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Quantification of push-off and collision work during step-to-step transition in amputees walking at self-selected speed: Effect of amputation level

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ABSTRACT

Maintaining forward walking during human locomotion requires mechanical joint work, mainly provided by the ankle–foot in non-amputees. In lower-limb amputees, their metabolic overconsumption is generally attributed to reduced propulsion. However, it remains unclear how altered walking patterns resulting from amputation affect energy exchange. The purpose of this retrospective study was to investigate the impact of self-selected walking speed (SSWS) on mechanical works generated by the ankle–foot and by the entire lower limbs depending on the level of amputation. 155 participants, including 47 non-amputees (NAs), 40 unilateral transtibial amputees (TTs) and 68 unilateral transfemoral amputees (TFs), walked at their SSWS. Positive push-off work done by the trailing limb (W^{+}_{StS}) and its associated ankle–foot ($W^{+}_{ankle-foot}$), as well as negative collision work done by the leading limb (W^{-}_{StS}) were analysed during the transition from prosthetic limb to contralateral limb. An ANCOVA was performed to assess the effect of amputation level on mechanical works, while controlling for SSWS effect. After adjusting for SSWS, NAs produce more push-off work with both their biological ankle–foot and trailing limb than amputees do on prosthetic side. Using the same type of prosthetic feet, TTs and TFs can generate the same amount of prosthetic $W^{+}_{ankle-foot}$, while prosthetic W^{+}_{StS} is significantly higher for TTs and remains constant with SSWS for TFs. Surprisingly and contrary to theoretical expectations, the lack of propulsion at TFs' prosthetic limb did not affect their contralateral W^{-}_{StS}, for which a difference is significant only between NAs and TTs. Further studies should investigate the relationship between the TFs' inability to increase prosthetic limb push-off work and metabolic expenditure.

1. Introduction

Despite significant advances in care and fitting, lower limb amputation still reduces quality of life, autonomy and mobility (van Velzen et al., 2006), particularly in transfemoral amputation (Davies and Datta, 2003; Sawers and Hafner, 2013). Amputees face elevated oxygen costs, with an increase around 50 % for transfemoral amputees (TFs) and 25 % for transtibial amputees (TTs) compared to non-amputees (NAs) (Ettema et al., 2021; van Schaik et al., 2019), leading to reduced walking speeds to maintain an oxygen consumption equivalent to NAs (Robert and Waters, 1992).

Modelling human walking as an inverted pendulum suggests that the step-to-step transition is an important determinant of the energy cost of walking (Kuo et al., 2005). Indeed, during this phase, mechanical work is needed to redirect the Body Center of Mass (BCoM) velocity and accelerate the BCoM following the loss of velocity when the leading limb collides with the ground. Donelan's collision model predicts a rate of collision work proportional to the fourth power of step length (Donelan et al., 2002a). In healthy subjects, the variation of kinetic energy at the BCoM results from the propulsive power done by the trailing limb in late stance. This is mainly provided by the ankle–foot which generates net positive joint work per gait cycle (Winter, 1983) and, to a lesser extent, by the hip (Zelik and Adamczyk, 2016). Trailing limb push-off work can help reducing collision losses and thus reduce metabolic demand (Houdijk et al., 2009; Kuo et al., 2005).

Based on this observation, authors have assumed that the oxygen

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overconsumption of amputees could be attributed to a lack of propulsion at the prosthetic ankle–foot system. Indeed, although the Energy-Storage-And-Return (ESAR) prostheses seek to reproduce the function of the ankle–foot system, they cannot provide positive net mechanical work. Consequently, there is a reduction of the push-off work at the prosthetic ankle–foot (Russell Esposito et al., 2015; Seroussi et al., 1996), leading to asymmetrical gait and augmented loading on the contralateral side (Morgenroth et al., 2011; Pröbsting et al., 2022). As a result, researchers and manufacturers have developed increasingly effective passive (Fey et al., 2012; Heitzmann et al., 2018; Zelik et al., 2011) and active ankle–foot devices (Pröbsting et al., 2022; Russell Esposito et al., 2015). It has been found that prosthetic energy storage and return as well as prosthetic limb push-off work increased as the stiffness of the prosthetic foot decreased (Fey et al., 2011; Zelik et al., 2011). Meanwhile, the link between ankle–foot power and metabolic expenditure has been investigated in TTs (Caputo and Collins, 2015; Ingraham et al., 2018; Kim et al., 2021; Quesada et al., 2016; Segal et al., 2012) and TFs (Graham et al., 2008; Macfarlane et al., 1997) but so far the results are inconclusive regarding the positive effect of an increased ankle–foot power. In their literature review, Müller et al. (2019) reported that 1. ankle–foot power was positively correlated with walking speed in NAs and TTs using ESAR or motorized feet and 2. motorized feet and some recent ESAR with specific designs can deliver as much power as NAs ankle–foot However, the methods used for calculating the push-off power vary between the studies included in this literature review which could bias comparisons and lead to incorrect conclusions (Zelik and Honert, 2018). Indeed, two methods are commonly used to compute ankle–foot power: Ankle Joint Method (AJM) with either 3 or 6 DOF (Buczek et al., 1994) and Distal Shank Method (DSM) (Prince et al., 1994; Takahashi et al., 2012). The AJM makes rigid body assumptions, whereas the DSM considers the ankle–foot as a deformable body. It is already known that the AJM tends to overestimate the power generated during the push-off phase for NAs and underestimate it for prosthetic feet, compared to DSM (Farinelli et al., 2019; Heitzmann et al., 2018; Zelik and Honert, 2018). Besides, Müller did not report any studies with TFs. As most prosthetic feet are not specific to any amputation level, it would be interesting to compare the prosthetic ankle–foot power of TFs and TTs wearing similar prosthetic feet.

In addition to the quantification of ankle–foot power, the Individual Limb Method (ILM) (Donelan et al., 2002) can be used to quantify the power and work performed by the entire leading and trailing legs during step-to-step transition. Using this computation mode, it has already been shown that push-off work done by trailing limb increases with speed for NAs (Donelan et al., 2002a). On average, about 50 % less push-off work is done by prosthetic limb (Adamczyk and Kuo, 2015), with an increase of push-off work with speed for TTs whereas the work remains nearly constant for TFs (Bonnet et al., 2014; Pinhey et al., 2022). The decrease of push-off work done by TTs' prosthetic limb leads to an increase of contralateral limb collision work (Adamczyk and Kuo, 2015; Herr and Grabowski, 2012), which was found to be correlated with an increase of metabolic cost (Houdijk et al., 2009).

The differences reported in the literature about the mechanical power and work done by the prosthetic ankle–foot, on one hand, and the prosthetic limb, on the other hand, between amputees and non-amputees may be due to 1. differences in calculation methods, 2. the level of amputation and/or 3. an effect of differences in walking speed between groups.

In this paper, we propose to study a large cohort of 155 subjects walking at self-selected walking speed (SSWS), including NAs, TTs and TFs. The inertial model and the methods for calculating mechanical power and work are common to all subjects. The aim of this study is to investigate the positive push-off work done by the trailing prosthetic limb and the prosthetic ankle–foot, as well as the negative collision work done by the leading contralateral limb for TTs and TFs. These mechanical works will be compared between groups and to the ones performed by NAs.

2. Methods

This study is a retrospective study. Data were previously collected at IBHGC Paris, INI\CERAH Créteil and IRR Nancy between 2010 and 2021 after approval by the Ethics Committee (*Comité de Protection des Personnes* CPP n°2011-A00133-38, n°2018-A01543-52 and n°NX06036) and written informed consent (Facione et al., 2019; Pillet et al., 2014).

2.1. Subjects

Analysis was performed on data from 47 non-amputee participants, 40 transtibial amputees and 68 transfemoral amputees for a total of 155 subjects. Participant demographic information are in Table 1; details in **Supplementary Table 4**. The amputees' mobility level (K-Level) was either 3 or 4 (Borrenpohl et al., 2016), also characterised by a level of activity greater than or equal to d4602 (World Health Organization, 2001). Amputee participants were unilateral amputees all walking with an ESAR prosthetic foot and a microprocessor-controlled prosthetic knee for TFs. Each amputee participant was clinically cleared for independent walking and had used their prosthesis for a minimum of four weeks prior to data collection.

2.2. Experimental protocol and instrumentation

Kinematic data were collected using an optoelectronic system (Vicon, Oxford, UK) with at least 10 infrared cameras sampled at 100 Hz. The Conventional Gait model (Leboeuf et al., 2019) was used as marker set and internal condyles and malleoli were added. For the prosthetic side, the midpoint of markers placed at the extremity of the mechanical axis was used to define the knee joint center. In the same way, the ankle joint center was the midpoint between markers placed laterally and medially at the distal end of the rigid pylon. Ground reaction forces (GRF) were measured at 1000 Hz using at least two force plates (AMTI) embedded in the middle of the walkway. Each participant adopted a SSWS and at least three successful walking cycle were

Table 1
Summary of characteristics of participants: 47 non-amputees (NAs), 40 unilateral transtibial amputees (TTs) and 68 unilateral transfemoral amputees (TFs). Continuous variables are expressed as mean values (±standard deviation) and group differences were evaluated using one-way ANOVA. If a significant difference (overall p-value < 0.05) was found, a multiple-comparison post-hoc was done.

Group	NAs N = 47	TTs N = 40	TFs N = 68	Overall p-value
Gender				
Male	33 (70 %)	38 (95 %)	62 (91 %)	
Female	14 (30 %)	2 (5 %)	6 (9 %)	
Age [years]				
Mean (±SD)	35 (±17)	45 (±12)	43 (±13)	< 0.05 [a, b]
Height [cm]				
Mean (±SD)	173 (±9)	176 (±7)	178 (±9)	< 0.05 [b]
Mass [kg]				
Mean (±SD)	69 (±13)	83 (±17)	80 (±15)	< 0.001 [a, b]
Side of amputation				
Right	N/A	19 (48 %)	22 (32 %)	
Left	N/A	21 (52 %)	46 (68 %)	
Cause of amputation				
Trauma	N/A	31 (78 %)	55 (81 %)	
Cancer	N/A	2 (5 %)	9 (13 %)	
Congenital	N/A	2 (5 %)	3 (4 %)	
Vascular	N/A	4 (10 %)	1 (1 %)	
N.D.	N/A	1 (3 %)	0 (0 %)	
Time since amputation [years]				
Mean (±SD)	N/A	7 (±7)	12 (±12)	< 0.05 [c]

N/A: Not Applicable. N.D.: No Data.
[a] Significant difference between NAs and TTs ($p < 0.05$).
[b] Significant difference between NAs and TFs ($p < 0.05$).
[c] Significant difference between TTs and TFs ($p < 0.05$).

recorded for each limb.

2.3. Data analysis

Ankle-foot power refers to the power generated by the part of the limb distal to the shank. We have used the DSM that we can summarize as the power associated to the 6 DOF link resulting from the components between shank and ground. The shank was considered to be a rigid solid. The distal part was considered as a deformable solid without mass as suggested by Zelik and Honert, 2018. This power $P_{ankle-foot}$ has been computed according to the equation given by Farinelli et al., 2019:

$$P_{ankle-foot} = \overrightarrow{F_{GRF/R}} \cdot \overrightarrow{V_{A\in shank/R}} + \left(\overrightarrow{M_{A,F_{GRF/R}}} + \overrightarrow{M_{free}}\right) \cdot \overrightarrow{\Omega_{shank/R}} \quad (1)$$

With R the global frame; A a point belonging to the shank and defined as the middle of the malleolar markers; $\overrightarrow{V_{A\in shank/R}}$ the velocity of A in R; $\overrightarrow{M_{A,F_{GRF/R}}}$ the moment due to the GRF in relation to A; $\overrightarrow{M_{free}}$ the free moment of GRF and $\overrightarrow{\Omega_{shank/R}}$ the rotational velocity of the shank segment.

For the computation of the limb power, the ILM (Donelan et al., 2002b) was used to directly quantify the power generated by the leading and the trailing limb individually:

$$P_{trail} = \overrightarrow{F_{ground\rightarrow trail\ \ limb}} \cdot \overrightarrow{V_{BCoM/R}} \quad (2)$$

$$P_{lead} = \overrightarrow{F_{ground\rightarrow leading\ \ limb}} \cdot \overrightarrow{V_{BCoM/R}} \quad (3)$$

With $\overrightarrow{F_{ground\rightarrow trailing\ limb}}$ and $\overrightarrow{F_{ground\rightarrow leading\ limb}}$ the GRF acting on the trailing and leading limbs, respectively. The BCoM velocity $\overrightarrow{V_{BCoM/R}}$ was computed as the time derivative of the BCoM trajectory, defined as the barycenter of the centers of mass of the body segments. Segmental center of mass were calculated from body segment inertial parameters as estimated by De Leva (1996). For amputees, the same method was used, replacing the shank and foot masses with those of the knee and foot prostheses used by each subject, respectively.

Mechanical work was computed as the integration of power over time and normalized by the participants' body mass, including prosthesis mass for amputees. During step-to-step transition, the negative work done by the leading limb during collision (approximately 0–15 % of stride) was noted W^-_{StS} and the positive work done by the trailing limb during push-off (45–65 %) was noted W^+_{StS} (Fig. 1). The push-off work done by the ankle–foot was noted $W^+_{ankle-foot}$. For amputees, only the transition from prosthetic limb (trailing limb) to contralateral limb (leading limb) was studied. Finally, a vector was defined between the midpoints of the first and fifth metatarsals of each foot at the moment of contralateral heel strike. The contralateral step length was computed as the norm of the projection of this vector along the antero-posterior axis defining the direction of walking.

2.4. Statistics

Differences in demographic variables (age, height and weight) and SSWS between groups were analysed using one-way ANOVA. If a significant difference (p-value < 0.05) was found, a multiple-comparison post-hoc was done. Linear regression models were fitted to SSWS-work and SSWS-step length relationships for each group of participants, removing outliers with residuals greater than 3 standard deviations in absolute value. Slope and intercept coefficients with their 95 % confidence intervals (CI), coefficient of determination R^2 and p-values are

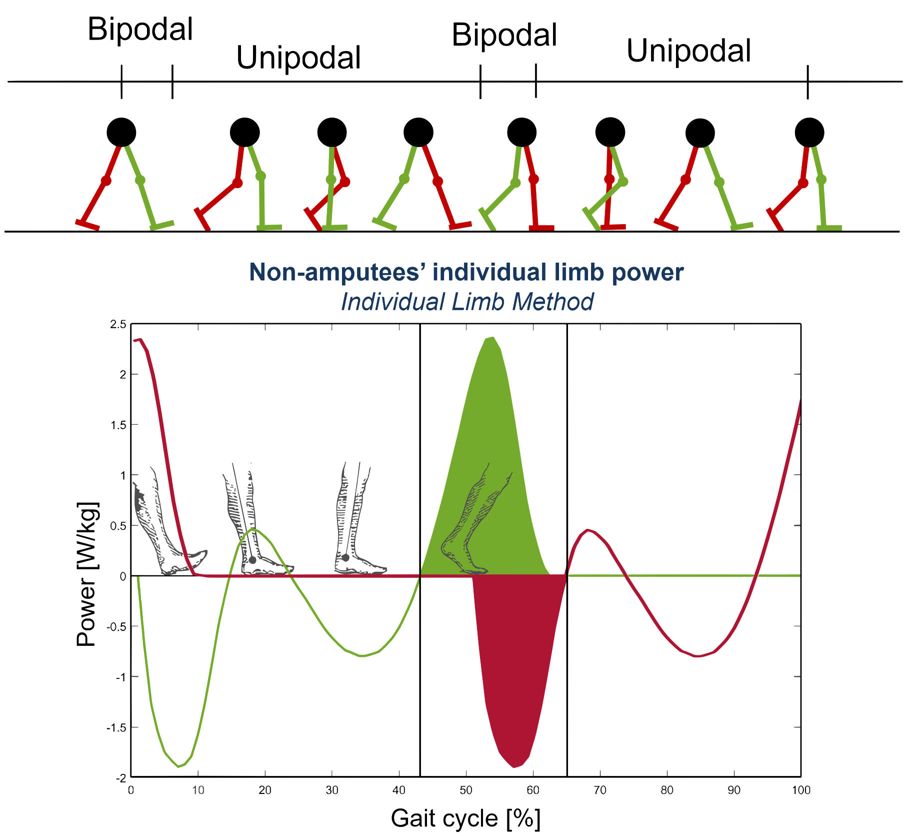

Fig. 1. Representation of non-amputees' individual limb power during gait cycle, computed by the Individual Limb Method (Donelan et al., 2002b). The green line represents the power developed by the trailing leg, and the red line represents the one done by the leading leg. During step-to-step transition (vertical lines), trailing leg generates positive push-off work (green area) and leading leg negative collision work (red area). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

given for each linear best fit approximation. An ANCOVA was performed to assess the effect of groups (categorical independent variable) on W^{-}_{StS}, W^{+}_{StS}, $W^{+}_{ankle-foot}$, and step length (continuous dependent variables) separately, while controlling for the effect of SSWS (continuous covariate). The assumptions of normality of residuals and homogeneity of variances have been verified beforehand with Shapiro-Wilk test and Levene's test, respectively. If the ANCOVA reveals a significant difference, a post-hoc test is performed with a Bonferroni adjustment and the estimated marginal means of each subgroup were compared.

3. Results

3.1. Demographic data

Table 1 presents the subject characteristics. TTs and TFs have no significant difference in terms of age, height and mass. However, NAs were significantly younger and lighter than TTs (respectively $p = 0.005$ and $p < 0.001$) and TFs ($p = 0.013$ and $p < 0.001$). TFs were taller than NAs ($p = 0.028$). The cause of amputation was mostly traumatic ($n_{TT} = 31/40$ and $n_{TF} = 55/68$) and most of the subjects were men ($n_{TT} = 38/40$ and $n_{TF} = 62/68$). The average time since amputation was significantly longer for TFs (12 ± 12 years) compared to TTs (7 ± 7 years).

3.2. Self-selected walking speed and step length

The self-selected walking speed ranged from 1.00 to 1.72 m/s, from 0.93 to 1.52 m/s and from 0.86 to 1.49 m/s for the NA (1.33 ± 0.15 m/s), TT (1.27 ± 0.13 m/s), and TF (1.20 ± 0.15 m/s) groups respectively (Table 2). Significant differences in the SSWS were established between NAs and TFs ($p < 0.001$) and between TTs and TFs ($p = 0.020$) but not between NAs and TTs ($p = 0.216$).

Significant positive correlations between SSWS and contralateral step length were found for all groups (Fig. 2, Table 3). ANCOVA results indicated that the effect of SSWS ($p < 0.001$) and the interaction effect ($p < 0.05$) were significant, while the effect of groups was not ($p = 0.541$), i.e., no significant difference in estimated marginal mean between amputation groups was found once the effect of SSWS has been considered (Table 2).

Table 2
Temporal parameters and mechanical works done during step-to-step transition for NAs, TTs and TFs. For amputees, only prosthetic-to-contralateral limb transition was evaluated. An ANCOVA was performed to assess the effect of groups on $W^{+}_{ankle-foot}$, W^{+}_{StS}, W^{-}_{StS} and contralateral step length separately, while controlling for the effect of self-selected walking speed. If a significant difference (overall p-value < 0.05) was found, a post-hoc test was performed with a Bonferroni adjustment. Step length and mechanical works are expressed as estimated marginal means (±standard deviation). Self-selected walking speeds are expressed as mean values (±standard deviation) and group differences were evaluated using one-way ANOVA. If a significant difference (overall p-value < 0.05) was found, a multiple-comparison post-hoc was done.

	NAs	TTs	TFs	Overall p-value
A: Temporal parameters				
Self-selected walking speed [m/s]				
Mean (±SD)	1.33 ± 0.15	1.27 ± 0.13	1.20 ± 0.15	< 0.001 [b, c]
Step length [m]				
Estimated marginal mean (±SD)	0.66 ± 0.02	0.66 ± 0.02	0.67 ± 0.01	0.541
B: Foot-Ankle (Distal Shank Method)				
Push-off work $W^{+}_{ankle-foot}$ [J/kg]				
Estimated marginal mean (±SD)	0.22 ± 0.02	0.18 ± 0.01	0.16 ± 0.01	< 0.001 [a, b]
C: Trailing leg (Individual Limb Method)				
Push-off work W^{+}_{StS} [J/kg]				
Estimated marginal mean (±SD)	0.28 ± 0.02	0.15 ± 0.01	0.12 ± 0.01	< 0.001 [a, b, c]
D: Leading leg (Individual Limb Method)				
Collision work W^{-}_{StS} [J/kg]				
Estimated marginal mean (±SD)	−0.14 ± 0.02	−0.19 ± 0.02	−0.17 ± 0.02	< 0.001 [a]

[a] Significant difference between NAs and TTs ($p < 0.05$).
[b] Significant difference between NAs and TFs ($p < 0.05$).
[c] Significant difference between TTs and TFs ($p < 0.05$).

3.3. Mechanical works done during prosthetic-to-contralateral limb transition

The push-off work done by the prosthetic ankle–foot ranged from 0.08 to 0.27 J/kg and from 0.07 to 0.27 J/kg for the TTs and TFs, compared to the one done by biological ankle–foot of NAs which ranged from 0.15 to 0.32 J/kg (Table 2). Small but significant positive correlations between walking speed and $W^{+}_{ankle-foot}$ were found for TTs (slope = 0.15, $R^2 = 0.118$, $p = 0.030$) and TFs (slope = 0.14, $R^2 = 0.200$, $p < 0.001$), but no correlation was found for NAs (slope = 0.08, $R^2 = 0.104$, $p = 0.068$) (Fig. 3, Table 3). ANCOVA results indicated that the effects of SSWS ($p < 0.001$) and groups ($p < 0.001$) were significant, while the interaction effect was not ($p = 0.517$). A significantly greater estimated marginal mean $W^{+}_{ankle-foot}$ was found for NAs (0.22 ± 0.02 J/kg) compared to TTs (0.18 ± 0.01 J/kg, $p < 0.010$) and TFs (0.16 ± 0.01 J/kg, $p < 0.001$) (Table 2). No significant difference was found between TTs and TFs ($p = 0.133$).

The push-off work done by the prosthetic limb of TTs and TFs ranged respectively from 0.09 to 0.25 J/kg and from 0.06 to 0.22 J/kg compared to the one done by individual limbs of NAs which ranged from 0.19 to 0.42 J/kg (Table 2). W^{+}_{StS} was positively correlated with SSWS for NAs (slope = 0.21, $R^2 = 0.283$, $p < 0.001$) as well as for TT (slope = 0.23, $R^2 = 0.380$, $p < 0.001$) (Fig. 3). However, no correlation was found for prosthetic TFs (slope = 0.01, $R^2 = 0.002$, $p = 0.710$) (Table 3). ANCOVA results indicated that the effects of SSWS ($p < 0.001$) and groups ($p < 0.001$) as well as the interaction effect ($p < 0.001$) were significant, with greater estimated marginal mean W^{+}_{StS} for NAs (0.28 ± 0.02 J/kg) than TTs (0.15 ± 0.01 J/kg, $p < 0.001$) and TFs (0.12 ± 0.01 J/kg, $p < 0.001$) (Table 2) and greater for TTs than TFs ($p < 0.010$) (Table 2).

Concerning the collision work done by contralateral limb, two data points from TTs and one from TFs were removed as outliers according to a Cook D's test. The collision work done by contralateral limb of TTs and TFs ranged respectively from −0.42 to −0.05 J/kg and from −0.44 to −0.01 J/kg compared to the one done by leading limb of NAs which ranged from −0.36 to −0.03 J/kg (Table 2). Negative correlations between SSWS and W^{-}_{StS} were found for all the groups (Fig. 3, Table 3). ANCOVA results indicated that the effects of SSWS ($p < 0.001$) and groups ($p < 0.001$) were significant, while the interaction effect was not ($p = 0.092$). A significantly greater estimated marginal mean W^{-}_{StS} was only found for TTs (−0.19 ± 0.02 J/kg) compared to NAs (−0.14 ± 0.02 J/kg, $p < 0.001$) (Table 2).

4. Discussion

We have investigated the positive push-off work done by the trailing prosthetic limb (W^{+}_{StS}) and the prosthetic ankle–foot ($W^{+}_{ankle-foot}$), as well as the negative collision work (W^{-}_{StS}) done by the leading contralateral limb for transtibial (TTs) and transfemoral (TFs) amputees. These mechanical works have been compared between groups and to the ones performed by non-amputees (NAs), while taking into account the effect of self-selected walking speed (SSWS). As expected from literature, TFs have a significantly lower SSWS than TTs and NAs (Robert and Waters,

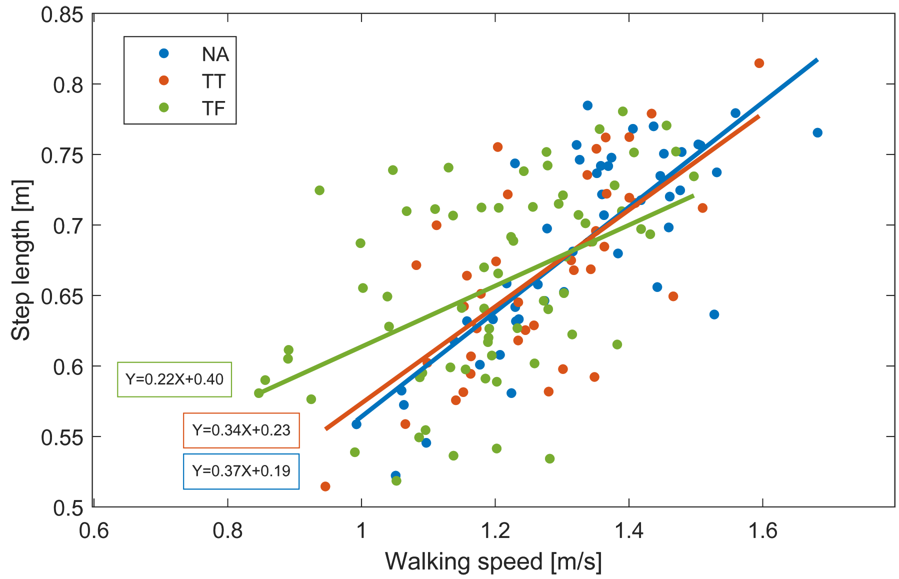

Fig. 2. Step length during the transition from the prosthetic (trailing leg) to the contralateral limb (leading leg) versus patient self-selected walking speed for the NA, TT and TF groups with their respective linear best fit.

Table 3
Statistical parameters of the linear best fit approximation of $W^+_{ankle-foot}$, W^+_{StS}, W^-_{StS} and contralateral step length versus self-selected walking speed during the transition from the prosthetic (trailing leg) to the contralateral limb (leading leg) for NAs, TTs and TFs. The slopes and intercept of the linear best fit are given with their confidence intervals (CI), the correlation coefficient R^2 and the p-value of the F-test on the linear model are also given (noted * when p-value < 0.05).

	$W^+_{ankle-foot}$ [J/kg]			
	Slope (±95 %CI)	Intercept (±95 %CI)	R^2	p-value
NAs	0.08 ± 0.08	0.12 ± 0.11	0.104	0.068
TTs	0.15 ± 0.14	−0.01 ± 0.18	0.118	0.030*
TFs	0.14 ± 0.07	−0.01 ± 0.08	0.200	1.50E-04*

	W^+_{StS} [J/kg]			
	Slope (±95 %CI)	Intercept (±95 %CI)	R^2	p-value
NAs	0.21 ± 0.11	0.01 ± 0.14	0.283	3.41E-04*
TTs	0.23 ± 0.09	−0.13 ± 0.12	0.380	2.31E-05*
TFs	0.01 ± 0.07	0.11 ± 0.08	0.002	0.710

	W^-_{StS} [J/kg]			
	Slope (±95 %CI)	Intercept (±95 %CI)	R^2	p-value
NAs	−0.31 ± 0.13	0.25 ± 0.17	0.371	2.37E-05*
TTs	−0.44 ± 0.16	0.36 ± 0.20	0.472	2.62E-06*
TFs	−0.24 ± 0.10	0.14 ± 0.11	0.288	3.47E-06*

	Step length [m]			
	Slope (±95 %CI)	Intercept (±95 %CI)	R^2	p-value
NAs	0.37 ± 0.08	0.19 ± 0.11	0.64	1.78E-11*
TTs	0.34 ± 0.12	0.23 ± 0.16	0.46	1.61E-12*
TFs	0.22 ± 0.10	0.40 ± 0.12	0.23	3.89E-05*

1992).

4.1. TTs and TFs have similar push-off work done by prosthetic ankle–foot

The estimated marginal means $W^+_{ankle-foot}$ were consistent with published values for NAs' ankle–foot (0.22 ± 0.02 J/kg vs 0.19 ± 0.03 J/kg from Takahashi et al., 2017) as well as for TTs prosthetic ankle–foot (0.18 ± 0.01 J/kg vs 0.15 ± 0.04 J/kg from Segal et al., 2012). To our knowledge, no values of prosthetic ankle–foot push-off work have been reported for TFs in the literature. After Bonferroni adjustment for SSWS effects, significant differences in the estimated marginal means $W^+_{ankle-foot}$ were only observed between NAs compared to both TTs and TFs' prosthetic ankle–foot. TTs and TFs exhibited similar increases of prosthetic $W^+_{ankle-foot}$ with increasing SSWS, indicating that they had similar ability to use ESAR feet at similar given SSWS.

4.2. Push-off work done by the prosthetic limb differs between TTs and TFs

The estimated marginal means W^+_{StS} done by NAs' trailing limb (0.28 ± 0.02 J/kg) was consistent with literature (0.28 ± 0.05 J/kg from Houdijk et al., 2009 and 0.28 ± 0.06 J/kg from Bonnet et al., 2014). Likewise, TTs' prosthetic W^+_{StS} (0.15 ± 0.01 J/kg) was similar to published values (0.16 ± 0.04 J/kg from Houdijk et al., 2009 and 0.17 ± 0.07 J/kg from Russell Esposito et al., 2015). Finally, TFs' prosthetic W^+_{StS} (0.12 ± 0.01 J/kg) were closed to those reported by Bonnet et al. (2014) (0.09 ± 0.03 J/kg) and the equation of linear best fit approximation linking it to the walking speed was similar to the one of Pinhey et al. (2022).

Significantly lower W^+_{StS} was done by the prosthetic limb of amputees compared to the one done by NAs and TFs have the lowest one. TFs was the only group in which the push-off work performed by the prosthetic limb did not increase with SSWS while prosthetic $W^+_{ankle-foot}$ did, which was consistent with Pinhey et al., 2022. Push-off work done by the prosthetic ankle–foot was higher than the one done by the prosthetic limb in amputees, at the opposite of non-amputees. In NAs and TTs, push-off work done by the ankle–foot increased less with SSWS than the one done by the trailing leg (respectively a slope of 0.08 vs 0.21, and 0.15 vs 0.23). This point suggests that the increase of W^+_{StS} can be

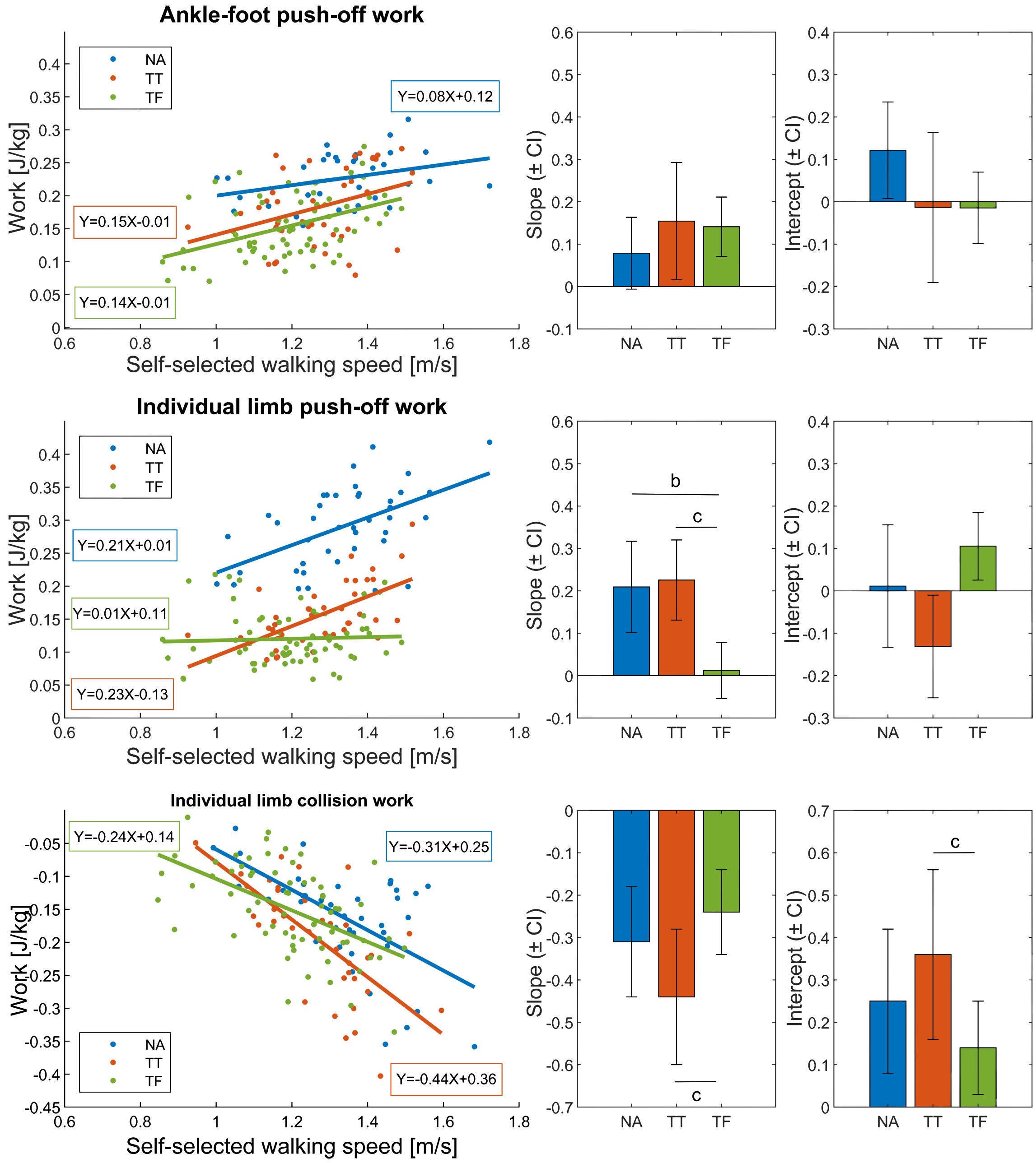

Fig. 3. (Left) $W^{+}_{ankle-foot}$, W^{+}_{StS} and W^{-}_{StS} versus self-selected walking speed during the transition from the prosthetic to the contralateral limb for TT and TF groups and during the transition from the left to the right limb for NA. Linear regression are plotted for each group. (Right) The slopes and intercept of the linear regressions are given with their confidence intervals (CI). Significant differences ($p < 0.05$) in slope and intercept between NAs and TTs, NAs and TFs and TTs and TFs are respectively indicated by a, b and c.

attributed to the contribution of another joint than the ankle, most likely the hip (Zelik and Adamczyk, 2016).

At the opposite, TFs' prosthetic $W^{+}_{ankle-foot}$ increased more than the prosthetic limb W^{+}_{StS} (respectively a slope of 0.14 and 0.01). One plausible explanation is that the work provided by the hip at the end of the stance phase to initiate the knee flexion (Koehler-McNicholas et al., 2016) does not contribute in increasing the prosthetic limb work. Indeed, this action of the hip results in a modification of the orientation of the GRF to redirect it behind the knee axis necessary to unlock knee flexion. As a consequence, the scalar product between GRF and the BCoM velocity is decreased, and so is W^{+}_{StS} (Bonnet et al., 2014; Pinhey et al., 2022). Thus, the initiation of flexion during propulsion seems to be determinant in the generation of work done by prosthetic limb, inducing significant alignment-sensitive residual hip work (Koehler-McNicholas et al., 2016) and impacting metabolic consumption (Schmalz et al., 2002).

In conclusion, our results suggest that the relationship between push-off work done by the prosthetic ankle–foot and the one done by the prosthetic limb, as a function of SSWS, depends on the level of amputation.

4.3. Reduced prosthetic limb push-off work doesn't result in an increase in contralateral limb collision work for above knee amputees

The estimated W^{-}_{StS} done by NAs' leading limb (-0.14 ± 0.02 J/kg) was consistent with literature (-0.14 J/kg from Adamczyk and Kuo, 2015 and -0.18 ± 0.05 J/kg from Houdijk et al., 2009). Likewise, TTs' contralateral W^{-}_{StS} (-0.19 ± 0.02 J/kg) was similar to published values (-0.23 J/kg from Adamczyk and Kuo, 2015 and -0.23 ± 0.07 J/kg from Houdijk et al., 2009). Finally, the equation of linear best fit approximation linking TFs' contralateral W^{-}_{StS} to speed was close to the one of Pinhey et al., 2022.

The increase in negative collision work as a function of SSWS was expected due to the increase in BCoM velocity and GRFs (Adamczyk and Kuo, 2015; Donelan et al., 2002b; Houdijk et al., 2009; Pinhey et al., 2022). The theorical expectation of dynamic walking as an inverse pendulum (Adamczyk and Kuo, 2015) suggests that weaker push-off work in the trailing leg will lead to higher collision work in the leading leg.

Thus, the higher TTs' contralateral W^{-}_{StS} compared to NAs was expected, unlike the non-increase of TFs' contralateral W^{-}_{StS} compared to TTs and NAs which was contrary to what the inverted pendulum model suggests. This could be due to a shorter contralateral step length as suggested by Donelan's collision model which predicts a rate of collision work proportional to the fourth power of step length (Donelan et al., 2002a). However, no difference in contralateral step length was found between groups, possibly due to the fact that the direction of step length asymmetry varied inconsistently across TFs (Roerdink et al., 2012).

4.4. Limitations

The study has some limitations that must be considered for interpretation. First, it focused on active participants with advanced prosthetics, potentially not reflecting less active amputees with basic devices (Pröbsting et al., 2022; Sawers and Hafner, 2013). Additionally, women and certain amputation causes are under-represented. Secondly, 6 TTs and 5 TFs took part in the data collection less than 6 months after their operation, possibly affecting recorded gait stability. While clinicians deemed walking levels stable, transitional strategies following amputation may introduce bias. Thirdly, other variables can have an effect on mechanical work, and therefore bias the results obtained if they are not taken into account, such as cause of amputation (Waters et al., 1976) length of residual limb (Bell et al., 2014), prosthetic alignment (Schmalz et al., 2002), prosthetic knee (Chin et al., 2006) or prosthetic foot (Heitzmann et al., 2018). It would be interesting to make subgroups for a more detailed study. Fourth, BCoM velocity was calculated kinematically from each patient's inertial model, which may introduce inaccuracies. This method demonstrated comparable values than the one using the integration of the GRFs in a TFs amputee population (Lansade et al., 2021). This latter method is to be preferred in the future, but requires measuring the forces under each of the two feet throughout the integration phase, which may require more than two force plates to address both the prosthetic-to-contralateral transition and the contralateral-to-prosthetic transition. Fifthly, complex musculoskeletal models (Dorn et al., 2015) propose an alternative approach to estimate walking-related mechanical work. Based on these models, some authors suggest that, in NAs, the major determinant of metabolic cost during walking may be the elevation of the BCoM in single-limb support (Neptune et al., 2004), which would account for 44 % of the total cost of walking while step-to-step transition would account for only 37 % (Umberger, 2010). Indeed, walking with a prosthesis necessitates adaptation at other instants than step-to-step transition, this is why further studies should explore the entirety of the gait cycle to better understand the energetic challenges of walking with a prosthesis. Finally, our study did not investigate the contralateral-to-prosthetic limb transition. Compensations have already been observed, including greater push-off work at contralateral limb compared to the prosthetic limb for both TTs (Houdijk et al., 2009) and TFs (Pinhey et al., 2022), exceeding that in NAs' limb (Bonnet et al., 2014). A combined study of both transitions should be addressed in future work to understand the overall energy balance during walking.

5. Conclusion

By using data obtained from a large cohort of 155 subjects, it became possible to investigate the mechanical work generated by the ankle–foot and lower limb in non-amputees, transtibial amputees and transfemoral amputees, by controlling the effect of SSWS. This analysis contributed to quantifying how individuals adapted their mechanical work to their SSWS, based on their level of amputation.

First, after adjusting for SSWS, our study emphasized that NAs produce more push-off work with both their ankle–foot and trailing limb than amputees do on prosthetic side. Secondly, our study highlighted that TTs and TFs can generate the same amount of prosthetic $W^{+}_{ankle-foot}$, while prosthetic W^{+}_{StS} is higher for TTs and remains constant with SSWS for TFs. This could be due to swing phase initiation of the prosthetic knee. Surprisingly and contrary to theoretical expectations, collision work on the TFs' contralateral limb was not affected by the decrease of prosthetic limb push-off work.

To better understand the impact of mechanical work on energetic efficiency during walking, further research has to investigate the link between mechanical work and metabolic expenditure of people with amputation.

CRediT authorship contribution statement

L. Sedran: Writing - original draft, Writing – review & editing, Formal analysis, Conceptualization. **X. Bonnet:** Writing – review & editing, Investigation, Formal analysis, Conceptualization. **M. Thomas-Pohl:** Supervision, Investigation, Conceptualization. **I. Loiret:** Supervision, Investigation, Conceptualization. **N. Martinet:** Supervision, Conceptualization. **H. Pillet:** Writing – review & editing, Supervision, Investigation, Formal analysis, Conceptualization. **J. Paysant:** Supervision, Conceptualization.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

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