Optimizing Prosthetic Gait

Walking with a lower limb prosthesis can be a challenge. Walking requires a balance between the available physical capacity and the physical load experienced when walking. This thesis gives insight into the available physical capacity of people with a prosthesis and the underlying factors causing the increased physical load while walking with a prosthesis. Based on the results recommendations are formulated that help to improve the quality of life of people after a lower limb amputation, optimize prosthetic development and prosthetic rehabilitation.
Optimizing Prosthetic Gait

Balancing capacity and load

Daphne Wezenberg
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Optimizing Prosthetic Gait

Balancing capacity and load

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General Introduction
Introduction

Lower limb amputation, the surgical ablation of a lower limb or part thereof, has been one of the first surgical procedures ever performed. The level at which a lower limb amputation can be performed can vary from the toes up until as high as the pelvis. In this thesis I will focus on the amputations performed at the level of the tibia and femoral. Not surprisingly, the presence of a lower limb amputation drastically affects the quality of life \[63, 164, 198\]. A difference in the quality of life between etiology groups exists; larger physical disability and greater social isolation have been reported for people amputated due to vascular deficiency compared to people amputated after trauma \[43\]. Impairments in body function due to the loss of part of a limb can substantially hamper daily activity and functioning, and hence, can restrict active participation in the community \[42\]. In recent years technological advancements in operational techniques, prosthetic developments, and post-operative care of people after amputation, have paved the way for significant improvements in functioning and the quality of life of these patients. These developments are mediated by scientific research yielding a continuous growing body of knowledge on the biomechanical abnormalities and compensational strategies used by amputees \[35\].

Despite these scientific and clinical developments, people with a lower limb amputation still face many problems and challenges, especially when it comes to walking with the prosthesis. In the following sections the rationale behind the research, presented in this thesis is outlined. In this section it will be argued why it is important to restore walking ability, and it will be elucidated which factors can serve as important determinants for walking ability with a lower limb prosthesis. Before elaborating on the problems people face regaining mobility, the following paragraph will briefly discuss the number and characteristics of the people faced with a lower limb amputation.
Prevalence and incidence

No one will dispute that undergoing a lower limb amputation is a life-changing surgery. Unfortunately, a substantial number of people are faced with this operation every year. Reported incidence rates vary largely between geographical areas \cite{105,153}, reported values range between 1 per 100,000 women in Japan \cite{65} up to 47.2 per 100,000 for inhabitants of the United States \cite{48}. In war torn countries, land mines account for the vast majority of all amputations performed \cite{199}. In developmental countries, 34\% of all extremity amputations performed are due this type of trauma \cite{204}. Conversely, in developed countries, the vast majority (87-94\%) of lower limb amputations is performed due to vascular deficiency \cite{48,167,179}, whereas a substantially smaller number of amputations are performed due to trauma (3-4.6\%) \cite{48,65,179}. An even smaller number of people undergo amputation due to congenital limb deficiency \cite{65}, or malignant cancer \cite{60}. The population studied in this thesis stems from developed countries (The Netherlands and Italy), and from here on forward this thesis will focus on this population. Most of the patients who undergo amputation due to vascular deficiency have been previously diagnosed with peripheral arterial disease or diabetes mellitus. In the UK one in every three amputations performed involves patients with diabetes mellitus, whilst almost half of the people in Australia who undergo amputation are diabetic \cite{153}. Unfortunately, the overall total number of people that undergo amputation has increased over the years. In western countries, this increase is primarily caused by an increase in vascular deficiency related amputation \cite{48}. Moreover, because incidence rates increase with age \cite{48} and more people are likely to survive to old age, the older vascular amputee has become an increasingly important subgroup within the total population in need of prosthetic rehabilitation \cite{74}.

In the developed countries, incidence rates are expected to aggravate even further in the coming years as lifestyle-related diseases like peripheral vascular disease and diabetes mellitus become more common \cite{242}. This increase is thought to happen despite recent improvements in surgical intervention techniques, aimed at salvaging the limb. While intuitively one might think that undergoing a lower limb amputation is a last resort, and more drastically reduces the quality of life than limb salvage surgery, scientific research has failed to substantiate this. Studies examining the quality of life between these two populations of patients show no differences in quality of life scores \cite{16,63,173,202,239}. Some studies actually report less pain when walking \cite{16}, less anxiety and depression \cite{119}, and less post-operative complications \cite{16,63,119,202} after amputation compared to limb salvage surgery. Furthermore, although subjects with an amputation have a lower subjective rating of mobility \cite{39,63,120}, objective measures of functional mobility in terms of walking speed and oxygen cost do not differ between both groups \cite{202}.
Mobility

The level of mobility is an important determinant for the quality of life as it strongly influences the extent to which patients are able to regain much of the daily activities performed prior to the amputation. The relationship between mobility and quality of life has been investigated by Pell and colleagues (1993) [164]. They found that, despite larger social isolation and emotional distress in people with an amputation, impaired mobility was the only parameter associated with the quality of life after an amputation. Furthermore, early postoperative mobilization is strongly advocated in order to avoid deleterious effects of immobility, especially in the older persons [34].

Mobility is a term that encompasses a plentitude of different features. Based on the categorical differentiation of items related to mobility reported by van Bennekom and colleagues (1995) [211], Bussmann and colleagues (1998) made mobility explicit by stating that: “Mobility is the process of moving oneself, and of changing and maintaining postures” [19]. This definition is fairly broad, and in this context mobility can imply simply moving a leg while lying in bed or performing an activity like walking. In this thesis I focus on mobility in the latter context, more specifically, mobility in terms of subjects’ ability to walk with a prosthesis. Throughout the literature different parameters have been used to quantify walking ability ranging from the timed-up-and-go test [188, 214], functional walking distance [80], adaptability [111], functional scales [75, 182], ambulatory activity monitors [17, 77, 108], to self-reported scales [150, 188]. In this thesis subjects’ walking ability has been quantified by walking speed and walking economy (i.e. the energy expenditure needed to walk a given distance).

There are a number of determinant factors negatively associated with subjects’ walking ability, for instance, a low cognitive capacity [184, 214], age [99, 188], functionality prior to amputation [99, 184, 214], and the level of amputation [99, 184, 214]. While some studies also suggest a likewise negative association between comorbidity and walking ability, a recent review by Sansam and co-workers (2009) did not support this [184]. The presence of a vascular amputation is associated with a lower walking ability after amputation, but whether this can be attributed to the older age or the presence of comorbidity in these patients is unknown [184]. In conjunction, this study and clinical experience indicate that the older patients with a vascular trans-femoral amputation are assumed to have the lowest probability to regain and maintain walking.

The framework

To classify the limitation in walking ability that people might face after lower limb amputation, the International Classification of Functional, Disability and
The International Classification of Functioning, Disability and Health (ICF) framework can be used. The ICF provides a universal language for understanding human functioning and disability, and as such, provides a guideline for the description and classification of functioning in the context of health. The ICF is based on three domains; 1) body functions and structures, 2) activities and 3) participation (Figure 1). The limitations in walking ability are classified as a limitation within the activity domain. The factors associated with a poor walking ability are predominantly related to impairments within the domain of body functions and structures, and to external factors like prosthetic design and environmental factors (Figure 1). Ergo, the current thesis focuses on the impairments in the body functions and structures domain (e.g. aerobic capacity) and external factors (prosthetic push-off power) and relates this to the walking ability in terms of walking speed, oxygen cost and balance. These measures are often used in both scientific and clinical practice economy, and provide an objective evaluation of subjects’ walking ability. Insight into factors associated with subjects’ walking ability (such as the aforementioned factors like age, cause and level of amputation) is pivotal for intervention management and determining the likelihood of successful outcomes.
ambulation after amputation. However, merely assessing correlations between these factors and walking ability does not provide information about the underlying mechanisms, and hence does not reveal direct targets for intervention. While there are different mechanisms that may limit walking ability (e.g. wounds on the stump or limited motivation), this thesis takes a more bio-physical approach in that it focuses on the importance of maintaining a balance between the physical capacity and the physical load imposed while walking.

**Capacity versus load: the balloon analogy**

Imagine a balloon being filled with water. The more water is in the balloon the more strain will be put on the balloon, up until the moment the stress on the walls of the balloon is too much and the balloon bursts. Importantly, the larger the balloon the more water it can hold before it bursts. This analogy readily illustrates the balance between the capacity humans possess (i.e. the size of the balloon) and the load that is imposed when walking (i.e. the amount of water in the balloon): the lower the available capacity, the lower the load that can be comfortably handled. When the load exceeds a reasonable portion of the available capacity, the balance will be disturbed and walking will take great effort and might even prove impossible (i.e. the balloon bursts).

The term capacity can encompass various aspects of human functioning. The focus in my thesis is on the physical capacity of the subject. Physical capacity is a complex concept and is sub-divided, by the American College of Sports Medicine (ACSM) into five different components: aerobic power, anaerobic power, muscle strength, flexibility and neuromuscular control. While all of these aspects are thought to be in some way affected in people with a lower limb amputation, the focus in this thesis lays on the aerobic capacity. The peak aerobic capacity is the maximal metabolic power that can be derived from the oxidative pathways and is defined as the maximal amount of oxygen that can be utilized during exercise. The term aerobic load is the amount of oxygen that is utilized while performing an activity, which in this thesis is walking.

**Relative aerobic load**

People walking with a lower limb amputation are not only at risk of having a reduced aerobic capacity (i.e. smaller balloon), they also have to deal with an increased aerobic load while walking (i.e. more water). The notion that the aerobic load while walking with a prosthesis is substantially increased is a notion firmly entrenched and supported by numerous studies. The high stresses on the balloon, through a combination of low
capacity and high loads, render walking with a prosthesis a formidable challenge. Subjects may adapt to the higher strain by reducing their walking speed since this reduces the imposed load and resultantly, makes walking less demanding. From the above we can deduce that subjects’ ability to ambulate at normal speed with a prosthesis might be related to whether sufficient aerobic capacity is available (i.e. whether the size of the balloon is adequate) to compensate for the higher aerobic load imposed when walking. The exact ratio between aerobic load and aerobic capacity is currently unknown. The imposed load divided by the available capacity provides us with a quantitative measure for this balance, namely the relative aerobic load. Valid information about the relative aerobic load is imperative as it provides a firm scientific ground on which the necessity and structure of training regimes can be based.

In order to pinpoint the mechanism causing the limitations in walking ability of people with an amputation, and to provide evidence for adequate rehabilitation, information about the balance between capacity and load must be explored. In the following paragraphs an outline will be given on how different impairments of the body potentially influence the relative aerobic load. Initially the importance of sufficient physical capacity is argued. Secondly, the focus is shifted away from the physical capacity and directed towards understanding the mechanisms responsible for the higher load when walking with a prosthesis.

**Figure 2. The balloon analogy.**

The balance between aerobic capacity and aerobic load while walking can be visualized using the concept of a water balloon. The more water in the balloon the more strain (represented by yellow stripes) will be put on the balloon, up until the moment the stress becomes too much and the balloon will burst. The left balloon represents the situation when there is sufficient capacity and the aerobic load while walking is low. The right balloon might be the situation amputees face; the capacity is low (small balloon) and the load imposed when walking (the amount of water) is high. Resultantly, walking takes great effort and is perceived as strenuous.
Chapter 1

**Aerobic capacity**

The aerobic capacity is subject to both the ability of the cardiovascular and respiratory system to supply oxygen to the muscles, as well as the ability of the skeletal muscles to utilize this oxygen. The magnitude of peak aerobic capacity is, aside from a genetic predisposition, dependent on subjects’ activity level, sex, and is known to decline with age. Most of the people undergoing lower limb amputation are older adults and have had long periods of immobility preoperatively followed by a long period of convalescence due to, for example, delayed wound healing. Hence, these patients are likely to have a severely reduced aerobic capacity compared to their non-amputated healthy counterparts. But also those amputated at a young age are confronted with a perioperative period of limited physical activity, which also puts them at risk for reduced aerobic capacity. The reduced aerobic capacity might predispose subjects to poor walking ability. This has long been acknowledged by rehabilitations physicians and many rehabilitation interventions incorporate some form of aerobic training. Remarkably though, only a limited number of scientific studies report on peak aerobic capacity in amputees, and even fewer have focused on the elderly amputee most at risk of a reduced aerobic capacity. Hence, it is difficult to establish which role aerobic capacity plays in the process of regaining and maintaining walking ability in people with a prosthesis.

The paucity of articles reporting measured aerobic capacity of people after lower limb amputation might be attributed to the problems encountered when performing a maximal exercise test in this specific patient group. The impaired motor system, reduced muscle mass, balance problems, local problems in the stump as well as the contra-lateral limb, and the high prevalence of secondary coronary arterial diseases makes standard graded exercise testing unsuitable. A valid and safe exercise test can provide the clinician with important information concerning the cardiac condition of these patients, and adjustments to the rehabilitation program or medication intake can be made accordingly.

**Aerobic load**

Aside from improvements in the aerobic capacity, efforts aimed at reducing the elevated aerobic load can improve the ability to ambulate, as this reduces the relative aerobic load. However, the mechanisms responsible for the higher aerobic load in prosthetic gait are poorly understood. Two possible explanations have been proposed: First, energetically costly biomechanical adaptations in the remaining joints are required to compensate for the absence of active ankle power during prosthetic push-off, and secondly, energy is needed for compensational muscular activity to ensure stability while walking with a prosthesis.
Aerobic load for propulsion

To understand the importance of sufficient push-off power on the overall metabolic energy cost, human gait may be modeled as a double inverted pendulum (Figure 3). This so-called dynamic walking model was originally described by McGeer and colleagues (1990)\textsuperscript{[144]} and has been further elaborated upon by Kuo and colleagues\textsuperscript{[129-131]}. According to this model, the overall metabolic energy cost while walking is closely related to the work performed during the step-to-step transition\textsuperscript{[57]}. Moreover, the cost for the step-to-step transition can partly explain the higher metabolic energy cost found in people walking with a lower limb prosthesis\textsuperscript{[110]}. The most efficient way of reducing the step-to-step transition cost, and consequently reducing the metabolic energy cost, is by generating a push-off power along the trailing leg through ankle plantar flexion\textsuperscript{[129]} immediately preceding heel strike of the leading leg\textsuperscript{[181]}. In normal walking, this push-off power is primarily generated by the biological ankle\textsuperscript{[158, 234]}.

Unfortunately, the push-off power that can be generated by the prosthetic ankle is profoundly reduced, hence, step-to-step transition cost is higher and subjects with an amputation need to revert to other, less efficient, strategies to remain walking at the same walking speed\textsuperscript{[6, 110, 152]}. In recent years, lower limb prosthetic developments have progressed exponentially. These new developments aim at designing a prosthesis that is able to more closely mimic human ankle power during gait, thereby, reducing the step-to-step transition cost and concomitantly the metabolic energy cost while walking. Insight into the efficacy of these newly developed prostheses can be obtained using dynamic walking models. These models are particularly attractive because of their simplicity and ability to provide insight into net mechanical work performed during the step-to-step transition; moreover, they provide helpful insight into the metabolic cost while walking with different prosthesis.

Figure 3. The dynamic walking model.

In this model human walking is modeled as an inverted pendulum. During the step-to-step transition the center of mass (COM) velocity has to be redirected. This is accomplished with negative impact work performed by the leading leg and positive push-off work performed by the trailing leg. The work is calculated as the dot product of the forces under each limb and the center of mass velocity.
Aerobic load for balance

Simplifying metabolic cost to the mechanical work associated with the step-to-step transition, though very insightful, overlooks important other contributors to the overall metabolic energy cost. An important factor overlooked is the energy required to ensure stability while walking (i.e. the ability to recover from and adapt to perturbations) \(^{[134]}\). Alteration in the energy required for the control of balance might explain why some studies find a reduction in step-to-step transition cost, while the associated reduction in metabolic energy cost is either absent \(^{[192]}\), or lower than expected \(^{[98]}\). The human ability to walk without falling, i.e. to remain in balance, depends on the interaction between the passive dynamics of the musculoskeletal system and the active control by the central nervous system. Human walking is, to some extent, passively stable in the direction of progression; however, it requires active balance control in lateral direction \(^{[57, 128]}\). Gross control of lateral stability is predominately ensured by proper lateral foot placement \(^{[128]}\), while more refined adaptations are realized by lateral ankle movements after placement of the foot \(^{[102]}\). While the use of the lateral foot placement strategy is not compromised in people with a lower limb amputation, they lack the ability for more refined adaptation in the prosthetic ankle \(^{[102]}\) and might be forced to use less economic balance control strategies. To sum, in order to determine which factors contribute to the aerobic load imposed on the body while walking with a lower limb amputation information about both the effect of prosthetic ankle power and balance control is eminent.

Aim and research questions

The general aim of this thesis is to enhance our knowledge regarding the aerobic capacity and aerobic load while walking of people with a unilateral lower limb amputation and provide insight into how these factors influence the ability to regain and maintain walking. By doing so, this thesis aims to provide recommendations that can optimize rehabilitation, prosthetic prescription and development, and improve functioning and the quality of life of people after lower limb amputation.

More specifically this thesis aims to answer the following research questions:

- What is the relative aerobic load of people walking with a lower limb amputation and how does this relative aerobic load affect the walking ability?
- What is the aerobic capacity of people with a lower limb amputation and how can this be reliably determined?
- Which factors can be influenced in order to reduce the aerobic load while walking with a prosthesis?
Outline of thesis

A schematic representation of how the aforementioned influencing factors may relate and interact with one another is depicted in Figure 4. In the following paragraphs, I will briefly delineate how these factors will be addressed in subsequent chapters. The data presented in Chapter 2 till 4 are collected during one experimental cohort study. In each of these chapters specific research questions were addressed separately.

In Chapter 2 of this thesis, a detailed description of a newly developed standardized test for assessing the peak aerobic capacity of people with a unilateral lower limb amputation is given. In addition, information is presented about the feasibility and validity of this test.

Based on the results obtained from the study presented in Chapter 2, the proposed test is used to determine the peak aerobic capacity of older adults with a lower limb amputation in Chapter 3. In this chapter the association between the presence, level and cause of amputation on peak aerobic capacity is determined. This study augments the limited body of knowledge on peak aerobic capacity of elderly amputees with a lower limb prosthesis due to either trauma or vascular deficiency.

In Chapter 4 the peak aerobic capacity of people with a lower limb amputation is combined with the aerobic load measured while walking at different speeds to gain insight into the relative aerobic load while walking with a lower limb prosthesis. Again the association between the presence, level and cause of the amputation was investigated. Moreover, the relationship between relative aerobic load and walking economy and walking speed was determined. Furthermore, a data-driven model is constructed to gain insight into the potential effect of an increased aerobic capacity on the relative aerobic load, walking economy and walking speed.

Chapters 5 and 6 concentrate on potential factors that contribute to the higher aerobic load while walking with a prosthesis. In Chapter 5 the mechanical cost of walking with the energy storage and return prosthesis is compared to the cost when walking with the more conventional soft ankle cushioned heel prosthesis. This has been realized by applying the principles of dynamic walking by calculating the step-to-step transition cost. Possible differences in step-to-step transition cost between both feet are explained by close examination of the following parameters: step length, push-off power and roll-over shape.

The aim in Chapter 6 is to determine whether strategies applied for balance control can contribute to the overall metabolic cost of walking in able-bodied people. As very little is known on the energy cost for balance control, this chapter
is more fundamental in nature and involves healthy able-bodied controls. The energy cost for balance control is studied by enforcing a walking pattern similar to subjects’ unconstrained gait. By enforcing a gait pattern subjects can no longer rely on a stepping strategy to control balance in lateral direction, and balance is perturbed. Because global gait characteristics are set, changes in metabolic energy cost are a consequence of changes in the balance control strategies applied.

In Chapter 7 the main findings of the previous chapters are recapitulated and discussed. Based on the information deduced from the reported studies, future research recommendations are proposed. Finally, implications for clinical practice are presented and possible suggestions on how these results may be incorporated in the clinical prosthetic rehabilitation are provided.

![Diagram](image)

**Figure 4. Schematic representation of factors influencing prosthetic walking.**

The imposed aerobic load while walking divided by the aerobic capacity provides us with a qualitative measure for the effort experienced while walking, namely the relative aerobic load. The topics studied in this thesis are indicated by their chapter number.
Feasibility and validity of a graded one-legged cycle exercise test to determine peak aerobic capacity in older people with a lower limb amputation

Feasibility and validity of a graded one-legged cycle exercise test to determine peak aerobic capacity in older people with a lower limb amputation

Abstract

Information concerning the exercise tolerance and aerobic capacity is imperative for generating effective and safe exercise programs. However, for older people with a lower limb amputation, a standard exercise test is not available. The primary aim of the present study was to determine whether a graded one-legged peak exercise test is feasible and provides a valid assessment of peak aerobic capacity in older people walking with a lower limb prosthesis. A total of 36 older people with a lower limb prosthesis and 21 people who were able-bodied (controls; overall mean age = 61.7 years, SD = 6.1) performed a discontinuous, graded, one-legged, exercise test. The peak respiratory exchange ratio (RERpeak) was used as an indicator of maximal effort. The controls performed an additional two-legged exercise test in provide insight into differences the testing modes. All participants were able to perform the exercise test. Electrocardiographic tracings and blood pressure were adequately monitored. The controls and the people with a lower limb amputation were able to stress the cardiovascular system to a similar extent. Analyses of construct validity revealed that the peak aerobic capacity measured with the one-legged exercise test was able to distinguish between participants on the basis of age, body mass index and sex to a similar extent as the conventional two-legged exercise test. The graded one-legged exercise test was feasible, and provided a valid assessment of peak aerobic capacity and exercise tolerance in older people walking with a lower limb prosthesis.
Feasibility and validity of the exercise test

Introduction

Graded exercise testing has been widely acknowledged and implemented as a valid and effective way to assess physical fitness, diagnose coronary artery disease, allow for risk stratification \(^{[83, 154]}\), and determine the presence of other factors that limit exercise \(^{[115, 138]}\). Additionally, graded exercise testing allows for the development of safe, effective and individualized exercise programs and enables the objective evaluation of the exercise capacity in several groups of patients \(^{[115, 138, 154]}\). People with an amputation at the level of the lower limb are more likely to have a reduced aerobic capacity and are at an increased risk of coronary artery disease as a result of preexisting comorbidities \(^{[3, 122]}\) or hemodynamic changes \(^{[156]}\). Exercise testing can provide clinicians with important information concerning the exercise capacity and the cardiac condition of these patients \(^{[66, 180]}\). In addition, previous research showed that graded exercise testing can be used to predict walking ability \(^{[24-25, 66]}\) in people with an amputation at the level of the lower limb.

For testing people with a lower limb amputation, special consideration concerning the exercise mode and protocol used is necessary. For example, the impaired motor system, reduced muscle mass, balance problems, and problems associated with prolonged stump loading make application of the commonly used and recommended treadmill exercise test \(^{[83]}\) unsuitable \(^{[66, 210]}\). Several alternative exercise modes have been used \(^{[3, 12, 21-25, 29, 37, 66, 70, 122, 133, 170]}\) and compared \(^{[118, 220]}\) in people with a lower limb amputation. The two most commonly used exercise modalities are arm ergometry and one-legged exercise. However, several arguments favor one-legged exercise test for people with a unilateral lower limb amputation. First, one-legged exercise targets the remaining, still relatively large muscle group of the intact limb, resulting in a higher measured peak aerobic capacity than exercise with the small muscle mass of the upper extremities (e.g. arm exercise). Second, one-legged exercise results in lower perceived exertion and lactate values than arm ergometry, making one-leg cycling an exercise modality that is better tolerated \(^{[160]}\). Third, one-legged exercise allows for a functional evaluation of exercise capacity because it targets the muscles that are active during walking – a strenuous activity for people with a prosthesis \(^{[226]}\). Finally, one-legged exercise allows for reliable electrocardiogram (ECG) tracing because of limited movements of the upper extremities \(^{[76, 197]}\).

Tolerability and safety can be further improved by applying a discontinuous protocol; rest intervals in between exercise steps allow for noise-free ECG and adequate blood pressure recordings. Furthermore, the reduced intramuscular pressure during the rest intervals allows for metabolites to be transported over the cell membrane, possibly leading to a reduced feeling of local fatigue.
A pilot study showed that a discontinuous protocol consisting of 90 seconds of exercise separated by 30 seconds of rest, did not alter obtained the peak oxygen consumption during one-legged exercise, and it was experienced by participants as more comfortable than continuous exercise [231].

In several studies, a one-legged exercise test has been used to either predict [118], or directly measure [21-25] the peak aerobic capacity in people with a lower limb amputation. However, the feasibility of a discontinuous protocol in older people has not been studied. More importantly, because none of the previously reported studies included an age-matched control group, information about possible differences in exercise response between the older people who are able-bodied (controls) and people with a lower limb amputation is lacking.

The primary aims of the present study were to describe a discontinuous, graded, one-legged, peak exercise test and to determine whether it is feasible and provides a valid assessment of the peak aerobic capacity in older people currently walking with a unilateral lower limb prosthesis. With respect to feasibility, we hypothesized that study participants with an amputation were able to perform this exercise without being limited by feelings of discomfort other than fatigue and that the exercise set-up would provide adequate monitoring of the participants to create a safe test environment. Concerning validity, we hypothesized that both participants that were able-bodied and the people with a lower limb amputation would reach the criteria for maximal aerobic exercise. A secondary aim was to determine the concurrent validity of the one-legged exercise test in the older subjects. This was done by comparing the peak aerobic capacity obtained by the same group of controls during the one-legged and the conventional two-legged exercise test. The construct validity of the one-legged exercise test was determined by analyzing whether the peak aerobic capacity measured with the one-legged exercise test was able to distinguish between study participants on the basis of age, body mass index (BMI) and sex to the same extent as when measured using the conventional two-legged exercise test.

**Methods**

**Subjects**

A total of 37 people with a lower limb prosthesis and 21 people who were able-bodied (controls) agreed to participate in this study (mean age 61.7, SD 6.1). One individual with a lower limb amputation had abnormalities on the resting ECG and complained of chest pain before the exercise test; therefore, this individual was excluded from the study. Of the remaining 36 people with an amputation,
Feasibility and validity of the exercise test

26 had undergone amputation because of trauma and 10 had undergone amputation because of vascular deficiency. Of the 37 people with an amputation, 14 had amputations at the trans-femoral level and 23 at the trans-tibial level. The participants were aged between 50 and 75 years of age (Table 1). All participants with an amputation were able to walk four minutes with their prosthesis. All participants had had their amputation for more than 1 year before participation (Table 1), but a great deal of variation existed among participants (average years since amputation 27.2, range 1-66; Table 1).

Exclusion criteria were an absolute contraindication for exercise \textsuperscript{[143, 146]}, an impairment of cognitive function that would hamper understanding of the instructions, and a history of neurological diseases. None of the participants was extremely active or engaged in competitive sports. Before participation in the study, all participants with an amputation underwent an examination by a physician and medication intake was noted. After both verbal and written clarifications of the test protocol, the participants gave written informed consent.

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<th>Table 1. Participants’ characteristics</th>
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<td>BMI</td>
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<td>Years since amputation</td>
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Abbreviations: BMI, Body Mass Index

Exercise test

Participants were asked to refrain from drinking coffee and eating large meals at the day of the test and to avoid excessive exercise in the 24 hours preceding the exercise test. Before the test, the resting ECG and blood pressure were recorded. The ECG electrodes (CardioPerfect, Welch Allyn Benelux, Delft, The Netherlands) were placed as described by Edenbrandt and colleagues (1989) \textsuperscript{[61]}. The ECG was continuously monitored throughout the test, and blood pressure was recorded during each rest phase and after completion of the test.

The exercise test was performed with an electronically braked cycle ergometer (Lode Corival, Lode B.V., Groningen, The Netherlands) while using a discontinuous protocol in which 90 seconds of exercise was followed by a 30-second rest phase. Participants with a lower limb amputation pedalled with their non-amputated limb, whereas for the participants who were able-bodied the pedalling limb was randomly assigned. The nonexercising leg was positioned on a cushioned
extension plateau with the hip in approximately 90 degrees of flexion. Participants with a lower limb amputation were free to choose whether they wore their prosthesis during the test (Figure 1). The exercising leg was placed on a pedal with a large footplate and secured with Velcro (Velcro USA Inc, Manchester, New Hampshire) straps (Figure 1, inset). Participants were encouraged to maintain the frequency of pedalling at 50 to 60 revolutions per minute. Contrary to two-legged cycling, one-legged cycling requires torque to be exerted during both the push and pull phases of pedaling. For some participants this characteristic led to a distorted cycling rhythm, especially at the end of the pull phase. Cycling rhythm improved when participants were manually guided through the pull phase. Sufficient time was given for participants to become familiar with the setup.

For ensuring a time to exhaustion of 8 to 12 minutes \[83, 146, 155\], the starting workload and incremental steps were individually determined on the basis of the rate of perceived exertion (RPE) while participants pedaled at a range of workloads. More specifically, when participants indicated that the workload was somewhat hard, this workload was set as the initial starting workload. Incremental steps were based both on participants’ daily activity level and the subjective prediction of exercise performance by the test leader. This resulted in incremental steps for participants who were somewhat active, between 15 and 20 W, and for participants with lower physical activity levels, between 5 and 10 W. During the rest phase, the workload was reduced to 10 W; during the subsequent
Feasibility and validity of the exercise test

Figure 2. The oxygen uptake and heart rate response.

(A) Oxygen uptake response (oxygen consumption [VO2]) and (B) heart rate (HR) of a typical participant with a lower limb amputation during the one-legged exercise protocol. The load during the exercise protocol is depicted as a gray area; each exercise bout of 90 seconds is followed by 30 seconds of rest. The start and end of the exercise test are marked with vertical dashed lines. Both peak oxygen consumption and peak heart rate are depicted by a black square. This participant was unable to finish the last exercise phase; therefore, the participant’s peak power output was 110 W.

Exercise phase, workload was gradually increased over a period of 20 seconds toward the target workload (Figure 2). Throughout the exercise test, participants were verbally encouraged. The test was ended when either pedaling frequency dropped below 50 revolutions per minute, a further increase in workload did not lead to an increase in oxygen uptake, irregularities in the ECG, or extreme blood pressure changes were noted, or when participants indicated that they wanted to stop. Directly after the test was stopped, blood pressure was recorded and participants were asked which factor limited further exercise.
Chapter 2

The control group performed an additional, two-legged exercise test with the same discontinuous protocol as used during the one-legged exercise test. However, because power output is higher when exercise is performed with two legs than when it is performed with one leg, the starting workload and incremental steps were higher during the two-legged exercise test. The tests were separated by a minimum of two days.

**Outcome parameters**

Oxygen consumption was measured breath-by-breath using open-circuit respirometry (Oxycon Delta, CareFusion, Houten, The Netherlands) Breath-by-breath variability was attenuated with a three-breath smoothing average filter. The peak aerobic capacity ($\dot{V}O_{2peak}$ ml·kg$^{-1}$·min$^{-1}$), peak respiratory exchange ratio (RER$_{peak}$) and peak heart rate (HR$_{peak}$ beats·min$^{-1}$) were determined as the highest value attained during the last or penultimate exercise phase. In addition, HR$_{peak}$ was presented as the percentage of the predicted HR$_{peak}$ (HR$_{predicted}$). The predicted HR$_{peak}$ was calculated using the equation proposed by Tanaka and colleagues (2001)\(^{201}\) Only data from participants not using beta-blockers were used to calculate HR$_{peak}$ and HR$_{predicted}$. The peak power output (P$_{peak}$) was presented as the highest workload completed by a participant. The criterion for maximal exercise effort was defined as a RER$_{peak}$ value of greater than 1.1\(^{113, 155}\). The performance-limiting factors, indicated by the participants at the end of the exercise test, were placed in three groups according to whether the indicated limited factor represented fatigue of the legs (e.g. local fatigue), general fatigue or other remaining factors (such as pain in the chest region or saddle discomfort).

**Data analysis**

Descriptive statistics were generated for characteristics of the participants, protocol, and physiological parameters. Differences between the controls and the participants who had an amputation with respect to participants’ characteristics, protocol and physiological outcome parameters were analyzed using an unpaired $t$ test (SPSS Version 16.0, SPSS inc., Chicago, IL, USA) To determine whether participants with an amputation and controls differed with respect to the indicated performance-limiting factors, a chi-square test was performed.

Differences in protocol and physiological parameters between the one-, and two-legged exercise test, both performed by the controls, were analyzed with a paired-sample $t$ test. To determine whether values reached on the one-legged exercise test were consistent with values reached in the two-legged exercise test, concurrent validity was assessed by calculating the two-way mixed single-measure intraclass correlation coefficient (ICC$(3,1)$, type consistency)\(^{196}\). The
Feasibility and validity of the exercise test

A criterion for agreement was set at .75 for the lower limit of the 95% confidence interval (CI) of the ICC(3,1)\textsuperscript{[135]}. The standard error of the estimate was calculated to determine the magnitude of the difference between the exercise tests. To test whether the performance-limiting factor indicated by the able-bodied controls was different between exercise modalities, a McNemar test for paired responses was performed. Construct validity was tested by use of multiple linear regression analyses with $\dot{V}O_2^{\text{peak}}$ as the dependent variable and age, BMI and sex as the independent variables. The associations found for the one-legged exercise test were compared with those found for the two-legged exercise test. Because only the controls performed the subsidiary two-legged exercise test, only data obtained from the controls were entered in the model. The dependent variable, $\dot{V}O_2^{\text{peak}}$, ml·kg\textsuperscript{-1}·min\textsuperscript{-1} was natural log transformed to improve model fit. The level of significance was set at $p < .05$.

### Results

#### Feasibility

All participants were able to complete the exercise test and no abnormalities on the ECG tracing were noted during or after the exercise test. All participants indicated that fatigue (and not any other factor) was the reason for stopping the test. Overall ($n = 57$) significantly more participants indicated fatigue of the leg (as opposed to general fatigue) to be the performance-limiting factor during one-legged exercise ($\chi^2 = 14.754, p < .001$; Table 2). There was no statistical difference between groups in the indicated limiting factors ($\chi^2 = 3.287, p = .07$). Although the initial starting workload and incremental steps were, on average, lower in participants with a lower limb amputation, the differences were not significant. Likewise, total exercise times did not differ between the groups. On average, total exercise times remained below the recommended 12 minutes (Table 3)\textsuperscript{[83, 155]}.

<table>
<thead>
<tr>
<th></th>
<th>One-legged</th>
<th>Two-legged</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>amputees</td>
<td>able-bodied controls</td>
</tr>
<tr>
<td></td>
<td>n = 36</td>
<td>n = 21</td>
</tr>
<tr>
<td>Fatigue of the leg*</td>
<td>83.3% (n = 30)</td>
<td>61.9% (n = 13)</td>
</tr>
<tr>
<td>General fatigue</td>
<td>16.7% (n = 6)</td>
<td>38.1% (n = 8)</td>
</tr>
<tr>
<td>Other</td>
<td>0% (n = 0)</td>
<td>0% (n = 0)</td>
</tr>
</tbody>
</table>

* Significantly more participants indicated fatigue of the legs as a limiting factor during the one-legged exercise ($p < .001$).

---

**Table 2. Limiting factors**

<table>
<thead>
<tr>
<th></th>
<th>One-legged</th>
<th>Two-legged</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>amputees</td>
<td>able-bodied controls</td>
</tr>
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<td></td>
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</tr>
<tr>
<td>Other</td>
<td>0% (n = 0)</td>
<td>0% (n = 0)</td>
</tr>
</tbody>
</table>
Overall (n = 57), the predefined criterion for maximal effort was reached (mean RERpeak was 1.3, SD = 0.1; Table 4) during the one-legged exercise test. However, four participants with an amputation failed to reach a RERpeak of greater than 1.1 (range = 1.02 - 1.08). Overall (n = 57), participants reached, on average 89.1% (SD = 11.4%) of their age-corrected predicted HRpeak value. Neither HRpeak and RERpeak values differed between the controls and the participants with a lower limb amputation (p = .40, p = .93, respectively), indicating that both groups

### Table 3. Mean protocol meters (±SD).

<table>
<thead>
<tr>
<th></th>
<th>One-legged</th>
<th>Two-legged</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>amputees n = 36</td>
<td>able-bodied controls n = 21</td>
</tr>
<tr>
<td>Start workload (W)</td>
<td>67.8 (29.0)</td>
<td>74.5 (15.4)</td>
</tr>
<tr>
<td>Increments (W)</td>
<td>15.1 (4.1)</td>
<td>17.2 (3.1)</td>
</tr>
<tr>
<td>Time to exhaustion (min:sec)</td>
<td>8:23 (2:57)</td>
<td>9:07 (3:00)</td>
</tr>
<tr>
<td>Peak power output (W)</td>
<td>121.0 (40.8)</td>
<td>141.9 (36.3)</td>
</tr>
</tbody>
</table>

* Significantly different from the value found using the one-legged exercise test (both tests are performed by controls; p < .05).

Abbreviation: $P_{peak}$, peak power output.

### Validity

Overall (n = 57), the predefined criterion for maximal effort was reached (mean RERpeak was 1.3, SD = 0.1; Table 4) during the one-legged exercise test. However, four participants with an amputation failed to reach a RERpeak of greater than 1.1 (range = 1.02 - 1.08). Overall (n = 57), participants reached, on average 89.1% (SD = 11.4%) of their age-corrected predicted HRpeak value. Neither HRpeak and RERpeak values differed between the controls and the participants with a lower limb amputation (p = .40, p = .93, respectively), indicating that both groups

### Table 4. Mean peak physiological values (±SD).

<table>
<thead>
<tr>
<th></th>
<th>One-legged</th>
<th>Two-legged</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>amputees n = 36</td>
<td>able-bodied controls n = 21</td>
</tr>
<tr>
<td>$\dot{V}O_{2peak}$ (ml·kg⁻¹·min⁻¹)</td>
<td>25.1 (7.9)*</td>
<td>30.8 (10.3)</td>
</tr>
<tr>
<td>$\dot{V}CO_{2peak}$ (ml·kg⁻¹·min⁻¹)</td>
<td>26.4 (8.8)*</td>
<td>33.3 (11.4)</td>
</tr>
<tr>
<td>RERpeak</td>
<td>1.3 (0.2)</td>
<td>1.3 (0.1)</td>
</tr>
<tr>
<td>RR (beats·min⁻¹)</td>
<td>44.5 (10.9)</td>
<td>48.0 (13.1)</td>
</tr>
<tr>
<td>VE (l·min⁻¹)</td>
<td>77.1 (22.7)*</td>
<td>91.1 (27.3)</td>
</tr>
<tr>
<td>HR (beats·min⁻¹)</td>
<td>146.7 (21.7)</td>
<td>151.6 (16.5)</td>
</tr>
<tr>
<td>HR%predicted (%)</td>
<td>88.9 (12.3)</td>
<td>91.6 (9.2)</td>
</tr>
</tbody>
</table>

Abbreviations: $\dot{V}O_{2peak}$, peak aerobic capacity (peak oxygen consumption); $\dot{V}CO_{2peak}$, peak carbon dioxide output; RERpeak, respiratory exchange rate; RR, respiratory rate; VE, minute ventilation; HRpeak, peak heart rate; HR%predicted, the peak heart rate expressed as a percentage of the predicted peak heart rate.

* Significantly different from controls (p < .05).
† Significantly different from one-legged exercise (both tests are performed by controls; p < .05).
‡ In total three participants were taking β-blockers and therefore, were excluded in the calculation of HRpeak and HR%predicted.
reached similar levels of exertion. Nonetheless, the averaged value for peak aerobic capacity, peak carbon dioxide output, and minute ventilation were significantly lower for participants with an amputation than for controls ($p = .02$, $p = .01$, $p = .04$, respectively; Table 4). Figure 2 shows the oxygen uptake response and heart rate of a typical participant with a lower limb amputation during the exercise test.

Analyses of the one- and two-legged exercise tests, both performed by the controls, revealed that peak power output was 34.3% lower in the one-legged exercise test than in the two-legged exercise ($p < .001$). In the one-legged exercise test, the controls reached an average peak aerobic capacity of 30.8 ml·kg$^{-1}$·min$^{-1}$ (SD = 10.3); this value was significantly lower than that reached in the two-legged exercise test (37.9 ml·kg$^{-1}$·min$^{-1}$, SD = 11.0, $p < .001$; Table 4). The HR$_{peak}$ value was 7.5% lower during the one-legged exercise test; evidently this value also resulted in a lower HR$_{%predicted}$ (91.6% versus 99%; Table 4). Despite the difference in the HR$_{peak}$ values, no difference in RER$_{peak}$ values was found between the one- and two-legged exercise tests. The ICC(3,1) for the one- and two-legged exercise tests was 0.89 (95% confidence interval = 0.75 - 0.95; $p < .001$; Figure 3). The standard error of the estimate calculated for the one-legged versus the two-legged exercise was 5.1 (Figure 3). The McNemar marginal homogeneity test revealed no statistical differences in limiting factors between the test modalities ($p = .34$; Table 2).

Multiple linear regression analyses revealed that the peak aerobic capacity measured using either the one-legged exercise test or the two-legged exercise test was associated with participants’ age, BMI and sex. As expected, age and BMI were inversely associated with peak aerobic capacity, and men reached higher peak aerobic capacities than women. The percentage change, as calculated from

![Regression line](image-url)
Table 5. Regression Models.

<table>
<thead>
<tr>
<th>Entered variables</th>
<th>β-coefficients</th>
<th>β-standardized</th>
<th>p</th>
<th>%change</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>β (95% CI)</td>
<td>β-standardized</td>
<td>p</td>
<td>%change</td>
</tr>
<tr>
<td>Model one-legged (n = 21)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>age</td>
<td>-0.017 (-0.032 - -0.003)</td>
<td>-316 p = .02</td>
<td>-1.7%</td>
<td></td>
</tr>
<tr>
<td>BMI</td>
<td>-0.052 (-0.075 - -0.029)</td>
<td>-593 p &lt; .001</td>
<td>-5.1%</td>
<td></td>
</tr>
<tr>
<td>Sex (men[1]/women[2])</td>
<td>-0.233 (-0.417 - -0.048)</td>
<td>-347 p = .02</td>
<td>-20.7%</td>
<td></td>
</tr>
<tr>
<td>Model two-legged (n = 21)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>age</td>
<td>-0.020 (-0.037 - -0.004)</td>
<td>-383 p = .02</td>
<td>-2.0%</td>
<td></td>
</tr>
<tr>
<td>BMI</td>
<td>-0.040 (-0.066 - -0.014)</td>
<td>-473 p = .004</td>
<td>-3.9%</td>
<td></td>
</tr>
<tr>
<td>sex (men[1]/women[2])</td>
<td>-0.226 (-0.430 - -0.022)</td>
<td>-350 p = .03</td>
<td>-20.2%</td>
<td></td>
</tr>
</tbody>
</table>

The dependent variable was the natural log-transformed $\overline{\text{VO}_{2\text{peak}}}$.

Abbreviations: $\beta$, unstandardized beta coefficient; 95% CI, 95% confidence interval; $\beta_{\text{stand}}$, standardized beta coefficient; %change, the percentage change in peak aerobic capacity when the independent parameter is changed with one unit (e.g. for the one-legged exercise a one year increase in age leads to a 1.7% reduction in peak aerobic capacity); $R^2_{\text{adj}}$, is the precision of estimate adjusted for the multiple explanatory variables in the model.

the regression analysis, revealed similar influences of age, BMI and sex on peak aerobic capacity in both the one-legged exercise and the two-legged exercise tests (Table 5).

Discussion

The aims of the present study were to describe a discontinuous, graded, one-legged, peak exercise test and to determine whether it is a feasible and valid method for assessing the peak aerobic capacity in older people walking with a lower limb prosthesis.

Feasibility

All participants were able to tolerate the exercise test, and none of the participants experienced post-exercise discomforts beyond short-term feelings of fatigue that were logically related to the exercise performed. The same proportions of participants that were able-bodied (controls) and participants who had an amputation indicated local fatigue to be the limiting factor during the exercise test, indicating that both groups experienced the same limitations. Even though upright cycling is a familiar activity for most people in The Netherlands, performing this activity with one-leg is not; hence, sufficient familiarization time is important. Although several participants needed guidance of the leg to
Feasibility and validity of the exercise test

improve cycling rhythm, none of the participants indicated this was a limiting factor during the exercise test. Whereas previous research has noted that when participants were asked to perform a more complex coordination pattern consisting of combined arm and leg activation, movement rhythm could become a limiting factor [197,220].

Concerning the exercise protocol, the starting workload and the sizes of the incremental steps resulted in an overall (n = 57) average time to exhaustion that was well within the recommended maximum of 12 minutes [83, 146, 155] (Table 3). Nevertheless, for six participants (three controls and three with an amputation) the time to exhaustion exceeded the recommended 12 minutes by one and half minutes. Fortunately, the elongated exercise time did not lead to associated discomfort that might have limited exercise performance in these participants.

With regard to safety, numerous studies have indicated the importance of blood pressure and ECG monitoring in people with an amputation [3, 180, 197]. As was hypothesized, the rest intervals enabled accurate and regular blood pressure recordings and noise-free ECG tracings, thereby allowing adequate screening for cardiac abnormalities. To the contrary, adequate monitoring is limited during arm ergometry or combined arm and leg exercise because movements of the upper limbs produce low signal-to-noise ratio in ECG tracings [76, 171, 197, 220]. Because the exercise test in the present study was performed with the legs, the ECG also could be relatively accurately monitored during the exercise phase. These findings are in agreement with those of previous studies in which a discontinuous protocol was used to adequately monitor people with an amputation [12, 37, 70, 171]. During the rest phases the participants could indicate any discomforts, and the test leader was able to encourage the participants to try and complete the next exercise phase. We conclude that the discontinuous, graded, one-legged, peak exercise test is feasible because it proved to be both tolerable and safe in older people with a unilateral lower limb amputation.

Validity

To obtain a valid assessment of peak aerobic capacity, an exercise test should stress the cardiopulmonary system to a sufficient extent. However, a substantial amount of literature states that peak aerobic capacity is strongly dependent on the exercise mode and the muscle mass involved. Because of the limited active muscle mass involved during cycling with one-leg, the cardiopulmonary system might not be stressed to its full potential and exercise might be limited by peripheral factors [38, 145, 159, 191]. Additionally, because of vascular deficiencies or hemodynamic changes, people with an amputation might respond differently to a one-legged exercise test than people who are able-bodied [156]. Different criteria for determining whether the cardiovascular system has been stressed to its full
potential have been used and discussed in the literature. In the present study we have chosen to use a RER$_{\text{peak}}$ value greater than 1.1 as the criterion for maximal effort. This choice was based on the grounds that the discontinuous nature of the protocol prevented a clear detection of a plateau in the oxygen uptake and because the considerable interindividual variability inherent in the age-corrected predicted maximal heart rate precludes the use of these parameters.$^{[113]}$ In total, 93.0% of the participants reached the a priori set criterion for maximal effort. These results, in addition to the fact that neither RER$_{\text{peak}}$ nor HR$_{\text{peak}}$ values differed between the groups, corroborated the notion that both older participants with an amputation and controls were able to stress the cardiopulmonary system to the same nearly maximal, extent during the one-legged exercise test.

Concurrent validity was determined by comparing the peak aerobic capacity reached in the one-legged test to that reached during the two-legged exercise test. In the one-legged exercise test, participants who were able-bodied (controls) were able to utilize 81.3% of the amount of oxygen utilized in the two-legged exercise test. This value is somewhat higher than the averaged value found for young participants.$^{[38, 123, 136, 145, 159-160]}$ Despite this systematic difference between both tests, the a priori set criterium (the lower bound of the confidence interval was $\leq .75$) for agreement between the tests was reached ($\text{ICC}_{[3,1]}$ 0.89, 95% confidence interval = 0.75 - 0.95)$^{[135]}$, indicating that the one-legged test is a good predictor for the peak aerobic capacity reached during the two-legged test. Nevertheless, there consisted a large interindividual variability in the relationship between the tests, as indicated by the rather high standard error of estimate. Interestingly, these rather high prediction errors corresponded to the magnitude of differences reported in other repeatability studies using a variety of exercise modes.$^{[197, 210]}$. These results indicated that the variation in agreement between one-, and two-legged exercise tests might depend on the test-retest variability and might be independent of the exercise mode used.

The associations between peak aerobic capacity and age, BMI and sex have been confirmed in numerous studies over the years. An analysis of construct validity revealed that the one-legged exercise test was able to replicate these associations. The nature of the relationship confirmed our expectation; age and BMI were negatively related, and men performed significantly better than women. A comparison of the established two-legged exercise test and the one-legged exercise test revealed that age, BMI and sex influenced the measured peak aerobic capacity to similar extents. For example, calculation of the percentage change (Table 5) revealed that when age was increased by one year, the peak aerobic capacity was predicted to be 1.7% lower when the equation based on the one-legged exercise test was used; this value is in agreement with the 2.0% reduction found using the regression equation based on the two-legged exercise test was used. These results demonstrated that the peak aerobic capacity
measured with the one-legged exercise test was able to distinguish participants on the basis of age, BMI and sex to a similar extent, as would be expected when using a conventional peak exercise test.

The question remains as to whether the one-legged exercise test is a valid measure for assessing peak aerobic capacity and exercise tolerance. The aforementioned characteristics of the one-legged exercise test revealed that the controls and participants who had an amputation performed equally well with regards to maximal effort and that the one-legged exercise test had both a high concurrent and a high construct validity. Albeit, the one-legged exercise test did not stress the controls to the same extent as a two-legged exercise test given the systematic difference found between the tests in peak aerobic capacity and HRpeak. However, for people walking with a prosthesis, the capacity predicted by the two-legged exercise test is purely hypothetical and might not be valid because the additional muscle mass utilizing the oxygen in two-legged exercise is, in part, absent. As a consequence, the peak aerobic capacity measured with the one-legged exercise test could be regarded as a good approximation of the theoretical ‘true’ maximal aerobic capacity in people walking with a lower limb amputation. At least it targets the same muscles that are active during strenuous aerobic activities of daily living, thereby providing a good assessment of aerobic tolerance during daily activities.

**Limitation and future research**

Current research is subject to some limitations. The major limitation is that the participants with a lower limb amputation were all healthy and relatively active; all participants were able to ambulate with their prosthesis and had had their amputation for more than one year. As a consequence, the results cannot be generalized to people who have undergone amputation more recently and who are less active and possibly not accustomed to exercise. Therefore, future studies are needed to determine the applicability of the exercise test described here for this population. Future studies also should focus on the sensitivity and reproducibility of the exercise test and on providing insight into the magnitude of the difference that can be reliably detected at the individual level.

Furthermore, in the present study we tested the feasibility and validity of a one-leg cycle test. Perhaps other exercise modalities, such as those using the arms or a combination of the arms and legs, might provide an equally effective functional evaluation.\(^{12, 76}\) However, arm ergometry might result in increased arterial pressure\(^ {76, 136, 191}\) and problems in obtaining reliable ECG tracings and blood pressure recordings\(^ {197}\), and exercise combining both the arms and legs might pose coordination difficulties which could hamper the feasibility of the exercise test. Because the present study focused merely on the applicability and
validity of the discontinuous, one-legged, exercise test, we cannot conclude that the protocol is superior to other modalities. Further studies comparing different exercise modalities are needed.

**Conclusion**

The results of the present study demonstrated that the discontinuous, graded, one-legged, peak exercise test is feasible and provides a valid mean for assessing peak aerobic capacity and exercise tolerance in people walking with a lower limb prosthesis. The exercise test proposed in the present study can be easily implemented as the necessary equipment and adaptations are relatively inexpensive and widely available. The exercise set-up described here could be used as a safe and effective way to test people walking with a lower limb prosthesis. The test can help to assess the peak aerobic capacity and exercise tolerance and it can be used to design safe, effective, and individualized exercise programs. Further research to determine the ability of the protocol to detect changes in peak aerobic capacity and to determine the feasibility of the exercise test for other populations is warranted.
Feasibility and validity of the exercise test
Peak oxygen consumption in older adults with a lower limb amputation

Peak oxygen consumption in older adults with a lower limb amputation

Abstract

The objective of this study was to investigate whether the aerobic capacity of older adults who underwent a lower limb amputation is associated with the presence, cause (traumatic or vascular), and level of amputation (trans-tibial or trans-femoral). A total of 36 older subjects who underwent lower limb amputation and age-matched (n=21) able-bodied controls participated. All subjects were able to walk for a minimum of 4 minutes. Peak oxygen consumption ($\bar{VO}_{2peak}$) was measured using open-circuit respirometry while performing a discontinuous, graded, one-legged, peak cycle exercise test. After correcting for age, body mass index, and sex, the multiple linear regression analysis revealed that subjects who underwent amputation had a 13.1% lower aerobic capacity compared with able-bodied controls ($p = .021$). Differentiation among etiologies revealed that subjects with a vascular amputation had a lower $\bar{VO}_{2peak}$ of 29.1% compared with able-bodied controls ($p < .001$), whereas traumatic amputees did not differ from able-bodied controls ($p = .127$). After correcting for etiology, no association between level of amputation and $\bar{VO}_{2peak}$ was found ($p = .534$). Older adults who underwent an amputation because of vascular deficiency had a lower aerobic capacity compared with able-bodied controls and people with a traumatic amputation. The level of amputation was not associated with $\bar{VO}_{2peak}$. 
Introduction

The problems associated with regaining walking ability after undergoing lower limb amputation can be multidimensional and, amongst others, be related to poor prosthetic fitting, stump problems, impaired cognitive ability \[184\], problems on the non-amputated side, and reduced balance \[216\]. In addition, there is evidence suggesting that a low peak oxygen consumption can negatively influence walking ability \[22, 24-25, 184\]. Peak aerobic capacity is defined as the highest peak oxygen consumption obtained during maximal effort exercise \[40\], and is known to decline with age \[115\]. This decline can be severely aggravated in people with an amputation because of a perioperative period of limited physical activity \[207\] or comorbidities. As a result, differences might exist between individuals who underwent amputation because of trauma or because of vascular deficiency. Additionally, peak aerobic capacity might be influenced by the level of amputation. It has been shown that the higher the level of amputation the lower the walking distance \[80\] and activity level in daily life \[184\], which might result in a lower peak aerobic capacity \[226\].

Numerous studies have investigated the oxygen demand for walking with a prosthesis \[78, 81, 110, 209, 226\]. In contrast, only a limited number of studies have focused on the available aerobic capacity \[216\]. To assess physical fitness, direct measurement of the oxygen consumption during a maximal exercise test remains the preferred method in not only in young and elderly people \[115\], but also in people after lower limb amputation. Chin and colleagues (2002 and 2006) \[24-25\] report the physical fitness of older adults with an amputation because of vascular deficiency by determining the proportion of the predicted maximum oxygen uptake that was reached during a maximal exercise test. Others have used similar prediction equations to determine the physical fitness of people with an amputation by using data obtained during submaximal exercise testing \[122, 133, 226\]. However, the exercise mode needs to be adapted when testing people with an amputation, and the amputee population differs from controls with respect to physical limitations, comorbidity, medication intake, and age. Therefore, using general prediction equations to evaluate aerobic capacity is not viewed as a valid approach when reporting results on peak aerobic capacity in older people with a lower limb amputation \[5\].

Even though a maximal exercise test is the most reliable method to determine subjects’ peak aerobic capacity, the impaired motor system, reduced muscle mass, comorbidities and local problems in the stump as well as the contralateral limb make standard graded exercise testing impossible. To overcome these difficulties, we previously developed a protocol that allows for the safe and reliable determination of the peak aerobic capacity of people with an amputation using a widely available cycle ergometer that is driven with the intact leg \[229\].
The aim of the present study was to investigate the VO$_{2peak}$ of older adults who underwent a lower limb amputation because of trauma or vascular deficiency by using a discontinuous, graded, one-legged, peak cycle exercise test. It was hypothesized that people who underwent lower limb amputation would have a lower VO$_{2peak}$ compared with controls, and that the lowest level would be found in the group with a vascular amputation. Additionally, it was hypothesized that the level of amputation would influence VO$_{2peak}$, with a more proximal amputation relating to a lower VO$_{2peak}$. Multiple linear regression analysis was used to determine to what extent the presence of an amputation, cause of amputation (traumatic vs vascular), or level of amputation (trans-tibial vs trans-femoral) was associated with VO$_{2peak}$.

**Methods**

**Subjects**

Potential subjects for this study underwent prosthetic rehabilitation after lower limb amputation between January 1992 and March 2009, and were currently known by the rehabilitation physician or prosthetist to walk with their prosthesis. From this list only those subjects were selected who (1) were aged between 50 and 75 years at the time of the study; (2) had undergone a unilateral amputation of the lower limb at the level of the lower leg (trans-tibial) or upper leg (trans-femoral); (3) had a time since amputation longer than one year; and (4) were able to ambulate for at least 4 minutes. Exclusion criteria were absolute contraindication for exercise$^{[143,146]}$, impairment of cognitive function, and neurological problems. Before participation, all subjects underwent clinical screening by a physician.

A total of 37 subjects with a lower limb amputation agreed to participate. One subject showed abnormalities on the resting electrocardiogram (ECG) and complained of chest pain, and was excluded from the study. Of the remaining 36 subjects, 26 had amputations after trauma, with 16 subjects having amputations at the trans-tibial level and 10 at the trans-femoral. Of the 10 subjects who had amputations because of vascular deficiency 7 had amputations at the trans-tibial level and 3 at the trans-femoral level. An additional group of 21 able-bodied older controls (aged between 50 and 75 y) were recruited through an advertisement in a local newspaper. After both verbal and written clarification of the test protocol, subjects gave written informed consent. This study carried the approval of the Medical Ethical Review Board of the VU University Medical Centre in Amsterdam.
Exercise protocol

Before exercise, a resting ECG was performed and the blood pressure was recorded. The VO_{peak} was determined using a graded, discontinuous, one-legged, exercise test, for which the feasibility and validity for this population had previously been verified [229]. During this test, subjects cycled with their nonamputated leg on an electronically braked cycle ergometer (Lode Corival, Lode BV, Groningen, The Netherlands), whereas in the able-bodied control group the exercising leg was randomly assigned. The graded peak exercise test was executed using a discontinuous protocol: each exercise phase lasted 90 seconds and was followed by 30 seconds of rest. After familiarization, start workload and increment steps were individually determined to ensure a test duration between 8 and 12 minutes [229]. Throughout the duration of the test, subjects were verbally encouraged. The test was ended when either pedaling frequency dropped below 50 revolutions per minute, a further rise in workload did not lead to an increase in oxygen uptake, irregularities on the ECG or extreme blood pressure alternations were noted, or when subjects indicated that they wanted to stop. The test protocol is described in more detail elsewhere [229]. Subjects were asked to refrain from drinking coffee and eating large meals on the day of the test, and not to be involved in excessive exercise in the 24 hours preceding the test.

Measurement

During the test, oxygen consumption was measured breath-by-breath using open-circuit respirometry (Oxycon Delta, CareFusion, Houten, The Netherlands). Breath-by-breath variability was attenuated using a three-breath smoothing average filter. The VO_{peak} (ml·kg\(^{-1}\)·min\(^{-1}\)), peak respiratory exchange ratio (RER_{peak}) and peak heart rate (HR_{peak} beats·min\(^{-1}\)) were determined as the highest value attained during the last or penultimate exercise phase. Predicted maximal heart rate was calculated using the equation proposed by Tanaka and colleagues (2001) [201], and used to determine the percentage of predicted maximal heart rate (HR_{%predicted}) reached during the test. Peak power output (W) was determined as the highest workload completed by the subject. Body weight was determined as the total weight of the subject excluding the weight of the subjects’ prosthesis.

Statistical analysis

All data were analyzed using a computerized statistical package (Version 16.0; Spss Inc. Chicago, IL, USA). Differences between groups in subject characteristics and outcome parameters were analyzed using a one-way analysis of variance, with a Bonferroni post hoc test to control for the multiple comparison. A
Pearson chi-square test was performed to determine whether the allocation to the different groups was independent of sex. To study whether $\hat{\text{VO}}_{\text{peak}}^2$ was associated with the presence of an amputation, cause of amputation, and level of amputation, a multiple linear regression analysis was performed in which $\hat{\text{VO}}_{\text{peak}}^2$ was the dependent variable and the possible influencing factors were the independent variables. As age, body mass index (BMI) and sex are independently related to $\hat{\text{VO}}_{\text{peak}}^2$, these three parameters were entered as confounders in the model. Visualization of the residuals of the regression model revealed skewness; therefore, $\hat{\text{VO}}_{\text{peak}}^2$ was natural log-transformed to improve model fit. The linear equations determined by the multiple regression analysis had the following general form:

$$\ln(\hat{\text{VO}}_{\text{peak}}^2) = \beta_0 + \beta_a \cdot \text{age}_i + \beta_b \cdot \text{BMI}_i + \beta_c \cdot \text{sex}_i + \sum_{j=1}^{m} \beta_j \cdot X_j + \varepsilon \quad (1)$$

where the coefficients $\beta_0$, $\beta_a$, $\beta_b$, and $\beta_c$ are constants representing the coefficients associated with the confounding independent variables age, BMI, and sex, respectively, and $\varepsilon$ indicates the random error. Subscript $i$ indicates the subject. $\beta_j$ is the coefficient associated with the independent variable of interest ($X_j$). The $m$ represent the number of independent variables of interest entered in the model. Four models were constructed. The first model was used to determine whether the presence of an amputation was associated with $\hat{\text{VO}}_{\text{peak}}^2$. This variable (amputation [yes/no]) was entered as an independent dummy variable in the first model ($n = 57$). To further determine whether differences in $\hat{\text{VO}}_{\text{peak}}^2$ between people with an amputation and able-bodied controls could be explained by a difference in the cause of amputation, a second model was constructed with two dummy variables (traumatic [yes/no] and vascular [yes/no], $n = 57$). A similar third model was constructed with both levels of amputation (trans-tibial [yes/no] and trans-femoral [yes/no]) as independent variables ($n = 57$). To gain

<table>
<thead>
<tr>
<th>Table 1. Subjects’ characteristics.</th>
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<tr>
<td></td>
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<tr>
<td>overall</td>
</tr>
<tr>
<td>Age (years)</td>
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<tr>
<td>Sex (men/women)</td>
</tr>
<tr>
<td>Bodyweight (kg)</td>
</tr>
<tr>
<td>BMI (kg·m$^{-2}$)</td>
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<tr>
<td>Years since amputation</td>
</tr>
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</table>

NOTE. Values are mean ± SD or n. Bodyweight is the weight of the person minus the weight of the prosthesis.
Abbreviations: n/a, not applicable; TF, trans-femoral; TT, trans-tibial.
* Significantly different from controls ($p < .05$).
† Significantly different from people with a traumatic amputation ($p < .05$).
insight into the possible effects of an uneven distribution of both cause and level of amputation among the different amputation groups, a final regression analysis (model 4, n = 36) was performed in which both the cause (traumatic or vascular) and level (trans-tibial or trans-femoral) were entered as independent dummy variables. For each associated independent variable, the percentage difference in VO\textsubscript{2peak} was calculated when the independent variable is changed by one unit.

**Results**

The people with a vascular amputation were older (p = .027) and had a higher body weight (p = .021) and BMI (p = .013) compared with the people with a traumatic amputation and able-bodied controls. No differences between the latter two groups were found. Time after amputation differed significantly between amputee groups (p < .001). The allocation of subjects to the different groups was independent of sex (χ\textsuperscript{2} = 0.596, p = .742; Table 1). All subjects successfully completed the exercise test to exhaustion and all but four subjects reached RER\textsubscript{peak} values of > 1.1, with a mean ± SD value over all subjects of 1.3 ±0.1 (Table 2). Peak power output was significantly lower in the group with a vascular amputation (p < .001). HR\textsubscript{peak} was also lower in the vascular group; however, when expressed as a HR\%predicted [201] no differences were found (p = .085) among groups.

When correcting for age, BMI, and sex, the multiple linear regression model revealed that having a lower limb amputation was inversely associated with VO\textsubscript{2peak} (p = .021, model 1). The presence of an amputation because of vascular deficiency was associated with a lower VO\textsubscript{2peak} of 29.1% (p < .001), whereas a traumatic amputation was not significantly associated with a difference in

<table>
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<tr>
<th>Table 2. Results for graded peak exercise test.</th>
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<tbody>
<tr>
<td>Peak values (mean ± SD)</td>
</tr>
<tr>
<td>VO\textsubscript{2peak} (ml·kg\textsuperscript{-1}·min\textsuperscript{-1})</td>
</tr>
<tr>
<td>RER\textsubscript{peak}</td>
</tr>
<tr>
<td>HR\textsubscript{peak} (beats·min\textsuperscript{-1})</td>
</tr>
<tr>
<td>HR%predicted (%)</td>
</tr>
<tr>
<td>P\textsubscript{peak} (watt)</td>
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</table>

**NOTE.** Values are mean ± SD or n. Two traumatic and one vascular subject were using β-blockers and were therefore excluded in the calculation of HR\textsubscript{peak} and HR\%predicted.

Abbreviation: P\textsubscript{peak}, peak power output; TF, trans-femoral; TT, trans-tibial.

* Significantly different from controls (p < .05).
† Significantly different from people with a traumatic amputation (p < .05).
Chapter 3

The third model with level of amputation as independent dummy variables revealed that an amputation at the trans-tibial level was associated with a significantly lower $\dot{V}O_{2peak}$ ($p = .021$, model 3; Table 3). However, when entering cause and level of amputation (model 4), level of amputation failed to explain an additional significant proportion of the variance in $\dot{V}O_{2peak}$ in the amputee subjects ($n = 36$). In this analysis, having a vascular amputation was associated with a 26.4% decrease in $\dot{V}O_{2peak}$ ($p = .004$) compared with having an amputation because of trauma (model 4; see Table 3). To determine to what extent the association of cause of amputation with $\dot{V}O_{2peak}$ was related to the level of amputation, and vice versa, we added an interaction term (level*cause) to the fourth model. This interaction term was not significantly associated with the dependent variable ($p = .488$). Therefore, the association found between cause of amputation and $\dot{V}O_{2peak}$ seems to be independent of the level of amputation.

Table 3. Multiple linear regression model.

<table>
<thead>
<tr>
<th>Entered variables (X_i)</th>
<th>$\beta$-coefficients</th>
<th>$F_{\text{change}}$(p)</th>
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<tbody>
<tr>
<td></td>
<td>$\beta_i$(SE)</td>
<td>p</td>
</tr>
<tr>
<td><strong>Model 1 (n = 57)</strong></td>
<td></td>
<td></td>
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<tr>
<td>-amputation</td>
<td>yes(1)/no(0)</td>
<td>-0.140 (0.059)</td>
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<tr>
<td><strong>Model 2 (n = 57)</strong></td>
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<tr>
<td>-traumatic</td>
<td>yes(1)/no(0)</td>
<td>-0.086 (0.058)</td>
</tr>
<tr>
<td>-vascular</td>
<td>yes(1)/no(0)</td>
<td>-0.344 (0.087)</td>
</tr>
<tr>
<td><strong>Model 3 (n = 57)</strong></td>
<td></td>
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<tr>
<td>-trans-tibial</td>
<td>yes(1)/no(0)</td>
<td>-0.156 (0.065)</td>
</tr>
<tr>
<td>-trans-femoral</td>
<td>yes(1)/no(0)</td>
<td>-0.113 (0.076)</td>
</tr>
<tr>
<td><strong>Model 4 (n = 36)</strong></td>
<td></td>
<td></td>
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<tr>
<td>-cause</td>
<td>traumatic(1)/vascular(2)</td>
<td>-0.306 (0.098)</td>
</tr>
<tr>
<td>-level</td>
<td>TT(1)/TF(2)</td>
<td>0.046 (0.074)</td>
</tr>
</tbody>
</table>

NOTE. The parameters of interest entered in the multiple linear regression models are presented with corresponding values. The total model had the following general form: $\ln(\dot{V}O_{2peak}) = \beta_0 + \beta_1\cdot\text{age} + \beta_2\cdot\text{BMI} + \beta_3\cdot\text{sex} + \sum\beta_i\cdot X_i + \epsilon$, where $X_i$ represent the $j$-th entered variable of interest and age, BMI and sex were entered as confounders. Model 4 only included the two amputation groups. Abbreviations: TT/TF, trans-tibial or trans-femoral; $\beta_i$ unstandardized beta coefficient; SE, standard error.

* The percentage change in $\dot{V}O_{2peak}$ when the independent variable is increased with one unit (e.g. those who underwent a lower limb amputation had a lower $\dot{V}O_{2peak}$ of 13.1%)
Discussion

We used a graded one-leg peak exercise test to determine the \( \dot{V}O_{2\text{peak}} \) of people with a lower limb amputation. While controlling for the possible confounding effects of age, BMI, and sex, the multiple linear regression analysis revealed that people who underwent a lower limb amputation had, in general, a 13.1% lower aerobic capacity compared to controls. An additional analysis, differentiating between etiologies, revealed that people with a traumatic amputation did not differ from controls, whereas people with a vascular amputation had a significantly lower \( \dot{V}O_{2\text{peak}} \) of 29.1%.

Chin and colleagues (1997, 2001 and 2002)\(^{[21-23]}\) report values of \( \dot{V}O_{2\text{peak}} \) between 18.0 and 20.1 \( \text{ml} \cdot \text{kg}^{-1} \cdot \text{min}^{-1} \) for people with a traumatic amputation, which are lower than the values found in this study (28.1 \( \text{ml} \cdot \text{kg}^{-1} \cdot \text{min}^{-1} \)). A possible explanation might be a difference in time since amputation. In the studies by Chin and colleagues (2001 and 2002)\(^{[22-23]}\) subjects underwent amputation more recently, which might explain the lower peak aerobic capacity found compared with our subjects. Pitetti and colleagues (1978)\(^{[171]}\) studied seven young subjects with a traumatic amputation with an average time since amputation of 19 years and found that the average aerobic capacity increased from 23.1 to 29.4 \( \text{ml} \cdot \text{kg}^{-1} \cdot \text{min}^{-1} \) after a 15-week training period. Remarkably, the values after training are comparable to the values found in the current study, even though subjects in the study by Pitetti and colleagues (1987)\(^{[171]}\) were substantially younger.

The higher values found in the current study compared with those in the literature might imply that our traumatic amputees were extremely fit compared with previously studied subjects, but because none of the amputees were abnormally active, we deem this unlikely. A more plausible explanation might be that with the current set-up, we were able to stress the cardiopulmonary system to a larger extent and more closely approximate subjects’ true maximal aerobic capacity. Compared to an upright bicycle, the recumbent bike used by Chin and colleagues (1997,2001 and 2002)\(^{[21-23]}\) might have led to a reduced gravity-induced hydrostatic pressure, which could result in inadequate perfusion and oxygen delivery to the active muscles\(^{[62, 227]}\). Furthermore, the continuous protocol used by Chin and colleagues (1997,2001 and 2002)\(^{[21-23]}\) might have caused premature feelings of local fatigue and consequently premature ending of the exercise test\(^{[21-23]}\). The notion that the cardiovascular system was stressed to its full potential in the current study is supported by the fact that the overall averaged RER\(_{\text{peak}}\) was 1.3 ± 0.1 indicating maximal aerobic effort.

Current findings corroborate the limited existing evidence and the intuitive notion that the aerobic capacity of people who have had lower limb amputation
because of vascular deficiency is lower. The 29.1\% lower \( \dot{V}O_{2\text{peak}} \) might be explained by the fact that comorbidities, stump problems resulting from delayed wound healing, or persisting pain caused by claudication limit an active lifestyle and, consequently, lead to a lower \( \dot{V}O_{2\text{peak}} \). In addition, people who undergo an amputation because of vascular deficiency might already have had a lower \( \dot{V}O_{2\text{peak}} \) as a result of a sedentary life, smoking, or preexisting medical conditions. To control for these influencing factors and to single out the effect of an amputation in this group of subjects, it would be of great value to in a future study to compare two group of subjects, both diagnosed with vascular deficiency, that differ only with respect to the existence of a lower limb amputation. With respect to the traumatic amputees, Chin and colleagues (2002) found initial differences at the start of rehabilitation between able-bodied controls and traumatic amputees, but these differences disappeared after a 6 week aerobic training regimen. The traumatic amputees in the current study already walked with their prosthesis for a number of years, and were able to walk comfortably for at least 4 minutes. The familiar and correctly fitted prosthesis apparently enabled the traumatic subjects to adopt a relatively active lifestyle, resulting in an aerobic capacity similar to that of able-bodied controls.

Although a regression model with level of amputation entered as a dummy variable (model 3), revealed that subjects who underwent a trans-tibial amputation had a lower aerobic capacity compared to able-bodied controls and people with a trans-femoral amputation, this difference disappeared when a regression analysis was performed in which both cause and level were entered as independent variables (model 4). This might be explained by the fact that in the trans-femoral group, 23\% of the subjects had amputations performed because of vascular deficiency, whereas in the trans-tibial group this was the case for 30\% of the subjects. Therefore, the higher number of subjects with a vascular amputation in the trans-tibial group could explain the association between level and \( \dot{V}O_{2\text{peak}} \) found in the third model.

The finding of no association between \( \dot{V}O_{2\text{peak}} \) and the level of amputation was found was surprising and contrary to what has been reported by Waters and colleagues (1976). They state that a more proximal amputation is related to a lower predicted peak aerobic capacity. However, their findings were not substantiated by a statistical test. Moreover, Waters and colleagues (1976) predicted the peak capacity based on data obtained during a fast-walking trial. Because walking efficiency differs between walking with a trans-tibial and a trans-femoral amputation, predicted oxygen consumption might not be reliably compared between both groups. The lack of an association between level of amputation and aerobic capacity suggests that in our population, the proposed limited activity level in people with a more proximal amputation was not related to a lower aerobic capacity. This could be because we included only
proficient ambulators. Another explanation might be a lack of statistical power in the current study. However, interestingly, the percentage difference associated with the level of amputation (4.7%) is substantially smaller than that associated with the cause of amputation (26.4%). This may imply that in clinical practice, the level of amputation only marginally influences the $\text{VO}_2\text{peak}$. The observed influences of presence, cause, and level of amputation on $\text{VO}_2\text{peak}$ were found after adjusting for the potential confounding effects of age, BMI and sex. Age, BMI and sex indeed explained a significant additional proportion of the total variance. When, for example, the confounders were excluded from the analysis in model 2, the adjusted explained variance decreased from 66.0% to 34.4% (data not shown). More importantly, part of the differences in average $\text{VO}_2\text{peak}$ between controls and people with a vascular amputation, presented in Table 2, can be explained by the higher age and BMI in the latter group.

Study limitations

All subjects were relatively active, healthy older adults with an amputation who were able to walk for at least 4 minutes, thereby limiting generalization. However, we believe that the lower $\text{VO}_2\text{peak}$ observed will only be aggravated in subjects of similar age who are currently inactive. Another limitation in the current study is the limited number of subjects with a vascular amputation and, specifically, the limited number of people who had a trans-femoral amputation in this group. This reduces the statistical power of this study and might conceal a potential association between $\text{VO}_2\text{peak}$ and the level of amputation. However, differences that were observed were consistent and large enough to provide statistically significant results. Furthermore, matching the participants on age inevitably resulted in the group with a traumatic amputation having a longer time since amputation. This could be a confounding factor influencing the differences found in peak aerobic capacity between both amputation groups.

Conclusion

The present study investigated the aerobic capacity of older adults who were ambulatory prosthetic users by means of a discontinuous, graded, one-legged, exercise test in which a regular cycle ergometer was used. $\text{VO}_2\text{peak}$ of vascular amputees was lower than that of able-bodied controls. After correcting for the confounding effects of age, BMI, and sex, $\text{VO}_2\text{peak}$ of the vascular group was 29.1% lower than that of able-bodied controls. The traumatic amputees had a $\text{VO}_2\text{peak}$ similar to that of able-bodied controls, and no effect of level of amputation was found. The lower aerobic capacity found in the vascular group, together with
the higher aerobic demand when walking with a prosthesis \cite{226}, might influence the walking ability in those walking with a lower limb amputation \cite{171} because of vascular deficiency. Although limited, some evidence indeed suggests that a relationship between walking ability and aerobic capacity exists \cite{29, 184, 224}; a lower aerobic capacity has been claimed to result in lower walking ability \cite{24-25}. Future research should focus on the question to what extent the lower aerobic capacity influences walking ability, and whether aerobic training can be beneficial in this respect.
Peak oxygen consumption
On the relationship between aerobic capacity and walking ability in older adults with a lower limb amputation

Wezenberg D, van der Woude LH, Faber WX, de Haan A, and Houdijk H,

"On the relationship between aerobic capacity and walking ability in older adults with a lower limb amputation",

Archives of physical medicine and rehabilitation, accepted February 2013.
On the relationship between aerobic capacity and walking ability in older adults with a lower limb amputation

Wezenberg D, van der Woude LH, Faber WX, de Haan A, and Houdijk H, "On the relationship between aerobic capacity and walking ability in older adults with a lower limb amputation”, Archives of physical medicine and rehabilitation, accepted February 2013.
Abstract

The objective of this study was to determine the relative aerobic load, walking speed and economy of older adults walking with a lower limb prosthesis, and to predict the effect of an increased aerobic capacity on their walking ability. A convenience sample of 36 older adults who underwent lower limb amputation due to vascular deficiency or trauma and 21 age-matched able-bodied controls participated. Peak aerobic capacity and the oxygen consumption while walking were determined. The relative aerobic load and walking economy were assessed as a function of walking speed, and a data-based model was constructed to predict the effect of an increased aerobic capacity on walking ability. Results showed that people with a vascular amputation walk at a substantial higher (45.2%) relative aerobic load than people with an amputation due to trauma. The preferred walking speed in both groups of amputees was slower than that of able-bodied controls, and it was below their most economical walking speed. We predicted that a 10% increase in peak aerobic capacity could potentially result in a reduction in the relative aerobic load of 9.1%, an increase in walking speed and it could lead to a 17.3% or 13.9%, and an improvement in the walking economy of 6.8% or 2.9%, for people after a vascular or traumatic amputation, respectively. Current findings corroborate the notion that, especially in people with a vascular amputation, the peak aerobic capacity is an important determinant for walking ability. The data provides quantitative predictions on the effect of aerobic training; however, future research is needed to experimentally confirm these predictions.
**Introduction**

Regaining or maintaining walking ability can be a challenge for people after undergoing lower limb amputation. Aside from problems with regards to pain, wounds or prosthetic fitting, walking with a prosthesis requires more metabolic energy than walking with two intact limbs \[110, 116, 225\]. The perceived exertion when walking can be aggravated when it coincides with a reduced aerobic capacity \[71, 216, 226\]. Unfortunately, the available aerobic capacity is known to reduce with age \[73\], and to be lower in older adults with a vascular amputation compared to age-matched controls \[230\]. Therefore, those most likely at risk of walking at an unfavorably high relative aerobic load (i.e. the aerobic energy requirement for walking relative to the available aerobic capacity) are the older adults with a vascular amputation.

When the relative aerobic load at a given walking speed exceeds a desirable limit, people can opt to reduce their walking speed, thereby reducing the metabolic energy expenditure \[71\]. Previous research has demonstrated that people with an amputation indeed lower their walking speed to such a degree that the metabolic energy expenditure per minute is equal to that of able-bodied controls \[85, 114, 116-117, 169, 225-226\]. There is, however, a tradeoff as reducing the walking speed will affect the ability to keep up with able-bodied peers and could result in a higher energy cost per distance traveled \[44, 81, 116, 203\]. The cost of walking is expressed as the energy expenditure per distance traveled and has a parabolic relationship with walking speed, with a minimum at a speed which is defined as the most economical walking speed \[9, 28, 81, 240\]. Able-bodied persons \[28, 141\] prefer to walk at, or close to, their most economical walking speed. By contrast, older adults with an amputation might not be able to walk at their most economical walking speed as that walking speed might coincide with an undesirably high relative aerobic load. Hence, a reduced peak aerobic capacity can influence walking ability as it can result in an increased relative aerobic load, reduced preferred walking speed and a reduced walking economy.

Whereas the notion that relative aerobic load is an important determinant of walking ability is recognized \[107\], it has been scarcely investigated in older adults walking with a lower limb prosthesis. Moreover, it is not known to what magnitude a reduced exercise capacity might affect walking effort, speed and walking economy, consequently possible effects of aerobic training on walking ability are unclear. The current study was designed to determine the relative aerobic load in a group of older adults walking with a prosthesis and a group of able-bodied controls, and establish the effect of etiology and level of amputation on relative aerobic load. In addition, we determined the relationship between relative aerobic load, walking economy and walking speed. Using the obtained results, we constructed a data-based model, to provide quantitative predictions.
of the potential effects of an increased peak aerobic capacity on the relative aerobic load, walking speed and walking economy.

Methods

Subjects

In total a convenience sample of 36 people, aged between 50 and 75 years with a unilateral lower limb amputation, participated in this study. Subjects had undergone amputation over one year prior to participation, and all were able to walk comfortably with their prosthesis for a minimum of four minutes. Of the 36 subjects, 26 had undergone amputation because of trauma, of which 16 at the trans-tibial and 10 at the trans-femoral level. The remaining 10 subjects had undergone amputation because of vascular deficiencies (7 trans-tibial and 3 trans-femoral). Subjects were ineligible for participation when they had any absolute contraindication for exercise \[^{[143]}\], impairments of cognitive function, wounds on the stump or a history of musculoskeletal or neurological diseases other than the amputation that might affect test performance. Prior to participating all subjects with an amputation were examined by a physician. An additional group of 21 able-bodied controls participated. Informed consent was obtained and the study carried the approval of the Medical Ethical committee of the VU University Medical Center in Amsterdam.

Experimental procedure

All subjects visited the rehabilitation centre twice; visits were separated by a minimum of 1 day (mean = 5.3, sd = 4.4). Subjects were asked to refrain from drinking coffee and eating large meals on the test days, and were asked not to be involved in excessive exercise in the 24 hours preceding test days.

The oxygen consumption during walking was determined breath-by-breath using open-circuit respirometry (Oxycon Delta; CareFusion, Houten, The Netherlands) while walking on a treadmill (3m by 1m, ForceLink©, Culemborg, The Netherlands). After a familiarization period, preferred walking speed (PWS) was determined according to the method previously reported by others \[^{[139-141]}\]. Subjects performed five walking trials at different walking speeds. Each of these trials lasted four minutes in order to allow steady state oxygen consumption to be attained. Subjects walked at PWS and at speeds 15 and 30 percent higher and lower than PWS. Imposing walking speeds relative to PWS was deemed essential due to large inter-individual differences in PWS. All trials were performed in random order, and were separated by 10 minutes of rest.
During the second visit peak aerobic capacity was measured using a graded one-legged cycle exercise test. The exercise test was performed using an electronically braked cycle ergometer (Lode Corival, Lode B.V., Groningen, The Netherlands). For a more detailed description of the exercise test and a validation of the protocol we refer to Wezenberg and colleagues (2012).

Data analysis

During the walking trials, steady state oxygen consumption was determined as the mean value of oxygen consumed during the last minute of data collection. This value was then normalized for body weight in order to attain the energy expenditure during walking ($\dot{V}O_{\text{walking}}$ [ml·kg$^{-1}$·min$^{-1}$]). Verification of steady state was done online by visual inspection of the oxygen consumption, and offline using the method previously described by Schwartz and colleagues (2007). Peak aerobic capacity ($\dot{V}O_{\text{peak}}$ [ml·kg$^{-1}$·min$^{-1}$]) was determined as the highest value attained during the last or penultimate exercise phase during the graded exercise test. Relative aerobic load ($\dot{V}O_{\text{rel}}$ [%]) was determined as the percentage of available $\dot{V}O_{\text{peak}}$ that was used when walking. The oxygen cost of walking at preferred walking speed ($C_{\text{pws}}$ [ml·kg$^{-1}$·m$^{-1}$]) was determined as the measured oxygen consumption while walking at PWS divided by the preferred walking speed.

Relative aerobic load and oxygen cost as a function of walking speed

For all groups, stratified according to etiology, the relative aerobic load as a function of walking speed ($v$) was described using a second order polynomial function $[9, 28]$

$$\dot{V}O_{\text{rel}} = \frac{\dot{V}O_{\text{walking}}}{\dot{V}O_{\text{peak}}} = a \cdot v^2 + b \cdot v + c$$  

Only subjects who were able to walk a minimum of three trials were included in this analysis. Similarly, a second polynomial function was fitted to describe the relationship between oxygen consumption (ml·kg$^{-1}$·min$^{-1}$) and walking speed.

$$\dot{V}O_{\text{walking}} = m \cdot v^2 + n \cdot v + p$$

This function was then divided by walking speed to obtain the oxygen cost (J·kg$^{-1}$·m$^{-1}$) as a function of walking speed (m·s$^{-1}$).

$$\text{oxygen cost} = \frac{\dot{V}O_{\text{walking}}}{v} = m \cdot v + n + \frac{p}{v}$$
Model predictions

The potential effect of a given increase in \( \dot{V}O_{\text{peak}} \) on relative aerobic load, while maintaining the same walking speed, is found by multiplying the relative aerobic load with a factor \( x \) equal to the inverse of the fractional change in \( \dot{V}O_{\text{peak}} \). The potential effect of a given increase in \( \dot{V}O_{\text{peak}} \) on walking speed can be determined by multiplying Equation 1 with the factor \( x \), and solving the equation for walking speed, while restricting the left side of the equation to the relative aerobic load measured at PWS. In other words, in this way we determined the change in walking speed when subjects would choose to walk at the same relative aerobic load as before but with a higher \( \dot{V}O_{\text{peak}} \). Finally, the corresponding effect on walking economy could subsequently be derived from the known relationship between oxygen cost and walking speed.

Statistics

All data were analyzed using a computerized statistical package (SPSS inc., Version 18.0; Chicago, Illinos). To evaluate whether the choice of PWS and \( \dot{V}O_{\text{walking}}, \dot{V}O_{\text{rel}} \) and \( C_{\text{pws}} \) at PWS (dependent variables) were influenced by the presence of an amputation, cause of amputation and level of amputation (independent variables) multiple linear regression analyses were performed. For each dependent variable a model was constructed to identify whether an association between the dependent variable and the presence of an amputation exists. An additional model, which included only the subjects with an amputation,
was constructed to determine the association between the dependent variables and both etiology and level of amputation. Age, BMI and sex were added as confounders in all regression models. After analysis of the initial constructed model we decided to natural log transform the dependent variable to improve the model fit. Prior to interpretation of the data, case-wise diagnostics and normality analysis were performed on all constructed models.

**Results**

The subject characteristics, stratified according to etiology, are shown in Table 1. The multiple linear regression analyses revealed that both etiology \((p = .033)\) and level of amputation \((p = .006)\) were negatively associated with PWS; with a more proximal and vascular amputation resulting in the lowest walking speed (Table 2).

A trend can be noted indicating that the presence of an amputation might be associated with a higher relative aerobic load at PWS compared to able-bodied controls \((p = .071)\). When entering only those subjects with an amputation and controlling for the difference in level of amputation, the regression analysis revealed that the presence of a vascular amputation resulted in a relative aerobic load at PWS that was 45.2 % higher than that of people with a traumatic amputation \((p = .001)\). No association between relative aerobic load and the level of amputation was found \((p = .803; \text{Table 2})\).

The oxygen cost of walking \((C_{pws})\) was related to both etiology \((p = .008)\) and level of amputation \((p = .002)\), with a more proximal and vascular amputation resulting in the highest energy cost of walking at PWS (Table 2).

The Figure shows the mean values found for the able-bodied controls and the subjects walking with a lower limb prosthesis grouped according to etiology. The vertical bars represent the standard deviation. Squares represent the averaged preferred walking speed.
Note: For all four dependent variables a regression analysis was performed. In addition to the entered variables presented here, age, BMI and sex were added as confounders. Whereas Model 1 contains also the able bodied control group, Model 2 only included subjects with a lower limb prosthesis.

Abbreviations: trau/vasc, traumatic or vascular; TT/TF, trans-tibial or trans-femoral; n/a; not applicable; PWS, preferred walking speed; \( \dot{V}O_{2\text{walking}} \), energy expenditure while walking at PWS; \( \dot{V}O_{2\text{rel}} \), relative aerobic load at PWS; \( C_{\text{PWS}} \), the oxygen cost of walking at PWS; \( F_{\text{change}} \), the \( F \) test statistics for the model with respect to the basic model with only age, BMI and sex included; \( R^2 \), the explained variance; \( \beta \), unstandardized beta coefficient; \%change, the percentage change in the dependent variable when the independent variable is increased with one unit, 95% confidence intervals are given.

### Table 2. Multiple linear regression models

<table>
<thead>
<tr>
<th></th>
<th>PWS  (m·sec⁻¹)</th>
<th>( \dot{V}O_{2\text{walking}} ) (ml·kg⁻¹·min⁻¹)</th>
<th>( \dot{V}O_{2\text{rel}} ) (%)</th>
<th>( C_{\text{PWS}} ) (ml·kg⁻¹·m⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>( F_{\text{change}} )</td>
<td>( R^2 )</td>
<td>( \beta ) (p)</td>
<td>%change (95%CI)</td>
</tr>
<tr>
<td><strong>Model 1 (n=57)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Amputation (yes/no)</td>
<td>39.5</td>
<td>.802</td>
<td>4</td>
<td>.52</td>
</tr>
<tr>
<td></td>
<td>(p&lt;.001)</td>
<td></td>
<td>(p=.517)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>-0.310</td>
<td>-26.7%</td>
<td>-0.029</td>
<td>n/a.</td>
</tr>
<tr>
<td></td>
<td>(p=.001)</td>
<td>(-33.6, -19.0)</td>
<td>(p=.517)</td>
<td>n/a.</td>
</tr>
<tr>
<td><strong>Model 2 (n=36)</strong></td>
<td>6.6</td>
<td>.797</td>
<td>5</td>
<td>.619</td>
</tr>
<tr>
<td></td>
<td>(p&lt;.001)</td>
<td></td>
<td>(p=.039)</td>
<td></td>
</tr>
<tr>
<td>Amputation (trau/vasc)</td>
<td>-0.191</td>
<td>-17.4%</td>
<td>0.059</td>
<td>n/a.</td>
</tr>
<tr>
<td></td>
<td>(p=.032)</td>
<td>(-30.5, -1.7)</td>
<td>(p=.435)</td>
<td>n/a.</td>
</tr>
<tr>
<td>level of amputation</td>
<td>-0.187</td>
<td>-17.1%</td>
<td>0.032</td>
<td>n/a.</td>
</tr>
<tr>
<td>TT(1)/TF(2)</td>
<td>(p=.006)</td>
<td>(-27.1, -5.6)</td>
<td>(p=.576)</td>
<td>n/a.</td>
</tr>
</tbody>
</table>
Relative aerobic load and oxygen cost as a function of walking speed

One participant with a traumatic and two with a vascular amputation were unable to complete a minimum of three speeds and were omitted in the analysis. For the remaining subjects, averaged relative aerobic load increased curvilinear with walking speed (Figure 1). The curvilinear relationship was shifted to the left for people with an amputation. An additional shift upwards was found in people with a vascular amputation. Figure 1 also shows that people with a traumatic amputation were able to walk at the same relative aerobic load as able-bodied controls but at a lower walking speed, this in contrast to people with a vascular amputation. The relative aerobic load while walking at PWS was only associated with etiology and not with the level of amputation (Table 2). However, it is important to acknowledge that the choice of PWS was related to the level of amputation; subjects with a trans-femoral amputation chose to walk at a lower PWS than subjects with a trans-tibial amputation. Consequently, people with a trans-femoral amputation walk at the same relative aerobic load as people with a trans-tibial amputation but attain this relative aerobic load while walking slower.

For able-bodied controls the averaged most economical walking speed is close to their PWS, stated differently; the speed at which the cost of walking is the lowest is close to the PWS. Conversely, subjects with an amputation chose a PWS which was substantially lower than their theoretically most economical walking speed (Figure 2). Increasing the walking speed to match the most economical walking speed coincides with a higher effort for walking. This entails that participants with a traumatic amputation would need to walk at 60.5% and those with a vascular amputation at 109.3% of their peak aerobic capacity in order for them to walk at their theoretically most economical walking speed (Table 3).

Figure 2. The relationship between the cost of walking and walking speed.

The values found for the able-bodied controls and the subjects walking with a lower limb prosthesis grouped according to etiology are shown. Squares represent the averaged preferred walking speed while the triangles represent the cost of walking while walking at the most theoretical economic walking speed. Only those subjects who were able to walk a minimum of three trials are included in the Figure.
Using the data-based model (Equation 1), we predict that for people with a vascular amputation to walk at a relative aerobic load similar to that of able-bodied controls (i.e. 48.7%) would require an increase in $\dot{V}O_{2\text{peak}}$ of 41.4%. When the aim is to walk at the theoretically most economical walking speed (while maintaining the same $\dot{V}O_{2\text{rel}}$), people with an amputation due to trauma would need to increase their $\dot{V}O_{2\text{peak}}$ by 19.2%, while those with an amputation due to vascular deficiency need to increase the $\dot{V}O_{2\text{peak}}$ by 60.1% (Table 3). In Figure 3 the relative changes in relative aerobic load, walking speed and cost of walking that can potentially be attained are depicted as a function of the percentage increase in $\dot{V}O_{2\text{peak}}$. A certain percentage increase in peak aerobic capacity will result in the same percentage difference in relative aerobic load for both etiologies. While the effect on walking speed and the associated cost of walking differs among etiologies.

**Discussion**

The principal finding of this study is that people with a lower limb amputation due to vascular deficiencies walked at a considerably higher $\dot{V}O_{2\text{rel}}$, despite walking at a much slower speed compared to able-bodied controls. People after traumatic amputation also walk slower; however, they are able to walk at a similar $\dot{V}O_{2\text{rel}}$ as the able-bodied controls. Both groups of amputees walked slower than their theoretically most economical speed.

In the current study the older able-bodied controls walked at a $\dot{V}O_{2\text{rel}}$ of 48.7% (Table 1); this value is in close agreement with previous published results for the same age group \cite{71, 107, 121, 139}. The few studies that report values for
Relative aerobic load in people with an amputation vary markedly (between 35 and 56%)\(^{[162, 226]}\). In contrast to the previous published studies\(^{[162, 226]}\), which used prediction equations, we have used a graded peak exercise test\(^{[229]}\). This provided us with a more reliable and comparable estimation of the individual relative aerobic load while walking.

**Relative aerobic load and oxygen cost as a function of walking speed**

Aerobic capacity declines with age\(^{[73]}\), consequently, healthy older subjects reduce their walking speed to ensure that \(\text{VO}_{2\text{rel}}\) remains within acceptable limits, thereby, reducing the feelings of fatigue during walking\(^{[31, 71, 139]}\). To walk at the same \(\text{VO}_{2\text{rel}}\), people with a traumatic amputation reduced their walking speed with 21.6% compared to that of able-bodied controls. Remarkably, for people with a vascular amputation to walk at a \(\text{VO}_{2\text{rel}}\) similar to controls seems not feasible as this would theoretically require speeds below zero (Figure 1).

For healthy able-bodied young\(^{[28]}\) and old adults\(^{[141]}\) the choice of walking speed is believed to be governed by the relationship between walking speed and walking economy. Our results support this as controls indeed walked at their most economic walking speed. However, both the subjects with a traumatic and vascular amputation generally did not (Figure 2). These findings are in agreement with some\(^{[44, 116]}\), but not all studies\(^{[79, 81, 85, 103]}\). It can be hypothesized that for people with an adequate aerobic capacity the most economic walking speed can be reached within an acceptable level of \(\text{VO}_{2\text{rel}}\). Presumably, people with an adequate aerobic capacity, like young people with a traumatic trans-tibial amputation\(^{[78, 208, 226]}\), the most economic walking speed can be reached at an acceptable level of relative aerobic load. While for other, and especially the older
people with a reduced capacity, this seems not within reach. For these people, the choice of walking speed seems to be governed by the relative aerobic load rather than walking economy.

Potential effects of an improvement in aerobic capacity

The current data set was used to quantitatively predict the potential effect of an improvement in peak aerobic capacity on the walking ability in terms of relative aerobic load, speed and economy. The reported increases in peak aerobic capacity of people with an amputation after aerobic training vary between 27.2% and 41.2% \[22-23, 170\]. Whereas these training studies were limited to relatively young people with a traumatic amputation, improvements in aerobic capacity have been reported for older subjects with a wide range of pathologies \[142\], suggesting that aerobic capacity can also be increased in older deconditioned people with a vascular amputation. From Figure 3 it can be seen that a relatively modest increase in peak aerobic capacity of 10% can, theoretically, result in clinical relevant changes with regards to relative aerobic load (9.1% lower), speed (17.3% higher) and walking economy (6.8% lower cost of walking) in people after a vascular amputation, and somewhat more modest changes in people after a traumatic amputation. Hence, the current model shows that even small increases in peak aerobic capacity can result in substantial improvements in walking ability in terms of relative aerobic load, speed and walking economy.

Alternatively to increasing the peak aerobic capacity, technological advancements have aimed at improving walking economy and, concomitantly, the relative aerobic load while walking. Although improvements in the field of prosthetics are promising, the current study underscores the importance that in order to allow a person to walk greater distances with less fatigue it is important to also look beyond the prosthetic innovations and optimize the physical capacity of the person walking with the prosthesis.

Study limitations

Information concerning the isolated effect of a potential increase in peak aerobic capacity can help to gain insight in the magnitude of change necessary when aiming to reduce the or increase the walking speed. However, there are some limitations with this approach. Firstly, it is important to realize that due to the large inter-individual variation, conclusions cannot be generalized to individual patients; it merely provides a quantitative description of an overall trend. Secondly, the novelty of this modeling approach is that we can alter one parameter (in this case $VO_{2\text{peak}}$). However, in reality, there is no isolated effect of aerobic training on $VO_{2\text{peak}}$ \[171\].
Previous research has shown that walking on a treadmill results in a lower PWS and is less economical in both healthy controls \[36\] and people walking with an prosthesis \[209\] compared to overground walking. Therefore subjects might have walked faster and more economical when tested while walking overground, however, conclusions drawn about the differences between groups will, most likely, remain the same.

Finally, a limitation in the current study is the small number and biased sample, especially with regards to the number of subjects with a vascular amputation. Furthermore, our set-up allowed us to only include subjects who could walk independently. Consequently, the participants selected in this study might have been biased to the physically fit which limits generalization of the results. Even though our results are evident, future research with larger groups will strengthen the conclusions.

**Conclusion**

To sum, people with a traumatic amputation walked at the same relative aerobic load as able-bodied controls but did so at a lower walking speed. People with a vascular amputation walked at an even slower speed, but nevertheless walked at a substantial higher relative aerobic load. Interestingly, both amputee groups chose a walking speed that was lower than the most economic walking speed.

The model predictions in this study indicate that improving $\dot{V}O_{2\text{peak}}$ through aerobic training can result in a clinically relevant reduction in relative aerobic load in people with a vascular amputation. Additionally, it can allow people to adopt a faster and more economical walking speed. Therefore, we propose that peak aerobic capacity ought to be considered as an important factor when aiming to improve the walking ability in people with an amputation. Future longitudinal studies are needed to empirically test the predicted effect of aerobic training on walking ability in older adults with a lower limb amputation.
Differentiation between prosthetic feet
based on in vivo center of mass mechanics

Wezenberg D, Cutti AG, Bruno A, Veronesi D, and Houdijk H,
"Differentiation between prosthetic feet based on in vivo center of mass mechanics",
Differentiation between prosthetic feet based on in vivo center of mass mechanics

Abstract

A decreased push-off power at the ankle, as seen in people walking with a prosthesis, will result in an increase in the energy dissipated during the step-to-step transition. Energy storage and return prosthetic feet have been developed to improve this push-off power, and therefore, are thought to reduce the step-to-step transition cost and concomitantly the metabolic energy required to walk. Interestingly however, no convincing evidence is found that support the notion that the metabolic energy cost is reduced compared to more conventional feet. This entails that either these feet are unable to reduce the step-to-step transition cost, or other disadvantageous factors attenuate the positive metabolic effect of the reduced step-to-step transition cost. The aim of this study was to determine whether walking with the energy storage and return foot reduce the step-to-step transition cost. Fifteen persons with a unilateral lower limb amputation walked with their prescribed energy storage and return foot and with a conventional solid ankle cushioned heel foot, while ground reaction forces and kinematics were recorded. The mechanical push-off power under the trailing limb was larger (33%) and the negative power under the leading limb was lower (13%) when push-off was performed by the energy storage foot compared to the conventional foot. The improved push-off power under the trailing prosthetic limb can be explained by the increased push-off power that was generated by the energy storage and return foot. Moreover, the longer forward progression of the center of pressure under the prosthetic foot augments the reduced step-to-step transition cost when walking with the ESAR prosthesis. Walking with the energy storage and return foot reduces the step-to-step transition cost. Hence, the limited effects on metabolic energy cost when walking with the ESAR feet should be attributed to metabolic costly side effects of these prosthetic feet.
Introduction

Walking with a lower limb prosthesis results in a higher metabolic energy cost than walking with two unimpaired legs \[228\]. With the introduction of the energy storage and return (ESAR) foot in the early 1980s, a passive-elastic prosthetic foot was marketed that was able to more closely mimic the human ankle by storing energy during stance and releasing this energy at push-off, which was assumed to reduce the metabolic energy cost while walking \[88-89, 244\]. Whereas several studies have shown that prosthetic users, subjectively choose the ESAR foot over the conventional soft ankle cushioned heel (SACH) foot \[88\], conflicting evidence is found with regard to its effect on metabolic energy cost \[104, 212, 219\]. Hence, it can be questioned whether the proposed mechanical effect of the ESAR foot is actually achieved.

In human walking the amount of metabolic energy needed to walk is to a large extent related to the mechanical work associated with the step-to-step transition \[56, 129, 131, 181\]. The double inverted pendulum model shows that negative mechanical work needs to be performed under the leading limb in order to redirect the body center of mass (COM) velocity from one circular arc to the next. In order to preserve walking speed, a similar amount of positive mechanical work needs to be performed. The most efficient way to produce this positive work is by generating push-off work at the ankle at, or prior to, heel contact of the contra-lateral limb. This will reduce the mechanical energy lost during collision, and therefore, the amount of mechanical work and metabolic energy required \[56, 129, 131, 181\]. Due to the absence of ankle musculature, people who walk with a lower limb prosthesis have a reduced push-off work at the prosthetic side and need to revert to other, less efficient, strategies to generate the required positive mechanical work during the step-to-step transition, resulting in an increase in metabolic energy cost \[6, 110, 152\].

The ESAR feet are specifically designed to improve push-off work, and thus are expected to reduce the mechanical energy lost during collision, and as such are expected to require less metabolic energy cost. Hence, the lack of evidence supporting a reduction in metabolic energy cost when walking with the ESAR foot is remarkable. Two possible causes for the limited effectiveness of the ESAR foot to reduce the metabolic energy cost can be postulated: 1) despite the fact that the ESAR foot is able to store and return energy it does not adequately reduce step-to-step transition cost, or 2) other disadvantageous factors (e.g. a compromised walking stability) are attenuating the positive effect of the reduced step-to-step transition cost.

In addition to push-off work, mechanical work at collision has been shown to depend on the roll-over shape of the feet \[1, 131, 181\]. When, instead of point feet,
the double inverted pendulum model is mounted with arc-shape feet, mimicking the human plantigrade foot, the center of pressure is able to move forward along the curved foot reducing the necessary directional change of the COM velocity, and concomitantly, the step-to-step transition cost [1]. Theoretically, step-to-step transition cost decreases quadratically with increasing roll-over radius, although, in terms of metabolic energy cost an optimal radius of 0.3 times the leg length is found [1, 95]. This corresponds to the values found in able-bodied subjects [94, 144] and people walking with a lower limb prosthesis [33]. Possible differences in shape can either diminish or augment the potential positive effect of an increased push-off work. Therefore, it is imperative to gain information about the roll-over shape characteristics when walking with the prosthesis in order to interpret the mechanical step-to-step transition cost. Whereas a number of studies have used quasi-static mechanical loading to characterize the roll-over shape of different prosthetic feet [32, 91, 93], information from in vivo studies in people walking with different prostheses is lacking.

To date, no studies have been published that combine in vivo measurements of center of mass mechanics with both prosthetic ankle power and roll-over characteristics of the prosthetic foot. To add to this limited body of knowledge, this study set out to firstly, determine whether walking with the widely prescribed ESAR foot reduces the step-to-step transition cost compared to the conventional SACH foot, and secondly, to relate the possible differences in transition cost to differences in the push-off work generated by the prosthetic foot and ankle and the roll-over characteristics of the two prosthetic ankles. We hypothesise that the ESAR foot can provide more push-off work, and together with a more favorable roll-over shape radius, can reduce the step-to-step transition cost during walking. If true, this would corroborate the notion that although the ESAR foot is able to reduce step-to-step transition cost, other disadvantageous factors are responsible for the limited effect on metabolic energy cost found in the literature.

**Methods**

**Subjects**

Patients were included who had walked with an ESAR prosthesis (Vari-Flex, Össur, Iceland) for at least the previous two years and were able to ambulate without walking aids. In total, 15 male subjects with a unilateral amputation at the level of the shank participated (age 55.8 years [SD = 11.1], weight 86.0 kg [SD = 12.6] and height 1.74 m [SD = 0.04]). All subjects had undergone amputation due to trauma and were free of any musculoskeletal disorder or neurological disease that could affect obtained results. After verbal and written clarification of the research procedure, informed consent was obtained.
Data acquisition

Subjects visited the prosthetic center twice; during the first visit data were collected while wearing their prescribed ESAR foot and walking at a fixed walking speed of 1.2 m·s⁻¹. At the end of the first visit subjects were mounted with the SACH foot (1D10, Ottobock, Germany) which was aligned by an experienced prosthetist. Subjects returned to the laboratory after on average 25.7 hours (range 21-28.4).

During both visits subjects walked on a 10 m walkway while segment trajectories were tracked using a 10-camera VICON System at 100Hz (VICON, Oxford, United Kingdom). Markers were placed on the anterior and posterior superior iliac spine, both the lateral and medial epicondyli and malleoli, and on the calcaneus. For the prosthetic side the malleoli markers were placed at a distal point at the rigid part of the prosthesis which approximated the sound limb malleoli position. Reapplication of the markers for the second measurement session was supported by images of the marker location during the first measurement session. Ground reaction forces were recorded at a 1000 Hz using two force plates (Kistler, Winterthur, Switzerland) embedded in the middle of the walkway. For each foot a minimum of five trials were collected in which walking velocity (measured using photocells [MICROGATE RaceTime 2, Bolzano, Italy]) was within 0.05 m·s⁻¹ of the target speed, i.e. 1.2 m·s⁻¹, and clean hits of the individual feet were recorded on the consecutive force plates.

Data analysis

Ground reaction force data were low-pass filtered at 100 Hz using a fourth-order zero lag Butterworth digital filter. If walking speed is kept equal over a step, total mechanical work ought to approximate zero. To ensure that walking speed remained constant over the step only the three trials with the lowest total net mechanical work were used. Basic spatio-temporal step parameters were calculated using a combination of the ground reaction forces and the location of calcaneus marker. The difference between the prosthetic minus the intact step length is used as the measure of step length asymmetry.

Mechanical work performed on the COM

The anterior and posterior superior iliac spine markers were used to estimate the position of the COM. The external mechanical power generated on the center of mass during the step-to-step transition was calculated as the dot product of the ground reaction force vector and the COM velocity vector for each limb independently [56]. The acceleration vector of the COM was calculated using the resultant of the ground reaction forces under both feet. COM velocity was then
calculated by integrating the acceleration vector of the COM over time while assuming periodic strides [56]. The mechanical positive work performed under the trailing limb during push-off ($W_{DS,\text{trail}}$ J·kg$^{-1}$) was calculated by integrating the trailing limb power during dual support. The negative mechanical work performed under the leading leg ($W_{DS,\text{lead}}$ J·kg$^{-1}$), was calculated as the integral of leading limb power between heel contact up until the instant the leading limb power became positive [131]. The leading limb power during the remaining time period was integrated to gain information about the net work performed during single stance ($W_{ss}$ J·kg$^{-1}$). Calculated power profiles and work per walked trial were subsequently averaged for each subject and foot type.

**Ankle work**

Because the prosthetic foot-ankle segment cannot be modeled as a rigid body with a hinge joint, using conventional inverse dynamics to calculate the mechanical power acting at the prosthetic ankle joint might introduce errors [186]. Therefore a different approach was adopted in which the power at the ankle was calculated by summing both the translational power and rotational power transferred from the foot to the shank [175]:

$$P_{\text{ankle}} = P_{\text{translation}} + P_{\text{rotational}} = F_{\text{ankle}} \times v_{\text{ankle}} + M_{\text{shank}} \times \omega_{\text{shank}}$$  

(1)

where $F_{\text{ankle}} \times v_{\text{ankle}}$ is the dot product of the ground reaction forces and linear velocities of the ankle. $M_{\text{shank}} \times \omega_{\text{shank}}$ and represent the moment at the ankle and the angular velocity of the shank at the distal point, respectively. Push-off work ($W_{\text{push}}$ J·kg$^{-1}$) was determined as the time integral of the positive power found prior to toe-off. By integrating the remaining power profile (e.g. excluding the positive work performed for push-off) the net amount of work that is either stored or dissipated prior to push-off was determined ($W_{\text{neg}}$ J·kg$^{-1}$). Again, outcome parameters were separately analyzed for each of the three trials, after which the trails were subsequently averaged to obtain a mean score for each subject and foot type.

**Roll-over characteristics**

For each subject and foot type, roll-over shapes were determined by transforming the center of pressure (COP) data from a laboratory-based reference frame to the three dimensional shank-based coordinate system [67, 223]. Because the COP data progresses forward from heel to toe during stance while the shank is rotated over the foot, the roll-over shapes can be modeled as a lower half of a circle in the shanks plane of progression. The characteristics of this arc were obtained for each subject by fitting the equation that represents the lower half of a circle on
the averaged data over the three trials performed for each foot. The data were fitted using a fitting algorithm with a non-holonomic constraint, ensuring that obtained radius was larger or equal to the maximal vertical displacements of the roll-over shape. As suggested by Hansen and colleagues (2004) \cite{94}, roll-over shape represents the COP during the time period from heel contact to contralateral heel contact. However, close examination revealed that during the first rocker movement roll-over shapes deviated strongly from circular. This problem has been noted previously by Hansen and colleagues (2004) \cite{91} and is also evident in the roll-over shapes presented by Miff and colleagues (2008) \cite{149}. This downward shift (as can be observed in Figure 3) is presumably caused by the compression of the heel at initial contact. Moreover forces are still low during this phase as the trailing limb is still in contact with the ground resulting in noisy forces and incorrect COP localization \cite{91}. Therefore, in this study the first rocker movement was excluded from the roll-over shape calculation, using the second derivative of the Savitzky-Golay filtered data. This resulted in disregarding on average 20 and 22 percent of the data points and improved the error of the circular fit (i.e. reduced it) with 49.1% and 50.1% for the ESAR and SACH foot, respectively. Logically, excluding part of the shank-based COP data results in a reduced total arc length of the roll-over shapes, but because in both the ESAR and SACH trials the same amount of data were excluded this did not alter conclusions. Reliable estimation of roll-over shape characteristics is based on the assumption

<table>
<thead>
<tr>
<th>Stride parameters</th>
<th>ESAR</th>
<th>SACH</th>
</tr>
</thead>
<tbody>
<tr>
<td>Speed</td>
<td>1.22(0.02)</td>
<td>1.22(0.02)</td>
</tr>
<tr>
<td>Stride length</td>
<td>1.38(0.06)</td>
<td>1.37(0.07)</td>
</tr>
<tr>
<td>Stride time</td>
<td>1.13(0.06)</td>
<td>1.12(0.05)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Step parameters</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Prosthetic side</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Step length</td>
<td>0.70(0.04)</td>
<td>0.72(0.04)*</td>
</tr>
<tr>
<td>Step time</td>
<td>0.57(0.03)</td>
<td>0.56(0.02)</td>
</tr>
<tr>
<td>Intact side</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Step length</td>
<td>0.68(0.03)</td>
<td>0.67(0.04)*</td>
</tr>
<tr>
<td>Step time</td>
<td>0.56(0.03)</td>
<td>0.56(0.03)</td>
</tr>
</tbody>
</table>

| Asymmetry         |             |          |
| Step length†      | 0.01(0.04)  | 0.05(0.04)* |

Table 1. Gait parameters (mean ± SD).

Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Össur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock, Germany).

* denotes statistical difference between prosthetic feet (p < .05).
† difference in step length calculated as the prosthetic minus the intact side step length.
that the shank-based COP can be represented by a lower half of an arc. However, the radius of both the biological, but also a prosthetic foot-ankle complex might not represent a perfect circular arc \cite{33, 92}. Consequently, important insight into the roll-over characteristic of the feet as the shank progresses forward might be lost. Therefore, we also calculated the forward travel of the COP on the ground (s) as a function of the angle between the shank and the vertical (\(\alpha\)) (Figure 3B) \cite{33}.

**Statistics**

Data were tested for normality using a Kolmogorov-Smirnov test and all parameters were normally distributed. Differences between prosthetic feet were tested using a paired sample \(t\) test. Significance was set a priori at the 5% level.

**Results**

All subjects were able to walk at the requested 1.2 m/s with both the ESAR and SACH foot. The averaged stride length and stride time did not differ between the two prosthetic feet (Table 1). Interestingly, the two prosthetic feet differed in step

![Figure 1. Center of mass mechanical power profiles.](image)

The center of mass mechanical power profiles of the step during which the prosthetic limb is the trailing limb (A) and the leading limb (B). Dashed lines represent center of mass mechanical power under the trailing limb, while solid lines represent the mechanical power under the leading limb. Hatched dashed and solid areas represent the part over which push-off work and collision loss was calculated, respectively.

Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Óssur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock, Germany).
length asymmetry \( (p < 0.001) \). Subjects took smaller steps with the intact limb compared to the prosthetic limb when walking with the SACH foot as compared to walking the ESAR foot (Table 1).

**Mechanical work performed on the COM**

Figure 1A shows that the external mechanical power performed by the trailing leg on the COM is larger, and the negative power under the leading limb during collision is lower when push-off is performed by the ESAR prosthesis compared to the SACH prosthesis. This is confirmed by a 33% larger \( W_{DS,\text{trail}} \) of the trailing prosthetic foot \( (p = 0.01) \) and a 13% lower \( W_{DS,\text{lead}} \) under the leading intact limb \( (p = 0.04) \) while walking with the ESAR compared to the SACH foot (Table 2). Additionally, net work performed on the COM over the subsequent single stance period \( (W_{ss}) \) was lower with the ESAR foot. Interestingly, while no difference in push-off work was found in the intact limb in either foot condition, more negative work under the leading prosthetic limb was found when walking with the SACH foot compared to the ESAR foot \( (16.7\%, p = 0.003; \text{Figure 1B}) \).

**Ankle work**

In Figure 2 the ankle power for the prosthetic and non-prosthetic limb is shown from heel contact until toe-off of that limb (i.e. stance period). It can be seen that when push-off was performed with the prosthetic limb (Figure 2A), substantial less positive work is generated at the ankle compared to the intact limb. The largest difference between intact and prosthetic push-off ankle power is seen when walking with the SACH foot. As expected, the \( W_{\text{neg}} \) was larger in the ESAR foot \( (-0.29 \pm 0.09 \text{ J} \cdot \text{kg}^{-1}) \) compared to the SACH foot \( (-0.19 \pm 0.06 \text{ J} \cdot \text{kg}^{-1}, p < 0.001; \text{Table 2}) \). No differences are found in push-off power of the intact limb (Figure 2B).

**Roll-over characteristics**

The COP data expressed in the shank-based coordinate system (thin line) and the fitted roll-over shape (thick line) are depicted in Figure 3A. The characteristics of this roll-over shape are summed in Table 2. No difference in roll-over shape radius was found between both prosthetic feet \( (p = 0.992) \), while the total arc length of the ESAR foot was larger compared to that of the SACH foot \( (p < 0.001) \). Figure 3B clearly shows that the forward travel of the COP as a function of shank angle is not circular. As opposed to a steady increasing line reflecting a constant curvature, the line shows the typical S-shape pattern previously reported by Curtze and colleagues (2009) \([32]\). The different phases can be understood using the three rocker phases as described by Perry and colleagues (1992) \([168]\). The
The initial relative flat period can be attributed to the heel rocker during which the tibia progresses forward while pivoting on the heel. The second period is steep and represents the ankle rocker during which the COP is moved along the foot while the tibia progresses further forward. The final flat period represents the forefoot rocker during which the heel is lifted from the ground while pivoting on the forefoot. Large similarities in shape are seen between both prosthetic feet during the heel and ankle rocker. However, when walking with the SACH foot the line flattened at an earlier shank angle, indicating an earlier onset of forefoot rocker when walking with the SACH foot. As a result, a lower total COP forward displacement in the SACH foot is seen compared to the ESAR foot (Figure 3B).

Table 2. Center of mass, ankle work, and roll-over shape characteristics (mean ± SD).

<table>
<thead>
<tr>
<th></th>
<th>ESAR</th>
<th>SACH</th>
</tr>
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<tbody>
<tr>
<td><strong>Center of mass mechanical work</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Prosthetic trailing</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$W_{DS,\text{lead}}$ (J·kg$^{-1}$)</td>
<td>-0.20 (0.10)</td>
<td>-0.23 (0.08)*</td>
</tr>
<tr>
<td>$W_{DS,\text{trail}}$ (J·kg$^{-1}$)</td>
<td>0.12 (0.06)</td>
<td>0.09 (0.04)*</td>
</tr>
<tr>
<td>$W_{SS}$ (net, J·kg$^{-1}$)</td>
<td>0.03 (0.15)</td>
<td>0.09 (0.08)*</td>
</tr>
<tr>
<td>Intact trailing</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$W_{DS,\text{lead}}$ (J·kg$^{-1}$)</td>
<td>-0.10 (0.07)</td>
<td>-0.12 (0.05)*</td>
</tr>
<tr>
<td>$W_{DS,\text{trail}}$ (J·kg$^{-1}$)</td>
<td>0.19 (0.06)</td>
<td>0.20 (0.05)</td>
</tr>
<tr>
<td>$W_{SS}$ (net, J·kg$^{-1}$)</td>
<td>-0.04 (0.11)</td>
<td>-0.03 (0.07)</td>
</tr>
<tr>
<td><strong>Ankle mechanical work</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Prosthetic limb</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$W_{\text{neg}}$ (J·kg$^{-1}$)</td>
<td>-0.29 (0.09)</td>
<td>-0.19 (0.06)*</td>
</tr>
<tr>
<td>$W_{\text{push}}$ (J·kg$^{-1}$)</td>
<td>0.11 (0.03)</td>
<td>0.05 (0.02)*</td>
</tr>
<tr>
<td>Intact limb</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$W_{\text{neg}}$ (J·kg$^{-1}$)</td>
<td>-0.24 (0.07)</td>
<td>-0.23 (0.07)</td>
</tr>
<tr>
<td>$W_{\text{push}}$ (J·kg$^{-1}$)</td>
<td>0.22 (0.06)</td>
<td>0.24 (0.06)</td>
</tr>
<tr>
<td><strong>Roll-over characteristics</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Radius†</td>
<td>0.24 (0.04)</td>
<td>0.24 (0.05)</td>
</tr>
<tr>
<td>Total arc length†</td>
<td>0.20 (0.03)</td>
<td>0.16 (0.02)*</td>
</tr>
</tbody>
</table>

Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Össur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock, Germany); $W_{DS,\text{lead}}$: center of mass mechanical work performed under the leading limb during dual support; $W_{DS,\text{trail}}$: center of mass mechanical work performed under the trailing limb during dual support; $W_{SS}$: center of mass work during single stance; $W_{\text{neg}}$: negative mechanical ankle work; $W_{\text{push}}$: ankle push-off mechanical work.

* denotes statistical difference between prosthetic feet (p < .05).

† parameters are normalized to center of mass height.
Discussion

This study showed that walking with the energy storage and return (ESAR) foot resulted in a reduced step-to-step mechanical cost compared to a conventional solid ankle cushioned heel (SACH) foot. The amount of mechanical work performed on the COM found in the current study is in close agreement with results from previous studies using similar prosthetic feet [98, 110, 152, 192, 241]. Although more positive COM work can be generated with the ESAR foot compared to the SACH foot, values are still substantially lower compared to positive COM work performed under the intact limb (36.8% lower) or values found in able-bodied controls walking at a similar speed (53.9% lower) [52].

Push-off work

In normal walking the work generated at the ankle is the major contributor to the total COM push-off work under the trailing limb during step-to-step transition. This factor can indeed explain part of the differences found in transition cost between prosthetic feet. Ankle push-off work was 120% higher with the ESAR prosthesis, however, it was still 50.0% less than that generated by the contra-lateral intact ankle. These values are in line with previous studies [183]. Because the ESAR prosthesis is a passive device, the amount of push-off work that can be generated is related to the amount of elastic-strain energy

Figure 2. Push-off power over the stance period.
Push-off power transferred from the foot to the shank of the prosthetic limb (A) and the intact limb (B). The filled areas represent the part over which ankle push-off work was calculated.
Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Össur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock, Germany).
that can be stored during the preceding stance period. Congruous with literature\textsuperscript{[84]}, negative work performed by the prosthetic foot and ankle was 55.5% higher during stance in the ESAR foot compared to the SACH foot (Table 2). Part of this higher negative work is stored to allow for the increased push-off work. The period during which this energy was stored was the same in both prostheses and predominantly occurred in mid-, to late stance (Figure 2). To sum, differences in step-to-step transition cost can be partly explained by the ability of the ESAR feet to store during stance and return this power during push-off, thereby, reducing the collision loss.

**Step length**

In addition to improved ankle push-off work, step-to-step transition cost is dependent on the step length; taking shorter steps will require less directional change of the COM, and as such, reduce the collision loss. Previous experimental results showed that step-to-step transition cost and metabolic energy cost will increase in proportion to the fourth power of step length\textsuperscript{[55]}. In the current study, no difference in stride length was found between prosthetic feet. However,
subjects demonstrated a larger step length asymmetry with the SACH foot compared to the ESAR foot. More specifically, subjects took relatively shorter steps when push-off was performed with the SACH prosthesis (i.e. shorter intact limb step length), thereby, potentially mitigating the increased step-to-step transition cost with the SACH foot. Conversely, subjects took a relatively larger consecutive prosthetic step (intact push-off). This increase in prosthetic step length with the SACH prosthesis could explain the higher collision loss (16.7%) found in that step (Figure 1B). The cause for the relatively smaller intact step when push off is performed with the SACH prosthesis could be related to the reduced push-off power that is generated with the SACH foot. Additional explanation can be sought in the inherent mechanical properties of the SACH prosthesis limiting long steps. The rigid ankle of the SACH foot restrains dorsal flexion during midstance resulting in an early heel rise and concomitantly shorter steps. Moreover, the shorter keel found in the SACH foot compared to the ESAR foot results in a highly flexible forefoot section and contributes to an earlier onset of forefoot rocker. Close examination of Figure 3B affirms the notion that with the SACH foot the forefoot rocker is initiated earlier (earlier flattening of the line).

Roll-over characteristics

The ability of the human ankle and foot to move in a controlled fashion while the center of pressure progresses forward under the foot does not only result in larger intact step length, it also greatly influences the directional change of the COM necessary during step-to-step transition, and thereby directly influences the transition cost. When taking into account the fact that radii of the roll-over shape were normalized using the measured COM height as opposed to leg length, roll-over shape radii found in this study approximate the aforementioned optimal radius of 0.3 times the leg length. Contrary to our hypothesis, roll-over shape radii found in the current study did not differ between the two prosthetic feet. These results seems to contradict previous findings by Curtze and colleagues (2009) who stated that when feet are tested using quasi-static mechanical loading markedly different radii are found between both prosthetic feet. However, inherent mechanical properties of the prosthetic feet might be subdued due to alignment alterations made by the prosthetist when fitting the prosthesis. Moreover, subjects might alter their kinematics in order to ensure a foot-ankle curvature with a radius of .3 times the leg length. As roll-over shape radii did not differ between feet during walking this factor does not affect the difference in step-to-step transition cost found between both prosthetic feet. However, in accordance with literature, the total arc length was shorter in the SACH foot as opposed to the ESAR foot. The shorter keel length of the SACH foot directly influences the total arc length as it limits COP forward progression during late stance. The shorter arc length, and concomitantly shorter COP
forward progression could possibly contribute to the larger collision loss found in the SACH foot as it increases the necessary directional change of the COM at a given step length \[124\]. Besides the shorter forward COP progression and the earlier onset of forefoot rocker in the SACH prosthesis, strikingly large similarities are found in the shape of the progression curve between both feet (Figure 3B), indicating that inherent differences in prosthetic roll-over shape characteristics are attenuated when walking with the prosthesis.

*The optimal prosthesis*

In order to achieve a reduction in metabolic energy cost the energy storage and release of the prosthetic foot needs to be of the right size and at the right instant during gait. For example, previous studies have shown that excessive push-off work can lead to compensational joint work requiring metabolic energy to ensure stability \[147, 192, 241\]. Additionally, the period in gait during which energy is stored \[192\] and the timing of energy release \[129, 181\] are factors that, if not optimal, can attenuate any positive effects of an increased push-off work on metabolic energy cost \[88\]. Recently an ESAR foot has been developed that is able to actively control the timing of the energy release \[26\]. This prosthesis has been shown to enable push-off work similar to that of a biological ankle. However, whereas positive results were found on metabolic energy cost in able-bodied subjects \[26\] it led to an unexpected increase in metabolic cost in people with a lower limb amputation \[192\]. Contrary to the passive devices, the new generation ‘bionic’ prostheses are able to generate net positive work during the stance phase of walking \[6, 219\]. A recent study showed that metabolic energy cost can be reduced towards that of able-bodied walkers \[98\]. Although promising, it is important to note that these prostheses use batteries to generate ankle work, which would, theoretically, result in a metabolic cost below that of able-bodied walkers.

Evidently, the efficacy of a prosthesis to reduce the metabolic energy cost is not merely related to the ability to reduce the step-to-step transition cost, other factors like for example the prosthetic-stump interface, an altered balance control \[110, 218, 228\], alterations in swing-leg kinematics \[53\], compensational proximal joint moments \[68, 192\], and the inability to use energy transfer through bi-articular muscles \[158\] could influence overall metabolic energy requirement. Current results underscore the importance to further elucidate how these factors affect metabolic energy cost and attenuate the positive effect of a decrease in step-to-step transition cost, in order to further optimize prosthetic design.
Limitations

One of the limitations of this study was the relative short accommodation period of one day. This could have amplified differences in the gait pattern between the prescribed ESAR and the SACH prosthesis. Unfortunately, longitudinal studies into adaptation time with a novel prosthesis are scarce. In a study by Grabowski and colleagues (2010) no differences on metabolic energy cost were found after three days while differences after 21 days were significant. This might indicate that a minimum of at least more than three days is required to detect differences in metabolic energy cost. Whether the suggested adaptation time also applies to mechanical outcome measures remains undetermined. With an adaptation time of one day we did find the anticipated differences, though it is unclear whether these would increase, or decrease after more adaptation time. Moreover, contrary to Grabowski and colleagues (2010) the subjects in our study were all in some degree familiar with the SACH prosthesis as either their bath prosthesis or as their first prosthesis after amputation.

Because outcome variables are greatly influenced by walking speed, we had subjects walk at a set walking speed of 1.20 m·s$^{-1}$. To compare this speed to subjects’ preferred walking speed we had subjects walk at their own preferred walking speed for 40 meters through a corridor while walking speed was determined based on the time it took to walk the middle 20 meters. Although the preferred speed (1.27 m·s$^{-1}$) differed significantly ($p = .032$) from the 1.2 m·s$^{-1}$ enforced during the measurement, differences are small and maybe clinically irrelevant. However, we cannot exclude that deviation from subjects’ preferred walking speed might have influenced the results.

Conclusion

This study showed that walking with the widely prescribed ESAR foot resulted in a lower step-to-step transition cost compared to walking with the conventional SACH prosthesis. This difference is explained by the higher amount of positive work performed by the ESAR foot during push-off and the larger arc length of the roll-over shape compared to the SACH foot. The lack of evidence supporting a reduction in metabolic energy cost while walking with an ESAR foot suggests that other factors outside those related to step-to-step transition cost contribute to the overall metabolic energy cost. In order to enhance prosthetic development, it remains a formidable challenge to disentangle and optimize these factors in order to reduce the metabolic energy cost while walking with a prosthesis.
Mind your step: Metabolic energy cost while walking an enforced gait pattern

Mind your step: Metabolic energy cost while walking an enforced gait pattern

Wezenberg D, de Haan A, van Bennekom CA, and Houdijk H,
"Mind your step: Metabolic energy cost while walking an enforced gait pattern",
Abstract

The energy cost of walking could be attributed to energy related to the walking movement and energy related to balance control. In order to differentiate between both components we investigated the energy cost of walking an enforced step pattern, thereby perturbing balance while the walking movement is preserved. Nine healthy subjects walked three times at comfortable walking speed on an instrumented treadmill. The first trial consisted of unconstrained walking. In the next two trials, subject walked while following a step pattern projected on the treadmill. The steps projected were either composed of the averaged step characteristics (periodic trial), or were an exact copy including the variability of the steps taken while walking unconstrained (variable trial). Metabolic energy cost was assessed and center of pressure profiles were analyzed to determine task performance, and to gain insight into the balance control strategies applied. Results showed that the metabolic energy cost was significantly higher in both the periodic and variable trial (8% and 13%, respectively) compared to unconstrained walking. The variation in center of pressure trajectories during single limb support was higher when a gait pattern was enforced, indicating a more active ankle strategy. The increased metabolic energy cost could originate from increased preparatory muscle activation to ensure proper foot placement and a more active ankle strategy to control for lateral balance. These results entail that metabolic energy cost of walking can be influenced significantly by control strategies that do not necessary alter global gait characteristics.
Introduction

Patients require more metabolic energy to walk at the same speed as their matched healthy controls \(225\). This increase has been attributed to an altered and less efficient movement pattern \(44, 110, 132, 140\), an increased effort for balance control \(109, 148\), or a combination of both \(103, 140, 148\). As early as 1986, Workman and Armstrong \(237\) described a model which explicitly differentiated between the energy cost associated with balance preservation and the energy cost needed to perform the walking movement. In subsequent years, however, many studies have focused predominantly on alterations in the walking movement in terms of external and internal mechanical work to explain differences in energy cost between normal and pathological gait \(44, 132\). In contrast, the energy cost associated with balance preservation has deserved less attention. Yet, research has demonstrated that, even during upright standing, balance perturbations result in higher metabolic energy expenditure \(109\). Moreover, it has been shown that metabolic energy cost while walking can be decreased by facilitating balance control by means of an external stabilizing device \(41, 57, 161\).

Differentiation between the energy cost needed to perform the walking movement and that associated with balance control is less clear than the model of Workman and Armstrong suggests. Frequently, balance perturbations directly lead to adaptations in the walking movement \(236\), for example, many patients walk with increased step-width to extend the margins of the base of support \(57, 102, 128\) resulting in a more stable gait. However, the dynamics of this wider gait pattern will also result in increased mechanical work and consequently higher metabolic energy expenditure \(54\). The question arises whether this extra metabolic energy should be assigned to the metabolic energy cost related to balance control or to that associated with the walking movement. Since many balance control strategies alter gait characteristics, the energy requirement of alternative mechanisms controlling balance, as for example muscle co-activation \(103\) and increased local dynamic stability \(51\), becomes indeterminable.

Information about the energetic demands of these alternative control strategies can be gained by perturbing balance, while restricting alterations in the averaged gait characteristics. Experimentally, this can be accomplished by enforcing the self-selected gait pattern on a treadmill, thereby restricting alteration in global gait characteristics like speed, step-width, step length and dual support time. Enforcing a gait pattern simultaneously imposes a balance perturbation. Normally, humans primarily use a stepping strategy to maintain stability when walking; balance perturbations in one step are corrected in consecutive steps by appropriate placement of the foot \(102\). However, when foot positions are enforced, the use of a stepping strategy will consequently be restricted. Therefore, subjects have to revert to an altered, possibly less efficient, balance control strategy. In
addition, metabolic energy cost might be influenced by the fact that placing the foot at a predefined location will require more conscious control \[^{7, 59}\] and associated preparatory muscle activation.

The primary objective of this study was to investigate the influence of walking at an enforced gait pattern on the metabolic energy cost in young healthy subjects. By enforcing a gait pattern balance is perturbed, without allowing global gait characteristics to change. It was hypothesized that enforcing a gait pattern, composed of the averaged self-selected gait pattern, would lead to a substantially higher metabolic energy cost compared to walking unconstrained. The increase in metabolic energy cost was thought to be even higher when an exact copy of the self-selected gait pattern is enforced, since the variability, naturally present when walking, result in a less predictable enforced walking pattern. In addition to the primary objective, gait parameters were measured to determine the accuracy of task performance and center of pressure (COP) profiles were analyzed to gain insights into the balance control strategies applied.

**Method**

**Subjects**

Nine subjects with an average age of 27.6 (SD = 9.9) years and a body mass index of 21.3 (SD = 1.8) participated in this study. Subjects were healthy, having no muscular, neurological or visual limitation, and were asked to refrain from drinking coffee and eating large meals at the day of the test. Subjects were informed about the research procedure before they gave informed consent. This study was approved by the local ethics committee.

**Equipment**

Subjects walked on an instrumented treadmill (C-Mill, ForceLink, Culemborg, The Netherlands). The treadmill was mounted with a force plate (1.5 by 1 m) from which step characteristics could be derived. Data acquisition was performed using a 32 bits A/D converter at 100 Hz. The instrumented treadmill was connected to a projector which could project step patterns onto the walking surface.
Walking trials

After a familiarization period on the treadmill, subjects’ comfortable walking speed was obtained using the method described by Martin and colleagues (1992) \[141\]. Thereafter, subjects walked three times 6 min at comfortable self-selected speed separated by a minimum of 10 min rest. The first trial consisted of unconstrained walking. In the next two randomly assigned trials, the gait pattern was enforced by means of a step pattern projected on the treadmill. Each projected step consisted of a yellow rectangular shape, measuring 30 by 10 cm, projected on the black treadmill belt. The projections moved at the same speed as the treadmill and started 3 m in front of the subject, which enabled appropriate anticipation of foot placement. Subjects were asked to place their feet within the projected surface. In one of the enforced trials, subjects were confronted with a gait pattern composed of the averaged stride length, spatial asymmetry and step-width determined using the data collected in the unconstrained trial. The projections in this trial were periodic and contained no variability; therefore, this trial will be referred to as the periodic enforced trial (Figure 1). In the second enforced trial, a gait pattern was projected that was an exact copy of the steps taken during the unconstrained trial. Although on average the same walking pattern was enforced compared to the periodic trial, subjects were now confronted with an amount of variation equal to the variation while walking unconstrained. This trial will be referred to as the variable enforced trial (Figure 1).
During the enforced trials task execution was visually monitored, and stimulating feedback was given in order to redirect attention towards appropriate foot placement whenever subjects showed a decline in task performance.

**Data analysis**

To ensure that steady state was reached, only data collected during the last minute of each walking trial was used for analysis. Oxygen consumption ($\dot{V}O_2$, ml · min$^{-1}$) was measured breath-by-breath using open circuit respirometry (Oxycon delta, Jaeger, Hoechberg, Germany). $\dot{V}O_2$ was converted to metabolic power ($P_{\text{met}}$) using the following equation:

$$P_{\text{met}} = (4.960 \cdot \text{RER} + 16.040) \cdot (\frac{\dot{V}O_2}{60})$$

where RER is the respiratory exchange ratio, which in all analyzed trials was below one. $P_{\text{met}}$ was normalized for bodyweight and walking speed to obtain metabolic energy cost ($C_{\text{met}}$, J kg$^{-1}$ · m$^{-1}$).

Using a local reference frame within the force plate, center of pressure (COP) profiles were collected during the last minute of the walking trial. The profiles resembled a butterfly pattern from which heel contact was defined as the local minima in anterior-posterior direction contra-lateral to the side of interest. Toe-off was determined by identifying a local minimum in the total force that occurred around the local maxima in the COP profile as previously outlined by Roerdink and colleagues (2008) \[177\].

From known toe-off and heel contact instants, stride length (cm), dual support time (%) and step-width (cm) were calculated. Concretely, stride length was defined as the distance between heel contact and consecutive heel contact of the same leg. Dual support time was expressed as a percentage of the gait cycle when both feet were in contact with the ground. Step-width was defined as the absolute distance in frontal plane of the COP at consecutive periods in the gait cycle when both feet were in mid-stance (Figure 1). Mid-stance was defined as the instant at which COP had travelled half the distance between toe-off and heel contact.

Compliance with the task was monitored by comparing the mean stride length, dual support time and step-width between trials. The accuracy with which the foot was positioned within the projected surface was calculated as the standard deviation around the averaged COP location within the projection in medio-lateral and anterior-posterior direction at mid-stance.

Differences in balance control strategies, applied during single limb support, were analyzed by investigating the COP displacement under the foot \[102\].
Walking an enforced gait pattern

Previous research has shown that a more variable COP displacement in medio-lateral direction during single limb support indicates a more pronounced ankle strategy\(^{[102]}\). The amount of COP displacement under the foot was determined as the standard deviation in medio-lateral COP position during single limb support. The averaged value over all single support phases during the last minute of data collection was used in the statistical analysis.

Statistics

Data were tested for normality using the Kolmogorov-Smirnov test. All parameters were normally distributed. Statistical analysis to test for differences between trials was performed using repeated measurement ANOVA (SPSS Inc., Chicago, IL, USA), with subjects as between factor and trial as the within factor. Multiple comparison correction was made using a Sidak post hoc test. The consistency between trials in global gait characteristics (stride length, dual support time and step-width) was tested using intra-class correlation coefficient (ICC\(_{(3,1)}\)) with a two-way mixed model for single measurements\(^{[196]}\). The accuracy of foot placement between the two enforced trials was tested using paired sample t test. Significance was set at the 5% level. Effect sizes were calculated using eta squared (\(\eta^2\))\(^{[69]}\).

Results

The metabolic energy cost (\(C_{\text{met}}\)) was significantly higher in both the periodic and variable walking trial (4.38 ± 0.34 and 4.59 ± 0.45 J·kg\(^{-1}\)·m\(^{-1}\), respectively) compared to unconstrained walking (4.05 ± 0.35 J·kg\(^{-1}\)·m\(^{-1}\), \(p < .001\); Figure 2). Effect sizes range between 0.42 and 0.62 indicating a medium to large effect (Figure 2)\(^{[69]}\). The difference between the periodic and variable trial failed to reach significance (\(p = .09\)).

Mean stride length and dual support time did not differ between trials (\(p = .724\) and \(p = .078\), respectively). This in contrast to mean step-width (\(p = .017\)). Step-

Table 1. Mean (± SD) step parameters.

<table>
<thead>
<tr>
<th></th>
<th>unconstrained</th>
<th>periodic</th>
<th>variable</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride length (cm)</td>
<td>135 (9.6)</td>
<td>135 (9.7)</td>
<td>135 (9.6)</td>
</tr>
<tr>
<td>Dual support time (%)</td>
<td>22 (3.4)</td>
<td>23 (3.4)</td>
<td>23 (3.5)</td>
</tr>
<tr>
<td>Step-width (cm)</td>
<td>11 (3.6)</td>
<td>13 (2.9)</td>
<td>11 (2.4)</td>
</tr>
</tbody>
</table>

NOTE. Mean (± SD) for stride length (cm), dual support time (%) and step-width (cm). The significant difference in step-width is denoted using an asterisk (*). Pair wise comparisons of step-width failed to reach significance.
width seemed to be larger in the periodic trial, although pair wise comparison failed to reach significance (p = .091 for periodic versus unconstrained and p = .083 for periodic versus variable; Table 1). Further analysis of the consistency over trials revealed that single measurement ICC(3,1) (95% CI) was 0.99 (0.98 - 0.99) for stride length, 0.94 (0.83 - 0.99) for dual support and 0.76 (0.45 to 0.93) for step-width.

The accuracy of foot placement was analyzed by calculating the amount of deviation around the averaged COP location within the projection at mid-stance. Larger deviations were found in medio-lateral direction in the variable trial compared to the periodic trial (right foot; p = .013, left foot; p = .022). In anterior-posterior direction no differences were found between trials (right foot; p = .432, left foot; p = .344; Figure 3).

The variation in COP displacement under the foot during single limb support differed between trials (left side; p = .002, right side; p = .004). Figure 4a illustrates the medio-lateral COP displacement under the left foot during single limb support for a number of consecutive steps of a typical subject. The overall mean variation (cm) in consecutive steps is illustrated in Figure 4b. The periodic and variable trials both showed a higher amount of variation (1.14 and 1.21 cm, respectively) in the medio-lateral direction when compared to the unconstrained trial (0.84 cm). However, differences between the unconstrained and periodic trial failed to reach the set level of significance (left side; p = .076, right side; p = .057), although a trend can be noted.
Discussion

This study investigated whether balance and movement control can contribute significantly to the metabolic energy cost of walking, independent from alterations in global gait characteristics. By enforcing subjects to walk at their self-selected gait pattern, alterations in the global gait characteristics were restricted while balance was perturbed. From the two enforced trials, the variable trial was hypothesized to elicit the largest effect on metabolic energy cost. In accordance with our hypothesis, \( C_{\text{met}} \) increased when a walking pattern was enforced compared to walking unconstrained. This increase reached 8% in the periodic trial and 13% in the variable trial. A similar (14%) increase in \( C_{\text{met}} \) was previously found by Donelan and colleagues (2001) when subjects were forced to follow their self-selected step-width \(^{[54]}\). An increase in metabolic cost of more than 10% is commonly regarded as a relevant change \(^{[13, 190, 206]}\). Although this value is partly based on subjective clinical opinion, it implies that our experimental manipulation elicited a substantial effect during the variable trial, and that the periodic trial failed to reach this criterion.

Comparing the enforced trials

It was hypothesized that the variable trial would elicit the highest effect on metabolic energy cost. The highest energy cost was indeed found in the variable trial, however, differences between the periodic and variable trial failed to reach the set level of significance. Calculation of the effect size between the periodic and variable trial \((r = 0.67)\) revealed that 45% of the total variance could be attributed to the task \(^{[69]}\). Due to the small sample size, the effect size might give...
a more objective measure of the importance of the observed effect. Additionally, analysis of the COP position at mid-stance showed more deviation in medio-lateral direction, with respect to the averaged COP position within the projected surface, in the variable trial compared to the periodic trial (Figure 3). This finding indicates that the error in foot placement was larger in the variable trial and further supports the notion that during the variable trial balance was perturbed more severely, consequently resulting in a higher metabolic load compared to the periodic trial.
Compliance with the task

A high level of compliance with the task is essential for a valid interpretation of the results. Subjects were able to accurately position the foot within the projected surface. The averaged COP position at mid-stance was well within the projection and deviated only to a small extent between subsequent steps (medio-lateral = 1.8 cm, anterior-posterior = 3.6 cm; Figure 3).

Considering that the enforced trials were based on the unconstrained walking trial, averaged step parameters should be equal to the values when walking unconstrained. This was true for the stride length and dual support time. Step-width was wider in the periodic trial compared to the unconstrained and variable trial, although, pair wise comparison failed to reach statistical significance and the analysis of the consistency over trials revealed a high consistency (ICC(3,1) = 0.76). The wider step-width could have amplified the found increase in metabolic energy cost. However, using the equation proposed by Donelan and colleagues (2001) only 1.2% of the increase in metabolic cost found in the periodic trial could be explained by the wider step-width. The small deviation in COP location within the projected surface at mid-stance and the comparable mean gait parameters seem to justify the conclusion that subjects complied with the task sufficiently.

Control strategies

Considering that subjects were able to comply with the task and walked at equal gait patterns over all trials, means that mechanical adaptations in the global walking pattern were limited. Therefore, observed increases in the metabolic energy cost should be related to changes in control strategies other than those altering global gait characteristics. A number of control strategies could have been employed to overcome the balance perturbation due to the enforced gait pattern, and consequently explain the higher oxygen cost found. First, the step pattern was projected well before foot contact, enabling sufficient time for subjects to anticipate on the upcoming step. Therefore, subjects possibly relied to a lesser extent on the passive dynamics of their swing leg but consciously made anticipatory modifications of their leg's trajectory. The precise placement of the foot is mostly regulated by activation of the responsible muscles just prior to foot contact (i.e. at the end of swing phase). This would, in contrast to walking unconstrained, result in more motor cortical involvement [7, 59]. In addition, the precision demands imposed eventuate in more muscle co-activation [193, 235]. These adaptations in motor control and preparatory muscle activity may partly explain the increased metabolic energy cost observed [106]. This notion is substantiated by the fact that the variable (less predictable) trial resulted in the highest metabolic energy cost. However, further research incorporating electromyography recordings are warranted to substantiate these conjectures.
Secondly, human gait is laterally unstable \cite{57, 128}. Normally, lateral stability is ensured by accurate placement of the foot (i.e. a stepping strategy) \cite{8, 102}. Due to the restrictions in foot placement a stepping strategy could not be used. Other strategies were needed, as for example a lateral ankle strategy. The lateral ankle strategy requires an active feedback loop and presumably more energy (i.e. is less efficient) than the stepping strategy \cite{8, 102}. Analysis of COP displacement during single limb support showed more variation in medio-lateral COP patterns during single limb support in the enforced trials compared to unconstrained walking, this effect was largest for the variable trial. This result supports the idea that, after positioning of the foot, a greater amount of fine tuning was required by means of an ankle strategy. In addition to an increased use of the ankle strategy, alterations in hip ad-/adduction and movements of the arms might have contributed to the measured increase in metabolic energy cost; however, these latter responses could not be assessed in the current experimental set-up.

**Conclusion**

Walking at an enforced gait pattern elicited a substantially higher metabolic energy cost. The greater variation in COP trajectories during single limb stance supports the idea that subjects relied more on an ankle strategy to overcome the inability to use the more efficient stepping strategy. This, in addition to altered preparatory muscle activation, is thought to have contributed to the increased metabolic energy cost found.

Obtained results have both a clinical and experimental implication. As for the clinical implication results demonstrate that increased metabolic energy cost in patients could, in addition to an altered gait pattern, originate from changes in balance control strategies that do not necessarily alter global gait characteristics. Experimentally, results underline that metabolic energy cost needs to be interpreted with caution when a gait pattern is enforced; the enforcement in itself could lead to alteration in the metabolic energy cost.
Walking an enforced gait pattern
Epilogue
Introduction

The overarching aim of this thesis was to enhance our knowledge about some of the factors that influence the ability to regain and maintain walking after unilateral lower limb amputation. While in the previous individual chapters specific aims related to this general aim have been investigated and discussed, this Chapter combines the obtained results and by critically reflecting on these results, determines if, and how, these studies have helped us to better understand the factors associated with prosthetic ambulation. Based on this synthesis, we will formulate recommendations for future research and clinical implications for prosthetic rehabilitation interventions and prosthetic design. To correctly value the conclusions drawn it is important to understand the considerations and implications of the methodology chosen. Hence, this Chapter starts by providing some general considerations regarding the research methods used.
Methodological considerations

In the Chapters 2, 3 and 4 a group of older subjects aged between 50 and 75 participated in all three studies. Using the same group of subjects in all three studies could have biased the results limiting generalizability. Rationale behind choosing older subjects was that these subjects typically experience the greatest difficulty regaining and maintaining walking after undergoing lower limb amputation. In part this is related to the fact that most of the older people are amputated due to vascular deficiency. However, to date surprisingly few experimental studies have been performed that specifically focused on the elderly vascular amputee. Possible reasons for the paucity might be that comorbidity is thought to influence the outcome parameters, moreover, mortality rates one year post-amputation are very high. Also a large group is not fitted with a prosthesis but will ambulate using a wheelchair. To amend the limited knowledge about the specific challenges that these people face, a group of vascular amputees was included in the first three studies described in this thesis. This group is compared to a group of traumatic amputees and able-bodied control subjects in each of the three Chapters. These groups were matched on age because age is known to substantially influence the outcome parameters (e.g. aerobic capacity) of the experiments summarized in this thesis. However, as most of the trauma related amputations are performed on adolescents or young adults, matching subjects on age yielded a significant difference between groups in the time since their first prosthesis; i.e. traumatic amputees had walked with a prosthesis for significantly more years. The time since their first prosthesis may have influenced the measured aerobic load while walking with the prosthesis, with higher physical strain values found in the relatively inexperienced vascular amputees. Unfortunately, longitudinal studies examining the time and process of adaptation to a prosthesis and the influence on biomechanical outcome parameters are scarce. Because the presence of a traumatic amputation and the time since first prosthesis are highly correlated, multicollinearity posed a problem in the regression analysis, rendering statistical adjustments impossible. However, all subjects in the current study were familiar with their prosthesis and able to ambulate for four consecutive minutes on a treadmill, therefore, it was believed that possible differences due to a different time since the first prosthesis were substantially smaller than the difference caused by level or etiology, and subsequently, did not influence the conclusions.

Subjects recruited to participate in the studies were either informed about the study by their physician or prosthetist, or responded to advertisements in a local magazine or on the Internet. As participation was voluntary, there will
have been a selection bias towards the more physically fit, limiting generalization to the overall population. However, the sample bias most likely resulted in an underestimation of the effects found in Chapters 2, 3 and 4. Conclusions drawn in Chapters 5 and 6 were neither influenced by a difference in time since amputation nor by a selection bias because in the repeated measurement approach subjects functioned as their own controls.

Aside from a possible biased sample in Chapters 2, 3 and 4, the number of subjects in each group was unequal. Moreover, the number of subjects with a vascular trans-femoral amputation was relatively low. This might have precluded detecting differences present among groups. In this regard it is important to mention that the differences found among groups for the different outcome parameters were large and indeed clinically relevant, while other outcome parameter that were borderline significant (e.g. the level of amputation) were substantially smaller and it is questionable whether these latter differences have clinical relevance.

Walking ability explained?

Whilst taking into account the methodological considerations, the question to answer in this thesis was how relative aerobic load affected subjects' walking ability in terms of walking speed and walking economy. From the experiments conducted in Chapter 4 one can derive that having sufficient aerobic capacity could be a prerequisite in order to regain walking at a certain speed. For example, vascular amputees are unable to walk at a walking speed similar to that of the able-bodied controls as this would require an aerobic effort that is beyond the aerobic capacity as measured during the cycling exercise test. Reducing walking speed is a method to ensure that the effort while walking (expressed as the relative aerobic load) does not surpass a predefined acceptable limit. In addition to being a prerequisite that enables walking at a certain speed it might be speculated that the relative aerobic load is an important factor governing the choice of walking speed. This speculation is based on the fact that subjects lower their walking speed even though it negatively affects walking economy (i.e. the slower walking speed requires more oxygen per distance walked) and prevents them from keeping up with able-bodied peers. Moreover, traumatic amputees reduced their walking speed to such an extent that it matches the relative aerobic load of the able-bodied controls. These results are interesting, as apparently the generally considered notion that walking speed is governed by walking economy is refuted for older people who walk with a lower limb prosthesis. However, these conclusions ought to be interpreted with care, as the association found between the relative aerobic load and the reduction in walking speed does not
necessary imply causality. More importantly, relative aerobic load might be only one of the total sum of factors defining subjects’ walking ability.

This study used a biophysical approach to understand the underlying mechanisms influencing subjects’ walking ability. This approach is mechanistic in origin and excludes any cognitive or psychosocial factors, and therefore, cannot fully explain subjects’ walking ability. Retrospective research provides strong evidence that the level and cause of amputation, cognition, functional level prior to amputation, and age are important predictive factors. While moderate evidence exists for the presence of stump and/or phantom pain, motivation, co-morbidity, smoking and social support as predictive factors for subjects’ ability to regain walking ability \[80, 99, 137, 165, 172, 184, 188, 214, 216\]. Though these factors might in different proportions predict subjects walking ability, the reason for their association with walking ability is not determined. On the other hand we know from the studies presented in this thesis that relative aerobic load is a prerequisite for regaining and maintaining walking ability.

To recapitulate, subjects’ walking ability is, amongst other factors, associated with the relative aerobic load. The higher the relative aerobic load while walking the smaller the spectrum of functional walking speeds at which an individual with an amputation can comfortably walk. Moreover, understanding the (relative) aerobic load provides relevant information about the importance of adequate aerobic capacity, and therefore, provides evidence that prosthetic rehabilitation ought to also focus on improving aerobic capacity. However, to determine causality future studies should focus on the prospective association between the relative aerobic load, aerobic capacity and subjects' walking ability.

Discovering aerobic capacity

Improvement of the relative aerobic load during walking in persons with a lower limb amputation can be accomplished by either reducing the absolute aerobic load or by increasing the aerobic capacity. To do so, requires fundamental knowledge about both the causes for the increased aerobic load (Chapters 5 and 6) and the peak aerobic capacity of the people walking with a prosthesis (Chapters 2 and 3). To obtain fundamental knowledge about the peak aerobic capacity a discontinuous, graded, one-legged, peak exercise test was developed. This exercise test proved to be both feasible and valid when testing older adults with either a traumatic or vascular amputation. Because it targets the lower limb muscles, it provides important information about the aerobic capacity while walking. However, it can be postulated that when subjects are to a large extent wheelchair bound and physical activity primarily involves the upper extremities,
an exercise test which uses both the arm and legs is a more functional evaluation [12, 76]. However, electrocardiogram disturbances and coordination difficulties might hamper the safety and feasibility of a combined arm-leg exercise test. Which test is most appropriate is dependent on the purpose of the test, the ability of the subject to perform the test and the available equipment.

As noted in Chapter 3, remarkably little information about the exercise capacity of amputees was known. Chapter 3 provides evidence for the intuitive notion that the peak aerobic capacity is substantially reduced in people with a lower limb amputation. In older subjects the reduced muscle mass [72] and mitochondrial capacity, result in a reduced muscle oxidative capacity [71] and concomitantly, a reduced peak aerobic capacity [73]. This reduction due to age is aggravated when combined with limited physical activity [200]. In healthy elderly a reduced peak aerobic capacity limits walking ability [71], but in older people with an amputation this natural decline in peak aerobic capacity is superimposed on the already present capacity limitation due to behavioral characteristics (i.e. food intake, sedentary lifestyle or smoking) and preexisting medical conditions [137]. Consequently, efforts to improve the aerobic capacity during rehabilitation must indeed be encouraged.

**Improving aerobic capacity**

It might be speculated that due to the deconditioned status of the older subjects, small improvements in the aerobic capacity might result in large effects on the functional performance and walking ability. In Chapter 4 a data-driven theoretical model was developed which helped to gain some insight into the magnitude of the potential effects that can be reached when aerobic capacity is theoretically improved. Based on this model it was predicted that an increase of 10% in peak aerobic capacity in the older vascular amputee resulted in a 9.1% reduction in walking effort, a 17.3% higher walking speed and an improvement in walking economy of 6.8%. These improvements are statistically significant and most likely lead to clinically relevant improvements in subjects’ walking ability. Moreover, these effects are larger than the effect reached by a reduction in aerobic load through the use of currently commercially available state-of-the-art prosthetic devices.

Though the above clearly demonstrates the importance of an optimal aerobic capacity, several guidelines either do not mention aerobic training as a rehabilitation intervention [14-15], or fail to provide specific guidelines for aerobic training [157, 205]. Fortunately, some awareness of the importance of physical
capacity exists as more than 80% of users and professionals in the field of prosthetics rate this aspect as an important predictor of prosthetic use. Despite the awareness, strikingly few longitudinal studies have been performed that investigate the effect of aerobic training on subjects’ peak aerobic capacity and walking ability \(^{[30, 216]}\). Those studies that have been performed involve traumatic amputees and report increases varying between 27.2\% and 41.2\% \(^{[22-23, 170]}\). Some evidence for the trainability of older vascular amputees might be based on studies in older patients diagnosed with symptomatic peripheral arterial disease. These studies have shown that aerobic exercise training, either through walking or arm- or leg cycling, resulted in an increase in walking distance through improvements in peripheral circulation, walking economy and peak aerobic capacity \(^{[4, 245]}\). These results, together with the knowledge that aerobic exercise is beneficial in older adults with a wide range of pathologies \(^{[142]}\), vindicates the notion that also in the vascular amputees improvements in peak aerobic capacity are possible. Future training studies specifically focusing on this growing group of patients are highly needed.

One might pose the question whether aerobic training is also beneficial for the traumatic amputee with no comorbidity. For people recently amputated due to trauma, the post-operative period of inactivity leaves them prone to a reduced peak aerobic capacity, and therefore, they are likely to benefit from aerobic training to ease prosthetic ambulation. Clearly, aerobic exercise training must not to be restricted to those recently amputated. Based on the results presented in Chapter 4, and the prediction equation presented therein, we know that although the relative aerobic load (i.e. the walking effort) is the same in the traumatic and able-bodied control subjects, traumatic amputees walk slower and below their theoretically most economical walking speed. Consequently, these people would benefit from having a peak aerobic capacity above that of healthy able-bodied controls as this will allow them to adopt a faster and more economical walking speed while maintaining to walk at the same relative aerobic load.

To conclude, the reduced peak aerobic capacity and the high relative aerobic load together with the substantial improvement in walking ability that can potentially be attained by small improvements in peak aerobic capacity, underscores the importance of aerobic exercise training as an integrated part of prosthetic rehabilitation in the vascular amputees. Positive, though more moderate, effects are also expected for the traumatic population. The current lack of knowledge about effective training regimes and the concurrent effect on walking ability is at least remarkable, and the need for future longitudinal studies targeted to improve subjects’ peak aerobic capacity are needed.
Factors influencing the aerobic load

In the final two Chapters the effect of different prosthetic designs on the mechanical cost while walking (Chapter 5) and the effect of different balance control strategies on the overall metabolic energy cost (Chapter 6) were investigated. In the following paragraphs the implications of these findings for overall walking ability in amputees are reviewed.

The prosthesis

Based on the results from Chapter 5 it is made evident that the energy storage and return (ESAR) foot reduces the mechanical energy cost necessary for the step-to-step transition. Unraveling of the causes for this reduction revealed that the improved push-off power and longer roll-over arc length of the ESAR prosthesis positively contributed to this reduction. A reduction in the step-to-step transition cost is assumed to be associated with a reduction in the overall metabolic cost required for walking \[55, 110, 192\]. Remarkably however, no convincing evidence is present in the literature supporting the notion that the metabolic energy cost is lower when walking with the ESAR foot \[88, 104, 212, 219\]. This discrepancy might be a result of the simplifications on which the dynamic walking approach is based. Apparently, the model overlooks important metabolic costly compensational strategies to deal with the demands imposed when walking with the ESAR prosthesis \[131, 148, 215\]. The dynamic walking model proved to be able to differentiate between different prosthetic feet and provides helpful and useful information about the center of mass work. However, it fails to fully explain the metabolic cost associated with prosthetic gait as improvements in step-to-step transition cost do not directly translate to corresponding reductions in the metabolic energy cost \[98, 192\].

Balance control

The challenge we are facing is to decipher which other factors may contribute to the overall metabolic energy cost. One possible candidate is the energy required for balance control. From Chapter 6 it can be deduced that conscious placement of the feet in a designated position has a distinct metabolic energy cost. This is the case even if the position of the feet and the associated mechanical constraints are no different than walking unconstrained. The question that arises is how this translates to the metabolic energy cost associated with balance control in people walking with a prosthesis. The mechanical restrictions and the loss
in motor control at the ankle precludes the use of the normally applied ankle strategy \[102\]. Moreover, people with a lower limb amputation have lost part of their somato-sensory system and, therefore, need to revert to other, possibly more costly control strategies to ensure adequate stability. Possible examples of these strategies are: placement of the prosthetic foot well within the margins that ensure stability \[102\], an increased activation of the contra-lateral intact ankle and hip, a wider stride and/or a slower walking speed \[50\]. Finally, it might be postulated that due to the lack of feedback provided by the somato-sensory system subjects more tightly control their gait in order to ensure a feeling of safety and to eventually prevent a fall despite a possible metabolic penalty \[233\].

In Chapter 6 the conscious control of walking is manipulated by turning a largely automatic task like walking into a task which requires constant cognitive guidance. For people with an amputation, and especially the older people \[96\], maintaining balance might already consume substantial cognitive resources. By redirecting part of these resources away from the walking task alteration in the gait pattern \[86\] and dynamic stability \[49\] are expected to occur. Some support for this hypothesis is provided in this patient group while studying standing balance. For example, Geurts and colleagues (1991) \[82\] concluded that at the start of rehabilitation, standing balance control requires substantial central information processing and during rehabilitation a central reorganization process takes place resulting in a restoration of the automaticity of balance control. In a similar experiment this conclusion was substantiated as dynamic measures of regularity (which are said to be indicative for the amount of automaticity of a task) decreased during rehabilitation \[178\]. Interestingly, Hof and colleagues demonstrated that in the small group of relatively young amputees performing a concurrent secondary task while walking did not affect spatio-temporal gait parameters \[102\]. However, unraveling the complex balance control system while walking might entail that analysis ought to surpass purely spatio-temporal signal analysis. It has been advocated that more dynamic measures ought to be applied to gain additional information about the inherent stochastic structure of the dynamic control of the human movement system \[58, 134\]. Previous studies in a small group of traumatic amputees revealed that amputees indeed responded differently to a perturbation than healthy controls on these dynamic measures \[134\]. Combining dynamic measures with measured of metabolic energy consumption while balance control is perturbed using a concurrent secondary task might provide valuable information about the metabolic cost related to balance control. Possible effects might be ameliorated when the secondary task involves a visual task which interferes with foot vision \[10-11\]. These speculations might form the basis of interesting fundamental scientific studies. Additionally, dual task research more closely mimics the daily life situation during which subjects are also continuously confronted with concurrent tasks that direct attention away from the primary walking task \[82, 238\].
To sum, the latter two studies described in this thesis provide evidence that the aerobic load while walking with a prosthesis originates, in part, from the inefficient and inadequate push-off power resulting in an increased step-to-step transition cost and from possible alterations in the balance control strategy applied. While improvements in subjects’ walking ability can be attained using state-of-the-art prostheses, improving subjects’ walking ability to that of healthy controls involves more than developing a prosthetic ankle and knee that provides appropriate propulsion. Amongst other prerequisites, inefficient balance control strategy, especially in challenging situations, can ameliorate any positive effect of an improved propulsion power.

The prosthesis of the future

To date, technical developments in the prosthetic design have resulted in a wide range of different prosthetic feet aside from the relatively simple energy storage and return prosthesis often prescribed and studied in this thesis. The majority of these newly developed prosthetic ankle and knee components aim to reduce the aerobic load while walking. Unfortunately, the prosthetic devices commercially available often show only marginal improvements in the required metabolic energy expenditure while walking. For example, literature investigating the effects of a micro-processor knee (C-leg®) on walking economy in people with a trans-femoral amputation varies between no differences [20] to a reduction in energy expenditure of 6-7% depending on the speed walked at [100, 187, 195]. Greater reductions in aerobic load might be expected from the so-called ‘bionic’ prostheses, which are powered systems who emulate the function of the biological ankle [98, 219]. To date, these devices are still being developed; however, studies are promising as reductions between 8% and 14% in the cost of walking have been found compared to conventional prostheses [6, 98]. Sophisticated and exciting new developments are currently happening involving prostheses developed for upper extremities. For example, upper arm prostheses have been developed that provide somato-sensory feedback which result in a reduction in phantom limb pain and an increase in functionality [45]. Another new and exciting advancement in the field of upper arm prosthetic rehabilitation, is targeted re-innervation of residual nerves that aims to amplify the information still present in the nerves and use this information to control the prosthetics [127], which might improve subjects balance control. There is hope that in the future these technologies can also be applied to lower limb amputees [2]. Merging these new technologies with the knowledge gathered from the current thesis and previously performed research, might provide fruitful soil for significant improvements in prosthetic design that can result in an improvement in the functioning and quality of life of people with a lower limb amputation.
Future research directions

Prosthetic development and rehabilitation is an exciting research area which has become increasingly important due to the unfortunate growing numbers of people currently with a lower limb amputation. This thesis provides a starting point for a number of additional scientific studies that can contribute to the overarching aim to improve our knowledge of prosthetic gait.

Using a longitudinal research design the proposed beneficial effects, as predicted in Chapter 4, on relative aerobic load and walking ability after aerobic training could (and should) be experimentally evaluated. Moreover, using this research design information about the effectiveness of specific training regimes can be investigated. Both from a scientific and clinical perspective it is interesting to investigate whether, aside from people currently undergoing rehabilitation, people who have finished clinical rehabilitation also experience beneficial results of aerobic training in terms of walking ability, functioning and quality of life.

Despite many years of research and an abundant number of articles published on the topic we still cannot fully explain the origin of the increased metabolic energy requirement when walking with a prosthesis. Applicability of relatively simple models like the dynamic walking model can provide valuable and interpretable information about the mechanical energy requirement when walking with the newly developed state-of-the-art prostheses. Moreover, future studies might determine whether these models can provide a useful tool to guide prosthetic prescription. However, quite possibly the greatest challenge we face is to disentangle the underlying mechanisms which cause the metabolic costly compensational strategies seen when walking with a prosthesis. To do so, implies altering the line of thought from a prosthetic oriented perspective to a more human orientated research perspective. More specifically, if we want subjects to restore their daily functioning without additional metabolic energy cost, a prosthesis needs to be developed that facilitates balance control and provides adequate adaptability to cope with the continuous changing environment in which subjects ambulate. To enable this we need to understand the importance of balance control to the overall metabolic energy cost. Investigations in which the attention, normally directed towards ensuring balance control, is diverted away from the walking task might provide helpful insights into the effect of conscious control on the walking ability and overall metabolic energy cost. Besides information about the underlying mechanisms and the implications on metabolic energy cost, future research ought to focus on how balance can best be improved in people with a lower limb amputation.
Implications for rehabilitation

Based on the results from this thesis it is made eminent that in order to walk subjects need to have sufficient aerobic exercise capacity. This is most pronounced in those patients whom are deconditioned due to a long period of inactivity, comorbidity and/or a sedentary lifestyle. Therefore, rehabilitation ought to include some form of aerobic training aimed to improve the peak aerobic capacity in these patients.

Due to the paucity of longitudinal studies conducted to date, limited information is available on the efficacy of general rehabilitation programs in this specific patient population. A recent study into the relative effort experienced by stroke, spinal cord injured and amputee subjects during rehabilitation revealed marked differences in aerobic strain between subjects [125]. This underscores the importance of tailor made exercise programs. To ensure continued successful prosthetic ambulation also after rehabilitation it is important to educate subjects on the importance of increasing and maintaining a good level of physical activity [213]. Moreover, in subjects with a lower limb amputation quality of life is to a high degree related to social acceptability [239]. Therefore, the likelihood of regaining and maintaining a physical active life style is largest when physical activity is implemented within a social supportive network [42].

In addition to improving the physical capacity, health care professionals ought to realize that balance control can influence subjects’ walking ability [188]. This study provides some evidence that an improved balance control might potentially reduce the metabolic energy cost while walking. Some weak evidence exists that balance training can be moderately effective in improving the balance of older subjects [112, 243]. With regards to the amputee population positive effects have been found in balance performance during mirror feedback training [101]. To conclude, balance control might be trained using real-life situations during which (like in daily life) conscious control is directed away from the walking task. Another approach is by providing biofeedback through, for example, a mirror or computer screen.


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<th>Reference</th>
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Introduction

Undergoing a lower limb amputation is a life-changing surgery. Unfortunately, a substantial number of people undergo amputation of part of the lower limb. Due to the aging population and a rising number of people diagnosed with diabetic mellitus or peripheral vascular disease this number is expected to grow in the coming years. The ability to walk greatly influences subject's functional independence and quality of life. Not surprisingly, regaining walking ability is one of the primary goals during prosthetic rehabilitation. Within the group of lower limb amputees it is the older person amputated due to vascular deficiency that often experiences the largest difficulties regaining and maintaining walking. In this thesis some of the underlying mechanisms causing these difficulties are investigated using a biophysical approach in which walking is regarded to require an adequate balance between aerobic capacity and aerobic load. The aim of this thesis was to gain insight into the aerobic capacity that subjects possess and relate this to the aerobic load imposed when walking. Additionally, insight is obtained about some of the underlying factors causing the increased aerobic load when walking with a prosthesis.
In Chapter 2 a method to measure peak aerobic capacity in people with a lower limb amputation is described. The peak aerobic capacity is defined as the highest peak oxygen consumption obtained during maximal exercise. In order to measure peak aerobic capacity in a safe and valid manner in people after lower limb amputation requires special consideration of the exercise mode and protocol used. In this Chapter the feasibility and validity of a discontinuous, graded, one-legged, peak exercise test was determined. In total 36 older amputees and 21 healthy controls performed the one-legged exercise test. The healthy controls performed an additional two-legged exercise test. All participants were able to complete the test and blood pressure and electrocardiogram tracings could be adequately monitored to allow for a safe assessment. The exercise test proved to stress the cardio-vascular system to a sufficient extent in both the amputee and control group. With regard to validity it was determined that the one-legged exercise test had both a high construct and a high concurrent validity. To sum, the results of the research presented in Chapter 2 show that the one-legged exercise test provides a feasible and valid assessment of the peak aerobic capacity in older people with a lower limb amputation.

The one-legged exercise test was used in Chapter 3 to determine the magnitude of the peak aerobic capacity and how this was related to the presence, level and cause of the amputation in people with a lower limb amputation. People with an amputation had a 13.1% lower peak aerobic capacity. Differentiation between etiologies revealed that traumatic amputees did not differ to controls, whereas the vascular amputees had a 29.1% lower peak aerobic capacity. Interestingly, no association between peak aerobic capacity and the level of amputation was found. The results corroborated the limited existing evidence and the intuitive notion that the peak aerobic capacity is reduced in people with a vascular amputation.

The lower peak aerobic capacity combined with the increased aerobic load while walking with a prosthesis can result in a high relative aerobic load. In Chapter 4 the relative aerobic load was investigated and the associated effects of level and cause of amputation were determined in the same group of subjects as investigated in Chapter 3. Based on the results, it was concluded that when walking at their preferred walking speed, older vascular amputees walked at a 44.6% higher relative aerobic load than healthy controls. Traumatic amputees compensated for the increased aerobic load by reducing their preferred walking speed to such an extent that the relative aerobic load equaled that of able-bodied controls. They did this even though this entailed walking at a lower walking economy. A data-based model was constructed to determine the potential effect of an increased aerobic capacity on subjects’ walking ability in terms of relative aerobic load, walking speed and walking economy. This model denotes that, for example, in vascular amputees a relatively modest increase in peak aerobic
capacity of 10% can result in 9.1% reduction in relative aerobic load, a 17.3% improvement in walking speed and a 6.8% improvement in walking economy. These results denote that aerobic training must indeed be considered an essential component of prosthetic rehabilitation.

Whereas in Chapters 2, 3 and 4 the focus was on determining the peak aerobic capacity and the associated relative aerobic load, in Chapters 5 and 6 information is obtained about potential factors causing the increased aerobic load while walking with a prosthesis. In Chapter 5 the mechanical energy cost while walking was investigated using a dynamic walking model. A total of 15 subjects walked both with an energy storage and return (ESAR) prosthetic foot and a solid ankle cushioned heel (SACH) prosthetic foot. Both prosthetic feet were compared with regard to the required step-to-step transition cost while walking. The ESAR foot required the least mechanical work during the step-to-step transition. This was explained by an increased push-off power and a larger forward progression of the center of pressure during single stance in the ESAR foot. Interestingly, previous studies comparing these two prosthetic feet found no convincing evidence that the ESAR feet also required less metabolic cost while walking. To sum, though the ESAR foot is able to reduce the step-to-step transition cost other factors outside those related to step-to-step transition cost influence the overall metabolic energy cost while walking with a prosthesis.

One of the possible factors influencing the overall metabolic energy cost while walking might be an altered balance control. Differentiating between the mechanical energy required for the walking movement and that associated with balance control is a challenge. In Chapter 6 we tried to do this by having nine able-bodied healthy controls walk at an instrumented treadmill while following an projected step pattern composed of either their averaged step pattern (periodic trial), or a step pattern that was an exact copy, including variability of their unconstrained walking trial (variable trial). The novelty of this approach is that we were able to investigate whether balance control strategies can contribute significantly to the overall metabolic energy cost, independent from alteration in the global gait characteristics. Results showed that walking an enforced gait pattern resulted in a metabolic oxygen increase of 8% in the periodic and 13% in the variable trial. It was postulated that the increased metabolic energy cost is related to increased preparatory muscle activation and a more active ankle strategy to control for lateral balance. This Chapter shows that metabolic energy cost can be associated with alterations in balance control strategies that do not alter global gait characteristics.
Conclusion

The overarching aim of this thesis was to enhance our understanding of some of the factors that influence the ability to regain and maintain walking after unilateral lower limb amputation. Based on the results presented in this thesis we can deduce that the peak aerobic capacity plays an important role in subjects’ walking ability. This peak aerobic capacity is reduced in vascular amputees, which makes walking a strenuous activity for these patients. The work presented in this thesis shows that relatively small improvements in peak aerobic capacity could potentially lead to significant and clinically relevant improvements in subject’s walking ability. Furthermore, this thesis showed that the development of prosthetic feet with adequate and correctly timed push-off power together with an optimal roll-over shape can decrease the step-to-step transition cost. However, the energy required for balance control seems an important factor contributing to the overall metabolic energy cost while walking.

Implication for rehabilitation

In this thesis we postulate that future research ought to involve longitudinal studies with a heterogeneous group of amputees in order to investigate the efficacy of different training protocols and to determine what effect an increased peak aerobic capacity has on subject’s walking ability. Additionally, simple dynamic models can be used to gain helpful insight into the mechanical work on the center of mass while walking on different state-of-the-art prosthesis. Though insightful, future studies must be designed that face the challenge of disentangling the metabolic energy cost associated with the mechanical work needed to perform the walking movement and that associated with preserving balance. Information about these underlying causes for the increased metabolic energy cost while walking with a prosthesis will help and aid the development of more efficient and functional prostheses. Moreover, health care professionals must realize that both aerobic and balance training can substantially improve subjects’ walking ability, and therefore, ought to be an integrated part of prosthetic rehabilitation.
Optimale loopvaardigheid met een prothese
Balanceren tussen capaciteit en belasting

Samenvatting
Optimale loopvaardigheid met een prothese
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Samenvatting
Introductie

Het ondergaan van een beenamputatie is een drastische chirurgische ingreep, die grote gevolgen heeft voor het functioneren. Jaarlijks ondergaan een aanzienlijk aantal mensen een amputatie van een deel van het been. Naast trauma, is vasculaire deficiëntie, vaak als gevolg van diabetes, de belangrijkste reden voor een beenamputatie. Omdat de levensverwachting in Nederland stijgt, en met veroudering de kans op diabetes en dus vasculaire deficiëntie toeneemt, zal het aantal mensen met een beenamputatie naar verwachting de komende jaren verder stijgen.

Mensen, die na een amputatie in staat zijn om te lopen met een prothese, zijn functioneel onafhankelijker, en hebben een hogere kwaliteit van leven dan mensen die in een rolstoel belanden. Het is daarom niet verrassend dat het herwinnen van de loopvaardigheid één van de voornaamste doelen is tijdens de revalidatie na een beenamputatie.

Het zijn de oudere mensen die een amputatie hebben ondergaan als gevolg van vasculaire deficiëntie, die de meeste problemen ondervinden in het herwinnen, dan wel handhaven van de loopvaardigheid. In dit proefschrift is vanuit een biofysische benadering een aantal van de mogelijke onderliggende oorzakelijke mechanismen voor de problemen bij deze categorie mensen onderzocht. In de gehanteerde biofysische benadering wordt er vanuit gegaan dat een goede loopvaardigheid een adequate balans vereist tussen de aerobe capaciteit en de aerobe belasting. Het doel van dit proefschrift was om inzicht te verkrijgen in de aerobe capaciteit van mensen die lopen met een beenprothese, en om deze aerobe capaciteit vervolgens te relateren aan de aerobe belasting die het lopen met een prothese vergt. Daarnaast geeft dit proefschrift inzicht in de onderliggende mechanismen die verantwoordelijk zijn voor de verhoogde aerobe belasting tijdens het lopen met een beenprothese. In de hierop volgende paragrafen wordt een samenvatting gegeven van de verschillende studies die in het kader van dit proefschrift zijn uitgevoerd. De samenvatting eindigt met een aantal concrete aanbevelingen voor de revalidatiepraktijk.
In hoofdstuk 2 van dit proefschrift wordt een methode beschreven waarmee de piek aerobe capaciteit van mensen met een beenamputatie bepaald kan worden. De piek aerobe capaciteit is gedefinieerd als de maximale hoeveelheid zuurstof die verbrijkt wordt tijdens een maximale inspanning. Het op een veilige en valide manier meten van de piek aerobe capaciteit bij mensen met een beenamputatie vergt een weloverwogen keuze wat betreft de te gebruiken inspanningstest. In hoofdstuk 2 wordt de uitvoerbaarheid en de validiteit van een discontinue, gradeerde, éénbenige, piek-inspanningstest beschreven. In totaal hebben 36 personen, allen tussen de 50 en 75 jaar, die een beenamputatie hebben ondergaan en 21 gezonde personen, die de controlegroep vormden, de éénbenige piek-inspanningstest uitgevoerd. Alle deelnemers waren in staat de éénbenige piek-inspanningstest te volbrengen. Zowel de bloeddruk als het elektrocardiogram kon accuraat worden afgenomen tijdens de test. De inspanningstest bleek zowel in de amputatie- als in de controlegroep het cardiovasculaire systeem in voldoende mate te activeren. Daarnaast was zowel de construct en concurrent validiteit hoog bij de éénbenige inspanningstest. De conclusie van het onderzoek, beschreven in hoofdstuk 2, is dan ook dat de éénbenige fietstest een uitvoerbare en valide methode is om de piek-aerobe capaciteit te bepalen van oudere mensen die een beenamputatie hebben ondergaan.

De éénbenige inspanningstest, zoals beschreven in hoofdstuk 2, is in hoofdstuk 3 toegepast om te bepalen of de piek aerobe capaciteit gerelateerd is aan het wel of niet hebben ondergaan van een amputatie, de hoogte van de amputatie of de oorzaak van de amputatie. Dezelfde mensen, beschreven in de vorige paragraaf, participeerde ook in dit onderzoek. Uit de resultaten blijkt dat de proefpersonen die een beenamputatie hebben ondergaan, gemiddeld een 13,1% lagere piek aerobe capaciteit hadden dan de controlegroep zonder beenamputatie. Wanneer onderscheid gemaakt werd naar de oorzaak van de amputatie, bleek dat de proefpersonen die een amputatie hebben ondergaan als gevolg van vasculaire deficiëntie, gemiddeld een 29,1% lagere piek aerobe capaciteit hadden dan de proefpersonen die vanwege een trauma een amputatie hebben ondergaan. De piek aerobe capaciteit van de door trauma geamputeerde groep verschilde niet met die van de controlegroep. Interessant is dat er geen relatie is gevonden tussen piek aerobe capaciteit en de hoogte van de amputatie. De resultaten van deze studie onderschrijven de beperkte bestaande bewijslast dat de piek aerobe capaciteit van personen met een amputatie door vasculaire deficiëntie lager is dan die van personen zonder beenamputatie.

Een lagere piek aerobe capaciteit, gecombineerd met de verhoogde aerobe belasting tijdens het lopen met een beenprothese, kan resulteren in een dermate verhoogde relatieve aerobe belasting dat de loopvaardigheid wordt beïnvloed. De relatieve aerobe belasting wordt bepaald door de aerobe belasting te delen door de aerobe capaciteit. In hoofdstuk 4 is de relatieve aerobe belasting
tijdens het lopen met een beenprothese bepaald. Vervolgens is er gekeken of de relatieve aerobe belasting gerelateerd is aan de hoogte, dan wel de oorzaak van de amputatie. De resultaten laten zien dat wanneer personen die een amputatie hebben ondergaan door vasculaire deficiëntie op hun voorkeursnelheid lopen, zij een relatieve aerobe belasting hebben welke gemiddeld 44,6% hoger is dan die van personen zonder een beenprothese. De personen met een aan trauma gerelateerde amputatie compenseerden voor de verhoogde aerobe belasting door hun loopsnelheid zodanig te verlagen dat de relatieve aerobe belasting gelijk werd aan die van de controlegroep. Dit deden ze ondanks dat de lagere loopsnelheid resulteerde in een lagere economie van het lopen.

Om te bepalen wat het mogelijke effect zou kunnen zijn van een verhoogde aerobe capaciteit op de loopvaardigheid in termen van relatieve aerobe belasting, loopsnelheid en economie van het lopen, is in hoofdstuk 4 een op data gebaseerd model beschreven. Dit model laat zien dat bij personen met een amputatie als gevolg van vasculaire deficiëntie, een relatief geringe verhoging van de piek aerobe capaciteit met 10% kan leiden tot een reductie in de relatieve belasting van 9,1%, een verhoging van de loopsnelheid met 17,3% en een verbetering in de economie van het lopen van 6,8%. Deze resultaten laten zien dat bij personen met een amputatie door vasculaire deficiëntie, aerobe training een essentieel onderdeel zou moeten zijn van het revalidatietraject als het gaat om het herwinnen van de loopvaardigheid.

Waar in hoofdstuk 2,3 en 4 de focus van het onderzoek lag op de piek aerobe capaciteit en de daaraan gerelateerde relatieve aerobe belasting, zijn in hoofdstuk 5 en 6 de mogelijke factoren onderzocht die bijdragen aan de verhoogde aerobe belasting tijdens het lopen met een beenprothese. In hoofdstuk 5 worden de mechanische kosten van het lopen met een beenprothese onderzocht door gebruik te maken van een dynamisch loopmodel. In totaal liepen 15 mensen met zowel een prothesevoet die energie kan opslaan en weer kan teruggeven (Energy Storage and Return; ESAR), als ook met een conventionele prothesevoet (Solid Ankle Cushioned Heel; SACH). Met behulp van het model kan de mechanische kosten die nodig zijn tijdens de dubbel support fase (de stap-naar-stap transitiekosten) worden berekend. Uit hoofdstuk 5 blijkt dat er met de ESAR-voet minder mechanische arbeid nodig is voor de stap-naar-stap transitie dan met de SACH-voet. Dit wordt verklaard door het feit dat er met de ESAR meer afzetkracht gegenereerd kan worden, en omdat er een meer optimale afwikkeling plaatsvindt. Desondanks biedt eerder onderzoek onvoldoende bewijs dat de ESAR voet ook daadwerkelijk de aerobe belasting tijdens het lopen reduceert ten opzichte van de conventionele SACH voet. Resumerend kan worden gesteld dat ondanks dat de ESAR-prothesevoet in staat is de mechanische arbeid van de stap-naar-stap transitie te reduceren, er andere factoren zijn die de aerobe belasting tijdens het lopen met de prothesevoet beïnvloeden.
Eén andere mogelijke factor, die het aerobe belasting tijdens het lopen met een beenprothese kan beïnvloeden, is de energie die nodig is voor de balanscontrole. Door een gebrek aan proprioceptieve informatie tijdens het lopen met een beenprothese, zijn mensen aangewezen op andere wellicht minder efficiënte mechanismen om de stabiliteit tijdens het lopen te waarborgen. Tot op heden is het lastig gebleken om de mechanische energie die nodig is voor loopbeweging te onderscheiden van de energie die nodig is om de balans te handhaven. In hoofdstuk 6 is getracht om inzicht te krijgen in het metabole energieverbruik voor balanscontrole door negen gezonde personen (27.6 jaar [SD = 9.9]) te laten lopen op een loopband waarbij ze een geprojecteerd looppatroon moesten volgen. Normaliter wordt een balansverstoring in de ene stap opgevangen door een corrigerende voetplaatsing tijdens de volgende stap. Maar wanneer de voetpositie wordt opgelegd worden mensen gedwongen om alternatieve, en wellicht minder efficiënte, balanscorrecties uit te voeren (bv. door activering van enkel musculatuur). Het geprojecteerde looppatroon welke de deelnemers moesten volgen, representeerde een gemiddelde van het eigen looppatroon (regelmatige conditie), of het was een exacte kopie was hun eigen looppatroon, dus inclusief de inherente variatie normaal aanwezig tijdens het lopen (variabele conditie). Door deze onderzoekskopzet was het mogelijk om te bepalen of een verandering in de balanscontrole leidt tot een significante verandering in het totale metabole energieverbruik, onafhankelijk van veranderingen in het globale looppatroon. De resultaten van het onderzoek laten zien dat tijdens het lopen op een geprojecteerd looppatroon het metabole energieverbruik met 8% en 13% toenam, in respectievelijk de regelmatige en de variabele conditie, vergeleken met het lopen zonder een opgelegd looppatroon. Mogelijke oorzaken voor dit verhoogde metabole energieverbruik zouden een verhoogde voorbereidende spieractiviteit, en een grotere bijdrage van de spieren rond het enkelgewricht voor het handhaven van vooral de laterale stabiliteit kunnen zijn. Hoofdstuk 6 laat zien dat een verhoging van het metabole energieverbruik tijdens het lopen met een beenprothese mogelijk veroorzaakt wordt door veranderingen in de balanscontrole, zonder dat dit direct invloed heeft op het globale looppatroon.

Conclusie

Het overkoepelende streven van dit proefschrift was om inzicht te krijgen in een aantal van de factoren die van invloed kunnen zijn op het herwinnen en behouden van de loopvaardigheid na een beenamputatie. Op basis van de resultaten, gepresenteerd in dit proefschrift, kunnen we stellen dat de piek aerobe capaciteit van personen die een beenamputatie hebben ondergaan door vasculaire deficiëntie, verlaagd is. Voor deze mensen is lopen een inspannende activiteit. Het gepresenteerde werk laat zien dat een relatief kleine verbetering
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in de piek aerobe capaciteit kan leiden tot significante en klinisch relevante veranderingen in de loopvaardigheid van personen met een beenprothese. Bovendien laat dit proefschrift zien dat een prothesevoet, die een adequaat getimede afzetkracht kan genereren en die een optimale afwikkeling van de voet geeft tijdens de standfase, de mechanische energie tijdens de stap-naar-stap transitie kan verlagen. Echter, de mechanische energetische kosten voor de stap-naar-stap transitie kunnen niet het verhoogde metabole energieverbruik van het lopen met een beenprothese verklaren. Mogelijk dragen ook de energetische kosten voor de balanscontrole bij aan het verhoogde energieverbruik tijdens het lopen met een beenprothese.

Implicaties voor de revalidatie

In dit proefschrift wordt de aanbeveling gedaan dat toekomstig onderzoek zich zou moeten richten op longitudinale studies met een heterogene groep van mensen met een beenprothese. Met behulp van deze longitudinale studies kan worden onderzocht wat de effectiviteit is van verschillende trainingsprogramma’s op de loopvaardigheid bij verschillende subgroepen binnen de populatie van mensen met een beenprothese. Daarnaast kunnen simpele dynamische modellen inzicht verschaffen in de mechanische arbeid tijdens het lopen met verschillende prothesevoeten. Voor toekomstig onderzoek ligt er een grote uitdaging om de energetische kosten voor de loopbeweging en de kosten die gerelateerd zijn aan de balanscontrole, van elkaar te scheiden. Informatie over de onderliggende oorzaken van de verhoogde energetische kosten tijdens het lopen met een beenprothese vormt de basis voor verdere ontwikkeling van meer efficiënte en functionele prothesen. Binnen de revalidatie na een beenamputatie is het van belang dat men zich realiseert dat zowel aerobe training als balanstraining een substantiële bijdrage kan leveren aan het optimaliseren van de loopvaardigheid. Om die reden zouden beide facetten dan ook een geïntegreerd onderdeel van de revalidatie moeten zijn.
Samenvatting
Dankwoord
No duty is more urgent than that of returning thanks.

- James Allen -
Oke, dit wordt een makkie! Ik mag een stuk schrijven waarbij ik me niet hoeft te houden aan enige consensus of wetenschappelijke regels. Dan breek ik gelijk maar met een gewoonte en start ik met het bedanken van alle mensen, die ik hieronder niet expliciet zal noemen. Bedankt, en sorry dat ik je niet bij naam noem. Je moet maar zo denken: nu ben je (soort van) alsnog als eerste genoemd.

Maar eigenlijk is er natuurlijk maar één iemand die ik als eerste wil en moet noemen. Han, zoals het promotietraject voor mij een sprong in het diepe was, heb jij je ook begeven op onbekend terrein. Ik ben je eerste promovendus en enorm trots dat ik samen met jou dit project heb mogen vormgeven. Jouw expertise als wetenschapper staat buiten kijf. Ik ben enorm dankbaar dat ik uit jouw bron van kennis heb mogen tappen. Je hebt, door je vertrouwen, je ongekend positieve instelling, enthousiasme en belangstelling, mij als wetenschapper en als persoon de ruimte gegeven om te groeien. Ik had me geen betere begeleider kunnen wensen. En je hebt inderdaad gelijk gekregen met dat vertrouwen! Mijn dank is enorm. Al jouw toekomstige promovendi mogen zich in de handen wrijven met jou als begeleider.

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En ja, ik weet dat dit het hoofdstuk is dat het meest gelezen wordt (dat is me vaak genoeg verteld tijdens de lunch [dank je Jeroen]). Vandaar dat ik dankbaar ben voor de hulp van de (door mij zo benoemde) expert in het becommentariëren van Nederlandse teksten a.k.a. Kirsten. Hierdoor maak ik in ieder geval niet de vergissing, die mijn co-promotor ooit beging! En wie denkt dat ik goed ben in tekenen, die kent mij niet goed genoeg; voor alle illustraties ben ik Dave en Theo enorm dankbaar: Theo de voorkant is geweldig! En Theo mijn oprechte spijt voor de stress die ik je bezorgd heb! Theo, maar ook Dennis, jullie zijn onmisbaar geweest in de afrondingsfase. En dat is vier ;) !

Zoals elke promovendus kan beamen, zijn er pieken en dalen tijdens het promotietraject. Zo stijgt soms de belasting terwijl de capaciteit daalt; m.a.w. de ballon komt steeds meer op spanning te staan.\(^1\) Op dit soort momenten kon ik altijd terugvallen op mijn vrienden. Iris, het leven is een achtbaan. Ik ben enorm dankbaar dat je samen met mij de rit wil maken, je bent een topvriendin! Trienke, heerlijk die nuchterheid van jouw kant en ik koester de (soms inhoudelijke) gesprekken en discussies samen. Meiden, bedankt dat jullie mijn paranimf willen zijn. En Katja, Achterhoeker, je bent mijn “oudste” vriendin en nu we zo dicht bij

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\(^1\) Snap je de vergelijking met de ballon niet? Dat krijg je ervan als je alleen het dankwoord leest!
elkaar wonen, hoop ik dat we steeds vaker de deur bij elkaar gaan platlopen! Ik
prijs me gelukkig met een groep vrienden waarmee ik kan sporten, feesten en
praten. Het liefst noem ik iedereen persoonlijk maar ja, dan doe ik de gewonnen
woorden (zie boven) weer teniet. Laat ik gewoon bij deze beloven dat we snel
weer eens ouderwets gezellig afspreken!

James, these words might be the most difficult ones to write of the whole thesis.
The completion of this thesis is to a large part dependent on your support and
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Hmm... zo makkelijk was het toch niet om dit te schrijven...
About the author
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Daphne Wezenberg (1982) was born in Doetinchem, the Netherlands. In 2009 Daphne graduated from the Windesheim University of Applied Sciences after attaining her Bachelor degree as a Physical Education Teacher. She continued her studies at the VU University Amsterdam where she attained her Masters degree in Human Movement Sciences in 2006. In the same year she finished the Teaching Training Program. In 2006 she started working at the University of Applied Sciences in Leiden at the department of Physiotherapy. In 2008 she returned to the Faculty of Human Movement Sciences to start her Ph.D. project on the relationship between walking ability and physical capacity in people with a lower limb amputation. This project was a collaboration between the VU University and Heliomare Rehabilitation. From September 2012 she is combining her research activities at the VU University with a position at the Hague University of Applied Sciences at the department of Human Movement Technology. In the future Daphne aims to conduct future research within the field of prosthetic ambulation and hopes to inspire both students and professionals to engage, and explore the field of rehabilitation and movement science.
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