

Effect of speed on the energy cost of walking in unilateral traumatic lower limb amputees

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Abstract In this work, the effect of walking speed on the energy expenditure in traumatic lower-limb amputees was studied. The oxygen consumption was measured in 10 transfemoral amputees, 9 transtibial amputees and 13 control subjects, while they stood and walked at different speeds from 0.3 m s⁻¹ to near their maximum sustainable speed. Standing energy expenditure rate was the same in lower-limb amputees and in control subjects (≈ 1.85 W kg⁻¹). On the contrary, during walking, the net energy expenditure rate was 30–60% greater in transfemoral amputees and 0–15% greater in transtibial amputees than in control subjects. The maximal sustainable speed was about 1.2 m s⁻¹ in transfemoral amputees and 1.6 m s⁻¹ in transtibial amputees, whereas it was above 2 m s⁻¹ in control subjects. Among these three groups, the cost of transport versus speed presented a U-shaped curve; the minimum cost increased with the level of amputation, and the speed at which this minimum occurred decreased.

Keywords Lower limb amputees · Walking · Energy expenditure · Oxygen consumption

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Introduction

In normal walking, the mass-specific energy cost of locomotion (C , in J kg⁻¹ m⁻¹), i.e., the cost of operating the locomotory machinery over a given distance, presents a U-shaped curve with a minimum at an ‘optimal’ speed in the range of 0.8–1.4 m s⁻¹ (i.e., 3–5 km h⁻¹). This relation between cost and walking speed has been observed in adults by many authors (e.g., Margaria 1938; Zarrugh et al. 1974) and in children older than six (De Jaeger et al. 2001).

It is often claimed that the energy expenditure during walking is significantly higher in lower limb amputees than in non-amputees (e.g., Waters et al. 1976; Fisher and Gullickson 1978; Schmalz et al. 2002). This increase in metabolic energy cost is related, among others, to the level of the amputation and the etiology: the energy cost is higher in transfemoral amputees than in transtibial amputees; and for the same level of amputation, vascular amputees have a higher energy cost than traumatic amputees (Torburn et al. 1995; Detrembleur et al. 2005).

It is also thought that amputees walk slower than healthy subjects and that their speed of progression is lower the higher the level of amputation (Waters et al. 1976; Sulzle et al. 1978). The higher energy expenditure observed in amputees could thus be due, at least in part, to the fact that they cannot reach the optimal walking speed. In other words, the increase in the cost of locomotion could be due not only to motion abnormalities but also to the low speed itself (Tesio et al. 1991; Detrembleur et al. 2005).

In this study, we have tested this hypothesis by investigating the effect of speed on the energy expenditure rate of transfemoral and transtibial traumatic amputees. Experiments were performed in Cambodia where, due to land mines, the prevalence of lower limb amputation is regrettably high (4 per 1,000 inhabitants, source: Handicap

International Belgium). Amputee and control subjects were requested to walk at different speeds ranging from very low speeds up to their maximal sustainable speed. The mass-specific energy cost of walking, i.e., the energy expended to move one unit of body mass a given distance, was then determined to show if an optimal speed exists, and whether this optimal speed changes with the level of amputation.

Materials and methods

Measurement of energy cost

The energy cost of walking was estimated from the oxygen consumption and the carbon-dioxide production measured with a K4[®] (Cosmed, Italy) telemetric system (Hauswirth et al. 1997) that also recorded the heart rate. The K4 system consisted of a portable unit worn by the subject, and a base station for recording the data. The portable unit weighed 1.5 kg and consisted of a silicon mask containing a flow-rate turbine, which was fixed on the subject's face, a processing unit containing the O₂ and CO₂ analyzers, which was placed on the subject's chest and a transmitter/battery pack, which was placed on the subject's back. Every day, the turbine was calibrated with a 3 l syringe, and a two-point calibration of the O₂ and CO₂ analyzers was carried out using ambient air and a standard calibration gas mixture (5% CO₂, 16% O₂, 79% N₂).

The mass-specific gross energy consumption rate (P_{gross} in W kg⁻¹) was obtained from the oxygen consumption rate using an energetic equivalent for oxygen, taking into account the measured respiratory exchange ratio (RER, i.e., the ratio between the carbon-dioxide production rate and the oxygen consumption rate) (McArdle et al. 1996):

$$\text{kJ} = (16.1 + 5 \text{ RER})/\text{O}_2. \quad (1)$$

Only trials with $\text{RER} \leq 1$ were recorded and analyzed.

The mass-specific net energy consumption rate (P_{net} in W kg⁻¹) is the energy consumption that can be attributed to walking per se. P_{net} was calculated from the energy consumption rate while walking minus energy consumption rate while standing (P_{stand}). The mass-specific gross and net costs of transport (C_{gross} and C_{net} in J kg⁻¹ m⁻¹), i.e., the gross and net costs of operating the locomotory system a given distance, were calculated, respectively, as the P_{gross} and P_{net} divided by the speed V_f .

Fusi et al. (2002) have defined the Locomotory Index (Index Locomotorius, IL) as the ratio between the C_{gross} of a patient walking at his self-selected speed (SSS) divided by the C_{gross} of the control subjects walking at the same speed. In this study, the IL sensu stricto could not be calculated because the SSS was not measured. For this reason, the definition of IL was adapted as follows. The IL_{gross} and

IL_{net} were defined, respectively, as the C_{gross} and C_{net} of the patients walking at their optimal speeds divided by the C_{gross} and C_{net} of the control subjects walking at the same speed than the patients.

Experimental procedure

Experiments were performed in the Physical Rehabilitation Center of Siem Reap (Cambodia) managed by Handicap International Belgium. The subjects were asked to walk along a nearly circular outdoor track (41-m long, 1-m wide) at different speeds ranging from 0.3 m s⁻¹ to 2.3 m s⁻¹, in classes of 0.28 m s⁻¹ (1 km h⁻¹). The walking speed was measured by ten pairs of photocells placed at neck level along the track. Subjects were given verbal commands in order to maintain their actual speed equal to the desired speed. The average variation in the walking speed within any test, measured every 3.5–4.5 m, was ± 0.03 m s⁻¹ (ranging between ± 0.01 and ± 0.08 m s⁻¹).

Each experiment began and ended with the measurement of the standing oxygen consumption rate (P_{stand}). Between these two measurements, a maximum of seven different speeds of walking were successively imposed in a random sequence. Each measurement was maintained as long as necessary to obtain a steady-state period of at least 3 min. When a subject was unable to maintain the imposed speed, or when the RER exceeded 1.0, the trial was stopped and the experiment was resumed at a lower speed. The highest speed a subject could maintain for at least 3 min with a $\text{RER} < 1$ was defined as the maximal sustainable speed.

For the amputees, the direction of walking (clockwise or counterclockwise) was chosen so that each subject walked alternatively with the prosthesis on the inside and on the outside of the curve. In each speed group, the number of subjects walking with the prosthesis inside and outside of the curve was equal. A two-factor ANOVA (SuperANOVA version 1.1) showed that the direction of walking around the track had no influence on the power spent both in transfemoral and transtibial amputees ($F = 2.884$, $P > 0.10$ and $F = 1.016$, $P > 0.32$, respectively).

Subjects

Experiments were carried out on 26 Cambodian adults: ten were amputated at the level of the femur, nine at the level of the tibia and seven were control subjects. The characteristics of the subjects are presented in Table 1. Informed consent of the subjects was obtained. Experiments were performed according to the Declaration of Helsinki.

Table 1 Anthropometric characteristics of the subjects

	n (♂–♀)	Age (years)	Height (m)	Mass (kg)	Stump length/segment length
Femoral amputees	10 (10–0)	34.7 ± 5.1	1.62 ± 0.04	57 ± 4.8	0.69 ± 0.19
Tibial amputees	9 (9–0)	35.3 ± 7.2	1.63 ± 0.03	58 ± 5.8	0.50 ± 0.12
Controls	13 (10–3)	27.8 ± 5.2	1.70 ± 0.10	67 ± 9.2	–
Caucasian	6 (3–3)	23.6 ± 3.4	1.77 ± 0.09	65 ± 8.2	–
Cambodian	7 (7–0)	31.4 ± 3.6	1.64 ± 0.05	69 ± 10.2	–

Values in the four last columns are mean ± SD. The control group was composed of six Caucasian subjects that participated to a previous study (De Jaeger et al. 2001) and of seven Cambodian subjects. Their characteristics are given separately in the two lower lines

In order to increase the robustness of the results, six Caucasian control subjects that participated in a previous study (De Jaeger et al. 2001) were added to the control group. Indeed, the power spent while standing and while walking at different speeds (P_{stand} and P_{gross} , respectively) of the seven Cambodian control subjects were not statistically different from the P_{stand} (one factor ANOVA, $F = 3.140$, $P > 0.10$) and the P_{gross} (two factors ANOVA, $F = 0.996$, $P > 0.32$) of six Caucasian control subjects. Table 1 shows that the Caucasian subjects were on the average 0.13 m taller than the Cambodian. On the average, a 13 cm taller subject walks at a given speed with a 6% lower stride rate; this difference was neglected since it is smaller than the variability of the stride rate of different subjects of the same size (Inman et al. 1981).

In all cases, the cause of amputation was a traumatic injury due to land mines. The amputation happened on the average 11.2 ± 4.2 years (mean ± SD, $n = 18$) before the experiments. All amputees have been wearing a prosthesis for at least 3 years, and on average 9.5 ± 4.2 years (mean ± SD, $n = 18$), before the experiments. None of the subjects had any signs or symptoms of cardiovascular, neurological or musculoskeletal disease other than the amputation. None of the amputated subjects had stump pain, swelling or pressure sores.

In transfemoral group, the patients were fitted with a CAT-CAM socket or a quadrilateral socket with a suction system. These sockets were combined with various prosthetic knees (Otto Bock 3R45, Endolite 160HI-activity) and feet (single axis foot, Flex-Foot). In the transtibial group, the patients were fitted with KBM sockets or with Iceross sockets. Multiflex (Endolite Blatchford) or Flex-Foot (Variflex and Sureflex) prosthetic feet were mounted on these sockets. On the average, the femoral and tibial prosthesis weighed, respectively, 2.7 and 1.8 kg, which represented about 5 and 3% of the body weight. The quality of fitting and alignment of the prostheses was assessed by an experienced examiner. The alignment procedure followed conventional clinical practice as described by Sin et al. (2001).

Results

The mass-specific power measured during standing (P_{stand} , Fig. 1) is about equal in each of the two groups of amputees and in the control subjects. A one-factor ANOVA analysis does not show any statistical difference between the three groups ($F = 0.251$, $P > 0.77$): the average of all the subjects of the three groups is 1.85 ± 0.24 W kg⁻¹ (mean ± SD, $n = 32$).

The mass-specific gross power, P_{gross} , is presented as a function of walking speed, V_f , in Fig. 2. Over the range of speed studied, the P_{gross} versus V_f relation is best described by a second-order polynomial function:

$$P_{\text{gross}} = aV_f^2 + bV_f + c. \quad (2)$$

The values of the coefficients a , b and c are given in Table 2. The coefficient a of the equations is greatest in

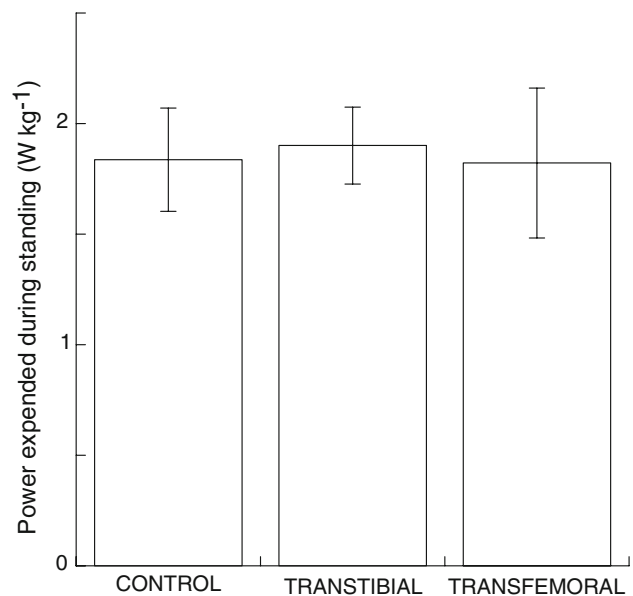


Fig. 1 The mass-specific power measured during standing (P_{stand}) is presented in each of the two groups of lower limb amputees and in the control subjects. The columns are the average of each group and the vertical bars represent the standard deviations

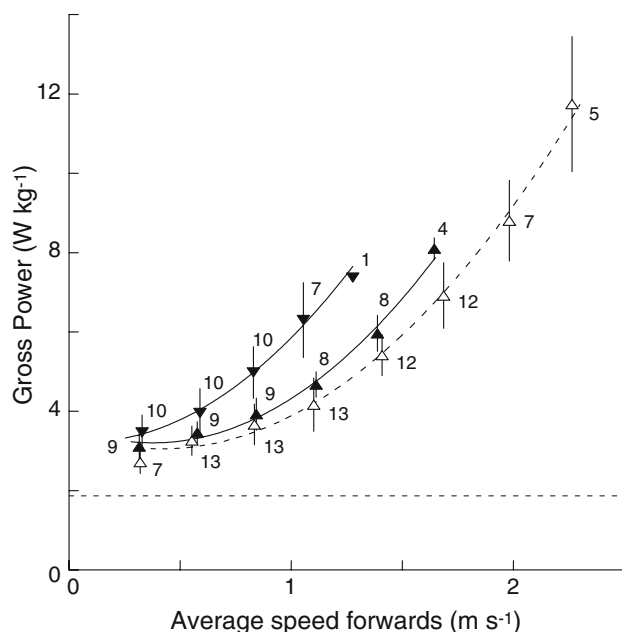


Fig. 2 The mass-specific gross metabolic power as a function of walking speed in control subjects (*open triangles*), transtibial (*filled upper triangles*) and transfemoral amputees (*filled lower triangles*). The *symbols* represent the mean values; *n* is given by the figures near each point; the standard deviation *bars* are drawn when they exceed the size of the symbol. The *curves* are second-order polynomial fits (Kaleidagraph 4.0) through all of the data points, coefficients of these equations are presented in Table 2. The *horizontal interrupted line* represents the average mass-specific power measured during standing for all the subjects together

transfemoral amputees, intermediate in transtibial amputees, and smallest in the control subjects, showing that the influence of the speed on P_{gross} is greater the higher the level of amputation.

In these three groups, when the walking speed approaches zero, P_{gross} tends to a value of about 3.45 W kg^{-1} (intercept c of the first equation in Table 2) which is greater than the power measured during standing (1.85 W kg^{-1}). The mass-specific net power (P_{net}), which represents the power spent for walking per se, was computed by subtracting P_{stand} (interrupted line in Fig. 2) from P_{gross} . The coefficients a and b of the P_{gross} versus speed equation are

thus about equal to ones of the P_{net} versus speed equation, whereas the intercept c is decreased from ~ 3.45 to $\sim 1.63 \text{ W kg}^{-1}$.

In order to determine the speed at which P_{net} is different in amputees and control subjects, a two-way ANOVA with contrasts is applied to the data. The difference in P_{net} between transtibial amputees and control subjects increases with speed but becomes statistically significant ($P < 0.032$) only at speeds above 1.1 m s^{-1} . In transfemoral amputees, P_{net} is significantly greater than in control subjects at all walking speeds ($P < 0.011$).

The maximal speed at which the subjects can walk with a RER lower than one decreases with the level of amputation: $1.4\text{--}1.7 \text{ m s}^{-1}$ in transtibial amputees and $1.1\text{--}1.3 \text{ m s}^{-1}$ in transfemoral amputees. All of the control subjects but one, could walk up to 2 m s^{-1} with a RER < 1 ; for five of them, the RER remained under one even at the highest speed tested (2.3 m s^{-1}). At their maximal sustainable speeds, P_{gross} (Fig. 2) is lower in amputees ($7\text{--}8 \text{ W kg}^{-1}$) than in control subjects ($10\text{--}12 \text{ W kg}^{-1}$).

The mass-specific gross cost of transport C_{gross} , i.e., the amount of energy spent while walking a unit distance, is shown as a function of speed for the different groups in Fig. 3a. Among these three groups, C_{gross} presents a U-shaped curve with a minimum occurring at an optimal speed (vertical interrupted lines in Fig. 3a). In transtibial amputees, the minimum occurs at a speed slightly lower than in the control group and the IL_{gross} is equal to 1.12. In transfemoral amputees, the minimum occurs at a speed which is $\sim 0.1 \text{ m s}^{-1}$ lower than in control subjects and the IL_{gross} is equal to 1.50.

Figure 3b presents the mass-specific net cost of transport C_{net} , i.e., the energy consumed per unit distance that can be attributed to the walking itself. As in C_{gross} , C_{net} presents as a function of speed a U-shaped curve. In each group, C_{net} at a given speed is smaller than C_{gross} because the former does not include the cost of standing. The optimal speed of C_{net} is also $\sim 0.3 \text{ m s}^{-1}$ lower than the optimal speed of C_{gross} . In transtibial and transfemoral amputees, the IL_{net} is, respectively, equal to 1.14 and 1.81. In amputees as in control subjects, when speed decreases

Table 2 Equations of gross and net mass-specific power (W kg^{-1}) versus speed (m s^{-1})

	$P_{\text{gross}} = a V_f^2 + b V_f + c$				$P_{\text{net}} = a V_f^2 + b V_f + c$			
	<i>a</i>	<i>b</i>	<i>c</i>	<i>r</i>	<i>a</i>	<i>b</i>	<i>c</i>	<i>r</i>
Femoral amputees	3.03	−0.42	3.24	0.87	3.24	−0.60	1.45	0.92
Tibial amputees	2.89	−2.19	3.62	0.97	2.96	−2.30	1.76	0.98
Controls	2.45	−2.05	3.48	0.96	2.42	−2.05	1.69	0.97

Values of the coefficients (a , b , c) and of correlation coefficient (r) of the second order polynomial fit that describes the relation between the mass-specific power and the walking speed

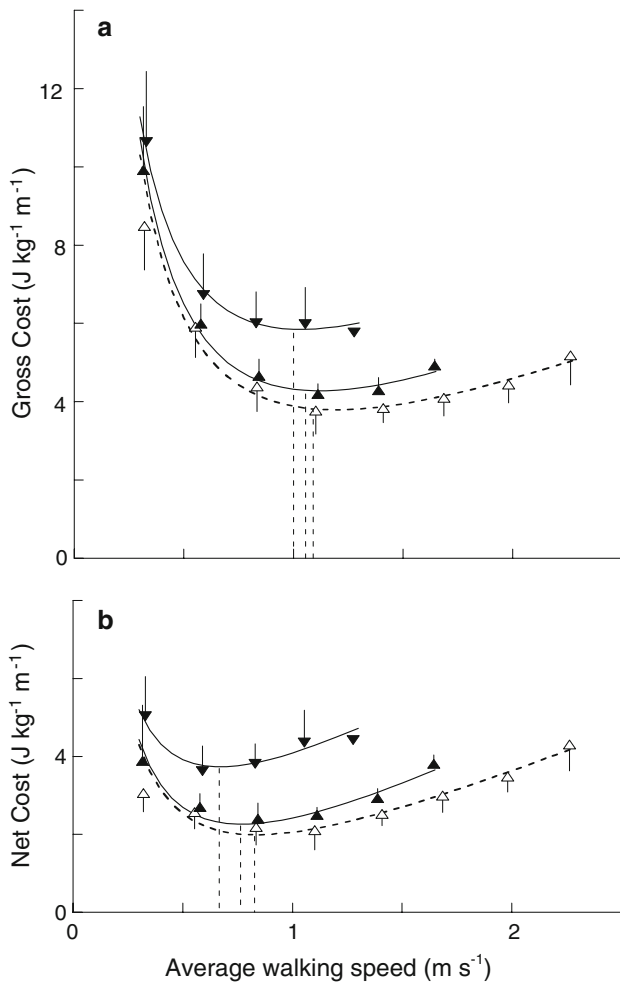


Fig. 3 The mass-specific gross cost of transport, C_{gross} , (a) and net cost of transport, C_{net} (b), are shown as a function of speed, for the three groups. The curves are drawn using Eq. 3 and the coefficients of Table 2. The vertical interrupted lines correspond to the theoretical values of V_{opt} of Table 3. Other indications as in Fig. 2

below the optimal speed, C_{gross} increases faster than C_{net} because, in C_{gross} , the standing energy cost becomes a larger fraction of the total cost of transport.

From Eq. 2, the cost can be expressed by (Bastien et al. 2005):

$$C = \frac{P}{V_f} = aV_f + b + \frac{c}{V_f} \tag{3}$$

Since C is minimal when its derivative with respect to V_f is nil:

$$\frac{dC}{dV_f} = aV_f^2 - c = 0, \tag{4}$$

the theoretical optimal speed V_{opt} can be computed as:

$$V_{opt} = \sqrt{c/a} \tag{5}$$

The theoretical minimal value of the cost, C_{min} , is calculated by substituting V_{opt} into Eq. 3:

Table 3 Minimal cost and the optimal speed for the three groups

	Gross cost		Net cost	
	C_{min} (J kg ⁻¹ m ⁻¹)	V_{opt} (m s ⁻¹)	C_{min} (J kg ⁻¹ m ⁻¹)	V_{opt} (m s ⁻¹)
Femoral amputees	5.8	1.00	4.3	0.70
Tibial amputees	4.3	1.10	2.3	0.80
Controls	3.8	1.20	2.0	0.85

Theoretical values of the minimal gross and net costs (C_{min}) and of the optimal speed (V_{opt}) at which the cost is minimal for the three groups. The values of C_{min} and V_{opt} are computed from Eqs. 5 and 6, respectively, using the coefficients of Table 2. They should be compared with the experimental values of Fig. 3

$$C_{min} = 2\sqrt{ac} + b. \tag{6}$$

In Fig. 3, the curves are computed from Eq. (3) using the coefficients of Table 2. The vertical interrupted lines correspond to the theoretical values of V_{opt} and C_{min} (Table 3), computed from Eqs. (5) and (6). It can be seen that these theoretical values match with the experimental data. When comparing the three experimental groups, it can also be seen that the minimum is greater and occurs at a lower speed, the higher the level of amputation.

Discussion

In this work, we have evaluated the effect of the walking speed and the level of amputation on the energy expenditure of unilateral traumatic lower limb amputees.

Below their optimal walking speed (1 m s⁻¹), transtibial amputees do not consume more energy than control subjects. However, when speed increases above 1 m s⁻¹, the mass-specific average power becomes significantly greater in transtibial amputees than in control subjects. At the fastest sustainable speed reached by transtibial amputees (~1.7 m s⁻¹), their P_{gross} is 20% greater and their P_{net} is 30% greater than that of the control subjects.

In contrast, transfemoral amputees consume more energy than control subjects at all speeds. At 0.3 m s⁻¹, P_{gross} and P_{net} are, respectively, 10 and 20% greater than in control subjects (Fig. 2). At the highest sustainable speed reached by the transfemoral amputees (~1.3 m s⁻¹), P_{gross} and P_{net} are, respectively, 50 and 100% greater than in control subjects.

These values are in good agreement with the literature both for P_{gross} (e.g., Molen 1973; Gonzalez et al. 1974; Sulzle et al. 1978; Scherer et al. 1999; Schmalz et al. 2002) and P_{net} (e.g., Detrembleur et al. 2005).

Energy consumption rate extrapolated at zero speed

During standing, traumatic lower limb amputees consume the same amount of energy per unit time as control

subjects, i.e., about 1.85 W kg^{-1} (Fig. 1). This value of standing power is consistent with those obtained in previous studies on non-amputated subjects (e.g., Passmore and Durnin 1955; De Jaeger et al. 2001). Thus, supporting body weight and maintaining equilibrium on one foot and one prosthesis does not cost measurably more energy than on 2 feet.

In amputees as well as in control subjects, the zero-speed intersect of the power versus speed function is about 90% higher than the standing energy consumption rate (Fig. 2). Although the three curves seem to converge to a similar value, the speeds investigated are not low enough to determine precisely if this is the case.

The fact that these regression functions do not extrapolate to the standing metabolic rate is most likely due to the metabolic cost of maintaining the walking pattern at very low speeds (walking in place). Indeed, in normal walking, as speed approaches zero, the intersect of the total mechanical power versus speed is about 0.3 W kg^{-1} (Cavagna et al. 1976). This non-zero intersect is mainly due to the mechanical power spent to maintain the vertical movements of the center of mass (COM) which does not tend towards zero (Cavagna et al. 1976). Indeed, if we assume a muscular efficiency of ~ 0.20 at very low walking speeds (Cavagna and Kaneko 1977), then the metabolic power spent to walk in place would be in the order of 1.5 W kg^{-1} , or about equal to the $\sim 1.6 \text{ W kg}^{-1}$ difference between the power spent during standing and the zero-speed intersect of the power versus speed function (Fig. 2). In conclusion, in non-amputated but also probably in amputated subjects, mainly due to the fact that the vertical movements of the COM are still present even at zero speed, the power spent during walking in place is greater than the power spent during standing.

Effect of speed and level of amputation on the energy consumption rate during walking

When the subject moves forward, the power spent increases non-linearly with the speed of progression. In normal walking, over the speed range studied, the power versus speed relation is best described by a second order polynomial function (Ralston 1958; Zarrugh et al. 1974). This is also the case in lower limb amputees (e.g., Molen 1973) although the effect of speed is more pronounced in amputees than in control subjects and increases with the level of amputation (Fig. 2). This is clearly shown by the increase in the coefficient a in Eq. 2 with amputation level (Table 2).

As pointed out by Tesio et al. (1998) and Detrembleur et al. (2005), the extra cost of the amputee locomotion cannot be explained by a single factor but must be attributed to multiple sources. The extra energy consumption

rate observed in amputees could be due to an increase in the muscular work done to move about and/or to a lower efficiency of positive work production by the muscle (Willems et al. 1995). In turn, the higher muscular work could be due to a higher external work (i.e., the work to maintain to movement of the COM relative to the surroundings) and/or to a higher internal work (i.e., the work done to move the limb segments relative to the COM and the work done by one leg against the other during the double contact phase). The possible sources of the greater energy consumption rate in amputees at a given speed are discussed hereunder.

At first glance, the wear of a prosthesis does not modify dramatically the characteristics of locomotion: at a given speed, the stride length and frequency of amputees are not modified (Hillary and Wallace 2000) and the external work done each stride is similar to the one done during normal walking (Tesio et al. 1998). These observations suggest that in amputees, walking can be powered by the unaffected leg only. However, a closer analysis of the stride shows an asymmetry between the two steps: the length of the step on the prosthesis is shorter than the one on the sound limb (Robinson et al. 1977), the stance on the prosthesis is briefer (Robinson et al. 1977; Hof et al. 2007) and its swing phase lasts longer (Frigo and Tesio 1986; Hof et al. 2007). These asymmetries increase the work done on the COM during some phases of the stride but decreases it during others (Tesio et al. 1998), with the result that the total external work done is about the same in amputees and in normal walking.

The higher energy expenditure can thus not be explained by a higher external work since, over a stride, the sound limb undergoes an overwork that is compensated by the “sparing” of the amputee limb (Tesio et al. 1998). However, in order to maintain the average speed of walking constant, the asymmetries observed in amputee’s walking may induce compensatory strategies that will reduce the efficiency of positive work production by the muscles. For example, these strategies may require isometric and/or antagonistic contractions, which result in an increase in energy expenditure without any increase in the muscular work. Such asymmetries may need to develop higher muscular power during some phases of the stride (Tesio et al. 1998), which could necessitate a higher energy expenditure than during normal walking.

The symmetry of the steps will first be affected by the unilateral absence of calf muscles. Indeed, in normal walking, especially at high speeds, these muscles play an important role in accelerating the body forwards at the end of the stance phase (Cavagna et al. 1976; Inman et al. 1981). The amputee’s lack of forward push on the prosthetic side of the body will require to double the power during the stance on the sound limb (Tesio et al. 1998). In

transtibial amputees, this asymmetry may be one of the principal sources of the increase in the metabolic cost of walking at speeds higher than 1 m s^{-1} .

Vertical movements of the COM are also perturbed during walking with a prosthesis. Indeed, the dorsiflexion of the ankle that normally occurs during swing phase is not possible with a prosthetic foot and therefore, the pelvis must be elevated to provide toe clearance. This pelvis movement will increase by $\approx 50\%$ the vertical displacement of the COM during the stance phase on the sound limb and, as a consequence, the power spent during this phase (Tesio et al. 1998; Detrembleur et al. 2005).

The lateral balance of the body is also altered during walking with a prosthesis: lower limb amputees have a less precise foot placement, which results in a wider stride (Hof et al. 2007). Even if the work done to move the COM laterally is more or less doubled in lower limb amputees, it remains a small part of the total work done during walking (Tesio et al. 1998).

The increased cost of walking with a prosthesis could also be due to a greater internal work done to move the limb segments relative to the COM. Indeed, the reduced time of contact on the prosthesis will induce faster movements of the sound limb during its swing forward. As a consequence, the step frequency is greater and the internal work is increased (Cavagna and Franzetti 1986). Furthermore, with a femoral prosthesis, a backward hip thrust is needed just before heel strike to extend the thigh stump with respect to the artificial leg. This movement will also increase the internal work (Frigo and Tesio 1986).

Another source of the over-cost could be an increased internal work done by one leg against the other (Bastien et al. 2003). Indeed, during the double contact phase of normal walking, both positive and negative work are done simultaneously: the back leg is accelerating the COM forward while the front leg is decelerating it backwards. In order to contain the positive work done by the back leg, combined movements of plantar flexion of the ankle and of flexion of the knee of the front leg reduce the shock at heel strike (Saunders et al. 1953). With a femoral prosthesis, the ankle joint is rigid and the knee remains extended during the double contact phase, with a consequence that the positive work done by the back sound leg against the prosthesis will be greater. Note that the enhanced lateral movements of the body in amputees could also increase significantly this internal work done by a leg against the other.

Finally, the metabolic efficiency greater than 0.25 in normal walking suggests the presence of an elastic mechanism during which mechanical energy is stored and released by the contracting muscles and tendons (Cavagna and Kaneko 1977; Willems et al. 1995). Such a mechanism is most likely perturbed in amputee's walking because the

rigidity of the prosthesis does not allow any storage of elastic energy. This phenomenon will reduce the efficiency of prosthetic walking as compared to normal walking.

Factors limiting the maximal aerobic speed of lower limb amputees

The maximal speed reached with a RER smaller than one is slower the higher the level of amputation (Fig. 2). At that speed, the average power spent is only $6\text{--}8 \text{ W kg}^{-1}$ in the amputee group as compared to $\sim 12 \text{ W kg}^{-1}$ in the control group. One might think that due to the loss of their lower limb, amputees move about less than normal subjects with the consequence that their general physical condition is degraded and their maximum sustainable capacity is reduced. In fact, this may not be the case. Indeed, at the maximal speed measured, the heart rate of the amputees is on average only $128 \text{ beats min}^{-1}$, which represents less than 70% of their maximal theoretical heart rate (computed as $220 - \text{age}$), suggesting that the amputees have not reached their overall maximum oxygen consumption rate. Most likely, the RER becomes > 1 because a few muscle groups are much more solicited than during normal walking, causing them to produce lactic acid which in turn will decrease the pH of the blood and induce a greater excretion of CO_2 .

However, in two of the ten transfemoral amputees, the RER remains below 1.0 even at the highest speed. In this case, the prosthesis could have been the limiting factor. Indeed, the artificial knee of the prostheses used in Cambodia has no spring or damping element in it, so the artificial leg swings freely. As pointed out by Frigo and Tesio (1986), the natural swing period of the prosthetic leg could be a main restraint to the maximal speed. Indeed, when the speed increases the stride duration is reduced. Above a critical speed, the swing period of the prosthetic leg would become too long to lock the knee before heel-strike.

The relation between the average power spent and the cost of locomotion

The gross cost of locomotion, C_{gross} , measures the intensity of the exercise; it represents the energy spent while walking over a unit distance. On the other hand, the net cost of locomotion, C_{net} , measures the amount of energy attributed to the activity itself, i.e., the energy needed to cover a unit distance.

In amputees, as in control subjects, the relation between the cost of locomotion and the walking speed shows a minimum at an intermediate walking speed (Fig. 3). As described in “Materials and methods”, the cost C (in $\text{J kg}^{-1} \text{ m}^{-1}$) is computed by dividing the power P (in

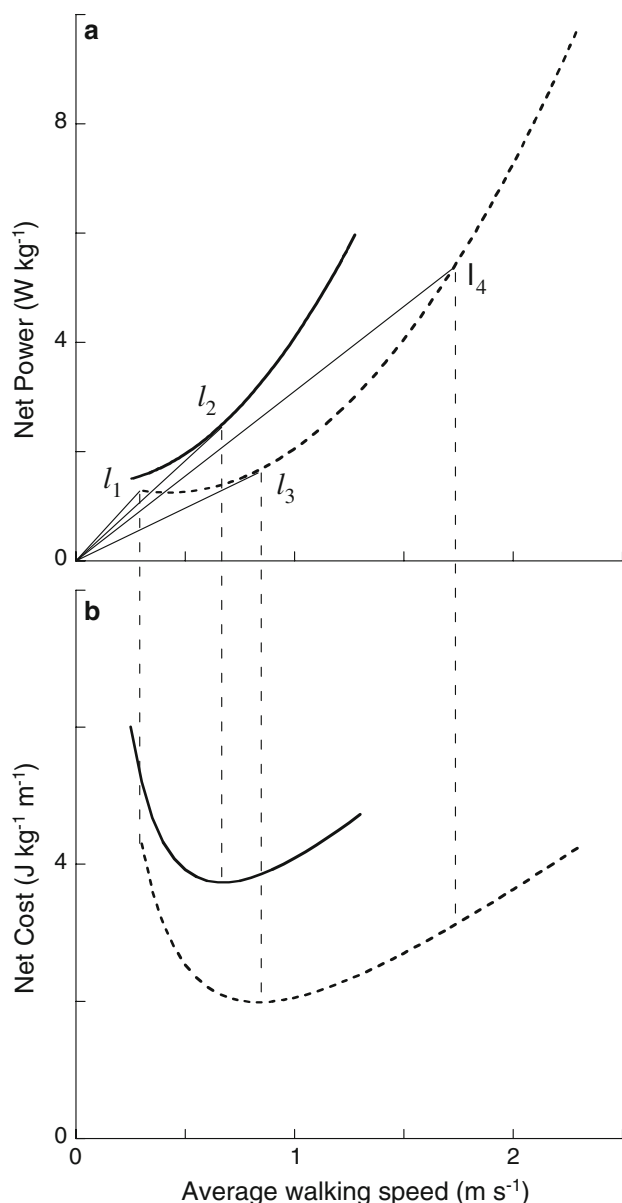


Fig. 4 The mass-specific net power, P_{net} (**a**) and net cost, C_{net} (**b**) as a function of speed in the transfemoral amputees (*continuous curves*) and in the control subjects (*interrupted curves*). The curves are drawn from Eq. 3 and coefficients of Table 2. In **a**, C_{net} at a given speed is the slope of the *straight line* joining the origin to P_{net} at that speed. This line is represented at a low (l_1) and high (l_4) walking speeds in control subjects, and at the optimal speed in transfemoral amputees (l_2) and in control subjects (l_3)

W kg^{-1}) by the speed of progression (V_f , in m s^{-1}), i.e., C is obtained by multiplying the power spent over a given distance by the time required to cover that distance. At low speeds, although P is low the time necessary to cover a distance is long, and as a result the total energy used is high. On the contrary, at high speeds, the time to cover a given distance is short, but P is high and as a consequence, C is again high.

Figure 4 illustrates as to how the relation between P_{net} and speed affects the shape of the C_{net} versus speed function; the same arguments apply to the P_{gross} and C_{gross} versus speed functions. As shown in Fig. 4a, C_{net} at a given speed is the slope of a line l joining the origin to the P_{net} at that speed. Thus, if the relation between P_{net} and V_f were a straight line passing through the origin, the slope of l (i.e., C_{net}) would be constant. However, as the speed approaches zero, P_{net} does not approach the origin (Fig. 2), and the slope of l tends to infinity (line l_1 in Fig. 4), explaining why C_{net} increases at low walking speeds. As walking speed increases, the relation between P_{net} and V_f cannot be simply a straight line, in which case the slope of l would decrease constantly. Because the relation between power and speed is curvilinear (Fig. 2), as speed increases the slope of l at first decreases, and then increases again (line l_4 in Fig. 4) above the optimal speed (line l_3).

The effect of the speed of progression on the power spent is greater the higher the level of amputation (Fig. 2). By comparing the P_{net} versus V_f curves of the transfemoral amputees and of the control subjects (lines l_2 and l_3 in Fig. 4), it can be seen that the steeper the relation between power and speed (i.e., the greater the coefficient a of the equation of Table 2), the lower the speed at which the slope of l is minimum (V_{opt} , Eq. 5) and the higher the value of that minimum slope (C_{min} , Eq. 6).

Conclusion

In conclusion, as in normal walking, the cost of locomotion of traumatic lower limb amputees presents a U-shaped curve with a minimal cost at an optimal speed. Traumatic lower limb amputees are able to walk at this speed and faster, disproving the hypothesis that their high cost is due to the fact that they cannot walk fast enough to attain the optimal speed. However, due to the fact that their energy expenditure rate increases faster with speed than in normal walking, the optimal speed is lower and the minimal cost is higher the higher the level of amputation.

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