation and
58 absorption in amputees. This has led to several avenues of research, particularly the design
59 and development of active (powered) prosthetic limbs [15-17]. However, given the majority
60 of amputees use passive prosthetic limbs, understanding how these devices interact with the
61 body during locomotion should remain a priority and may lead to improved passive devices
62 with better energy storage and return characteristics, perhaps utilising intelligent control.

63 An efficient gait will likely be dependent on energy conserving exchanges between
64 limb segments and also on energy storage and return mechanisms, typically utilising strain
65 energy in tendons and prosthetic components. In unilateral amputees, it is known that the
66 intact limb often compensates for deficiencies on the prosthetic side, which leads to
67 characteristic gait asymmetries of reduced stance time and increased swing time and step
68 length on the prosthetic side [10, 11]. However, despite these asymmetries, fit individuals

¹ Abbreviations: BTF – Bilateral Trans-Femoral, Con – Control, DMRC – Defence Medical Rehabilitation Centre, DoF – Degree of Freedom, ESR – Energy Storage and Return, JFP - Joint Force Power, STP - Segment Torque Power, UTF – Unilateral Trans-Femoral, UTT – Unilateral Trans-Tibial

69 with a trans-tibial amputation as a result of trauma often have a metabolic cost of walking
70 that is close to that of healthy able-bodied controls [4, 18], suggesting that, in certain cases at
71 least, it is possible to overcome the deficiencies associated with the loss of limb. While
72 unilateral amputees are able to compensate with their intact limb, this is not possible in
73 bilateral amputees, who are known to have a significantly increased metabolic cost of
74 walking [4, 19, 20].

75 While standard gait analysis techniques have been able to identify joint-level
76 differences between amputee and able-bodied gait, it remains unclear what impact the
77 inability to produce active ankle power has on the way in which energy is transmitted through
78 the limb as a whole. Quantifying the energy exchanges that occur between limb segments in
79 amputee gait may result in a better understanding of the underlying causes of inefficient gait
80 in unilateral and bilateral amputees, as this would provide a more complete picture of lower
81 limb amputee biomechanics during walking. Such an approach has previously been used to
82 assess the energy exchanges in the lower limbs in both healthy and pathological gait [21-26]
83 and in trans-tibial amputees to assess energy exchanges at the ankle [8, 27]. However, whole
84 limb energy flow analyses of trans-femoral and trans-tibial amputee populations have not
85 been previously performed, which has previously limited our ability to characterise amputee
86 gait to joint-level measures. Therefore, it was the aim of this study to investigate how
87 reduced ankle push-off power alters lower limb energy flows during amputee walking. The
88 hypothesis of this study is that changes in energy flows in the lower limbs of amputees during
89 walking can help to explain the substantial increases in metabolic demands commonly
90 reported for this population.

91

92 Materials and Methods

93 **Subject Details and Protocol**

94 In a previously published study [4], 30 amputees were recruited to form three cohorts
 95 of 10 unilateral trans-tibial (UTT), 10 unilateral trans-femoral (UTF) and 10 bilateral trans-
 96 femoral (BTF) amputees (Table 1). Study inclusion criteria were: Aged 18 to 40, amputation
 97 as a result of lower limb trauma, attending Defence Medical Rehabilitation Centre (DMRC)
 98 Headley Court for routine prosthetic appointment, at least 6 months after fitting of definitive
 99 prosthesis, no pain consequent to prosthetic fitting or alignment (minor “discomfort” was
 100 acceptable) and capable of walking comfortably for 10 minutes continuously. Study
 101 exclusion criteria were any neuromusculoskeletal pathology (aside from the amputated limb)
 102 which would likely affect the participants’ walking. All amputees were fitted with energy
 103 storage and return (ESR) feet, trans-femoral amputees with micro-processor knees (Table 2),
 104 and had undergone similar rehabilitation regimes at Headley Court. 10 healthy military
 105 personnel were also recruited to provide age- and height-matched control data for
 106 comparative purposes.

107

108 **Table 1:** Participant demographic information.

Groups	Mass [kg]	Height [m]	Age [years]
Control	78.0 (7.6)	1.82 (0.05)	30 (6)
UTT	89.8 (14.3)	1.82 (0.05)	28 (4)
UTF	88.3 (6.5)	1.80 (0.07)	29 (3)
BTF	86.7 (19.2)	1.81 (0.08)	29 (4)

109 Note: values are presented as mean (s.d.)

110

Level	Socket type	Socket interface	Suspension	Knee	Foot	Torque	Shock
UTT	total surface bearing	roll-on liner	pin	–	Echelon VT	yes	yes
	patella tendon bearing	pelite liner	friction	–	Re-Flex VSP	–	yes
	patella tendon bearing	pelite liner	friction	–	Vari-Flex XC	yes	yes
	patella tendon bearing	roll-on liner	friction	–	Echelon VT	yes	yes
	total surface bearing	roll-on liner	pin	–	Re-Flex Shock	–	yes
	patella tendon bearing	roll-on liner	pin	–	Re-Flex Shock	–	yes
	total surface bearing	roll-on liner	pin	–	Echelon VT	yes	yes
	total surface bearing	roll-on liner	pin	–	Vari-Flex XC	yes	yes
	total surface bearing	roll-on liner	pin	–	Echelon VT	yes	yes
	total surface bearing	roll-on liner	pin	–	Echelon VT	yes	yes
UTF	ischial bearing	roll-on liner	seal-in	C-leg	Axtion	–	–
	end bearing	roll-on liner	seal-in	KX06	Vari-Flex XC	yes	yes
	ischial containment	roll-on liner	seal-in	KX06	Re-Flex Shock	–	yes
	end bearing	roll-on liner	seal-in	Plie	LP Rotate	yes	yes
	end bearing	–	friction	KX06	Vari-Flex XC	yes	yes
	end bearing	–	friction	KX06	LP Rotate	yes	yes
	ischial bearing	–	skin suction	KX06	Elite VT	yes	yes
	ischial bearing	roll-on liner	seal-in	Genium X3	Triton Heavy Duty	–	–
	end bearing	roll-on liner	friction	KX06	Echelon	–	–
	ischial bearing	roll-on liner	seal-in	KX06	Vari-Flex XC	yes	yes
BTF	L – end bearing, R – ischial containment	roll-on liner	L – friction, R – seal-in	Genium	Triton Low Profile	–	–
	ischial bearing	roll-on liner	seal-in	C-leg	Axtion	–	–
	ischial bearing	roll-on liner	seal-in	C-leg	Axtion	–	–
	ischial containment	roll-on liner	L – friction, R – seal-in	Genium	Triton Vertical Shock	yes	yes
	ischial containment	roll-on liner	seal-in	Genium	LP Rotate	yes	yes
	ischial bearing	roll-on liner	seal-in	Genium	LP Rotate	yes	yes
	end bearing	–	friction	Genium	LP Rotate	yes	yes
	ischial bearing	roll-on liner	seal-in	Genium	LP Rotate	yes	yes
	end bearing	L – sock, R – roll-on liner	L – friction, R – seal-in	Genium	LP Rotate	yes	yes
	ischial containment	roll-on liner	seal-in	Genium	Triton Vertical Shock	yes	yes

111 **Table 2:** Summary of each amputee’s prosthetic prescription.

112 All participants performed the same protocol, which began with walking for 2
113 minutes up and down the gait laboratory walkway to establish their self-selected walking
114 speed, before 5 minutes of walking at the established self-selected walking speed. The
115 instrumented gait laboratory walkway was 10m in length, and the participants turned around
116 at the ends of the laboratory before returning again. This was repeated for the duration of the
117 data collection while whole-body optical motion (Vicon, Oxford, U.K.) and forceplate
118 (Kistler, Winterthur, Switzerland) data were recorded at 100Hz and 1000Hz respectively.

119

120 **Energy Flow Analysis**

121 A custom-written lower limb model comprised of a pelvis and bilateral thighs, shanks
122 and feet all linked by 6 degree of freedom (DoF) joints was used for inverse dynamics
123 analysis to provide the necessary data for the subsequent energy flow calculations [21, 27]
124 and was implemented in Matlab 2014b (The Mathworks Inc., Natick, MA, U.S.A.). Body
125 segment parameters for both the intact and prosthetic limb were scaled according to subject
126 mass and height using the anthropometric measures of de Leva [28]. Optical marker clusters
127 attached to a rigid base were used to track each body segment's motion, and individual
128 optical markers placed bilaterally on the following landmarks were used to determine
129 segment end points and scale the model to each participant: posterior and anterior superior
130 iliac spine, medial and lateral femoral epicondyles, medial and lateral malleoli, posterior and
131 lateral calcaneus, on the dorsal surface of the 1st, 2nd and 5th metatarsal heads. Optical motion
132 and forceplate data were used to calculate inter-segmental angles and moments at the ankle,
133 knee and hip joints of each limb separately following established inverse dynamics utilising
134 Newton-Euler equations of motion for the segment dynamics [29].

135 The approach of Winter & Robertson [21] was used to calculate energy exchanges
136 across the ankle, knee and hip joints. In summary, this approach uses inter-segmental moment

137 (\mathbf{M}_{Joint}) and force (\mathbf{F}_{Joint}), derived from the inverse dynamics, and the segment's angular
 138 velocity ($\boldsymbol{\omega}_{Seg}$) and translational joint velocity (\mathbf{V}_{Joint}), all of which are vector quantities
 139 expressed in the global coordinate frame, as inputs to calculate the power transferred between
 140 segments. Referring to Figure 1, at a joint the power flows into the two segments ($\mathbf{P}_{S2,Dist}$
 141 and $\mathbf{P}_{S1,Prox}$) are each the sum of the Segment Torque Power (STP) and Joint Force Power
 142 (JFP) which are given by:

$$143 \quad STP = \mathbf{M}_{Joint} \cdot \boldsymbol{\omega}_{Seg} \text{ (Eq. 1)}$$

$$144 \quad JFP = \mathbf{F}_{Joint} \cdot \mathbf{V}_{Joint} \text{ (Eq. 2)}$$

145 Assuming no loss of energy at a joint, the net muscle power generated or absorbed at the joint
 146 (hereafter referred to as joint power) is given by:

$$147 \quad \mathbf{P}_{Joint} = \mathbf{P}_{S2,Dist} + \mathbf{P}_{S1,Prox} \text{ (Eq. 3)}$$

148 We define the directions of positive power flows to be as shown in Figure 1.

149

150 **Statistical Analyses**

151 From the 5 minutes of walking at a self-selected walking speed, a minimum of 5 full
 152 gait cycles (with clean force plate contacts) were recorded for each limb were used as data
 153 inputs for the full inverse dynamics and energy flow analysis. Each trial and each gait cycle
 154 was analysed separately, with outputs from each gait cycle time-normalised to 100%. A clean
 155 foot contact was defined as fully within the boundary of the forceplate. A gait cycle was
 156 defined as the time between ipsi-lateral heel contacts, with heel contact being defined by a
 157 vertical force greater than 20N applied to the forceplates within the walkway. For the control
 158 and BTF groups, data from left and right legs were averaged, and for unilateral groups (UTT
 159 and UTF) prosthetic and intact data were grouped for subsequent comparisons.

160 Joint and segment power data at peak ankle push-off for each group were checked for
 161 normality using a Kolmogorov-Smirnov test, and two-tailed t-tests were used to compare

162 joint and segment powers at peak ankle power generation (subsequently referred to as “Push-
163 off”). To reduce the risk of type 1 errors only the following clinically relevant comparisons
164 were made: control vs. UTT/UTF intact side and BTF; UTT/UTF prosthetic side vs. BTF;
165 and prosthetic side vs. intact side for UTT and UTF. The threshold where a difference was
166 considered statistically significant was set at 0.016 to account for the three comparisons
167 performed for each group. Statistical analysis of the results was performed in Matlab 2014b
168 using the Statistical Analysis Toolbox (Version 9.1, The Mathworks Inc., Natick, MA,
169 U.S.A.).

170

171 Results²

172 **Qualitative Description of Energy Exchanges**

173 The periods of: first double-support, single-support and swing phase of the gait cycle
174 are illustrated in Figure 2. As all groups displayed similar trends in energy flow patterns for
175 much of the gait cycle, a qualitative description of the underlying patterns during the gait
176 cycle is provided first. As substantial differences between groups in energy exchanges were
177 observed primarily at push-off (period 3 in Figure 2), a quantitative comparison of the energy
178 exchanges at this period of the gait cycle was performed in the context of energy flows across
179 the entire limb (Table 3).

180

181 *Controls*

182 Able-bodied controls exhibited biphasic power transfer into and out of the trunk in
183 single support and swing with the two double support periods representing transitions between
184 these mechanisms (Figure 3). In the first half of single support, power is primarily transferred
185 to the trunk from the hip extensors, augmented to a small extent by transfer of power from the

² Source data will be made available at: <https://dx.doi.org/10.17866/rd.salford.2082871.v1>

186 thigh segment. In the second half, power is transferred out of the trunk with some being
187 absorbed by the hip flexors and the majority being transferred to the thigh. In swing there is
188 little power generation or absorption by the hip muscles and as such, power is simply
189 transferred from the trunk to the thigh in early swing and from the thigh to the trunk in late
190 swing. It should be noted that the overall pattern is such that there is an exchange of energy
191 from the stance thigh through the pelvis to the swing thigh in early single support and in the
192 opposite direction in late single support. In first double support there is a brief transfer of power
193 from the thigh into the trunk. During second double support, power is primarily transferred
194 from the hip flexors into the thigh.

195 In the first half of double support, power is transferred from the thigh to the shank as
196 the knee flexes, and then back into the thigh as the knee extends (Figure 3). In late single
197 support there is little power generation or absorption but energy flows across the joint from
198 thigh to shank. In second double support the shank transfers energy in roughly equal
199 proportions to the thigh and the knee muscles. In early swing there is little power generation
200 or absorption but a transfer of energy from the thigh to the shank. In late swing the shank
201 loses this energy with part of it being absorbed by the knee and the rest transferred to the
202 thigh.

203 At the ankle, the shank was observed to lose energy during early stance partly to the
204 foot and partly to power absorption in the ankle muscles after which there is a quiescent
205 phase for most of the first half of single support (Figure 3). In late single support the shank
206 loses energy which is absorbed by the ankle muscles. In second double support the ankle
207 generates substantial power with most of this energy passing to the shank. There is no power
208 absorption or generation in swing with energy flowing from shank to foot over the first half
209 and in the other direction over the second half.

210

211 *Unilateral Trans-Tibial Amputees*

212 Both limbs of the UTT amputees displayed similar energy flows for much of the gait
213 cycle, except in second double support of the intact limb, where hip power is transferred to
214 the trunk rather than the thigh (Figure 4). In the intact limb, power transfer from the thigh to
215 the pelvis in late swing finishes before initial contact whereas in the prosthetic limb that
216 power transfer continues until initial contact.

217 At the knee, energy flow patterns are broadly similar to the control group for the intact
218 knee, but these amputees appear to transfer more energy from thigh to shank in late single
219 support and there is a greater transfer of energy from shank to thigh in second double support
220 (Figure 4). There is a reduction in knee activity on the prosthetic side. In stance, the prosthetic
221 side knee thus acts largely as a mechanism through which energy is simply transferred (with
222 no augmenting by muscle activity) between the shank and thigh. Throughout swing the knee
223 functions in a broadly similar manner to that of the control group (with some energy absorption
224 in late swing).

225 At the ankle, the patterns are generally similar to those of the control group for the
226 intact limbs of the UTT amputees. On the prosthetic side there is less transfer of energy to the
227 foot in early stance and diminished (but not significantly so) power generation during second
228 double support.

229

230 *Bilateral and Unilateral Trans-Femoral Amputees*

231 We consider the UTF group with some references to the BTF group in brackets. In the
232 intact limb of UTF amputees increased power generation at the hip in early single support is
233 transferred almost exclusively to the trunk followed by a much smaller transfer from the
234 trunk to the thigh in late single support (Figure 5). Whereas controls show greater power

235 transfer out of the trunk in late single support than in early swing, this is reversed on the
236 prosthetic side of the trans-femoral amputees (for both UTF and BTF amputees) (Figure 5
237 and Figure 6). This peak transfer of power from the trunk to swing thigh occurs at a similar
238 time to the increased power generation by the hip muscles suggesting a general pattern of
239 increased power transfer from intact side hip muscles to swing thigh in early prosthetic
240 swing. In the intact limb, exchange of energy from the thigh to pelvis in late swing finishes
241 before initial contact whereas in the prosthetic limb that power transfer continues until initial
242 contact (as was observed in UTT amputees).

243 Energy flow patterns at the intact knee are broadly similar to the control group, but
244 these amputees appear to transfer less energy from thigh to shank in late single support
245 (Figure 5). There is a greater transfer of energy from shank to thigh in second double support
246 on the intact side, but large transfers from the thigh to shank on the prosthetic side. There is a
247 reduction in power generation or absorption at the prosthetic knee, which is particularly
248 apparent in the first half of the gait cycle for both UTF and BTF amputees. In stance, the
249 prosthetic knee acts largely as a mechanism through which energy is exchanged between
250 thigh and shank (Figure 5 and Figure 6). As before, throughout swing the prosthetic knee
251 functions in a broadly similar manner to that of the control group (with some energy
252 absorption in late swing).

253 Energy flow at the ankle was generally similar to those of the control group for the
254 intact limb of the UTF amputees. On the prosthetic side there is less transfer of energy to the
255 foot in early stance and considerably diminished power generation during second double
256 support, particularly in the trans-femoral groups (Figure 5 and Figure 6).

257

258 **Quantitative Differences at Push-off**

259 Figure 7 shows the mean power flows at the moment of peak ankle power generation,
260 subsequently referred to as push-off (mean variability in power flows is shown in Figure 8);
261 and a summary of all statistically significant results at push-off can be found in Table 3. At
262 push-off, a substantial proximally-directed transfer of power was observed across both the
263 hip and knee in the intact limb of unilateral amputees, while the prosthetic limb of unilateral
264 amputees was found to have a distally-directed transfer of power (Figure 7). This was
265 particularly apparent when considering the transfers across the knees and hips of UTT and
266 UTF amputees. BTF amputees were also found to have a distally-directed transfer of power
267 across the hip and knee, but this was only significantly different to the prosthetic side of UTF
268 amputees at the hip.

269 In the intact ankle, controls and UTT amputees generated significantly more ankle
270 power and transferred significantly more power to the shank compared to BTF amputees.
271 UTT amputees were also able to generate more ankle power on the prosthetic side compared
272 to UTF amputees on their prosthetic side, but this was not significant.

273

274 **Table 3:** Summary of all results that were statistically different at push-off

Hip Joint			Knee Joint			Ankle Joint			
Segment / Joint	Groups	p-value	Segment / Joint	Groups	p-value	Segment / Joint	Groups	p-value	
Distal Pelvis	UTT_P vs. UTF_P	0.002	Distal Thigh	Con vs. BTF	0.008	Distal Shank	Con vs. BTF	< 0.001	
	UTF_I vs. UTF_P	< 0.001		UTT_I vs. UTT_P	0.006		UTT vs. BTF	< 0.001	
	UTF vs. BTF	< 0.001		UTF_I vs. UTF_P	< 0.001		UTT_P vs. UTF_P	0.012	
				UTF_I vs. UTF_P	< 0.001				
Joint Power	UTT_P vs. UTF_P	0.009	Joint Power	Con vs. BTF	< 0.001	Joint Power	Con vs. BTF	< 0.001	
	UTT vs. BTF	0.004					UTT vs. BTF	< 0.001	
Proximal Thigh	Con vs. UTF	0.010	Proximal Shank	Con vs. BTF	< 0.001		Proximal Foot	None	0.92
	UTT_I vs. UTT_P	0.013		UTT_P vs. UTF_P	0.003				
	UTT_P vs. UTF_P	0.008		UTT_I vs. UTT_P	0.003				
	UTF_I vs. UTF_P	< 0.001		UTF_I vs. UTF_P	< 0.001				
	UTF vs. BTF	0.014							

275 Note: The suffix “_P” and “_I” denotes the prosthetic or intact side respectively. “None” signifies no

276 statistically significant difference was observed between any of the cohort, and the mains effect

277 ANOVA is given.

278

279 Discussion

280 **Energy Flow between Segments**

281 This study has quantified the energy transfers in the lower limbs of unilateral and

282 bilateral amputees to investigate the underlying mechanisms which lead to inefficient

283 amputee gait.

284 The key finding of this study was the change in direction of energy transfer across the

285 hip and knee at push-off from a proximally-directed transfer in intact limbs, to a distally-

286 directed transfer in prosthetic limbs. Despite both unilateral groups displaying this distally-

287 directed energy transfer on their prosthetic side, the source of this energy was not the same,

288 with the hip muscles providing the majority of power to the prosthetic limb for UTTs,

289 compared to the pelvis providing most of the power in UTFs (Figure 7). Critically, although

290 relatively short, push-off is a particularly energetic period compared to the rest of the gait

291 cycle. On the prosthetic side, the substantial reduction in peak ankle power generation and the

292 distally-directed power transfer during push-off (Figure 7) highlights the importance of this
293 particular phase of the gait cycle for the energy efficiency of amputee gait.

294 Comparing the phases either side of push-off showed differences in magnitude, rather
295 than direction, of energy transfer between trans-femoral and trans-tibial amputees and
296 controls. In late single support, UTTs absorb more energy at both their intact and prosthetic
297 ankle than controls. On the intact side, this could be linked to their increased push-off power,
298 which utilises elastic energy stored in the Achilles in late single support. On the prosthetic
299 side, this could be to increase the elastic energy available to be returned during push-off and
300 thus limit the reduction in push-off power associated with the loss of musculature that can
301 actively generate power. Conversely, UTFs and BTFs have reduced ankle power absorption
302 in late single support, and also reduced power transfer from the pelvis through the limb to the
303 ankle during push-off. This leads to substantially reduced prosthetic ankle power at push-off
304 as a result of the loss of musculature that can actively generate power (Figure 7).

305 Push-off and initial swing are the periods when power flows differ most between
306 groups, with trans-femoral amputees transferring more power from the pelvis and hip
307 muscles to their thigh and shank on their prosthetic side compared to trans-tibial amputees or
308 controls. This increased transfer of power to the thigh and shank segments during initial
309 swing, despite the slower walking speed, is likely to be a direct result of the lack of push-off
310 power at the prosthetic ankle of the trans-femoral amputees.

311 It should be noted that for much of the gait cycle (particularly for late single support
312 and swing) energy is transferred into a segment at one joint and out of it at the other joint.
313 The overall effect is thus for energy to be transferred through the segment.

314

315 **Segment Powers**

316 The observation that controls and unilateral amputees have similar segment powers
317 during late stance and, in particular, at push-off (Figure 7) is note-worthy as it implies that,
318 regardless of which joints are able to actively produce power, the segments of the lower limb
319 still require a certain amount of energy to be propelled into swing. Whether this energy
320 comes from a proximal or distal direction is irrelevant in terms of segment energetics, but
321 clearly has implications for lower limb amputees, who are unable to actively generate power
322 distal to their amputation due to the loss of muscles. In controls, at push-off the ankle plantar
323 flexors provide all of the energy transferred to the foot and shank and a large proportion of
324 the energy transferred to the thigh. But as the capacity to produce power at the ankle is
325 reduced in amputees, greater contributions from the proximal joints are required to meet the
326 energy requirements of the segments, which as mentioned are relatively invariant.

327 Understanding how an individual walks inefficiently is not an issue confined to
328 amputees, but has relevance to other clinical populations such as stroke survivors, as well as
329 to able-bodied individuals. Indeed, understanding how the human musculoskeletal system
330 functions during gait and trying to replicate this function has been of interest to the
331 biomechanics community for many years, resulting in conflicting opinions regarding the
332 importance of ankle push-off power [30] and controversy about the role of the ankle
333 plantarflexors during gait [31, 32]. While the observation of similar segment energies could
334 have been made using a segment energetics approach (considering only potential and kinetic
335 energy), the key finding of a directional change in energy transfers would not have been
336 possible and is indeed one of the strengths of the approach used in this study and will likely
337 promote further discussion around the role of push-off during gait.

338

339 **Limitations**

340 The main limitation of this study is the assumption that the body segments are all
341 rigid bodies. This is a problem common to most lower limb inverse dynamics models and has
342 been previously investigated in amputees [27, 33] in components that do not necessarily
343 remain rigid as is the case in many flexible ESR prosthetic feet. These studies found
344 translational powers contributed significantly to the overall joint power, particularly in early
345 stance, and highlight the importance of considering all 6 DoF when modelling the power
346 transfers across a prosthetic foot and ankle system. While a limitation, particularly at the foot
347 and ankle, the modelling approach used here allows for internal consistency between our
348 energy flow calculations and the inverse dynamics calculations and does not detract from the
349 main findings of this study. A second limitation of the study is the use of able-bodied
350 anthropometric regression equations to derive the inertial parameters of the prosthetic
351 segments (CoM positions and inertial tensors). While the use of device-specific values would
352 be more desirable, the effect of using able-bodied equations likely resulted in only minor
353 differences in power flow calculations at push-off due to the small differences between the
354 prosthetic and anatomic inertial parameters used in the inverse dynamics calculations. A final
355 limitation of the study was collecting data at each individual's self-selected walking speed.
356 As with all gait analysis studies, two confounding influences must be considered: controlling
357 walking speed on the one hand or, if one common walking speed was imposed, the degree to
358 which that was unnatural for each amputee. As the purpose of this study was to understand
359 power flows in normal amputee walking, the Authors felt that it was important that the
360 participants walked as naturally as possible. While walking speed is known to influence gait
361 analysis measures, the Authors feel this is an acceptable limitation as it provides the most
362 clinically relevant setting in which to investigate the energy transfers across the limbs.

363

364 Conclusions

365 In conclusion, this study used an energy flow analysis to investigate the underlying
366 causes of inefficiency in amputee gait. The key findings of this study were that, although
367 thigh and shank segment powers are consistent between amputee groups, in order to meet the
368 energy requirements of these segments at push-off, amputees must utilise a distally-directed
369 transfer of power on their prosthetic side, whereas control subjects and the intact side of
370 unilateral amputees retain a proximally-directed transfer of power. This change in direction
371 of energy transfer is likely to be a direct result of the lack of push-off power at the prosthetic
372 ankle, particularly in trans-femoral amputees, and leads to their increased metabolic cost of
373 walking, with greater demands placed on the trunk and remaining hip musculature. The
374 practical implications of this are that both clinicians and prosthesis designers should focus on
375 restoring more natural push-off. Firstly, biomechanical measures of push-off could be used
376 by clinicians to assess the effectiveness of rehabilitation and physiotherapy protocols.
377 Secondly, knowing how energy is transferred across the limb in amputees is relevant to the
378 design of future prosthetic devices, particularly with regards to providing greater ankle push-
379 off power in passive devices.

380

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385 presented.

386

387 Conflicts of Interest

388 None

389

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391

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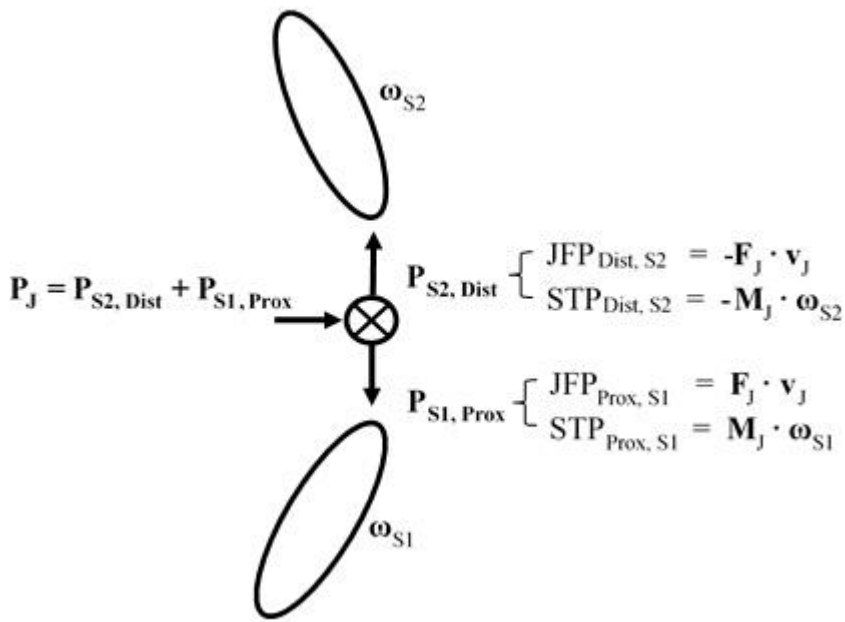
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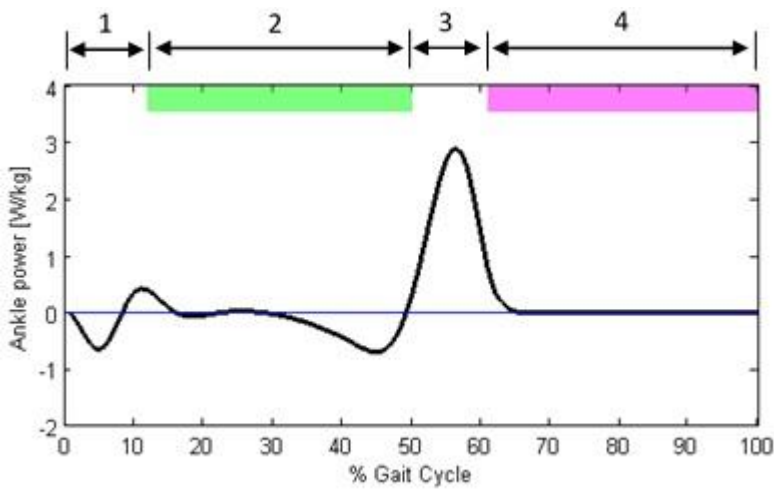
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484

485 **Figure 1:** Illustration of the power transfers across a joint due to segment torques (STP) and
 486 joint forces (JFP) at the ends of a segment.

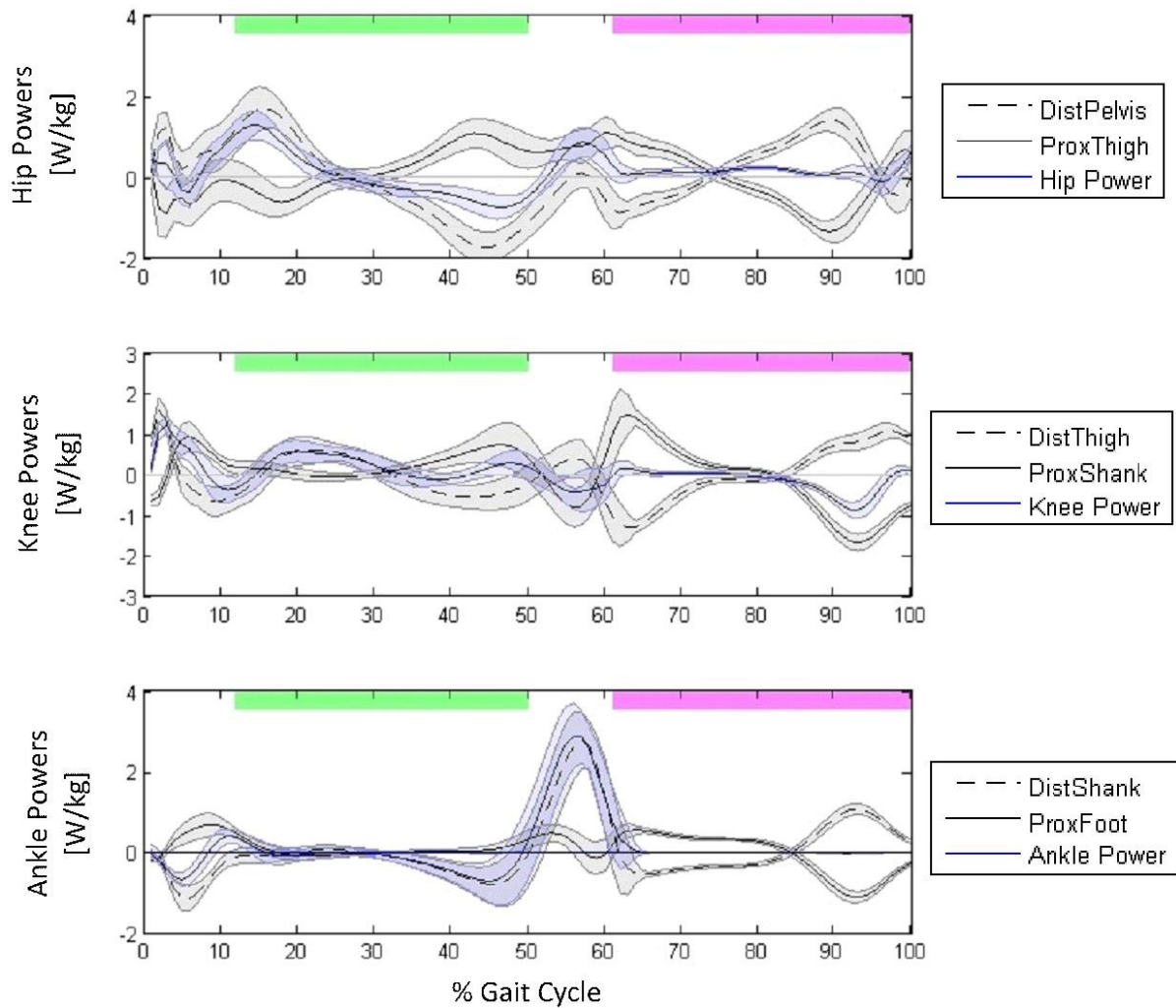
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488

489 **Figure 2:** Illustration of: (1) first double support; (2) single support; (3) second double support
 490 and (4) swing phase of the gait cycle based on the timings of ipsi-lateral heel-strike and toe-
 491 off, and contralateral heel-strike and toe-off.

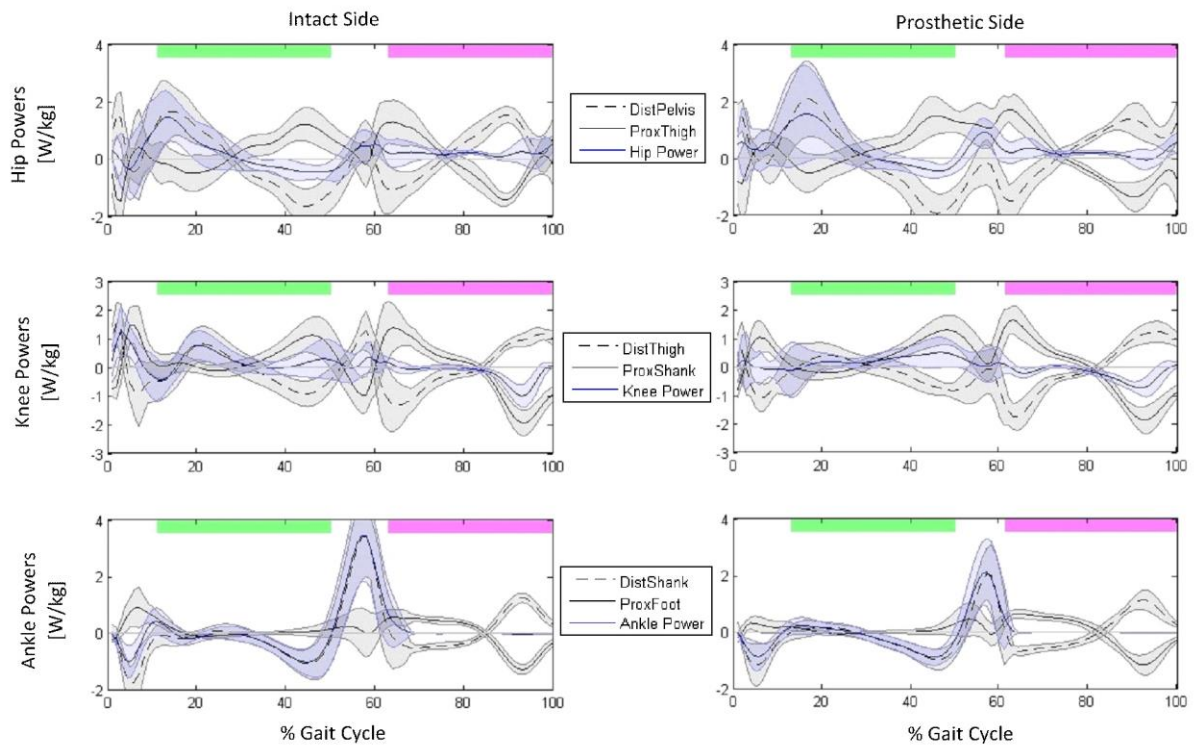
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493

494 **Figure 3:** Comparison of joint and segment power transfers for the control group. Note: Shaded
 495 green and purple regions at the top of each graph indicate single-support and swing phase
 496 respectively, with unshaded regions corresponding to the double-support periods. Shaded
 497 regions around each line indicates 1 standard deviation. “Dist” and “Prox” denote distal and
 498 proximal ends of a segment respectively.

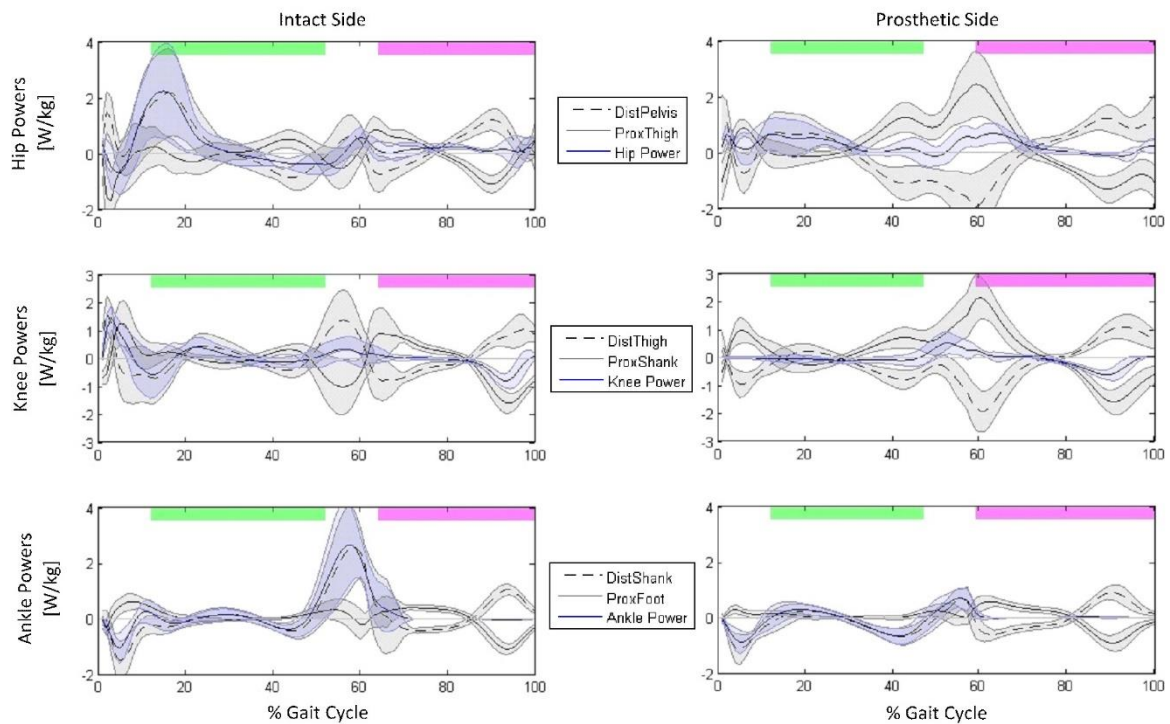
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501 **Figure 4:** Comparison of joint and segment power transfers for the UTT group. Note: Shaded
 502 green and purple regions at the top of each graph indicate single-support and swing phase
 503 respectively, with unshaded regions corresponding to the double-support periods. Shaded
 504 regions around each line indicates 1 standard deviation. “Dist” and “Prox” denote distal and
 505 proximal ends of a segment respectively.

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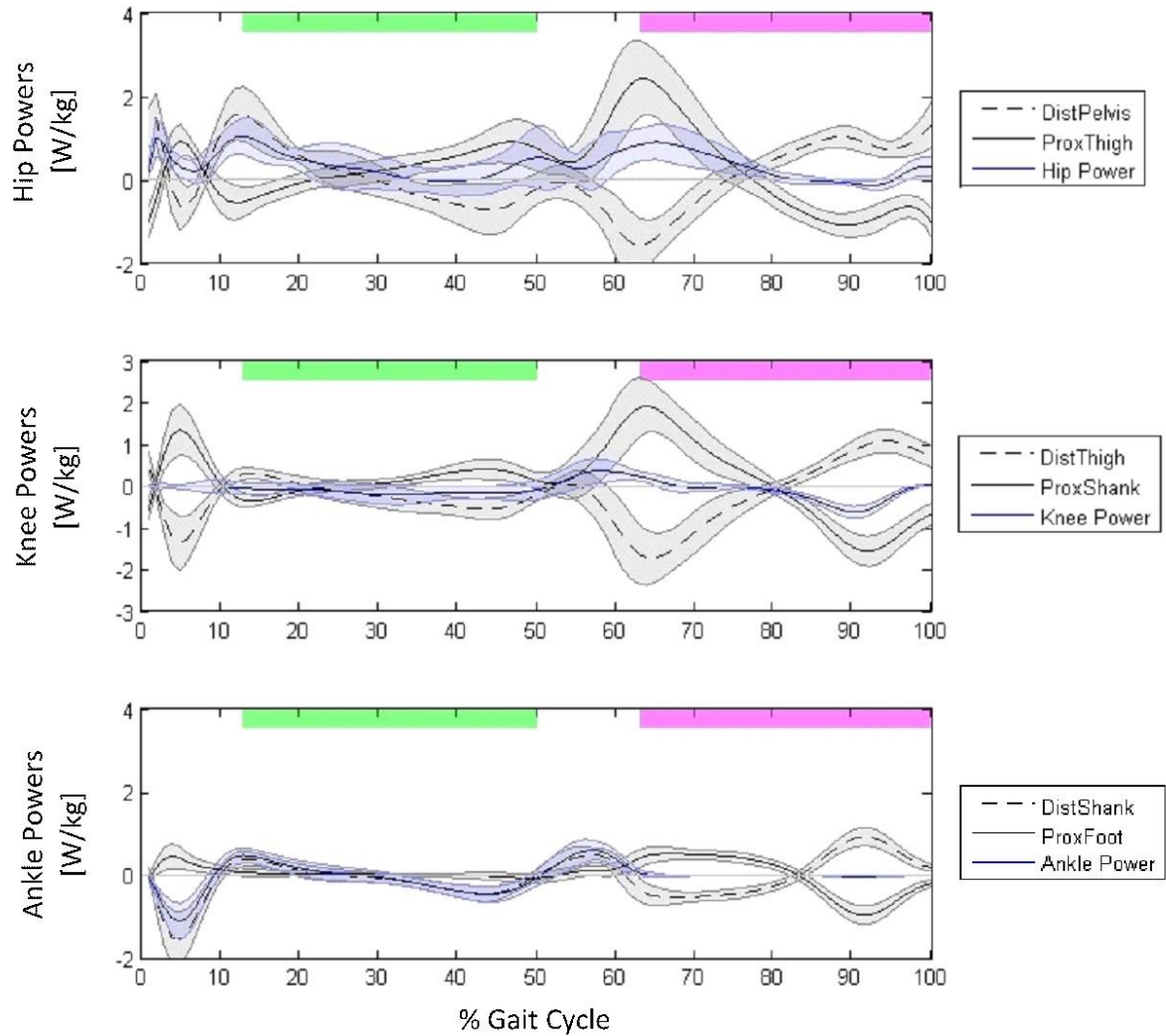


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508 **Figure 5:** Comparison of joint and segment power transfers for the UTF group. Note: Shaded
 509 green and purple regions at the top of each graph indicate single-support and swing phase
 510 respectively, with unshaded regions corresponding to the double-support periods. Shaded
 511 regions around each line indicates 1 standard deviation. “Dist” and “Prox” denote distal and
 512 proximal ends of a segment respectively.

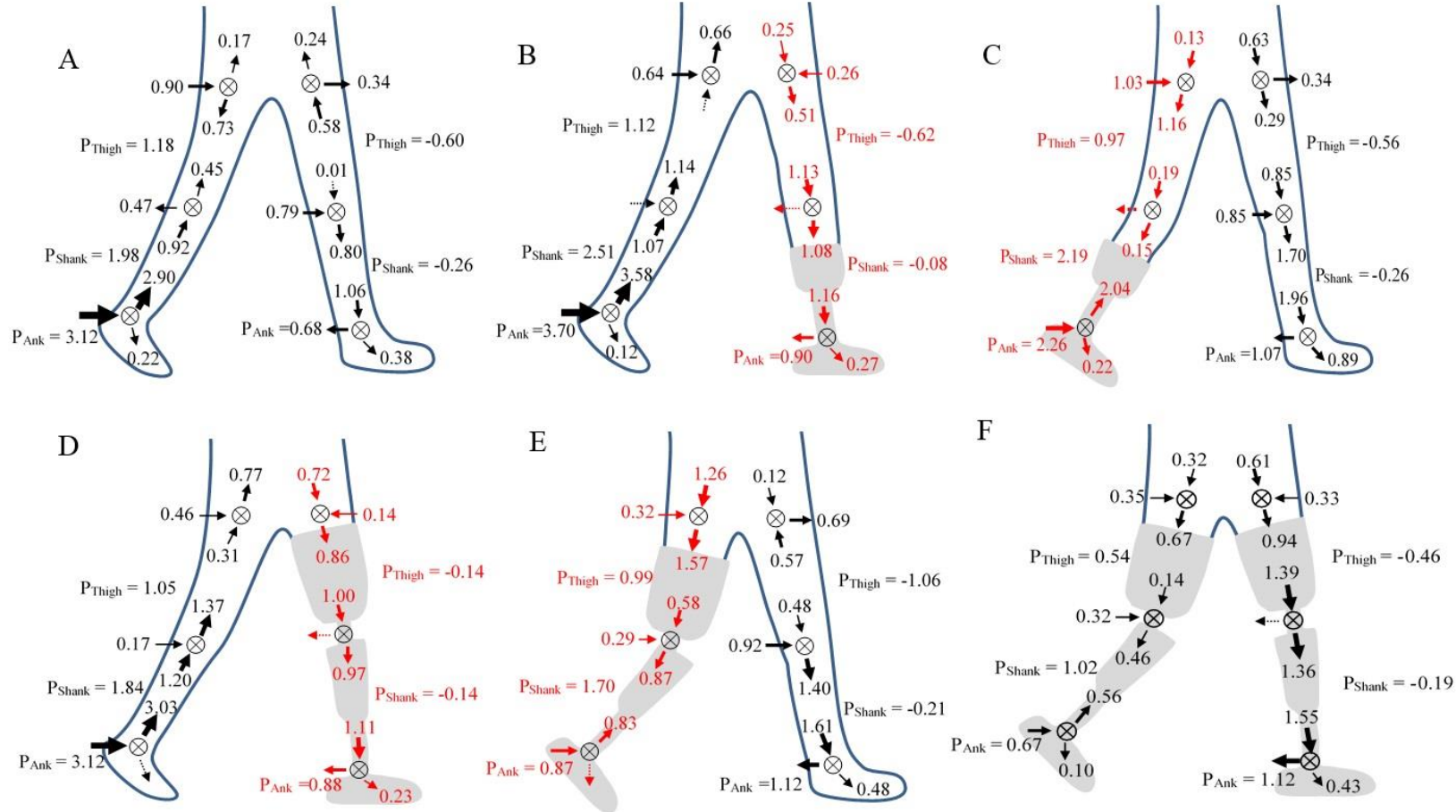
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516 **Figure 6:** Comparison of joint and segment power transfers for the BTF group. Note: Shaded
 517 green and purple regions at the top of each graph indicate single-support and swing phase
 518 respectively, with unshaded regions corresponding to the double-support periods. Shaded
 519 regions around each line indicates 1 standard deviation. “Dist” and “Prox” denote distal and
 520 proximal ends of a segment respectively.

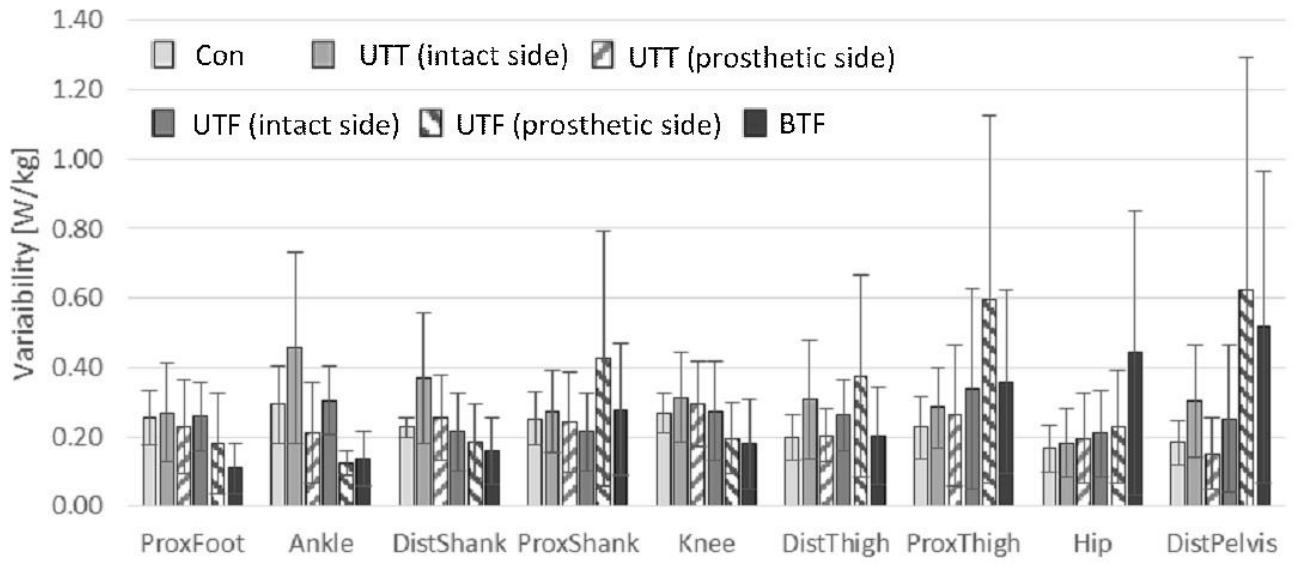


521

522 **Figure 7:** Comparison of segment powers at push-off on the intact side for: A) Controls, B) UTT, C) UTF; and the prosthetic side for: D) UTT,

523 E) UTF, F) BTF. Note: Arrows indicate the direction and relative magnitude of the power being transferred across each joint. Dotted arrows

524 indicate a power less than 0.1 W/kg.



525

526 **Figure 8:** Mean variability in joint and segment powers at push-off for all groups. Note: Error

527 bars indicate 1 standard deviation.

528

529 Highlights

530

531 Able-bodied and amputee lower limb energy exchanges were calculated during walking

532 Thigh and shank segment energies were consistent between groups

533 Intact limbs used a proximal flow of energy to propel the limb into swing

534 Reduced prosthetic ankle power generation was observed in amputees

535 Prosthetic side required a distal flow of energy to provide enough energy to the limb

536