Original Article

Distribution of joint work during walking on slopes among persons with transfemoral amputation

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Abstract

Persons with above-knee amputation have increased energy consumption and greater difficulty in negotiating uphill and downhill slopes. Walking on slopes requires an adaptation of the positive and negative work performed by the joints of the lower limb to propel the center of mass. Modern prosthetic feet and knees can only partially adapt to changes in inclination and the redistribution of joint work among persons with above-knee amputation is not described in the literature.

Level, upslope and downslope walking (at 5% and 12% inclinations) were investigated for twelve subjects with transfemoral amputation fitted with an Energy Storing And Return foot (ESAR) and Microprocessor controlled Prosthetic Knee (MPK) versus a control group of seventeen asymptomatic subjects. Lower limb joint and individual limb power and work were compared between prosthetic, contralateral and control limbs.

The prosthesis dissipates less energy than the joints of the lower limb of the control group when descending the slope, but the demand on the contralateral limb is limited by a lower speed and step length. The huge deficit of positive work produced by the prosthetic ankle cannot be compensated by the residual hip during level and slope ascent which transfers the demand for energy production to the contralateral limb up to 40% on a 12% slope.

This study highlights that prosthetic devices (ESAR foot and MPK) for persons with above-knee amputation present some limitations during slope walking that cannot be compensated by the residual hip and increase the work performed by the contralateral limb.

Keywords

Transfemoral amputation, lower limb, slope, locomotion, biomechanics

1. Introduction

Ascending and descending slopes and stairs is a greater biomechanical challenge than walking over ground. To raise or lower the body's center of mass (BCOM) at each step, humans must perform net positive mechanical work to walk uphill and net negative mechanical work to walk downhill (Franz et al., 2012). These variations of work can be quantified from the mechanical powers mainly performed by the lower limb. Two approaches were proposed in the literature. The first approach consists in computing joint work and each joint's contribution to total lower limb work (Alexander et al., 2017). Alexander quantified the lower limb joint work and the joint work contributions of asymptomatic subjects at different inclinations. The ankle, knee and hip joints respectively contributed between 40-62%, 17-28%, and 18-32% to the total positive joint work whereas the negative work was mainly performed by the knee (35-70%) especially during slope descent. But, the sum of the lower limb joint work fails to capture significant work performed elsewhere in the body, partly due to the rigid body assumption (Zelik and Kuo, 2010).

The second approach, the individual limb method was proposed by Donelan to directly quantify the work performed by the leading and the trailing leg during step-to-step transition in level walking. The mechanical work computed with this method was highly correlated to the metabolic energy consumption under varying walking speeds and step lengths conditions (Donelan et al., 2002) for asymptomatic subjects. According to Jeffers, the mechanical power during step-to-step transitions accounts for 65% of metabolic power in varying slopes and velocities conditions (Jeffers et al., 2015).

Due the loss of one of their lower limbs, individuals with unilateral amputation must rely on their prosthetic device to provide the necessary mechanical work when walking up and down a slope. Energy Storing and Return (ESAR) feet present a limited range of motion and a reduced ankle propulsion power. People with transtibial amputation (TTA) can increase the residual knee flexion at the prosthetic side to adjust to the slope gradient in descent and largely use hip and knee extensors to lift their BCOM in ascent (Langlois et al., 2014). On the contrary, powered feet can achieve a significant ankle propulsion and improve mechanical work symmetry between the legs during uphill and downhill walking (Jeffers and Grabowski, 2017). Thus, powered ankle feet have the potential to reduce metabolic costs during uphill walking (Montgomery and Grabowski, 2018), but clinical trials investigating powered ankle feet has resulted in controversial outcomes (Ingraham et al., 2018). This could be because the ankle–foot prosthesis does not span the knee joint and fails to replicate the function of the biarticular gastrocnemius (Müller et al., 2019; Pickle et al., 2017; Quesada et al., 2016).

Slope walking has been less investigated for people with TransFemoral Amputation (TFA). Vrieling et al. quantified the lower limb joint kinematics for seven TFA and showed that their prosthetic knee joint was not able to adapt to the slope gradient (Vrieling et al., 2008). The use of a microprocessor controlled prosthetic knee (MPK) was evaluated by Burnfield et al., who showed that MPKs permit a faster ascent and descent of slope and that users reported more confidence and stability (Burnfield et al., 2012). Advanced MPK improved knee flexion during stance and swing and decreased the use of handrails suggesting more confidence in the device (Bell et al., 2016). One active powered knee (PWRK) was evaluated by Wolf et al., but this device did not showed improvement in slope ascent or descent compared to MPK (Wolf et al., 2012). During slope descent, subjects with TFA absorbed less energy in their trailing prosthetic limb and maintained a normal intact limb loading by decreasing their walking speed and their intact step length (Morgenroth et al., 2018).

To compensate for the lack of some functionalities of their prosthetic device compared to native joints, compensation strategies have been reported during both the stance and the swing phases.

A larger range of motion of the lateral pelvis tilt was reported by Acasio for people with TFA walking upslope indicating a hip-hiking strategy in which they raise the pelvis on the prosthetic side. Moreover, a larger axial rotation of the trunk could be a compensating strategy to generate power (Acasio et al., 2019). During the prosthetic swing phase, to increase the distance between the prosthetic foot and the ground, people with TFA could perform hip circumduction or vaulting on their contralateral side (Drevelle et al., 2014).

To date, lower limb joint power has only been reported on slopes by (Wolf et al., 2012) for people with TFA, but neither individual limb power nor joint work contributions have previously been investigated. Our aim was to quantify joint powers (and works) for both the prosthetic and the contralateral limbs during slope walking for subjects with TFA wearing MPK and ESAR feet. The main hypothesis is that the prosthetic joints will perform less work than the intact ones. Our purpose is to understand how people with TFA compensate for the limits of their prosthetic limb with the other parts of their body. This was estimated by computing joint powers (and works) and using the individual limb method to quantify how these works can contribute to dynamics of the body center of mass.

2. Materials and methods

2.1 Subjects

The protocol was approved by the local ethics committee and written informed consents were obtained from all participants. Twelve subjects with transfemoral amputation participated in the study (TFA Group). The population is presented in detail in Table 1. All participants underwent clinical evaluation to check for the absence of pain or any gait problems before recruitment. All participants used a MPK knee and an ESAR foot and they all have more than one year of experience using a prosthesis. All subjects were experienced prosthetic users walking with their prosthesis all day without using walking aids on slope and on level ground. Seventeen able-bodied participants, who did not have any orthopedic or neurologic disorders (age: mean 42 years SD 19 years, height: mean 176 cm SD 11 cm, body mass: 72 kg SD 15 kg) were recruited as a control population (AB Group).

PLEASE INSERT TABLE 1

2.2 Protocol

All subjects followed the same protocol. They were equipped with a set of 54 reflective markers placed on body landmarks (Pillet et al., 2014). 3D positions of these markers during motion were captured with an optoelectronic system (Vicon 8i, 100 Hz, Oxford Metrics, Oxford, UK) with two force platforms (AMTI, 100 Hz, Watertown, MA, USA). Subjects walked at their comfortable self-selected speed on a flat surface (level walking), on a 5% inclined ramp device (gentle slope) and on a 12% inclined ramp device (steep slope). For the ramps devices, modular structures with independent blocks adapted to the geometry of the force plates were designed according to the principles proposed by Dixon and Pearsall (Dixon and Pearsall, 2010). At least three valid trials were recorded. A trial was considered as successful when each foot of the participant was in full contact with each force platform.

2.3 Data processing

A 13 segments model was created (foots, shanks, thighs, pelvis, trunk, head, arms, lower arms). Anatomical frames were defined for each segment of the model in Matlab (MathWorks Inc, Natick, MA, USA). Spatiotemporal parameters and lower limb joint kinematics and kinetics in the frontal, transverse and sagittal planes were computed in each walking situation (flat surface, gentle slope ascent & descent and steep slope ascent & descent). For prosthetic segments, anatomical landmarks and frames were placed by symmetry with the contralateral limb. An anthropometric model of each subject was also built and personalized using the method described by (Pillet et al., 2010). This model allowed the estimation of body segment inertial parameters useful for inverse dynamic calculation. Joint powers were then computed as the dot product of joint moment and relative angular velocity of the distal segment relatively to the proximal segment of the joint (Gordon et al., 1980). The sum of the ankle, knee and hip power was also assessed as the summed lower limb joints power.

At the same time, position and velocity (by derivation) of the body center of mass could also be obtained in the global reference frame. Then, the mechanical power of each individual limb was assessed by the dot product of the velocity of the body center of mass and the resulting ground reaction forces on the considered lower limb. Individual limb power was compared to the above mentioned summed lower limb joints power. Positive and negative works were calculated by numerical integration of powers over the whole gait cycle.

2.4 Statistics

Repeated measures 2x2 ANOVAs (limb x slope), using subject as a random variable, were used to determine limb effects (intact, prosthetic, contralateral) and slope effects on spatiotemporal and work measures. Estimated marginal means post-hoc tests were used to delineate significant differences in these measures with a threshold of significance level of $p < 0.05$.

3. Results

The results are reported separately for the control group and for both limbs of the TFA group. In all figures and tables, the results represent the means and standard deviations computed across all subjects of each group. Participants with TFA presented lower speeds than the AB subjects. The difference reaches statistical significance, except during the 5% downhill walking (table 2). Contralateral step length is significantly smaller than the step length of AB participants except during the 5% downhill walking. The prosthetic step length is also significantly smaller than the step length of AB participants for the 12% slope (table 3).

PLEASE INSERT TABLE 2 and 3

Lower limb joint power curves are presented for each slope for the AB group and for both limbs for the TFA group (figure 1). For the AB limb, the global pattern of the joint power curves as a function of the percentage of gait cycle is not affected by the slope. Generally, for each joint, the value of the positive peaks increases while climbing the slope and the value of the negative ones increases while descending the slope. For the TFA group, these curves show a strong asymmetry for all joints between the prosthetic limb and the contralateral limb.

PLEASE INSERT FIGURE 1

The ankle of the AB subjects was the joint that produced the most positive work in all walking conditions (figure 2). This positive work ranged from 0.31 +/- 0.08 J/kg for level walking to 0.49 +/- 0.14 J/kg during the 12% uphill walking. The positive work of the ankle increased by 32% for the contralateral leg reaching 0.61 +/- 0.17 J/kg during the 12% uphill walking as compared to level walking while the prosthetic ankle provided only 0.11 +/- 0.03 J/kg and remained quite constant regardless of the degree of the slope.

The knee of the AB subjects produced only 0.14 +/- 0.04 J/kg of positive work and -0.34 +/- 0.07 J/kg of negative work during level walking. The positive work increased to 0.23 +/- 0.09 J/kg during 12% uphill walking as compared to level walking and the negative work reached -0.66 +/- 0.16 J/kg during downhill walking. Positive work was statistically greater for the contralateral limb than for the AB subjects and absent for the prosthetic limb. Negative work was statistically lower for the prosthetic limb than for the AB subjects.

The positive work of the hip in the AB group increased with the degree of slope from 0.21 +/- 0.08 J/kg during level walking to 0.48 +/- 0.19 J/kg during upslope walking while the negative work remained constant across the walking situations. This work showed a tendency to be lower for the hip of the prosthetic limb compared to AB but the difference is only significant for the 12% incline.

PLEASE INSERT FIGURE 2

The summed lower limb joints powers curve is very similar to the external mechanical power performed by each individual leg on the COM (individual limb power) for the AB limb and for the contralateral limb (figure 3). As for the individual joint power, the values of the positive peaks increased during walking upslope and the values of negative peaks increased during walking downslope. Both power curves (summed lower limb joints and individual limb) highlight the asymmetry between the prosthetic and the contralateral limbs. For the prosthetic limb, however, differences appear between the summed ankle–knee–hip power and the external mechanical power performed by the prosthetic limb, which could suggest that other joints contribute to the mechanical work performed at the center of mass.

PLEASE INSERT FIGURE 3

The positive and negative works computed along the stance phase were compared for the summed ankle–knee–hip power and the individual limb power and are presented in figure 4. For the AB group, the individual limb method estimated equal positive and negative works (0.34 +/- 0.08 J/kg) during level walking. In Comparison, the summed ankle–knee–hip power method estimated higher positive work (0.41 +/- 0.11 J/kg) and lower negative work (-0.30 +/- 0.08 J/kg) respectively. As expected, when inclination increased, the individual leg positive work increased, and the individual leg negative work decreased in accordance with the variation of potential energy. This is also the case for the TFA group, but with an increasing asymmetry between the prosthetic and the contralateral limbs with the inclination of the slope.

PLEASE INSERT FIGURE 4

4. Discussion

Walking on level ground results in the production of the same amount of positive and negative work. Walking on a slope causes a variation in potential energy resulting in an imbalance between the required positive and negative works to be produced. The joints of the lower limb adapt to fit these requirements. Our findings are in agreement with previously-reported results for the able-bodied subjects (Alexander et al., 2017; Shepherd et al., 2020). The values of positive and negative joint power peaks increased while walking upslope and downslope, respectively. The joint that produced the larger part of positive power during late stance was the ankle, but the increase of power performed during upslope walking was more pronounced for the hip joint. During the 12% uphill, the positive work produced by the hip (0.48 +/- 0.19 J/kg) was close to the work produced by the ankle (0.49 +/-0.13 J/kg). The knee joint produced the larger negative power and the amount of power performed by the knee was the most affected compared to other joints by the slope gradient during downhill walking. Knee negative work during 12% downhill reached 0.66 +/-0.16 J/kg.

As hypothesized in the introduction, the prosthetic limbs were not able to reproduce the power patterns observed in the AB group. The prosthetic ankle feet presented a lower propulsive power compared to sound ankles in all walking situations. The maximum of positive work produced by the prosthetic ankles reached 0.12 +/-0.04 J/kg during 12% uphill which was four times less than the work produced by the AB ankle (0.49 +/-0.13 J/kg). This value is lower than the results reported in the literature for below knee amputees for equivalent energy storing and return (ESAR) feet (Jeffers and Grabowski, 2017). This work has never been reported before for transfemoral amputees walking on slopes. The decreased propulsive work produced at the ankle joint can be explained by the absence of flexion of the prosthetic knee during stance when climbing a slope. Indeed, in this position of the limb, the ability of the prosthetic foot to store and release energy could be hindered. It should be noticed that even with a motorized knee and ankle prosthesis, this strategy (extended knee joint during stance) is still observed (Azocar et al., 2020). In the literature, one active powered knee (PWRK) was evaluated by Wolf et al., but the study did not conclude in an improvement in slope ascent or descent compared to MPK (Wolf et al., 2012). In addition, microprocessor controlled prosthetic knees do not produce any positive power increasing the loss of propulsive work especially during uphill walking. Negative power from the prosthetic knee was increased only during the 12% downhill. In this situation, the energy dissipated in the prosthetic knee reached 0.45 +/- 0.16 J/kg which was less than for the one estimated in the AB group (0.66 +/- 0.16 J/kg).

During upslope walking, the positive hip work of the residual limb was 68% of the work produced by the hip of AB subjects. This is consistent with force deficit at the hip of transfemoral amputees which has been quantified to about 30% of the maximum isometric force compared to able bodied subjects (Heitzmann et al., 2020). The huge deficit of positive work produced by the prosthetic ankle cannot be compensated by the residual hip which led to a summed ankle–knee–hip work of 0.12 +/- 0.07 J/kg on level ground and 0.26 +/- 0.14 J/kg during upslope walking. The summed lower limb joints work produced by TFA is thus about four times lower compared to the one for AB limb (AB limb: 0.41+/- 0.11 J/kg for level ground and 0.96 +/- 0.16 J/kg during upslope walking). In addition, the work estimated using the summed ankle–knee–hip power is clearly lower than the work estimated using the individual limb method which suggests that mechanical work is performed on the center of mass by other joints than the ones of the lower limb during the prosthetic stance. In particular, the literature reports compensation at the level of the pelvis and the trunk (Acasio et al., 2019), that could be at the origin of this discrepancy.

The important demand on the contralateral limb was reported for level walking by Nolan and Lees (Nolan and Lees, 2000). In the present study, the positive work estimated by the individual limb method for the contralateral limb reached 1.32 +/- 0.23 J/kg during upslope walking which corresponds to an increase of 40% compared to AB subject in the same situation despite of a lower gait speed. A major part of this increase comes from the ankle which can present a premature positive power at midstance which is specific of vaulting gait (Villa et al., 2015). Positive and negative hip work at the contralateral limb were similar to the ones in the AB group whereas the negative work performed by the contralateral knee was lower (0.55 +/-0.16 J/kg) which could be due to the lower gait speed (Morgenroth et al., 2018). This is of interest because long-term exposure of the contralateral limb to high repetitive demands can lead to the degeneration of weight-bearing joints and subsequent joint pain (Gailey et al., 2008).

The limitations of available prosthetic components for above-knee amputees result in significant compensations during gait. The present study highlights these limitations by quantifying the mechanical work performed in slope situations despite the various technological developments. The reduced prosthetic push-off and the asymmetry between limbs may also explain the increased metabolic consumption of people with transfemoral amputation (Caputo and Collins, 2014; Huang et al., 2015). The technological improvements of prosthetic devices have brought safety by limiting the risk of falling and improving the quality of life but it is not clear if these improvements allowed to mitigate the increase in the energy cost of gait and overloading of the contralateral limb. Thus, there is a clear need to develop more energetically efficient solutions. Providing energy via motorization could allow to improve propulsion by the prosthetic leg (Pröbsting et al., 2020). However, the constraints related to the complexity of motorized systems still limit their use in real life. Different alternative approaches to full motorization are emerging today to improve the adaptation of prosthetic components to different locomotion situations (Lecomte et al., 2021; Lenzi et al., 2017). However, to propose efficient solutions beneficial to the locomotion, specific attention must be paid to the way the power generated at the ankle is transmitted to the body via the prosthetic knee. This requires the development of solutions that must be specific to femoral amputees. This study can be used as a basis to objectivize an improvement on the power and the work done by the prosthesis, the residual limb, or the contralateral limb.

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Table 1 : Detailed information of the participants with a transfemoral amputation (TFA) and their current prostheses.

Person with TFA	Cause for Amputation	Sex	Age (y)	Height (cm)	Body mass (kg)	Prosthetic knee	Prosthetic foot
1	Trauma	M	59	169	77	C-Leg (Ottobock)	1C40 (Ottobock)
2	Trauma	M	47	173	92	C-Leg (Ottobock)	Variflex LP (Ossur)
3	Trauma	M	46	185	81	C-Leg (Ottobock)	1C60 (Ottobock)
4	Tumor	M	38	175	79	C-Leg (Ottobock)	Variflex LP (Ossur)
5	Trauma	M	49	181	84	C-Leg (Ottobock)	1C40 (Ottobock)
6	Trauma	M	32	171	78	C-Leg (Ottobock)	Variflex LP (Ossur)
7	Trauma	M	47	170	85	C-Leg (Ottobock)	Variflex LP (Ossur)
8	Tumor	F	30	169	50	C-Leg (Ottobock)	1C40 (Ottobock)
9	Trauma	M	28	181	82	C-Leg (Ottobock)	Variflex LP (Ossur)
10	Trauma	M	27	174	64	Hybrid Knee (Nabtesco)	Variflex (Ossur)
11	Trauma	M	40	165	52	RheoKnee (Ossur)	Reflex (Ossur)
12	Trauma	M	33	177	63	Hybrid Knee (Nabtesco)	Variflex LP (Ossur)

Table 1 : Walking speed comparison mean (sd) for the participants with a transfemoral amputation (TFA) and the able-bodied participants (AB group)

Slope	**Walking speed (m/s)**		p value
	TFA Group	**AB Group**	
-12%	1.09 (0.17)	1.29 (0.23)	0.009
-5%	1.18 (0.20)	1.31 (0.21)	0.086
0%	1.24 (0.16)	1.38 (0.14)	0.009
5%	1.16 (0.17)	1.30 (0.18)	0.036
12%	1.04 (0.13)	1.28 (0.19)	0.001

Table 2 :Step length comparison. * indicates a significant difference ($p<0.05$) between AB and prosthetic limb, $^{\$}$ between AB and contralateral limb and $^{\#}$ between prosthetic and contralateral limb

Slope	**Step length (m)**		
	prosthetic limb	**contralateral limb**	**AB limb**
-12%	0.62 (0.09)$^{\#}$	0.55 (0.08) $^{\$\#}$	0.65 (0.09) $^{\$}$
-5%	0.64 (0.10)	0.62 (0.09)	0.67 (0.07)
0%	0.68 (0.07)	0.63 (0.07) $^{\$}$	0.70 (0.06)$^{\$}$
5%	0.68 (0.09)	0.62 (0.11) $^{\$}$	0.70 (0.07)$^{\$}$
12%	0.65 (0.09) *	0.62 (0.07) $^{\$}$	0.71(0.08)$^{*\$}$

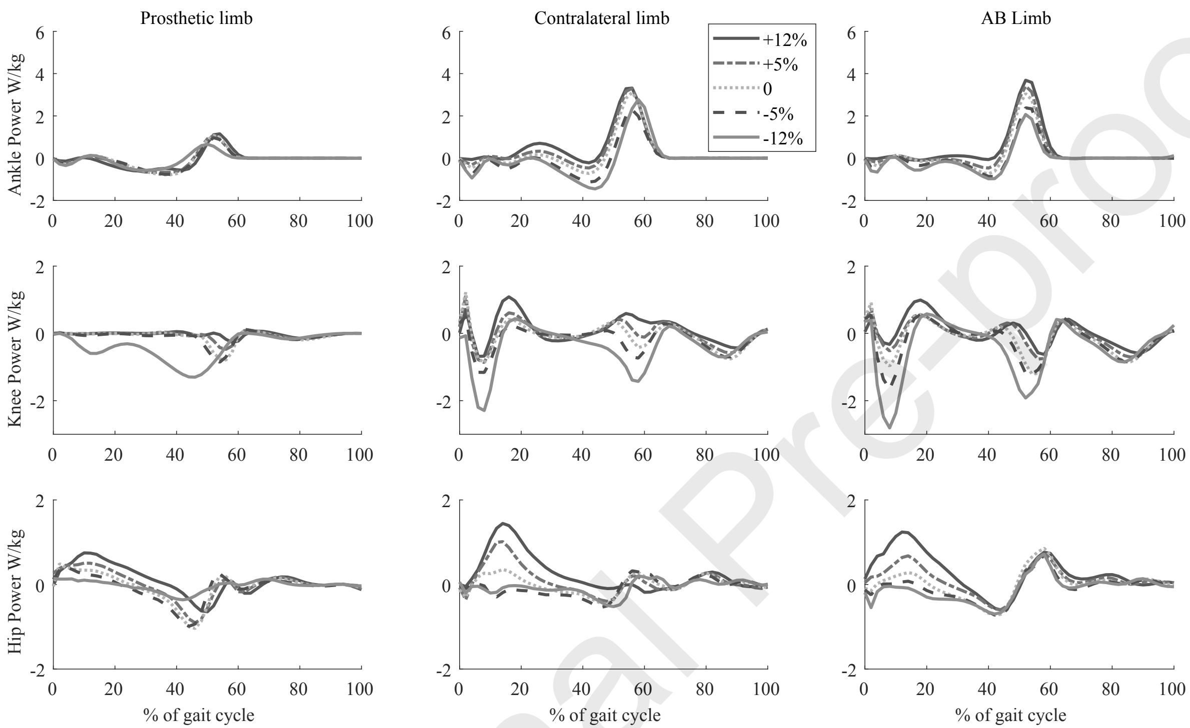

Figure 1: Ankle, knee and hip joint power during downhill and uphill walking throughout gait cycle. Mean curves obtained for the prosthetic (left), the contralateral (middle) and the asymptomatic limb (right) on all slope inclination: +12% (dark grey solid line), +5% (dash-dotted line), 0% (dotted line), -5% (dashed line), -12% (light grey solid line).

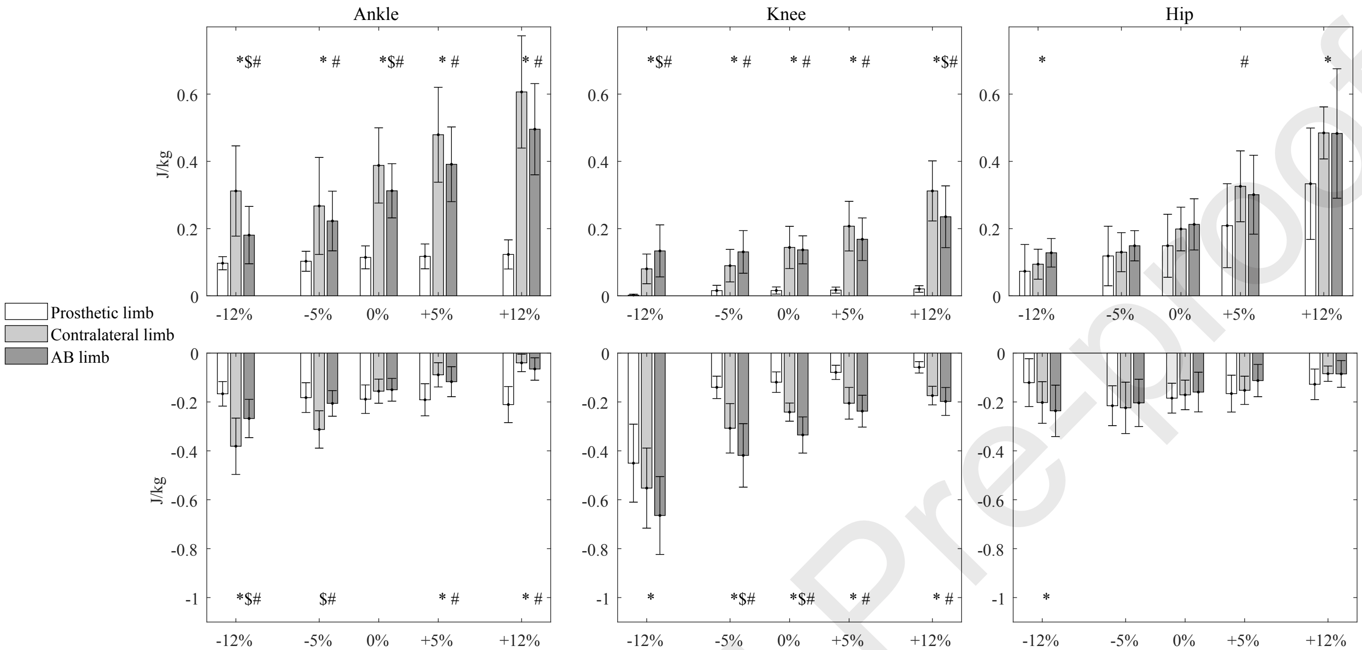

Figure 2: Ankle, knee and hip positive and negative work during downhill and uphill walking throughout the gait cycle for the prosthetic (white), contralateral (light grey) and AB limb (dark grey). * indicates a significant difference between AB and prosthetic limb, $^{\$}$ between AB and contralateral limb and $^{\#}$ between prosthetic and contralateral limb

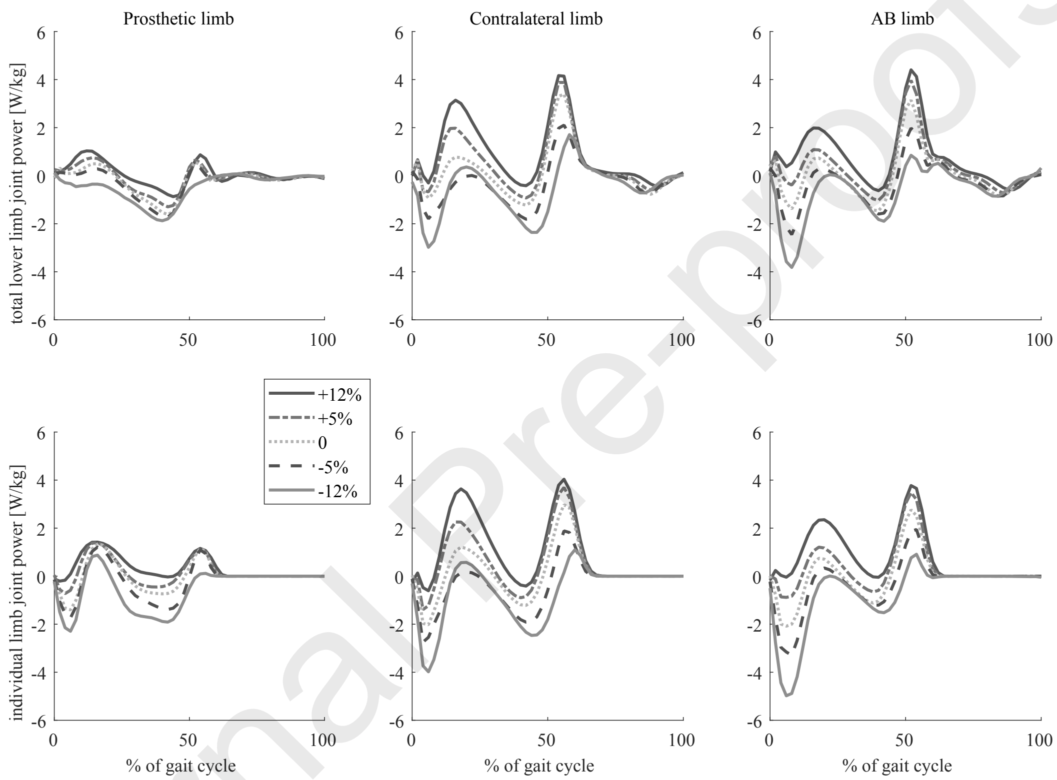

Figure 3: Summed lower limb joints power and Individual limb power during downhill and uphill walking.

Mean curves obtained for the prosthetic (left), the contralateral (middle) and the asymptomatic limb (right) on all slope inclination: +12% (dark grey solid line), +5% (dash-dotted line), 0% (dotted line), -5% (dashed line), -12% (light grey solid line).

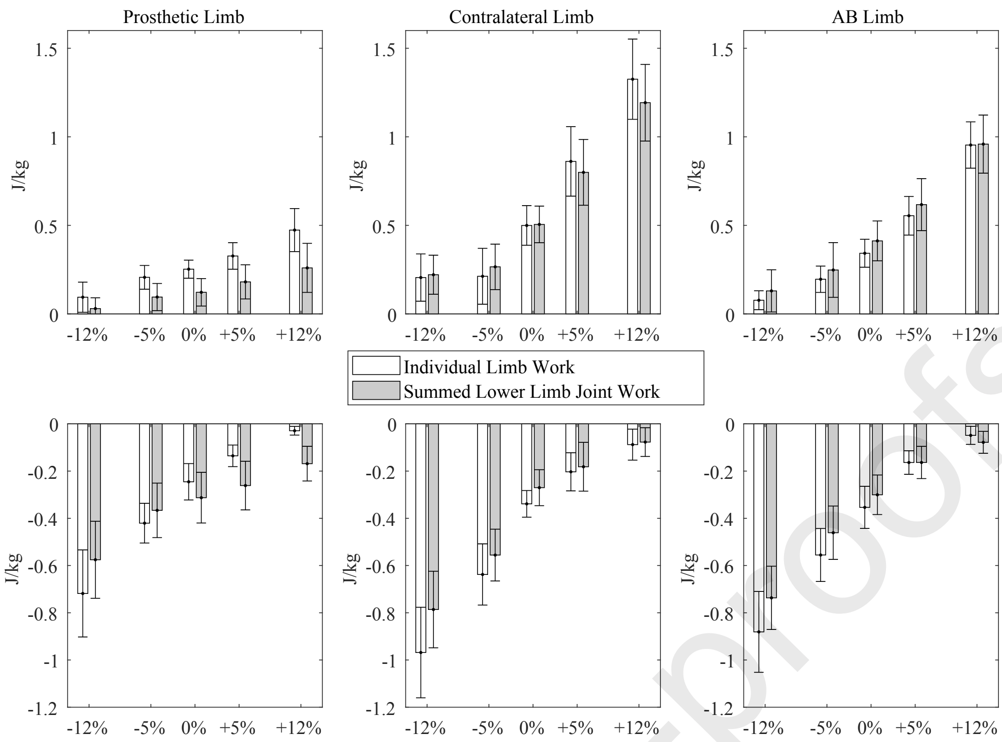

Figure 4: Positive and negative work obtained via the individual limb method (white) and using the summed lower limb joints power curve (light grey) during downhill and uphill walking throughout the gait cycle.