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The effects of prosthesis mass on metabolic cost of ambulation in non-vascular trans-tibial amputees

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Abstract

The effect of prosthesis mass on the metabolic cost of steady-state walking was studied in ten male non-vascular trans-tibial amputees (TTAs) and ten non-amputee controls. The subjects underwent four trials of treadmill ambulation, with each trial performed for nine minutes at level grade and 76 m/min. Twenty minutes of seated rest followed each trial. During trials numbers one and two, TTAs ambulated without mass added to their prosthesis. During the third and fourth trials, either 454 or 907 grammes mass (1 or 21bs mass respectively) were randomly assigned and added to either the prosthesis or the leg of the non-amputee control. Subjects were blinded to the amount of mass added to their limb. Within-group comparisons across the four trials showed significant differences in oxygen consumption ($VO2$) and heart rate (HR) between the two non "mass added" trials, but no effect for addition of mass. The VO_2 of TTAs was only 0.6ml/kg/min (4.7 percent) greater during walking following the addition of 907 grammes to the prosthesis than without mass addition at all, while HR averaged only 1.4 beats/min. higher under the same testing condition. Pearson-product moment correlations echoed these findings, as moderate, but in all cases, negative correlations were observed for associations among the factors of subject age, stump length, and prosthesis-shoe weight, and both VO_2 and HR. It was concluded that adding up to 907 grammes mass to a non-vascular TTA's prosthesis will not significantly increase the energy expenditure or HR at a normal walking speed, and that elevated energy cost of ambulation in repeated measures testing without mass added may reflect task familiarisation and not an added burden of prosthesis mass.

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Introduction

An estimated 105,000 to 115,000 Americans lose a lower limb to amputation each year, with approximately 30,000 trans-tibial amputations being performed (Skinner and Effeney, 1985). Although the published literature lacks consensus concerning the distribution of causes for amputation in North America, it has been estimated that 70 to 90% are the result of disease, between 10 to 20% are for traumatic reasons, approximately 4% because of tumour, and 4% are congenital (Gailey *et al,* 1994; Ganguli *et al.,* 1974; Smidt, 1990).

Previous studies have reported that the metabolic cost of ambulation in trans-tibial amputees (TTA) is 15-55% higher than that of non-amputees, while ambulation velocity is 10-40% slower (Gailey *et al,* 1994; Ganguli *et al,* 1974; Gonzalez *et al.,* 1974; Huang *et al,* 1979; Pagliarulo *et al,* 1979; Waters *et al,* 1976 and 1988) (Table 1). It has been reported that traumatic TTAs walk at a faster pace while expending greater energy than vascular TTAs (Ganguli *et al,* 1974; Pagliarulo *et al,* 1979; Waters *et al,* 1976 and 1988). By comparison

Table 1. Metabolic cost and velocity of amputee ambulation.

Author	Cause of amputation	Number of subjects	Velocity (m/min)	Rate of oxygen uptake (ml O_2/kg/min)
Gonzalez *et al.* (1974)	Trauma	n=4	71	12.9
Waters *et al.* (1976)	Trauma	n=14	71	15.5
Pagliarulo *et al.* (1979)	Trauma	n=15	71	15.5
Gailey *et al.* (1974)	Trauma	n=39	70	12.9
Gonzalez *et al.* (1974)	PVD	n=4	60	11.1
Waters *et al.* (1976)	PVD	n=13	45	11.7

non-amputee controls have an ambulation VO_2 ranging between 10.9-12.95 ml/kg/min at speeds between 60-99m/min (Blessey *et al,* 1976; Corcoran and Brengelmann, 1970; Gailey *et al,* 1994; Perry, 1992; Smidt, 1990; Waters *et al,* 1976 and 1988). (Table 2). The mean comfortable walking speed for non-amputee subjects is observed to be 76-80 m/min (Finely and Cody, 1970; Waters *et al,* 1976). In contrast the mean walking speed of TTAs is 71 m/min. approximately 10% slower than non-amputee subjects (Gailey *et al,* 1994; Pagliarulo *et al,* 1979; Waters *et al.*, 1978).

Many elements contribute to altered gait mechanics, slow walking pace and elevated energy cost of ambulation in amputees. Among the most critical of these elements are: 1) the degree of displacement of the centre of mass over the base of support in all three planes of motion (Engsberg *et al,* 1990; Peizer *et al,* 1969; Saunders *et al,* 1937); 2) asymmetry of motion secondary to an imbalance of the muscle group actions of the lower limbs (Mensch and Ellis, 1986; Skinner and Effeney, 1985); 3) diminished coordinated movement between the ankle and knee joints consequent to the loss of proprioceptive feedback and musculature of the prosthetic joints (Ganguli *et al,* 1974; Mensch and Ellis, 1986; Skinner and Effeney, 1985; Waters *et al,* 1976); 4) the inability of the prosthesis to simulate the normal biomechanics and functions of the anatomical ankle and foot (Fisher and Gullickson, 1978; Radcliffe, 1961) thus altering the normal biomechanics of locomotion (Fisher and Gullickson, 1978); 5) the influence of the prosthetic design on the mechanics of gait, which may cause the amputee to deviate from a normal gait pattern and increase energy expenditure during walking (Murphy and Wilson, 1962; Radcliffe, 1955, 1957 and 1961[a,b]); 6) the loss of kinetic energy normally stored as potential energy in the anatomical limb during gait (Ehara *et al,* 1993; Ganguli *et al,* 1974; Gitter *et al,* 1991); 7) the loss of the skeletal lever arm, thus forcing proximal muscle groups acting on the remaining bone length to compensate for a longer lever arm and control the entire lower limb during the gait sequence (Ganguli *et al,* 1974; Gitter *et al,* 1991; Inman, 1967); 8) the loss of absolute amounts of contractile tissue mass, changes in insertion site, or altered functional capacity which will result in diminished potential strength (Eberhart *et al,* 1954; Ganguli *et al,* 1974; Klopsteg and Wilson, 1954; Winter and Sienko, 1988); 9) changes in body temperature regulation from loss of skin surface area which may disrupt the body's natural homeostasis

Table 2. Metabolic cost and velocity of normal ambulation

Author	Number of subjects	Velocity (m/min)	Rate of oxygen uptake (ml O_2/kg/min)
Blessey *et al.*, (1976)	n=40	82	12.95
Waters *et al.*, (1978)	n=20	82	12.95
Gailey *et al.*, (1994)	n=21	75	10.9

(Levy, 1983). Interestingly, the mass of the prosthesis is surprisingly missing from this list.

Both logic and a previous report (Inman, 1967) suggest that prosthesis mass should be included among the factors that influence energy expenditure and speed of ambulation in TTAs. However, the relationships among prosthetic mass, speed of ambulation, and energy cost of walking are not clearly defined. A recent report examined the effects of prosthesis mass on ambulation VO_2 in TTAs who used heavy prostheses operationally defined as greater than 2.27 kg and those who used light prostheses, defined as being of mass 2.27 kg or less. After controlling for stump length, age, speed of ambulation and baseline VO_2, there was no significant difference in ambulation VO_2 between subjects who used high and low mass prostheses (Gailey *et al*, 1994). Additionally, no significant correlation was observed between prosthesis mass and either ambulation VO_2 or heart rate (HR), and the best predictors of the physiological responses to walking were the subjects' resting (i.e. pre-ambulation non-exercise) VO_2, HR, and stump length. It is possible, however, that the findings of this study may have been influenced by its retrospective nature and cross-sectional design. Thus, the purpose of this prospective, randomised, control-design study was two-fold: 1) to compare the VO_2 and HR of TTAs and non-amputee control at a steady state walking speed of 76 m/min and 2) to determine the effects of mass on VO_2 and HR when added to the limb of TTAs and non-amputee controls.

Methods

Subjects

Subjects were ten mesomorphic males aged 24 to 52 years (x = 37.8 ± 10.4) with unilateral TTA. Ten non-amputee control subjects were matched with the TTAs for age (range = 23-51 years, x = 34.0±12.9 years), gender, and somatotype. The subjects with TTA were at least one year post-amputation from trauma or tumour but not vascular disease, while their intact limb was without injury or disability. All had a minimal stump tibial length of five centimetres and had used their current prosthesis for at least six months without skin irritation, pain, or other complication, the stump length was measured from the medial tibial plateau to the distal tibia. The prosthesis mass, both with and without the shoe, was recorded to the nearest gramme on a calibrated scale. Consent to undergo study was obtained from all subjects in accordance with the guidelines of the Institutional Medical Sciences Sub-committee for the Protection of Human Subjects. Descriptive characteristics of the subjects and their prostheses are shown in Table 3.

Ambulation trials

The subjects underwent four consecutive ambulation trials on a motorized treadmill (Trackmaster JAS Fitness System). Each trial involved ambulation at 76m/min for nine minutes at level grade followed by 20 minutes of seated rest. The first trial was conducted without addition of mass to the prosthesis, and

Table 3. Amputee demographic data

Subject	Age	Cause	Years since Amputation	Stump length (cm)	Prosthetic mass (g)
1	44	Congenital	34	2.4	1927.77
2	52	Trauma	24	2.68	2041.17
3	42	Trauma	8	1.08	1304.08
4	25	Trauma	11	4.33	1247.38
5	25	Sarcoma	10	2.26	1304.08
6	49	Trauma	44	2.07	1871.07
7	24	Trauma	1	2.56	1474.18
8	45	Trauma	17	2.56	2267.96
9	38	Trauma	6	3.35	1927.77
10	34	Trauma	4	3.54	2154.56
Mean	37.8		15.9	2.7	1752

was intended to familiarise subjects with treadmill ambulation and testing procedures. The second trial was a duplicate of the first. For the third and fourth trials, masses of either 454 or 907 grammes (1 or 2 lbs mass respectively) were added to their prosthesis in a randomly selected order. The subjects were blinded to the amount of mass added to the limb. Mass was affixed to the limb using a custom Velcro sleeve which has pockets designed to evenly distribute the added mass over the length of the intact limb or shank of the prosthesis. The sleeve was secured to the prosthesis shank of the amputee subjects, and was randomly assigned and then secured to the right or left shank for the non-TTA control subjects.

Measurements

Heart rate (HR) and oxygen uptake (VO_2) measurements were taken in the sitting position before testing. Exercise HR and VO_2 were collected very 15 seconds throughout the nine minute ambulation period with minutes 6 to 9 data averaged for the test results. Expired gases were collected in a mixing chamber from which the minute ventilation was computed by a turbine flowmeter. The fractional expired O_2 was continuously sampled from this chamber and averaged over the same 15 second period to compute the VO2. Heart rate was quantified by a Vantage Performance Monitor.[1] Oxygen uptake was measured on expired gases collected by a Hans Rudolf non-rebreathing valve[2] with assay of the gases performed on a Horizon System V Metabolic Measurements Analyzer[3] calibrated to known gas volumes and concentrations. Reliability of equipment and methodology were established by performing four trials on ten non-amputees matched with the TTAs for age, gender and somatotype.

[1]Polar Electro, Inc , Hartland, WI. USA
[2]Hans Rudolph Inc., Kansas Cily, MO, USA
[3]Sensormedics, Inc , Loma Linda, CA, USA

Data analysis

Data were analysed using a 3 x 2 x 2 design Analysis of Variance (ANOVA) for repeated measures, with main effects tested for trial (0g [first trial], 0g [second trial], 454g and 907g), sample time (resting pre-ambulation, post-ambulation) and condition (TTA vs non-TTA). Post-hoc analysis was conducted with simple effects tests. A Pearson product moment correlation was used to explore possible associations among selective descriptive characteristics (age, stump length, and prosthesis-shoe mass) and post-ambulation HR and VO_2. The criterion for statistical significant was set at the 0.05 level for all tests.

Results

All subjects reported feeling comfortable while walking on the treadmill. Between-group comparisons were significant for both HR and VO2 under all of the three testing trials (i.e. 0g, 454g and 907 g) ($p<0.05$). Oxygen consumption for TTAs was 14.3, 11.0, and 14.5 percent greater than control subjects when weighted with 0g, 454g, and 907g respectively. Likewise, HR values for TTAs were 30.8, 30.4, and 31.2 percent higher than those of control subjects when masses were added of 0, 454 and 907g respectively (Table 4). With respect to within group comparisons (i.e. the comparison of HR and VO2 across the three testing trials, each for TTA and control subjects), HR and VO_2 differed between the two trials with no added mass for amputee subjects ($p<0.05$). Otherwise, all within-group comparisons were non-significant. The VO_2 of TTAs was only 0.06

Table 4. Amputee and control group VO_2 and HR means.

Group	Means	Trial 1 (0g)	Trial 2 (0g)	Trial 3 (454g)	Trial 4 (907g)
Amputee group (n=10)	VO_2 (ml/kg/min) HR (bpm)	13.7±2.3 108.5±17.8	12.8±2.1 105.4±17.7	13.1±2.5 105.8±17.7	13.4±2.6 106.8±18.8
Control group (n=10)	VO_2 (ml/kg/min) HR (bpm)	11.4±1.2 80.6±9.3	11.2±1.2 80.6±9.3	11.8±1.3 81.1±9.1	11.7±1.0 81.4±9.2

ml/kg/min greater following the addition of 907 grammes to the prosthesis than with no added mass at all, while HR averaged only 1.4 beats/min higher. Pearson-product moment correlations showed highly positive associations between HR and VO_2 for all four ambulation trials (Table 5). Moderate, but in all cases negative, correlations were observed for associations among age, stump length, and prosthesis-shoe mass, and both HR and VO_2.

Discussion

The mass of an amputee's prosthesis has long been of concern to clinicians and amputees. Inman (1967) wrote that a prosthesis had to be of a minimum mass in order to: 1) maintain adequate momentum during the swing phase of gait; 2) carry the limb through stance; and 3) assist with push-off. Over the past two decades there has been an industry-wide movement to reduce the mass of prosthetic devices in an attempt to create ultralight prostheses. Given that the average mass of a human limb distal to

Table 5. Correlation for prosthetic mass, stump length, and age with respect to VO_2 and Heart Rate

	Trial 1 VO_2 (0g)	Trial 2 VO_2 (0g)	Trial 3 VO_2 (454g)	Trial 4 VO_2 (907g)	Trial 1 HR (0g)	Trial 2 HR (0g)	Trial 3 HR (454g)	Trial 4 HR (907g)
Trial 1 VO_2 (0g)	1							
Trial 2 VO_2 (0g)	0.959	1						
Trial 3 VO_2 (454g)	0.918	0.964	1					
Trial 4 VO_2 (907g)	0.85	0.933	0.948	1				
Trial 1 HR (0g)	0.817	0.794	0.724	0.729	1			
Trial 2 HR (0g)	0.766	0.767	0.689	0.703	0.973	1		
Trial 3 HR (454g)	0.812	0.825	0.754	0.791	0.977	0.988	1	
Trial 4 HR (907g)	0.777	0.805	0.736	0.798	0.955	0.978	0.995	1
Age	-0.295	-0.167	-0.212	-0.222	-0.366	-0.311	-0.339	-0.342
Length	-0.252	-0.47	-0.388	-0.451	-0.227	-0.303	-0.333	-0.349
Mass	-0.503	-0.442	-0.319	-0.437	-0.412	-0.44	-0.492	-0.516

the knee joint is approximately four percent of total body mass (Mensch and Ellis, 1986) an adult male of 80 kg mass ought to have a lower limb mass of 3.2 kg. By comparison, prostheses currently used by TTAs typically have a mass less than 2.3 kg.

The reduction in the mass of modern prostheses is primarily due to the increased selection of lightweight materials for fabrication, such as titanium, carbon fibre composites and copolymer plastics. Even though prostheses masses, in general, have been considerably reduced from just a decade ago, many prosthetists will often forgo the use of accessories such as rotators, shock absorption devices or durable covers to minimise the mass of the prosthesis.

The findings of this study suggest that small additional increments of mass will not significantly increase the energy cost of steady-state ambulation in TTAs, and that measured increases in this energy cost represents short-term acclimatisation to the testing conditions. With respect to the former finding, the authors have previously reported that prosthesis mass is unrelated to the energy cost of ambulation in subjects with traumatic TTA, although energy cost was influenced by the stump length when this length was stratified by long and short trans-tibial amputation (Gailey *et al,* 1994). With respect to the later finding, it is consistent with reports that observe an immediate reduction of submaximal oxygen consumption in subjects who begin an exercise training programme (Ekblom *et al,* 1968; Fox *et al,* 1975). This reduction in energy cost is attributed to motor skill familiarisation necessary to perform the exercise task and not physiological adaptation of the cardio-circulatory or muscular system (Ekblom *et al,* 1968; Fox *et al,* 1975).

The masses used for this trial were selected with the knowledge that most TTA prostheses do not vary greatly in total mass. The mean mass of the subjects prostheses was 1.77±0.43 kg. Coincidentally, the added test masses of 454 and 907 g reflected approximately one and two standard deviations above the mean, respectively, for the average prosthesis masses of the subjects. The addition of mass greater than 907 g would probably exceed the prosthesis mass a clinician might realistically encounter in practice. It is noted that the placement method for the additional mass in this study was important, as this was added throughout the entire length of the prosthesis to avoid concentrating mass at the distal end. This was accomplished by uniformly adding mass over the length of the shank to decrease the possibility of altering the swing phase momentum, and likely better represents a clinical situation in which a heavier prosthesis would not have all of the additional mass located in its distal segment.

This study tested subjects at a uniform walking speed of 76 m/min. Several studies report that the ambulation speed of non-amputees varies from 67-80 m/min (Corcoran and Brenglemann, 1970; Gailey *et al,* 1994; Pagliarulo *et al,* 1979; Peizer *et al,* 1969), while the average walking velocity of traumatic TTAs is approximately 71 m/min (Gailey *et al,* 1994; Pagliarulo *et al,* 1979). The subjects in this trial walked at a speed reflecting the average of these upper and lower limits. Ralston (1958) demonstrated that between 65-85 m/min there is no appreciable change in metabolic cost during ambulation. Moreover, none of the subjects in the trial reported feeling uncomfortable with the speed of their treadmill ambulation. The VO2 of amputee subjects during treadmill ambulation was 13% greater than that of the non-amputee controls. This observation is consistent with prior reports of Pagliarulo *et al.* (1979) and Gailey *et al.* (1994) who reported a 16% difference in the metabolic cost of ambulation at self-selected pace, between traumatic TTA males and matched non-amputee controls.

It is important to note that the findings of this study may not be generally applicable to all amputee subjects and that the results of added mass may vary with stump length. However, stump lengths for subjects tested in this study were well within the accepted length for TTA (Epps, 190; Levy, 1983). Levy (1983) reported that 12.7-17.0 cm is the ideal TTA stump length. While most subjects in this trial had similar stump lengths (x = 15.9±5.8cm), the stump of one subject was much shorter (6.99 cm) and one much longer (27.94 cm). Nevertheless, observation suggests that the measured results for these subjects did not differ from those of other subjects. Also, young, healthy, adult male amputees who lost their limbs from non-vascular reasons were selected for this study, as the intent was to measure the energy cost of their ambulation in the absence of confounding influences of vascular disease. In cases of vascular amputation, or procedures involving, trans-femoral or hip disarticulation amputations, the effect of prosthetic mass on energy cost may be more pronounced.

The authors have previous suggested that prosthetists have a "window of mass" in which to work when designing a prosthesis, and might use this window to fabricate limbs that include additional prosthetic components or accessories (Gailey *et al.,* 1994). This opinion is supported by the findings of this study. Prosthetic components such as rotators, mechanical ankle joints and shock absorbers may remain options for improved function and comfort without concern that the increased mass will adversely influence energy cost. Other options for prosthetic design include use of more durable covers or exoskeletal prostheses that may extend overall life of the prosthesis, and/or reduce their cost. While the findings of this study should not remove all concerns about designing heavier prosthesis, neither do they suggest that the mass of a prosthesis must reach minimum proportions to be considered "quality" or "high tech". The addition of more functional, comfortable and durable products to a prostheses, if acceptable to the amputee however, could become a greater consideration.

Conclusion

In conclusion, the addition of up to 907g of mass to the prostheses of non-vascular trans-tibial amputees does not alter the energy cost of ambulation at steady-state pace. Differences between unloaded trials result from habituation to the testing conditions. These findings suggest that prosthetists can design limbs using heavier components without significant increasing the amount of energy necessary to ambulate. The inclusion of components such as rotators, ankle joints, shock absorbers or more durable covers including exoskeletal designed limbs may allow traumatic trans-tibial amputees to improve their gait and function without compromising the energy cost of ambulation.

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