Title: 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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Abstract

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

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

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

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

Introduction[1]

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

An efficient gait will likely be dependent on energy conserving exchanges between limb segments and also on energy storage and return mechanisms, typically utilising strain energy in tendons and prosthetic components. In unilateral amputees, it is known that the intact limb often compensates for deficiencies on the prosthetic side, which leads to characteristic gait asymmetries of reduced stance time and increased swing time and step 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

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

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

Materials and Methods

Subject Details and Protocol

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

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)

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

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

Table 2: Summary of each amputee's prosthetic prescription.

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

Energy Flow Analysis

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

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

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

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

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

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

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

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

Statistical Analyses

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

Joint and segment power data at peak ankle push-off for each group were checked for normality using a Kolmogorov-Smirnov test, and two-tailed t-tests were used to compare joint and segment powers at peak ankle power generation (subsequently referred to as "Push-off"). To reduce the risk of type 1 errors only the following clinically relevant comparisons were made: control vs. UTT/UTF intact side and BTF; UTT/UTF prosthetic side vs. BTF; and prosthetic side vs. intact side for UTT and UTF. The threshold where a difference was considered statistically significant was set at 0.016 to account for the three comparisons performed for each group. Statistical analysis of the results was performed in Matlab 2014b using the Statistical Analysis Toolbox (Version 9.1, The Mathworks Inc., Natick, MA, U.S.A.).

Results[2]

Qualitative Description of Energy Exchanges

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

Controls

Able-bodied controls exhibited biphasic power transfer into and out of the trunk in single support and swing with the two double support periods representing transitions between these mechanisms (Figure 3). In the first half of single support, power is primarily transferred 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

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

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

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

Unilateral Trans-Tibial Amputees

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

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

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

Bilateral and Unilateral Trans-Femoral Amputees

We consider the UTF group with some references to the BTF group in brackets. In the intact limb of UTF amputees increased power generation at the hip in early single support is transferred almost exclusively to the trunk followed by a much smaller transfer from the trunk to the thigh in late single support (Figure 5). Whereas controls show greater power transfer out of the trunk in late single support than in early swing, this is reversed on the prosthetic side of the trans-femoral amputees (for both UTF and BTF amputees) (Figure 5 and Figure 6). This peak transfer of power from the trunk to swing thigh occurs at a similar time to the increased power generation by the hip muscles suggesting a general pattern of increased power transfer from intact side hip muscles to swing thigh in early prosthetic swing. In the intact limb, exchange of energy from the thigh to pelvis in late swing finishes before initial contact whereas in the prosthetic limb that power transfer continues until initial contact (as was observed in UTT amputees).

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

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

Quantitative Differences at Push-off

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

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

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 UTF_I vs. UTF_P UTF vs. BTF	0.002 < 0.001 < 0.001	Distal Thigh	Con vs. BTF UTT_I vs. UTT_P UTF_I vs. UTF_P	0.008 0.006 < 0.001	Distal Shank	Con vs. BTF UTT vs. BTF UTT_P vs. UTF_P UTF_I vs. UTF_P	< 0.001 < 0.001 0.012 < 0.001
Joint Power	UTT_P vs. UTF_P UTT vs. BTF	0.009 0.004	Joint Power	Con vs. BTF	< 0.001	Joint Power	Con vs. BTF UTT vs. BTF UTT_P vs. UTF_P UTF_I vs. UTF_P	< 0.001 < 0.001 0.006 < 0.001
Proximal Thigh	Con vs. UTF UTT_I vs. UTT_P UTT_P vs. UTF_P UTF_I vs. UTF_P UTF vs. BTF	0.010 0.013 0.008 < 0.001 0.014	Proximal Shank	Con vs. BTF UTT_P vs. UTF_P UTT_I vs. UTT_P UTF_I vs. UTF_P	< 0.001 0.003 0.003 < 0.001	Proximal Foot	None	0.92

Note: The suffix "_P" and "_I" denotes the prosthetic or intact side respectively. "None" signifies no statistically significant difference was observed between any of the cohort, and the mains effect ANOVA is given.

Discussion

Energy Flow between Segments

This study has quantified the energy transfers in the lower limbs of unilateral and bilateral amputees to investigate the underlying mechanisms which lead to inefficient amputee gait.

The key finding of this study was the change in direction of energy transfer across the hip and knee at push-off from a proximally-directed transfer in intact limbs, to a distally-directed transfer in prosthetic limbs. Despite both unilateral groups displaying this distally-directed energy transfer on their prosthetic side, the source of this energy was not the same, with the hip muscles providing the majority of power to the prosthetic limb for UTTs, compared to the pelvis providing most of the power in UTFs (Figure 7). Critically, although relatively short, push-off is a particularly energetic period compared to the rest of the gait cycle. On the prosthetic side, the substantial reduction in peak ankle power generation and the distally-directed power transfer during push-off (Figure 7) highlights the importance of this particular phase of the gait cycle for the energy efficiency of amputee gait.

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

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

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

Segment Powers

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

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

Limitations

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

Conclusions

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

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Conflicts of Interest

None

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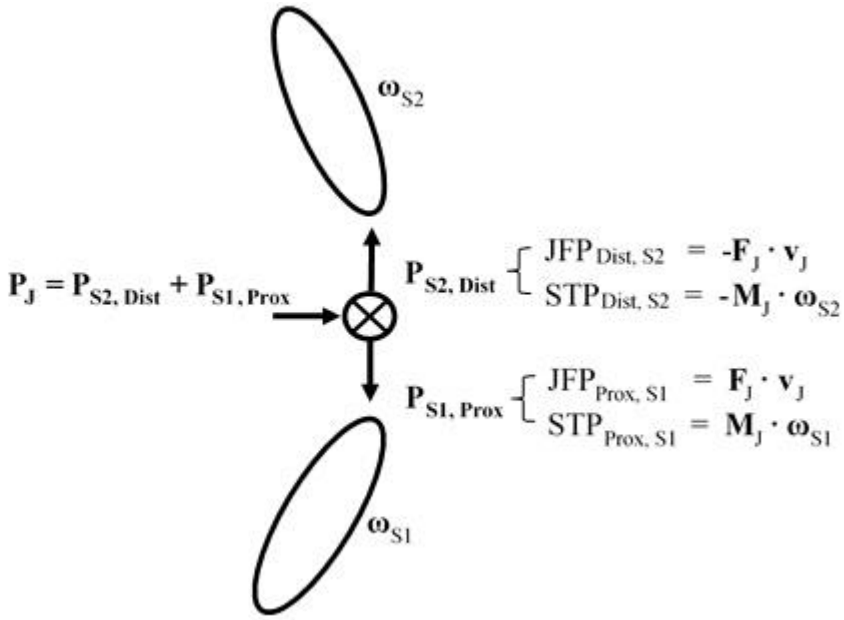

Figure 1: Illustration of the power transfers across a joint due to segment torques (STP) and joint forces (JFP) at the ends of a segment.

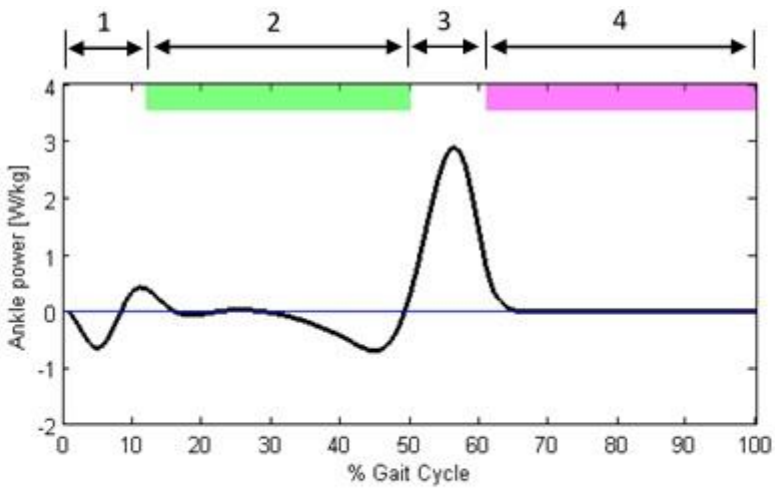

Figure 2: Illustration of: (1) first double support; (2) single support; (3) second double support and (4) swing phase of the gait cycle based on the timings of ipsi-lateral heel-strike and toe-off, and contral-lateral heel-strike and toe-off.

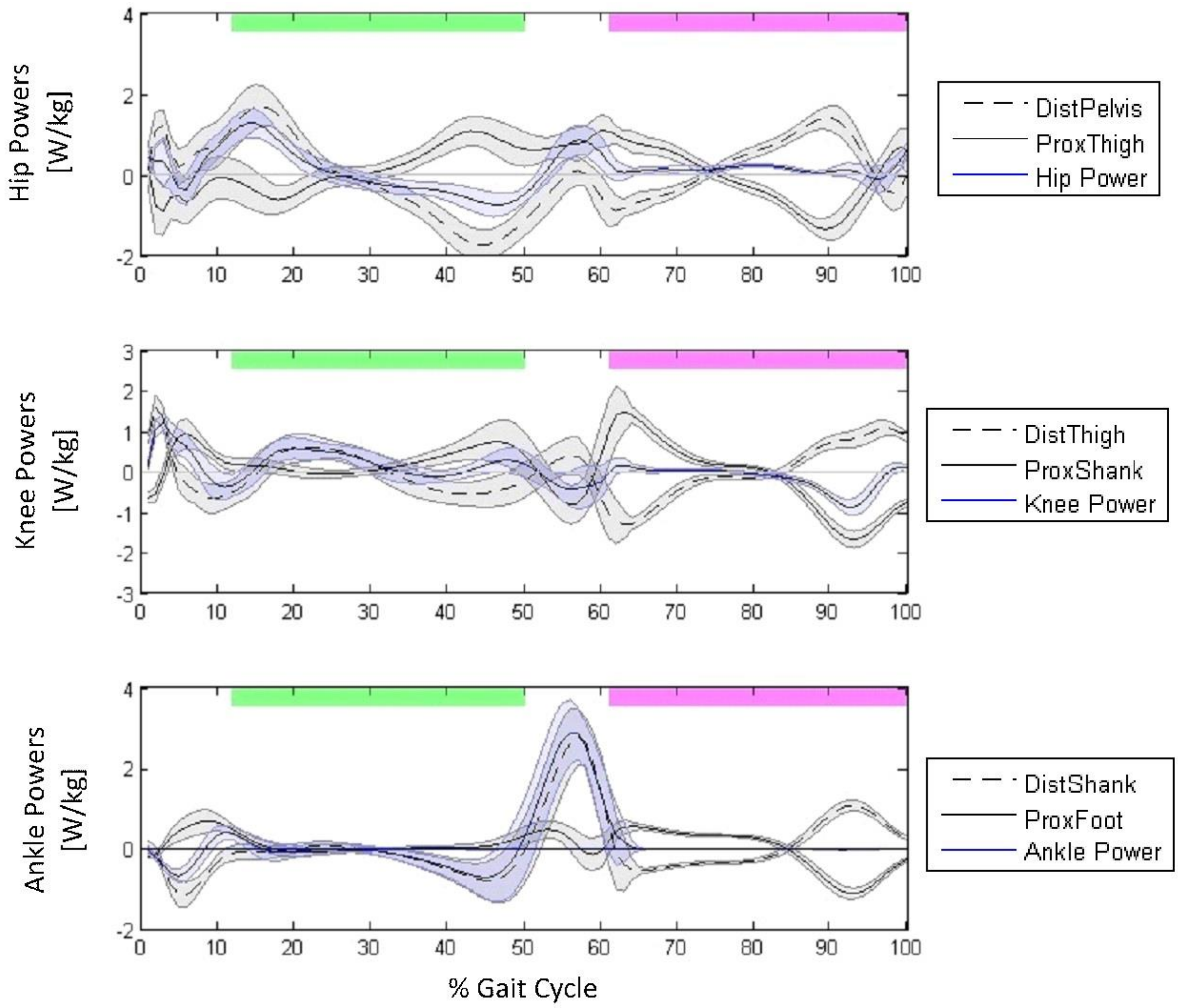

Figure 3: Comparison of joint and segment power transfers for the control group. Note: Shaded green and purple regions at the top of each graph indicate single-support and swing phase respectively, with unshaded regions corresponding to the double-support periods. Shaded regions around each line indicates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

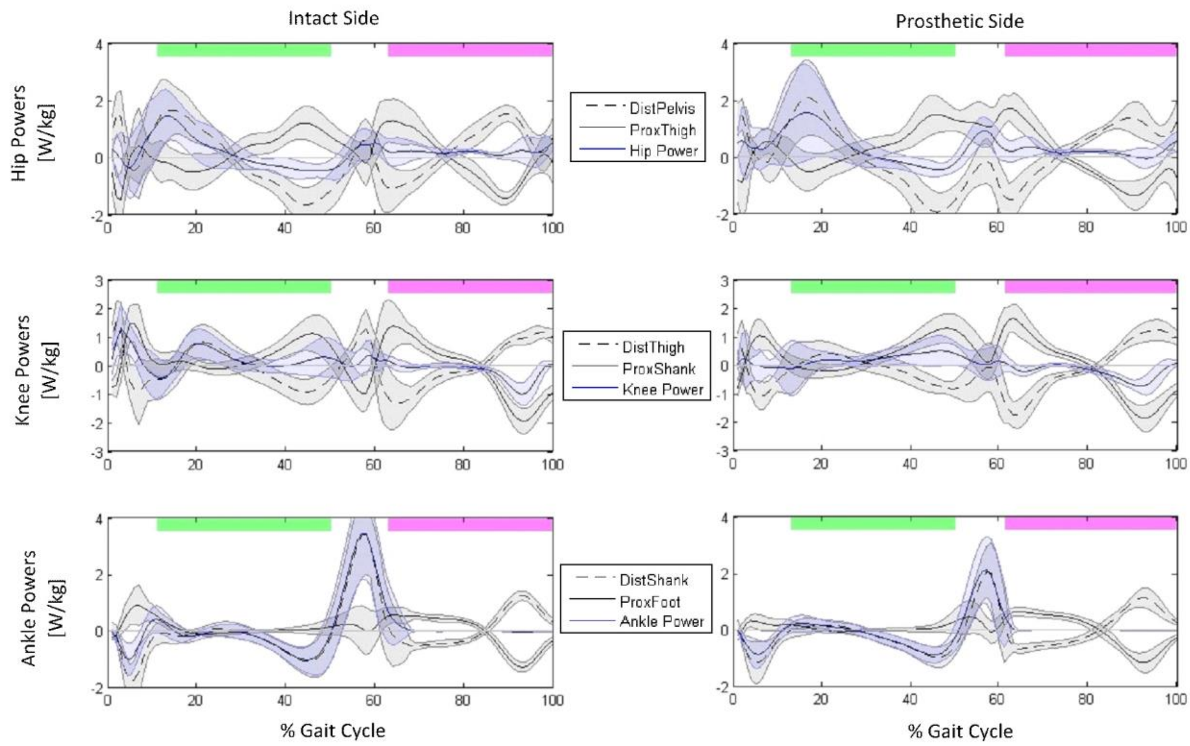

Figure 4: Comparison of joint and segment power transfers for the UTT group. Note: Shaded green and purple regions at the top of each graph indicate single-support and swing phase respectively, with unshaded regions corresponding to the double-support periods. Shaded regions around each line indicates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

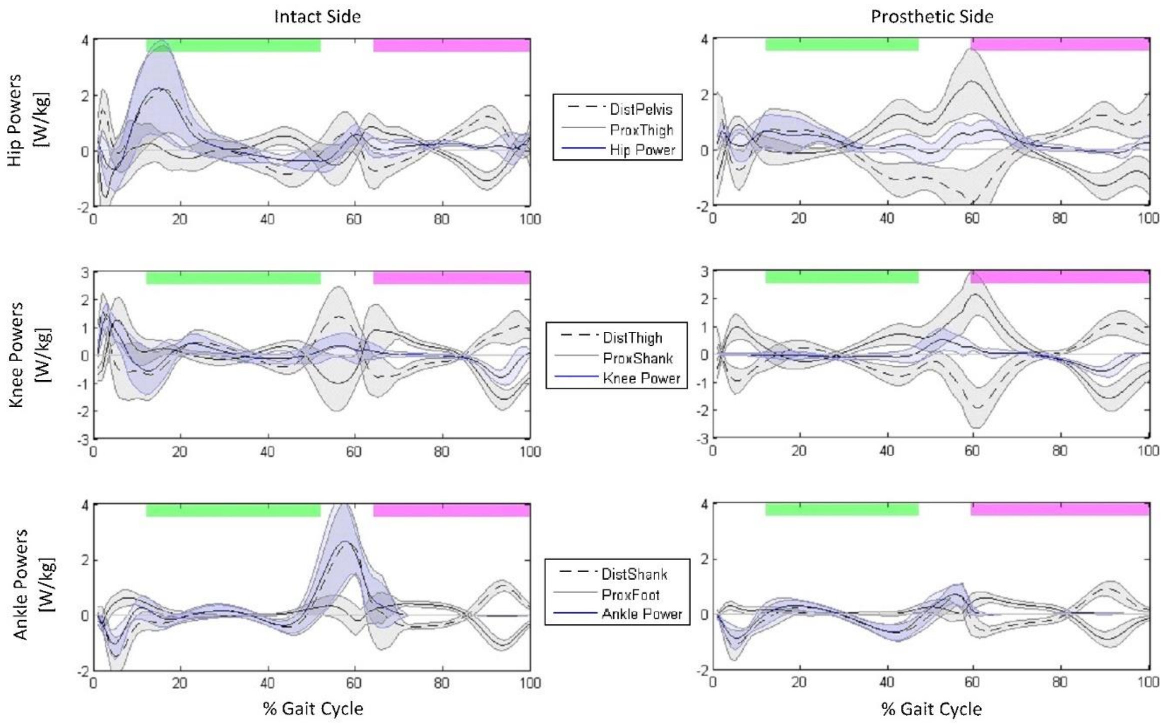

Figure 5: Comparison of joint and segment power transfers for the UTF group. Note: Shaded green and purple regions at the top of each graph indicate single-support and swing phase respectively, with unshaded regions corresponding to the double-support periods. Shaded regions around each line indicates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

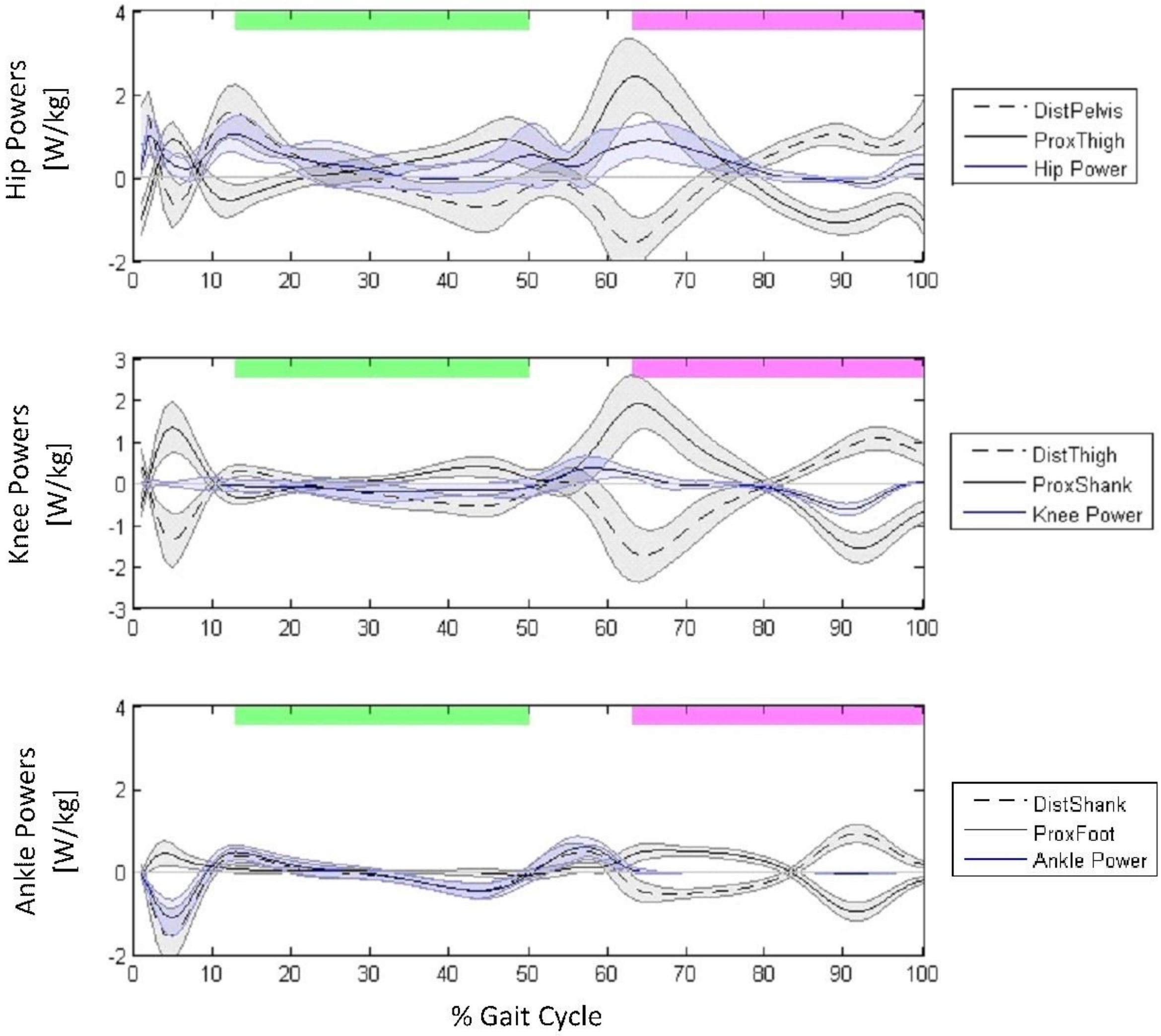

Figure 6: Comparison of joint and segment power transfers for the BTF group. Note: Shaded green and purple regions at the top of each graph indicate single-support and swing phase respectively, with unshaded regions corresponding to the double-support periods. Shaded regions around each line indicates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

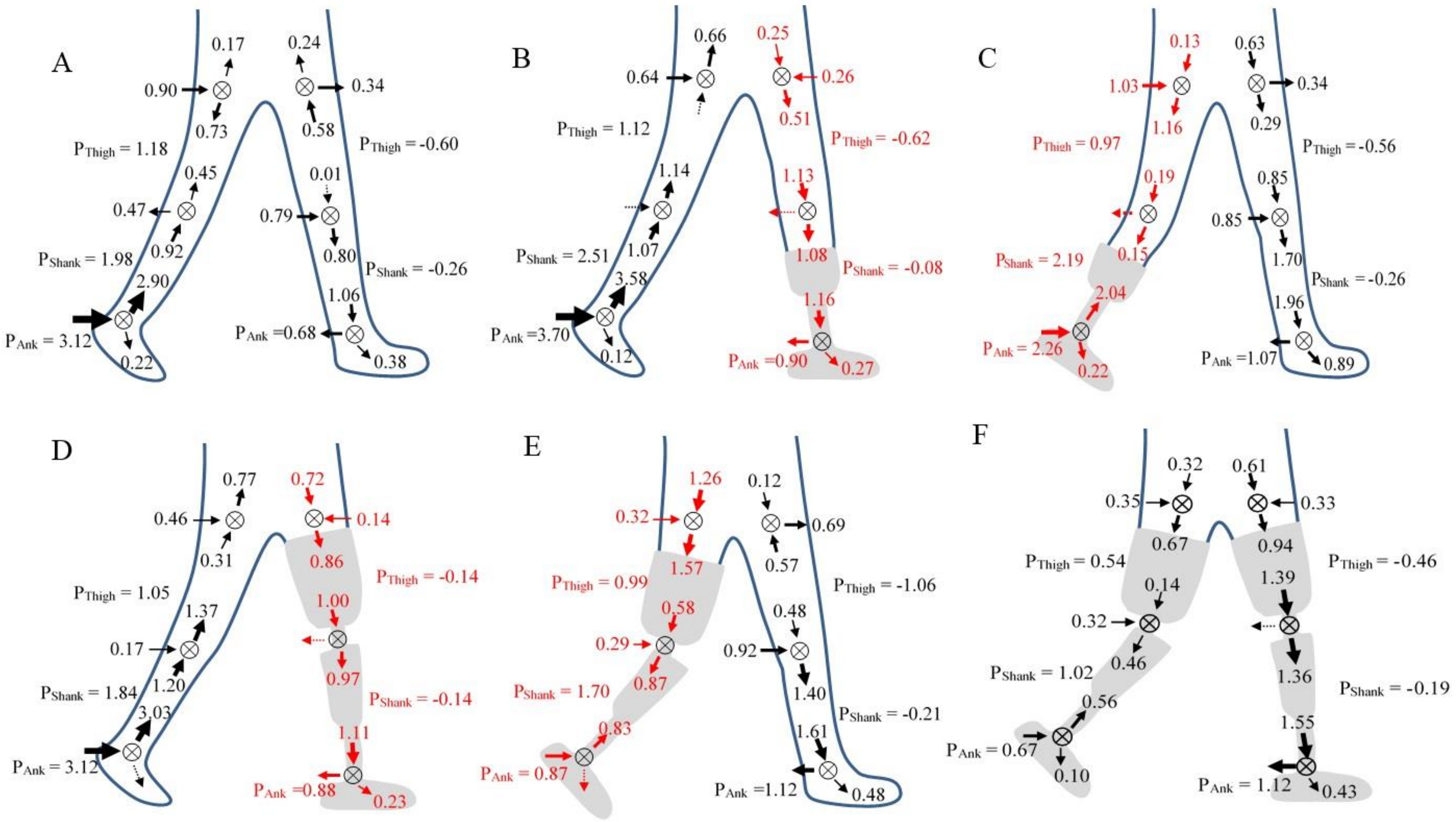

Figure 7: Comparison of segment powers at push-off on the intact side for: A) Controls, B) UTT, C) UTF; and the prosthetic side for: D) UTT, E) UTF, F) BTF. Note: Arrows indicate the direction and relative magnitude of the power being transferred across each joint. Dotted arrows indicate a power less than 0.1 W/kg.

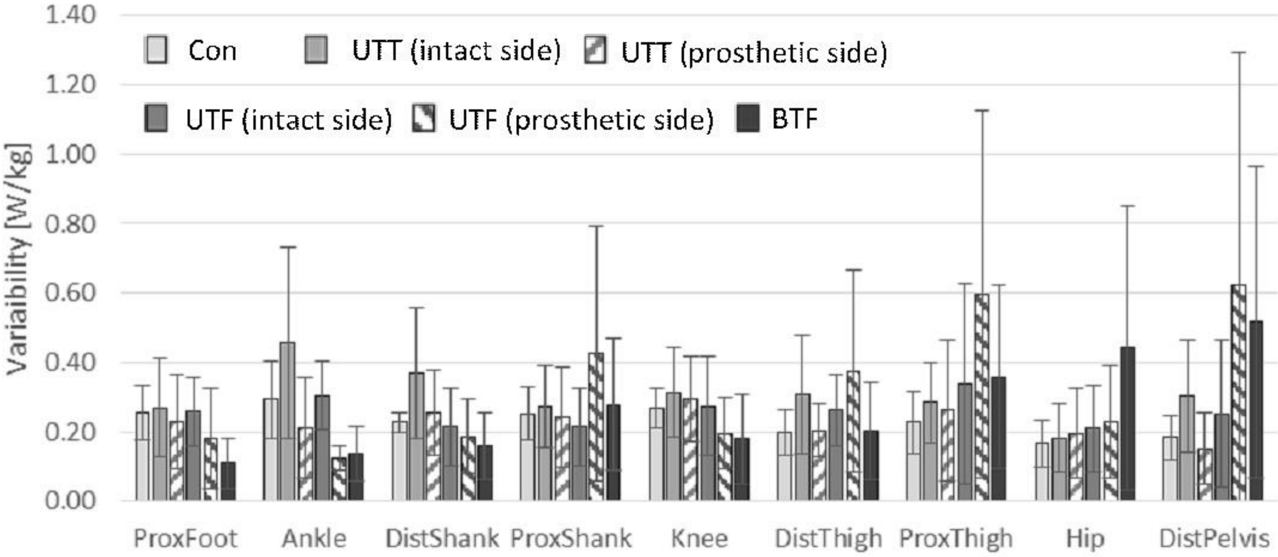

Figure 8: Mean variability in joint and segment powers at push-off for all groups. Note: Error bars indicate 1 standard deviation.

Highlights

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

Thigh and shank segment energies were consistent between groups

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

Reduced prosthetic ankle power generation was observed in amputees

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