Movement and balance control in lower limb amputees

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Stellingen

De afwezigheid van een actieve knie- en enkelfunctie in het prothesebeen van patiënten met een amputatie resulteert in het gebruik van diverse aanpassingsstrategieën om complexe motorische taken uit te voeren. (dit proefschrift)

De functionele mogelijkheid van een protheseknie om te flecteren wordt bij het stappen over een obstakel onvoldoende gebruikt door patiënten met een transfemorale amputatie. (dit proefschrift)

De keuze welk been wordt gebruikt als leidend been is een belangrijke aanpassingsstrategie voor patiënten met een beenamputatie maar is in de praktijk niet altijd bruikbaar doordat de afstand tot een object of locatie de keuze mede bepaald. (dit proefschrift)

Om te stoppen met lopen zet een amputatiepatiënt bij voorkeur zijn beste beentje voor. (dit proefschrift)

Aangezien het niet-aangedane been een belangrijke rol speelt in de uitvoering van aanpassingsstrategieën, is het van belang om de spierkracht en de spierbeheersing in dit been te trainen. (dit proefschrift)

Microprocessorgestuurde en/of adapterende prothesemechanismen zouden de uitvoering van complexe motorische taken kunnen verbeteren. Het huidige probleem is dat met de hedendaagse sensoren en actuatoren het (nog) niet mogelijk is om de dynamische eigenschappen van een protheseknie of prothesevoet binnen één stap aan te passen. (dit proefschrift)

Niet een symmetrisch looppatroon moet het doel zijn in de revalidatie van patiënten met een beenprothese maar optimaal functioneel herstel.

Het is niet eenvoudig om in een enkel moment te begrijpen wat een enkelmoment inhoudt.

Het meest gegeven antwoord in de wetenschap is dat er meer onderzoek nodig is om de vraag te beantwoorden.

Naar verluidt zijn er kunstmatige ledematen uitgevonden, die door de hersenen geleid kunnen worden. Misschien zal er ooit een tijd aanbreken dat ook mensen met gezonde ledematen zich door hun hersenen zullen laten leiden. (Gabriel Laub)

Met zorgen over morgen verpruts je de dag van vandaag.

Je leert om te leven en leeft om te leren.

Het feit dat vrouwen jaarlijks meer dan 12000 calorieën verbranden door te shoppen rechtvaardigt een overvolle klerenkast.

Aline Vrieling, 28 april 2009

Cmb.

RIJKSUNIVERSITEIT GRONINGEN

Movement and balance control in lower limb amputees

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aan de Rijksuniversiteit Groningen
op gezag van de
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Chapter I

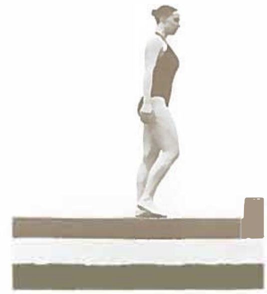
Introduction

Background of the thesis

In the Netherlands about 50% of all amputees are provided with a prosthetic device[1], and in this group regaining the ability of standing and walking with a prosthesis is an important goal in rehabilitation. As a result of the loss of somatosensory input and absent musculature and joint(s), a prosthetic limb does not possess the same functional properties as a non-affected limb. In most prosthetic knees and feet active muscle power and control are lacking, range of motion is limited and afferent sensory information is not available. Moreover, an asymmetry in body mass occurs since the prosthetic limb weighs less than the non-affected limb.

These limitations in function cause difficulties in the control of balance and movements. From amputees, it is known that they fall more often than healthy persons do.[2] The gait pattern of amputees is influenced by many factors, such as pain at the stump, level of amputation, physical capacity, fitting of the socket, alignment of the prosthesis and the properties of the prosthetic knee and ankle. The major part of the gait abnormalities in transtibial amputees are caused by the stiffness of the prosthetic foot and the inability of active dorsal and plantar flexion, whereas in transfemoral amputees the lack of active knee function adds to the asymmetric gait pattern.

In order to stand and walk with a prosthetic limb it is necessary for an amputee to compensate for the functional limitations by using adjustments strategies. The central nervous system generates motor programs for the execution of motor tasks.[3] Adaptation of balance and movement control after an amputation is a learning process. The prosthetic device has to be integrated in the motor cortex by changing the existing motor strategies and by developing new motor strategies.[4] Adjustment strategies can be originated in the locomotor system, such as the non-affected limb or the residual part of the amputated limb. For instance, in level walking the reduced push off power of the prosthetic limb is partially compensated for by an increase in muscle activity in the hip extensors and flexors in the prosthetic limb and the hip extensors and ankle plantar flexors in the non-affected limb.[5-10] An important adjustment strategy in standing is the increased loading of the non-affected limb compared to the prosthetic limb.[11-14]



The ability to maintain balance and to walk efficiently also depends on the integrity of the cognitive, visual, vestibular, and somatosensory system.[15, 16] Deficiencies in the somatosensory system after an amputation can be compensated for by increasing the role of the other control systems. In amputees standing balance is decreased compared to able-bodied subjects, especially when vision is deprived and while performing a dual task.[17-24] The dependency on the visual and conscious control of balance and movements is most pronounced in the early period of rehabilitation when the adjustments strategies are not yet integrated in an automated motor program.[12, 20, 21, 25]

Safe and independent locomotion with a prosthetic limb requires the ability to perform motor tasks in complex circumstances and environments. The majority of research into motor tasks in amputees has focused on balance control in bipedal quiet standing and on unimpeded steady-state level walking. Such research is not in accordance with normal daily activities and can only give limited insight in the functional limitations and adjustment strategies of amputees.[26] Increasing the demands on the motor system by complicating the motor task is one option to mimic the daily life situation more closely.[16] Gait studies in amputees have reported on gait initiation[27-30], gait termination[31], obstacle crossing[32, 33], adding an attention demanding task[34-36], running[37-39], stepping up and down a level[40], stair ascent and descent[41-44], declining a slope[45], and locomotion on uneven terrain or an obstacle course.[46, 47] As mentioned earlier, balance tasks in amputees were often complicated by adding a dual task or by depriving vision, which challenges respectively the cognitive and perceptual system. Other studies have challenged amputee's balance by internal perturbations that arise from voluntary movements such as standing in single-stance on the prosthetic limb[48], raising one leg sideways[49-51], leaning forward and to one side[11, 52], performing shoulder movements[4], catching a load[4] or standing on a mediolateral and anteroposterior pivoting stabilometer.[53]

Aim and outline of the thesis

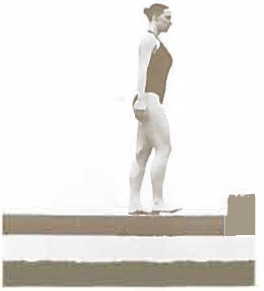
In the Center for Rehabilitation of the University Medical Center Groningen (UMCG) research into amputation, prosthetics and orthotics is an important feature. The present research project is the result of collaboration between the Center for Rehabilitation UMCG and the Center for Human Movement Sciences of the University of Groningen.

The limitations in function in the prosthetic limb present a difficult challenge to amputees in being able to continue with their daily activities. Amputees have to make adjustments in the gait pattern to manoeuvre safely in all complex situations and environments. The main aim of this study is to determine the functional limitations and the applied adjustment strategies in amputees when performing difficult motor tasks. The studied motor tasks were standing on a movable floor, obstacle crossing, gait initiation and termination and walking up and down a slope, which are commonly performed tasks in daily life activities. New insights in the biomechanical characteristics of complex motor tasks in amputees and into the adjustment strategies that are involved in the recovery of these tasks may help to improve rehabilitation methods and prosthetic design and to identify amputees who are at risk for falls.

The following research questions will be answered in this thesis:

1. Which are the differences in functional limitations and adjustment strategies between experienced amputees and able-bodied subjects while performing complex motor tasks?
2. In what way do these limitations in function and adjustment strategies influence the gait pattern and balance control in amputees while performing complex motor tasks?
3. Which changes appear in the limitations in function and adjustment strategies in obstacle crossing and gait initiation and termination during the rehabilitation process of patients with a recent lower limb amputation?

In the first part of the thesis (chapters 2-6) we focus on subjects who were amputated in the past and were experienced prosthetic users. Chapter 2 focuses on obstacle crossing in amputees. Flexion of the knee is an important strategy to step safely over an obstacle. The restricted knee flexion in amputees, especially after a transfemoral amputation, may lead to gait adjustments in stepping over an obstacle. In chapter 3 and 4 we present the results of respectively gait initiation and gait termination. In these motor tasks the ankle musculature has an important role in shifting the centre of pressure and regulating the ground reaction force. The lack of ankle strategy may result in an impaired ability to start and stop walking. In chapter 5 we study up- and downhill walking. Because an intact knee function is important to adjust to a slope gradient, amputees who use a prosthetic knee may have to use other adjustment strategies in uphill and downhill walking. In chapter 6 balance control was described while swaying in an anteroposterior direction on a moving platform, with and without blindfolding and a dual task. Due to the absence of



an active ankle strategy amputees may have difficulties to adjust to the perturbations in balance, especially when vision is deprived and attention distracted.

In the second part of the thesis (chapter 7) we describe a prospective study on patients with a recent lower limb amputation in order to gain more insight in the adaptation process of amputees to difficult motor tasks. During the rehabilitation process muscle strength, coordination and control are improved and amputees become more familiar with the prosthetic device. As a result the limitations in function in the prosthetic limb may become less restrictive and amputees may use different adjustment strategies to perform the motor tasks when the rehabilitation process progresses. In Chapter 7 we describe the adjustments in gait characteristics of obstacle crossing, gait initiation and gait termination that occur in subjects after a recent lower limb amputation.

In the final chapter (chapter 8) we give an overview of the limitations in function and adjustment strategies in the different motor tasks that were performed for this study. Furthermore, we discuss the clinical implications of this study, give advice on training methods and prosthetic design and provide recommendations for further research.

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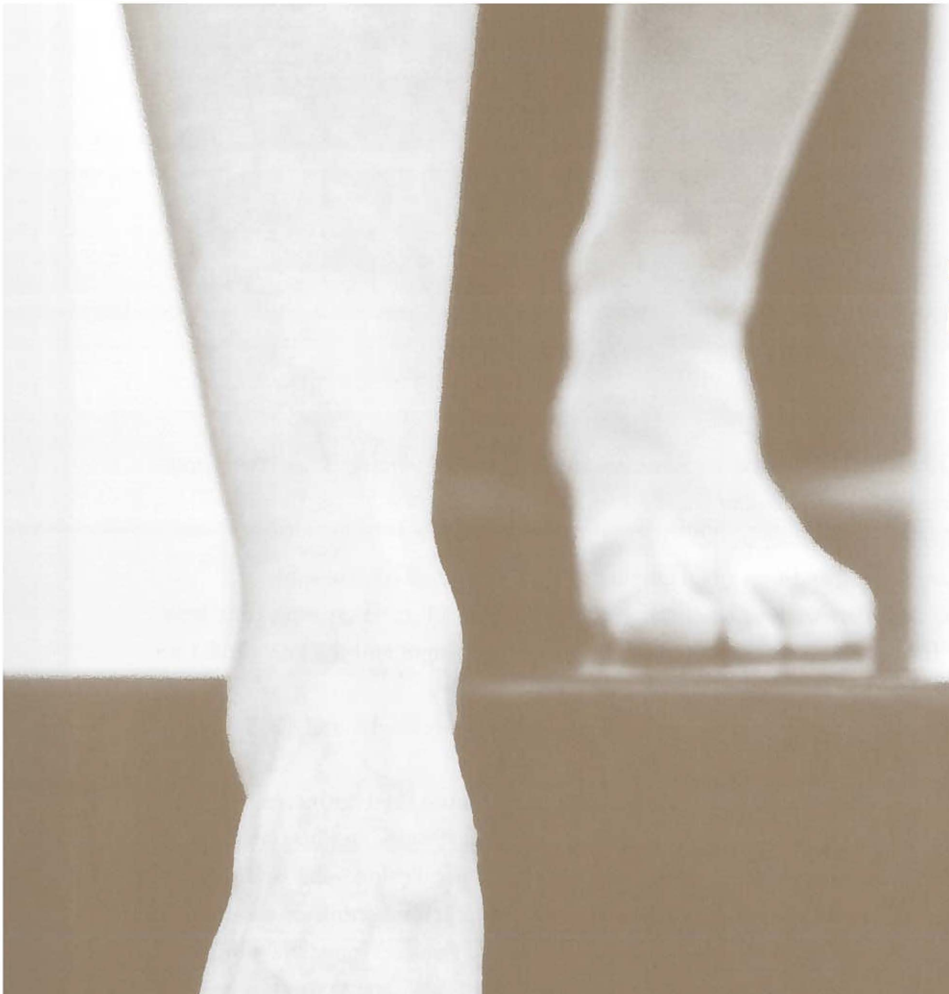

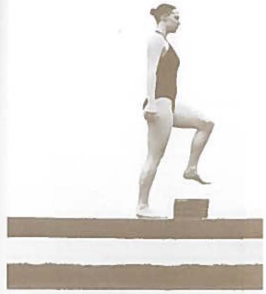

Chapter

2

Obstacle crossing in lower limb amputees

Aline Vrieling, Helco van Keeken, Tanneke Schoppen, Bert Otten, Jan Halbertsma, At Hof, Klaas Postema

Gait & Posture, Volume 26, Issue 4, October 2007, Pages 587-594

Abstract

Objective: To study limitations in function and adjustment strategies in lower limb amputees during obstacle crossing.
Design: Observational cohort study.
Subjects: Transfemoral and transtibial amputees and able-bodied control subjects.
Methods: In a motion analysis laboratory unimpeded and obstacle crossing runs were performed. The subjects stepped over an obstacle of 0.1 m height and thickness and 1 m width.
Outcome measures: Gait velocity, hip, knee and ankle joint angles and leading limb preference.
Results: Whereas able-bodied and transtibial subjects demonstrated an increase in knee flexion during obstacle crossing compared to unimpeded walking, in transfemoral amputees the opposite was seen, namely a decrease in knee flexion. The lack of knee strategy in transfemoral amputees was compensated by circumduction at the hip on the prosthetic side and plantar flexion of the non-affected ankle. Transtibial amputees preferred to cross the obstacle with the prosthetic limb first, while transfemoral amputees preferred the non-affected limb.
Conclusion: The different leading limb strategy in transfemoral and transtibial amputees could be explained by the restricted flexion and propulsion properties of the prosthetic knee. Training of obstacle crossing tasks during rehabilitation and improvement of prosthetic design may contribute to safe obstacle crossing.

Introduction

For control and execution of locomotion, it is essential that a person can adjust the gait pattern to the environment and to diverse situations.[1-3] Obstacle crossing is one of many complex tasks associated with mobility in everyday life.[4] Stepping safely over an obstacle requires skills of both lower limbs; the swing foot has to clear the obstacle to avoid tripping, while the stance limb has to establish a stable base of support.[5,6] Studies in able-bodied (AB) persons show that foot clearance during obstacle crossing is achieved by increasing joint flexion of the swing limb.[1,7-10] Flexion of the knee is the most important motor strategy and is achieved by an increase in activation of knee flexors.[1,5,7,8,11,12] Secondarily, this knee strategy leads to an increase in hip flexion.

An amputation of a lower limb results in a deficiency in sensory input and the absence of muscles and joint(s). A person with a lower limb prosthesis has to adapt to a mechanical device to become functionally independent again.[13] A prosthesis does not posses the same functional abilities as the non-affected limb. In transtibial (TT) amputees muscles in the posterior part of the knee that participate in knee flexion, such as the gastrocnemius muscle, are affected. Besides, knee function may be limited due to the posterior shell of the socket.[14,15] In transfemoral (TF) amputees the hamstrings muscles are divided and knee flexion is the result of a passive swing motion, which can not be actively controlled. Due to the absence of ankle dorsal flexors and the stiffness of the prosthetic foot, amputees are not able to shorten the swing limb by ankle dorsiflexion. In TT amputees a decrease in knee flexion and ankle dorsiflexion of the prosthetic limb in swing was shown in obstacle crossing.[14] So far, no studies have been published concerning obstacle crossing in TF amputees. The first objective of this study was to assess the limitations in function of the prosthetic limb in obstacle crossing. We hypothesized that swing limb flexion of the knee and ankle would be reduced in the prosthetic limb.

The human central nervous system contains motor strategies for the performance of motor tasks.[16] After a lower limb amputation the original motor strategies used for obstacle crossing have to be modified or new strategies have to be learned.[11,17] The second objective of this study was to determine which adjustment strategies amputees are using to compensate for the above-mentioned limitations in function. We hypothesized three possible adjustment strategies: (1) an increase in flexion in a more proximal located non-affected joint in the prosthetic limb during swing may contribute to shortening of this limb, (2) an increase in the length of the non-affected limb during stance by straightening of the hip or knee, or plantar flexion of the ankle may enhance clearance of the prosthetic foot, and (3) the choice to cross the obstacle first with a specific limb may ease obstacle crossing, since the movement trajectories of the leading and trailing limb are different.[7, 18]

Methods

Subjects

Subjects were selected via a prosthetic workshop in the north of the Netherlands. Registered clients with a unilateral TF or TT amputation at least eight months earlier and due to trauma, vascular disease or tumour were asked to participate in the study. Subjects had to use the prosthesis on a daily basis and be able to walk without walking aids more than 50 m. Subjects were excluded if they had any medical conditions that could affect their mobility or balance, such as neurological, orthopaedic or rheumatic disorders, the use of antipsychotic drugs, antidepressants or tranquillisers, cognitive problems, severely impaired vision or reduced sensation of the non-affected limb. In addition, subjects with problems of the amputation limb or prosthesis were excluded. A control group of AB subjects was also selected. They were recruited through advertisements at the regional blood bank and hospital, as well as the local television and radio. Exclusion criteria for AB subjects were: neurological, orthopaedic or rheumatic disorders, reduced sensation in the lower limbs and the use of antipsychotic drugs, antidepressants or tranquillisers.

Eight TF amputees, twelve TT amputees and ten AB subjects fulfilled the selection criteria and agreed to participate in the study. Informed consent was obtained from all subjects before testing and the Medical Ethics Committee approved the study protocol. The amputees used different types of prosthetic feet and knees. All TF amputees used freely mobile prosthetic knees. Characteristics of the study groups are provided in Table 1. No statistically significant differences in age, body weight (including prosthesis) and height were found between the groups. TF and TT amputees were comparable for the cause of the amputation and the time passed since the amputation.

Table 1. Subject characteristics.

	TF (n = 8)	TT (n = 12)	AB (n = 10)	F	p- value
Gender (M / F)	7 / 1	10 / 2	9 / 1		
Age (years)	46.4 ± 14.7	49.6 ± 11.6	45.2 ± 9.4	0.40	0.67
Weight (kg)	82.5 ± 11.9	84.2 ± 8.2	86.5 ± 9.1	0.40	0.68
Height (cm)	183.3 ± 6.1	180.9 ± 8.5	184.4 ± 6.7	0.65	0.53
Time (months)	185.5 ± 163.0	207.8 ± 169.4	-	0.09	0.77
Prosthetic foot	2 C-walk[1], 2 SACH[1], 4 Multiflex[2]	4 C-walk[1], 3 SACH[1], 1 Greissinger plus[1], 1 Multiflex[2], 2 Quantum[5], 1 S.A.F.E. II[6]			
Prosthetic knee	3 Graph-lite[7], 1 C-leg[1], 1 3R60[1], 1 Total knee[3], 2 SafeLife[4]				
Cause	4 trauma, 1 vascular, 3 tumour	6 trauma, 2 vascular, 4 tumour			
Side (R / L)	5 / 3	6 / 6			
AAS	30.3 ± 29.5	33.8 ± 26.1	-	0.08	0.78
ABC	80.1 ± 17.6	88.4 ± 5.4	98.7 ± 1.0	8.52	0.00 *; 0.00 #; 0.02 $

Mean values and standard deviations of age, weight, height, time since amputation, AAS and ABC in TF, TT and AB study groups. Gender, side, cause of amputation, prosthetic feet and knees in absolute numbers. Prosthetic devices by [1] Otto Bock, Duderstadt, Germany, [2] Endolite, Centerville, USA, [3] Össur, Reykjavik, Iceland, [4] Proteval, Valenton, France, [5] Hosmer, Campbell, USA, [6] Foresee, Oakdale, USA, [7] The Lin, Kuala Lumpur, Malaysia. * Statistically significant p-values ($p \leq 0.05$) of group differences. # Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. $ Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects.

Apparatus

The study was performed at a motion analysis laboratory. The subjects walked over a walkway of 8.0 m long and 1.2 m wide. The beginning and end of the walkway were fitted with infrared beams, which registered the passing of a subject. The starting and finishing position were just before and after the infrared beams, which allowed only a small distance for acceleration and deceleration. A wood block of 0.1 m height and thickness and 1 m width was used as obstacle. A width of 1 m was chosen to force the subjects to bring the limb forward over the obstacle instead of around it. A height and thickness of 0.1 m was selected because this dimension was considered to be safe and at the same time large enough to influence the gait pattern. The runs were videotaped using two video cameras with a sampling frequency of 25 Hz. One camera scanned the coronal plane, the other the sagittal plane. The camera in the sagittal plane was moved manually with the subject along the walkway. Subjects were provided with six SG 150 Penny & Giles goniometers (Biometrics, Gwent, UK) to measure the joint angles. In these goniometers a high accuracy and repeatability have been established. The goniometers were placed bilaterally on the (prosthetic) ankle, (prosthetic) knee, and hip joint, measuring the angles in the sagittal plane. Zero position of the goniometers was defined by asking the subjects to stand in an upright position with the hips and knees extended and the feet in a plantigrade position. The subjects wore their own shoes because a prosthesis is adjusted to the height of the soles. The signals of the goniometers were recorded by a portable data acquisition system (TMSI, Enschede, NL) at a sampling frequency of 800 Hz. Recording, synchronizing and analysis of all measurements was performed by using a custom-developed Gait Analysis System (GAS, UMCG, Groningen, NL) at 100 Hz.

Procedure

The ability to perform a difficult motor task, such as obstacle crossing, may be influenced by balance control and activity level, especially in amputees. Two self-report questionnaires were selected to evaluate these subject qualities: the modified Amputee Activity Score (AAS) and the Activities-specific Balance Confidence scale (ABC). The AAS is a specific measure developed for outpatient amputees with a prosthetic limb. This test was validated and has shown to have good test-retest reliability.[19, 20] The score lies between - 70 and + 50. A higher score on the AAS represents a higher activity level. Since the AAS is specific for amputees, AB subjects did not fill out this questionnaire. The ABC is a self-efficacy measure that assesses confidence across 16 situation-specific activities. The ABC scale is reliable and valid in a clinical sample of amputees and in elderly persons.[21-23] A higher score on the ABC scale indicates a higher confidence in balance control and the maximum score is 100.

All subjects were instructed to walk along the walkway at their self-selected comfortable gait velocity. The subjects performed eight runs; four unimpeded runs and four obstacle crossing runs. The starting condition (unimpeded or obstacle) was randomised.

Randomisation was done for each subject with two envelopes containing the two walking conditions. In obstacle crossing the subjects were instructed to step over the obstacle without touching it. Since a true to nature observation was pursued, subjects were free to use either limb for leading over the obstacle. In two runs the obstacle was placed at a distance of 3.5 m from the starting position and in the other two runs at 4.5 m. We chose two distances to prevent subjects from emerging in front of the obstacle with the similar limb in all runs. The study design allowed subjects to adjust their gait pattern prior to obstacle crossing. By changing the step length or gait velocity subjects were allowed to cross the obstacle with their preferred leading limb.

Outcome parameters

The average gait velocity was calculated from the length of the walkway divided by the time that the subject needed to walk over the walkway. The video images were used to obtain a leading limb preference. In amputees the percentage of leading prosthetic limb runs was determined, while in the AB group the percentage of leading right limb runs was scored.

Amputees had two possible limb sequences to step over the obstacle: leading with the prosthetic limb, followed by trailing with the non-affected limb or leading with the non-affected limb, followed by trailing with the prosthetic limb. Since we expected different movement trajectories for the leading and trailing limb, and for the prosthetic and non-affected limb, the joint angles were analysed separately for the leading prosthetic, the trailing prosthetic, the leading non-affected, and the trailing non-affected limb. When amputees used the same limb as leading limb in all four obstacle crossing runs, we were not able to measure the joint angles in the other leading limb condition. In the AB group we used the mean outcome of the right and left limb in the data analysis to compensate for any asymmetry in the limbs. Similar to amputees the results for the leading and trailing limb in AB subjects were analysed separately, resulting in data of two limb conditions; the leading and trailing non-affected limb.

Lower limb joint angles were measured at the instant of obstacle crossing. The video images were used to determine the instant that the middle of the foot, either leading or trailing, was right above the obstacle. In this position the joint angles of the hip, knee and ankle in the swing and stance limb were derived from the GAS. The analysed joint angles are shown in Figure 1. In amputees the joint angles of the prosthetic limb were assessed in swing and of the non-affected limb in stance. This method was chosen because differences among the study groups for the specific limbs were expected in the selected phases of the gait cycle. In unimpeded walking and obstacle crossing runs the mean maximum hip, knee and ankle flexion of the combined leading and trailing limb in swing was measured. In unimpeded walking runs the maximum flexion joint angles were assessed in three successive strides in the middle of the walkway and the mean was used in the data analysis.

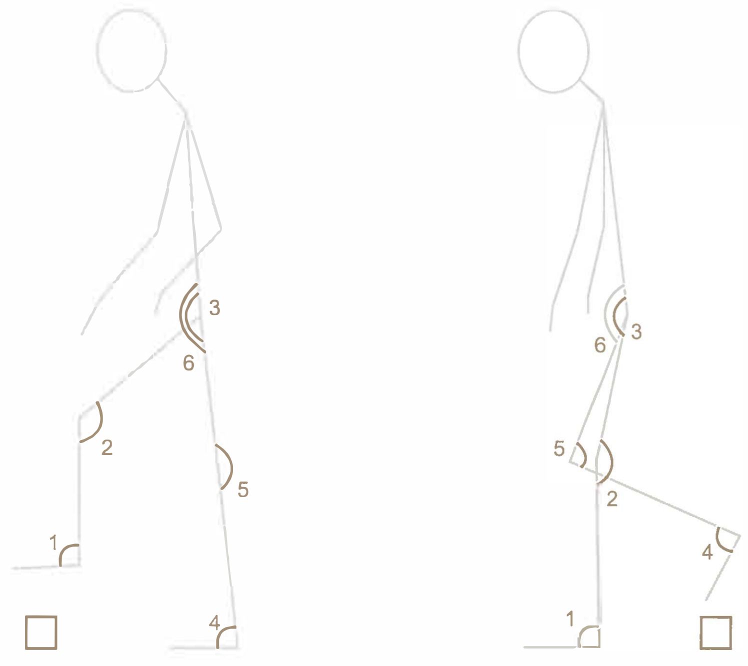

Figure 1.

Stick diagram showing the analysed hip, knee and ankle angles. The left figure shows the instant that the leading foot was above the obstacle and the right figure the instant that the trailing foot was above the obstacle.

Statistical analysis

For each limb condition the mean value of the outcome variables was calculated. Normality of the variables within the groups was tested with the Kolmogorov-Smirnov test. Differences in outcome parameters among the three groups were analysed by using an ANOVA followed by post-hoc analysis according to the least-significant difference (LSD) method. Differences in leading limb preference, AAS and time since amputation were only tested between TF and TT amputees. The paired t-test was used to analyse the differences in gait velocity and maximum joint angles between the unimpeded walking and obstacle crossing conditions within the groups. The level of significance was set on $p \leq 0.05$.

Results

The scores on the AAS and ABC questionnaires are provided in Table 1. TF and TT were almost equally active in daily life. AB had significantly more confidence in their balance than both amputee groups did. Gait velocity in unimpeded and obstacle walking (Table 2) in TF was significantly lower than in TT and AB. The presence of the obstacle reduced gait velocity significantly in all study groups. The leading limb preference in TF and TT was statistically different. (Table 3) TF showed a preference for the non-affected limb, whereas TT demonstrated a preference for the prosthetic limb.

Table 2. Gait velocity.

Gait velocity (ms^{-1})	TF (n = 8)	TT (n = 12)	AB (n = 10)	F	p-value
UW	0.99 ± 0.22	1.23 ± 0.15	1.35 ± 0.13	11.08	0.00 *; 0.00 #; 0.00 $
OC	0.90 ± 0.23	1.18 ± 0.17	1.31 ± 0.11	13.24	0.00 *; 0.00 #; 0.00 $
p-value	0.00 †	0.00 †	0.03 †		

Mean values and standard deviations in TF, TT and AB subjects in unimpeded walking (UW) and obstacle crossing (OC). * Statistically significant p-values ($p \leq 0.05$) of group differences. # Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. $ Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences between UW and OC.

Table 3. Leading limb preference.

Leading limb preference (%)	TF (n = 8)	TT (n = 12)	F	p-value	AB (n = 10)
Prosthetic / right	40.6 ± 11.0 *	67.7 ± 6.2 *	5.31	0.03 *	57.5 ± 8.4

Mean values and standard error for the prosthetic limb in TF and TT amputees and for the right limb in AB subjects. * Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects.

The results of the swing limb joint angles are demonstrated in Table 4. A group effect was seen in the hip, knee and ankle angles of the leading and trailing prosthetic limb. In TF hip and knee flexion in the leading prosthetic limb and knee flexion in the trailing prosthetic limb were significantly reduced in comparison to TT and AB. Hip flexion in the trailing prosthetic limb was significantly lower in TF than in TT. Ankle dorsiflexion in the leading prosthetic foot was significantly decreased in TF and TT compared to AB. The trailing prosthetic ankle of TF was significantly less plantar flexed than in AB. The results of the joint angles in the stance limb are presented in Table 5. A group effect was only seen in the trailing non-affected ankle. In TF this ankle showed a significant increase in plantar flexion in stance compared to TT and AB.

The results of the maximum flexion angles during swing in obstacle crossing and unimpeded walking are presented in Table 6. A group effect during obstacle crossing was seen in the hip, knee and ankle angles. In obstacle crossing maximal hip and knee flexion of the prosthetic limb in TF were significantly reduced compared with TT and AB. In TF and TT the maximum dorsiflexion of the ankle was significantly reduced compared to AB, especially in TT. In unimpeded walking no group differences in maximum flexion were shown. TT and AB were able to significantly increase hip and knee flexion in obstacle crossing compared to unimpeded walking. Maximum knee and hip flexion increased approximately 20° in TT and AB when crossing the obstacle. In TF maximum hip flexion increased only a few degrees in obstacle crossing, whereas maximum knee flexion in TF was significantly lower in obstacle crossing than in unimpeded walking. Ankle dorsiflexion was significantly increased by the obstacle in TF and AB, but not in TT.

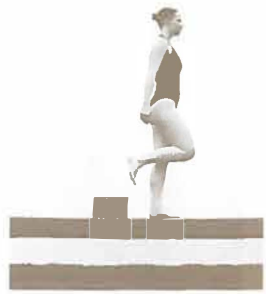

Table 4. Joint angles of the swing limb.

Angles (°)		TF (n = 8)	TT (n = 12)	AB (n = 10)	F	p-value
Hip	LP - LN	30 ± 13.8	44 ± 9.9	42 ± 5.5	3.44	0.05 *; 0.04 #; 0.02 $$
	TP - TN	7 ± 9.5	16 ± 5.8	12 ± 5.6	3.75	0.04 *; 0.01 $$
Knee	LP - LN	32 ± 29.4	62 ± 12.5	67 ± 12.8	7.78	0.00 *; 0.00 #; 0.00 $$
	TP - TN	24 ± 30.0	72 ± 12.5	64 ± 6.6	18.87	0.00 *; 0.00 #; 0.00 $$
Ankle	LP - LN	1 ± 5.7	−3 ± 2.5	7 ± 3.3	26.81	0.00 *; 0.00 #; 0.00 $
	TP - TN	1 ± 2.4	−2 ± 2.6	−7 ± 7.1	4.09	0.03 *; 0.13 #

Mean values and standard deviations of the leading prosthetic (LP) and trailing prosthetic (TP) limb in TF and TT amputees and of the leading non-affected (LN) and trailing non-affected (TN) limb in AB subjects during swing phase at the instant that the foot is right above the obstacle. Ankle dorsiflexion is positive, plantar flexion is negative. * Statistically significant p-values ($p \leq 0.05$) of group differences. # Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. $ Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. $$ Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects.

Table 5. Joint angles of the stance limb.

Angles (°)		TF (n = 8)	TT (n = 12)	AB (n = 10)	F	p-value
Hip	LN	20 ± 7.3	21 ± 7.4	19 ± 3.9	0.33	0.72
	TN	8 ± 10.1	9 ± 5.2	6 ± 3.8	1.02	0.38
Knee	LN	9 ± 7.4	13 ± 6.4	16 ± 3.7	2.92	0.07
	TN	6 ± 6.9	8 ± 2.8	6 ± 4.5	0.87	0.43
Ankle	LN	−5 ± 12.3	1 ± 2.4	−1 ± 3.1	1.15	0.34
	TN	−5 ± 11.1	2 ± 4.2	3 ± 2.5	3.55	0.05 *; 0.02 #; 0.03 $

Mean values and standard deviations of the leading non-affected (LN) and trailing non-affected (TN) limb in TF, TT and AB subjects during stance phase at the instant that the contralateral foot is in swing phase right above the obstacle. Ankle dorsiflexion is positive, plantar flexion is negative. * Statistically significant p-values ($p \leq 0.05$) of group differences. # Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. $ Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects.

Table 6. Maximum joint angles during swing phase.

Angles (°)		TF (n = 8)	TT (n = 12)	AB (n = 10)	F	p-value
Hip	UW	23 ± 5.7	25 ± 4.6	25 ± 3.3	0.85	0.44
	OC	30 ± 13.0	44 ± 6.4	41 ± 4.4	7.22	0.00 *; 0.01 #; 0.00 $$
	p-value	0.12	0.00 †	0.00 †		
Knee	UW	54 ± 11.7	59 ± 10.9	57 ± 4.3	0.64	0.53
	OC	31 ± 31.6	77 ± 13.2	77 ± 11.0	16.60	0.00 *; 0.00 #; 0.00 $$
	p-value	0.04 †	0.00 †	0.00 †		
Ankle	UW	2 ± 3.0	−1 ± 2.9	0 ± 2.1	2.04	0.15
	OC	4 ± 3.2	0 ± 2.4	9 ± 3.6	22.56	0.00 *; 0.00 #; 0.00 $; 0.02 $$
	p-value	0.01 †	0.32	0.00 †		

Mean values and standard deviations of the prosthetic leading and trailing limb in TF and TT amputees and of the non-affected leading and trailing limb in AB subjects during obstacle crossing (OC) and unimpeded walking (UW). Ankle dorsiflexion is positive, plantar flexion is negative. * Statistically significant p-values ($p \leq 0.05$) of group differences. # Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. $ Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. $$ Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences between UW and OC.

Discussion

For safe obstacle crossing, it is important that the swing limb reaches enough clearance to prevent contact between the foot and obstacle.[5, 6] An increase in knee flexion is the most important strategy to achieve adequate flexion and shortening of the swing limb.[1, 5, 7, 8, 11, 12] In our study prosthetic knee flexion in obstacle crossing in TF amputees was reduced in comparison with unimpeded walking, and compared to TT amputees and AB subjects as well. Normally, TF amputees initiate the swing movement of the prosthetic knee by generating activity in the flexors of the hip, followed by a totally passive motion of the prosthesis forward, like a pendulum. The results in our study suggest that during obstacle crossing TF amputees do not use this pendulum motion.

Another explanation for the decreased knee flexion in TF amputees is the reduced gait velocity. It is known that increased gait velocity results in more joint flexion in the lower limbs during swing.[24,25] Walking with a lower gait velocity also impedes the initiation of the pendulum motion. Apart from a negative effect on knee flexion, lower gait velocity can reflect an adjustment strategy to minimize the adverse consequences of contact with the obstacle.[4] In our study TF, TT and AB subjects all walked slower in obstacle crossing than in unimpeded walking, which may represent a safety measure in case of falling.

In AB subjects and TT amputees the knee strategy was an important adjustment strategy during obstacle crossing. In other studies an increased swing knee flexion in AB subjects was also seen in obstacle crossing.[1,5,8,11,12] However, in TT amputees a previous study showed a decreased knee flexion on the prosthetic side.[14] In this study the prosthetic limb was compared with the non-affected limb, whereas in our study a comparison was made with an AB control group. Besides, the relative decrease in knee flexion on the prosthetic side was only seen in stepping over obstacles higher than those used in our study.

To ensure foot clearance TF amputees used adjustment strategies to compensate for the lack of knee flexion. The video images showed that TF amputees externally rotated and abducted the hip on the prosthetic side during obstacle crossing. Seven out of eight TF amputees used this circumduction strategy consistently. Apart from foot clearance, the circumduction strategy also eases the transition from swing to stance. In consequence of the lack of pendulum motion of the prosthetic limb and the lower gait velocity, it is more difficult for TF amputees to extend the prosthetic limb. A possible strategy to ensure extension in the knee at initial contact is using the circumduction movement.

TF amputees shortened their prosthetic swing limb by increasing maximum hip flexion on the prosthetic side in obstacle crossing compared to unimpeded walking. However, the increase in hip flexion in TF amputees was much smaller than in TT and AB subjects, which confirms the lack of initiation of the prosthetic swing pendulum from the hip in TF amputees. Also other factors, like the lowered gait velocity and restrained hip motion due to the prosthetic socket, may have been contributing to the decreased hip flexion in TF amputees.

Tf amputees also used adjustment strategies to increase the height of the hip joint, which contributes to an easier obstacle clearance. When the leading prosthetic foot of TF amputees was above the obstacle, the trailing non-affected ankle was more plantar flexed. Two out of eight TF amputees increased ankle plantar flexion in the leading and trailing non-affected stance limb in all obstacle runs. Another study also found increased ankle plantar flexion in the non-affected limb in TT amputees.[14] This strategy was not seen in TT amputees in our study and this may be explained by to the non-restricted knee flexion in our TT group.

One TF amputee who was provided with a microprocessor-controlled knee was able to use the same adjustment strategy as AB and TT subjects. This subject showed an increased swing knee flexion in obstacle crossing in comparison with unimpeded walking. Conventional prostheses are aligned anteriorly, which means that the vertical ground reaction force is positioned in front of the centre of rotation of the prosthetic knee in stance. Anterior alignment increases stability by limiting the tendency of knee flexion at initial contact. In microprocessor-controlled knees stability in stance is ensured by damping of the knee. Microprocessor-controlled knees are more easily flexed, because the alignment is shifted posteriorly and the damping at the moment of swing initiation is automatically minimized.[26] Apart from the advances of the properties of the knee itself, amputees provided with a microprocessor-controlled knee may be more active and physically capable to perform difficult motor activities. Further research may lead to more insight into the influence of a microprocessor-controlled knee on adjustment strategies in obstacle crossing.

The choice of the subjects to lead obstacle crossing with a preferred limb could reflect an adjustment strategy. A clear leading limb preference was only found in TT amputees; they favoured the prosthetic limb. In contrast, TF amputees showed a tendency of preference for the non-affected limb. Since AB subjects did not show a leading limb preference, we may conclude that the variable distance from the starting position to the obstacle assured an alternate use of both limbs. Leading with the prosthetic limb has advantages. During crossing with the leading limb the movement can be controlled by visual feedback, while the trailing limb trajectory can not be guided by vision.[4, 14, 24, 27, 28] Moreover, at toe-off prior to obstacle crossing the foot of the leading limb is one step length further away from

the obstacle than the foot of the trailing limb. This provides more time for the leading limb to achieve adequate limb elevation.[5,24]

Amputees can benefit from leading with the non-affected limb for additional reasons. The trailing limb needs less limb flexion to avoid obstacle contact than the leading limb does. Furthermore, the risk of a trip or a fall is increased when the leading limb contacts the obstacle, because during crossing with the leading limb the body's centre of mass is moving away from the stance foot, resulting in a position of the centre of mass anterior to the base of support.[4,18] As the trailing limb clears the obstacle, the centre of mass is moving towards the stance foot, which represents a more stable situation.[7]

The most likely explanation for the difference in leading limb strategy between TF and TT amputees is the absence of the knee joint and adjacent musculature in the prosthetic limb in TF amputees. TT amputees can flex the knee on the prosthetic side sufficiently to ensure foot clearance, and only risk a very limited chance of contact with the obstacle when leading with the prosthetic limb. TT amputees use the advantages of leading with the prosthetic limb, which are visual control and increased time and distance to lift the foot. In TF amputees the prosthetic knee function is too restrained to shorten the swing limb adequately. Because the leading limb needs more flexion and obstacle contact with the leading limb has a greater risk of falling, TF amputees may prefer leading with the non-affected limb.

In order to reduce falls in amputees, it is important to train complex motor tasks, such as stepping over an obstacle, during the rehabilitation period. This training should be aimed at improving knee flexion or the execution of adjustment strategies. Since the limited flexion of the prosthetic knee mainly causes the limited function in TF amputees, prosthetic design should concentrate on the progress of this property in the prosthetic knee. The development of the microprocessor-controlled prosthetic knee is a promising example.

The study was limited by the different types of prosthetic knees and feet that the amputees used, which may have led to the use of different adjustment strategies. Due to the small sample size we were not able to study the influence of the diverse properties of the prostheses on the outcome measures. As a result of technical limitations we were not able to measure spatial variables, like step length and foot clearance, which are important features in obstacle crossing. Other studies have shown that the distance of the foot to the obstacle at toe-off influences the trajectory of the leading and trailing limb.[4, 27, 29] Finally, we only used one obstacle height and thickness. Other studies have shown that joint angles change with different dimensions of the obstacles.[5, 10, 14, 29, 30] Stepping over higher or thicker obstacles may result in the usage of other adjustment strategies or change the above-mentioned strategies.

Conclusion

Whereas AB subjects and TT amputees use a knee strategy in obstacle crossing, TF amputees have to apply other adjustment strategies to step safely over the obstacle. The different leading limb strategy in TF and TT amputees could be explained by the restricted flexion and propulsion properties of the prosthetic knee. Training of obstacle crossing tasks during rehabilitation and improvement of prosthetic design may contribute to safe obstacle crossing.

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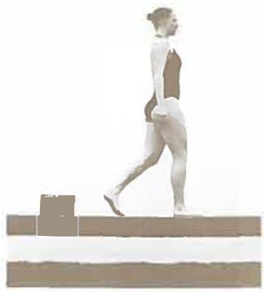

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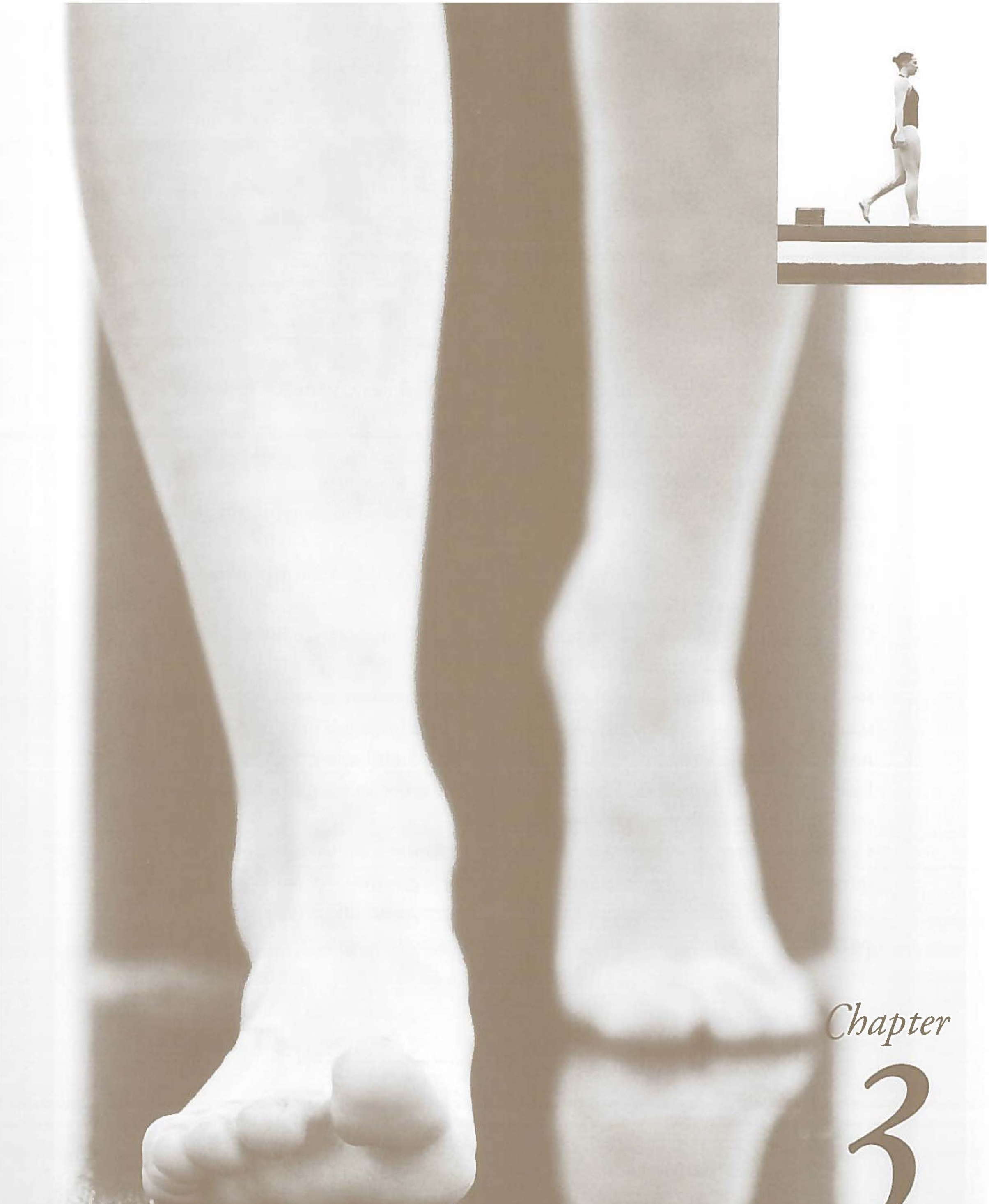

Chapter
3

Gait initiation in lower limb amputees

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Abstract

Objective: To study limitations in function and adjustment strategies in lower limb amputees during gait initiation.
Design: Observational cohort study.
Setting: University Medical Center.
Subjects: Amputees with a unilateral transfemoral or transtibial amputation and able-bodied subjects.
Methods: In a motion analysis laboratory subjects performed twelve gait initiation runs on a force plate and walkway.
Outcome measures: Leading limb preference, temporal variables, ground reaction forces and centre of pressure shift.
Results: Amputees demonstrated a decrease in peak propulsive ground reaction force, a smaller or absent posterior centre of pressure shift and a lower gait initiation velocity. The main adjustments strategies in amputees were more limb loading on the non-affected limb, prolonging the period of propulsive force production in the non-affected limb and initiating gait preferably with the prosthetic limb.
Conclusion: Since an intact ankle joint and musculature is of major importance in gait initiation, functional limitations and adjustment strategies in transfemoral and transtibial amputees were similar. Improving prosthetic foot properties and initiating gait with the prosthetic limb may facilitate the gait initiation process in amputees.

Introduction

Most studies concerning human gait have focused on steady state walking. However, for safe independent locomotion other aspects of gait are important as well. The transition from standing to walking is a task which is often required in daily life and challenges balance control.[1, 2] Compared to steady state walking, the requirements on the neuromuscular system are increased in gait initiation, since a complex integration of neural mechanisms, muscle activity and biomechanical forces is necessary.[1]

Postural adjustments and muscle activity at the ankle and hip are needed to initiate gait. The limb that moves forward first is called the leading limb and the other limb is termed the trailing one. Able-bodied (AB) individuals activate the tibialis anterior muscle and inhibit the soleus muscle activity to shift the centre of pressure (COP) posteriorly and to accelerate the centre of mass (COM) anteriorly.[3, 4] As a result, the propulsive ground reaction force (GRF) increases, thereby generating a forward momentum.[3, 5] Simultaneously, abductor muscles in the leading limb shift the COP toward this limb.[3, 6] Prior to heel-rise of the leading limb the COM is shifted toward the trailing limb, which unloads the leading limb and creates a stable base for balance control in single-limb stance.[7] Finally, a burst of soleus muscle activity initiates push-off of the leading limb[8, 9], whereas the COM is accelerated further in a forward and medial direction.[10]

The amputated limb is affected by sensory loss, while muscles and joint(s) are absent. Gait initiation requires two skills that may be limited in amputees, propulsion and balance control. Previous studies in amputees have shown inconclusive results concerning the COP trajectory, which is an important outcome measure in gait initiation.[1, 8, 11] In only two studies transfemoral (TF) amputees were tested next to transtibial (TT) amputees[8, 12], and one study included an AB control group.[11] Moreover, in all studies gait was initiated in response to a starting signal.[1, 8, 11, 12]

The first goal of this study was to determine limitations in function in the prosthetic limb of TF and TT amputees during self-initiated gait. We hypothesised that the posterior COP shift and propulsive GRF in the prosthetic limb will be reduced which results in a lower gait initiation velocity in amputees. The second purpose of this study was to identify adjustment strategies used by amputees to compensate for the limitations in function.

To enhance propulsion amputees may produce a larger and prolonged propulsive GRF in the non-affected limb. To ease balance control amputees may increase limb loading on the non-affected limb and shorten single-limb stance duration on the prosthetic limb. Finally, amputees may prefer to initiate gait with the prosthetic limb which serves both propulsion and balance.

Methods

Subjects

Subjects with a unilateral TF or TT amputation were recruited from a prosthetics workshop. Inclusion criteria included an amputation for at least one year, daily use of a prosthesis and the ability to walk more than 50 m without walking aids. A control group of AB subjects was also selected. They were recruited via advertisements at the local blood bank, hospital, television and radio station. Subjects were excluded if they had any medical conditions that could affect their mobility or balance, such as neurological, orthopaedic or rheumatic disorders, cognitive problems, severely impaired vision, or reduced sensory loss in the non-affected limb. Furthermore, amputees with pain or wounds at the amputation limb or prosthetic fitting problems were excluded.

The study group consisted of seven TF amputees, twelve TT amputees and ten AB subjects. The Medical Ethics Committee approved the study protocol. All subjects signed informed consent before testing. Amputees used different prosthetic feet and all TF amputees used free moveable prosthetic knees. Characteristics of the subjects are shown in Table 1.

Apparatus

The study was performed in a motion analysis laboratory, which is equipped with an 8.0 m long aluminium walkway and a force plate (Bertec, Columbus, USA) of 0.4 x 0.6 m. We recorded the gait initiation process with video cameras. The sampling frequency was 25 Hz. Flexible self-adhesive aluminium strips were attached at the heel and forefoot of the soles of the shoes. Contact of the strips with the conductive walkway detected the onset of initial contact and toe-off. Signals of the foot contacts were recorded on a portable data acquisition system (TMSI, Enschede, NL) at a sampling frequency of 800 Hz. The force plate measured the force and COP data. Recording, synchronising and analysis of all the measurements were undertaken with a custom-developed Gait Analysis System (UMCG, Groningen, NL). The sampling frequency was 100 Hz.

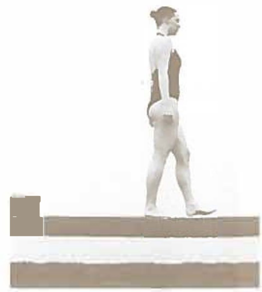


Table 1. Subject characteristics, leading limb preference, gait initiation velocity and single-limb stance duration.

Group		TF (n = 7)	TT (n = 12)	AB (n = 10)
Gender (M / F)		6 / 1	10 / 2	9 / 1
Age (years)		44.0 ± 14.1	49.6 ± 11.6	45.2 ± 9.4
Weight (kg)		81.4 ± 12.4	84.2 ± 8.2	86.5 ± 9.1
Height (cm)		182.6 ± 6.2	180.9 ± 8.5	184.4 ± 6.7
Time (months)		210.7 ± 158.1	207.8 ± 169.4	
Side (R / L)		5 / 2	6 / 6	
Cause		4 trauma, 3 oncology	6 trauma, 2 vascular, 4 oncology	
Prosthetic foot		2 C-walk[1], 2 SACH[1], 3 Multiflex[2]	4 C-walk[1], 3 SACH[1], 1 Greissinger plus[1], 1 Multiflex[2], 2 Quantum[5], 1 S.A.F.E. II[6]	
Prosthetic knee		3 Graph-lite[7], 1 C-leg[1], 1 3R60[1], 1 Total knee[3], 1 SafeLife[4]		
AAS		35.9 ± 26.9	33.8 ± 26.1	
ABC		83.5 ± 15.9 *a	88.4 ± 5.4 *b	98.7 ± 1.0 *a,b
Leading limb pref (%)		71.4 ± 14.9	90.0 ± 4.7	62.5 ± 11.9
Gait initiation velocity (ms^{-1})	LP	0.71 ± 0.16 *a,c	0.91 ± 0.15 *c	
	LN	0.69 ± 0.19 *a	0.83 ± 0.13 *b	1.03 ± 0.17 *a,b
Single-limb stance (s)	TP	0.43 ± 0.18 †	0.34 ± 0.05 †	
	TN	0.61 ± 0.10 *a,c†	0.43 ± 0.06 *c†	0.43 ± 0.05 *a

Mean values and standard deviations of age, weight, height, time since amputation, AAS, ABC, gait initiation velocity in the leading prosthetic (LP) and in the leading non-affected (LN) limb condition, and single-limb stance duration of the trailing prosthetic (TP) and the trailing non-affected (TN) limb in the TF, TT and AB study groups. Mean values and standard error of leading limb preference for the prosthetic limb in TF and TT subjects and for the right limb in AB subjects. Gender, side and cause of amputation and the used prosthetic feet and knees in absolute numbers. Prosthetic devices by [1] Otto Bock, Duderstadt, Germany, [2] Endolite, Centerville, USA, [3] Össur, Reykjavik, Iceland, [4] Proteval, Valenton, France, [5] Hosmer, Campbell, USA, [6] Foresee, Oakdale, USA, [7] The Lin, Kuala Lumpur, Malaysia. *a Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. *b Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. *c Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences between the limb (condition) in TF and TT subjects.

Procedure
Subjects filled out the Activities-specific Balance Confidence scale (ABC) to obtain information on balance control.[13-15] The ABC scale is a self-efficacy measure that assesses confidence in balance control across 16 activities. A higher score indicates more balance confidence and the maximum score is 100. TF and TT amputees filled in the modified Amputee Activity Score (AAS) to provide insight into their activity level.[16, 17] The score lies between - 70 and + 50. A higher score represents a higher activity level.

Subjects performed twelve runs: eight on the walkway and four on the force plate. In the walkway runs we assessed leading limb preference and single-limb stance duration. In the force plate runs we measured GRF, COP and gait initiation velocity. Subjects started walking from a bipedal standing position on their own initiative. In the first four walkway runs no instructions were given on which limb should be used as leading limb. In the following four walkway runs subjects had to alternate the leading limb to make sure that each limb was used as the leading one in four runs. In the force plate runs subjects started with both limbs placed on the force plate. The position of the feet on the force plate was self-selected. The subjects performed two force plate runs with the prosthetic limb leading and two runs with the non-affected limb leading.

Outcome parameters
We determined leading limb preference from the video images of the first four runs, in which the leading limb was self-selected. In amputees the percentage of prosthetic leading limb runs was determined, and in AB subjects the percentage of right leading limb runs. Toe-off of the leading limb divided gait initiation into a period of bipedal and single-limb stance. Single-limb stance duration in the trailing limb started at toe-off of the leading limb and ended at initial contact of the leading limb. In AB subjects the mean of the right and left limb was used in the data analysis to minimize the influence of asymmetry between these limbs, whereas in amputees the prosthetic and non-affected limbs were analysed separately.

GRF and COP data were obtained from a single force plate. Consequently, in bipedal stance the resultant GRF and COP of the leading and trailing limbs together was assessed. In amputees two limb conditions were distinguished in bipedal stance: (1) leading with

the prosthetic and trailing with the non-affected limb and (2) leading with the non-affected and trailing with the prosthetic limb. In single-limb stance GRF and COP were executed by the trailing limb alone, resulting in data on the trailing prosthetic and trailing non-affected limb in amputees. In AB subjects in bipedal stance data of the leading and trailing non-affected limb condition were collected, and in single-limb stance of the trailing non-affected limb alone. The peak amplitudes of GRF in the vertical (FZ), anteroposterior (FY) and mediolateral (FX) directions were obtained. (Figure 1) The first peak GRF (FZ,Y,X1) was assessed at the end of bipedal stance at push-off of the leading limb and was produced by the leading and trailing limb together. The second peak GRF (FZ,Y,X2) was assessed at the end of single-limb stance at the instant of push-off of the trailing limb and was executed by the trailing limb alone. We expressed FZ, FY and FX as percentages of body weight (% BW). Gait initiation velocity was calculated by integration of the anterior acceleration from FY2. The trajectory of the resultant COP in bipedal stance was described by four measuring points. (Figure 1)

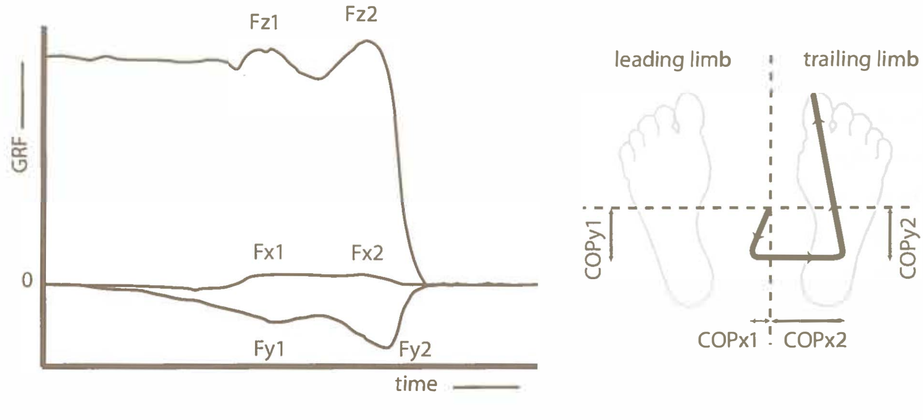


Figure 1.

Schematic representation of the analysed peak components of the GRF (left) and the measuring points of the COP trajectory (right). The first peak amplitudes of the GRF in the vertical (Fz1), anteroposterior (Fy1) and mediolateral (Fx1) direction were produced in bipedal stance and the second peak amplitudes of GRF in the vertical (Fz2), anteroposterior (Fy2) and mediolateral (Fx2) direction in single-limb stance. The COP trajectory was described by four distances from the bipedal starting position to: COPy1, the most posterior position on the leading limb side; COPx1, the most lateral position on the leading limb side; COPy2, the most posterior position on the trailing limb side; COPx2, the most lateral position on the trailing limb side.

Statistical analysis

Normality of the outcome parameters within groups was tested with the Kolmogorov-Smirnov test. For each limb (condition) differences between groups were analysed by using an ANOVA with study group as main factor, followed by post-hoc analysis according to the least-significant difference (LSD) method. Differences in time since amputation, AAS and leading limb preference were only tested between TF and TT amputees. The paired t-test was used to analyse differences between the non-affected and the prosthetic limb or between the leading prosthetic and leading non-affected limb condition within amputee groups. Level of significance was set on $p \leq 0.05$.

Results

Results of the AAS and ABC questionnaires, leading limb preference, gait initiation velocity and single-limb stance duration are presented in Table 1. AAS was similar in TT and TF. AB showed a higher score on the ABC scale than TF and TT. In AB there was a preference for the right limb and in TF and TT for the prosthetic limb. Eight out of twelve TT and four out of seven TF started walking with the prosthetic limb consistently in all four runs. Compared to AB, gait initiation velocity in the leading prosthetic limb condition was lower in TF and in the leading non-affected limb condition in TF and TT. Which limb initiated gait did not affect velocity in TF and TT. In TF the duration of single-limb stance was prolonged in the trailing non-affected limb compared to AB, TT and the trailing prosthetic limb, whereas in TT the single-limb stance duration in the trailing prosthetic limb was shorter than in the trailing non-affected limb.

The results of FZ, FY and FX are presented in Figure 2. TT showed a lower FZ1 in the leading prosthetic limb condition compared to AB, TF, and to the FZ1 in the leading non-affected condition. The FZ2 of the trailing prosthetic limb was decreased in TF and TT compared to AB. In addition, the FZ2 of the trailing prosthetic limb in TT was lower compared to TF and to the FZ2 of the trailing non-affected limb. The FY1 in TF and TT was decreased compared to AB in both limb conditions. In the leading prosthetic limb condition the FY1 in TT was higher than in TF and the FY1 in the leading non-affected limb condition. The FY2 of the trailing prosthetic limb was decreased in TF and TT compared to AB and to the FY2 of the trailing non-affected limb. The FX1 in the leading prosthetic limb condition in TT was decreased compared to AB. In TF and TT FX1 in the leading non-affected limb

condition was increased compared to AB and to the FX1 in the leading prosthetic limb condition.

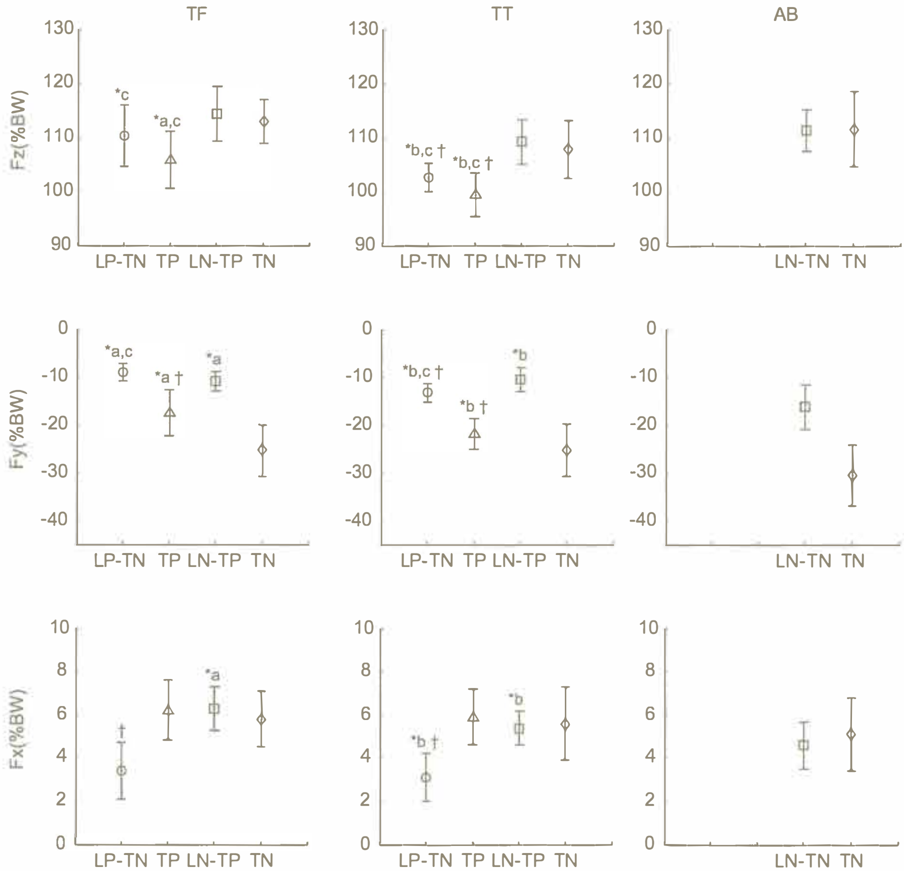


Figure 2.
Mean values and standard deviations of Fz,y,x1 produced in bipedal stance in the leading prosthetic and trailing non-affected limb condition (LP-TN) and in the leading non-affected and trailing prosthetic limb condition (LN-TP) of TF and TT subjects and in the leading and trailing non-affected limb condition (LN-TN) of AB subjects, and of Fz,y,x2 produced in single-limb stance by the trailing prosthetic limb (TP) and the trailing non-affected limb (TN) of TF, TT and AB subjects. Fz is positive in the upward direction, Fy in the posterior direction and Fx in the trailing limb direction. *a Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. *b Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. *c Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences between the limb (condition) in TF and TT subjects.

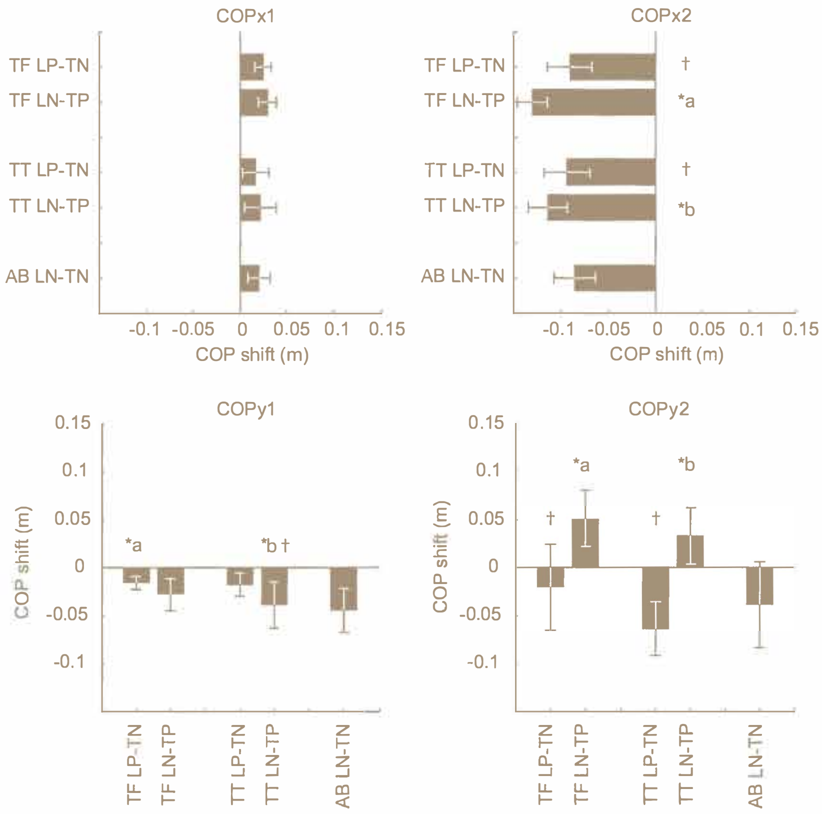


Figure 3.

Mean values and standard deviations of COPx,y1 in early bipedal stance and of COPx,y2 in late bipedal stance in the leading prosthetic and trailing non-affected limb condition (LP-TN) and in the leading non-affected and trailing prosthetic limb condition (LN-TP) of TF and TT subjects and in the leading and trailing non-affected limb condition (LN-TN) of AB subjects. COPx is positive in the direction of the leading limb and COPy in the anterior direction. *a Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. *b Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences between the limb (condition) in TF and TT subjects.

In Figure 3 the data of the COP are shown. No differences were seen in the COPX1 among the groups. In the leading non-affected limb condition the COPX2 in TF and TT shifted more lateral compared to AB and to the COPX2 in the leading prosthetic limb condition. In the leading prosthetic limb condition the COPY1 in TF and TT was shifted less posteriorly than in AB, and in TT the COPY1 was also decreased compared to the COPY1 in the leading non-affected limb condition. In TF and TT the COPY2 in the leading non-affected limb condition was located anteriorly of the starting position, whereas the COPY2 in AB and in the leading prosthetic limb condition in TF and TT was shifted posteriorly. A typical example of the COP trajectories in TF is presented in Figure 4.

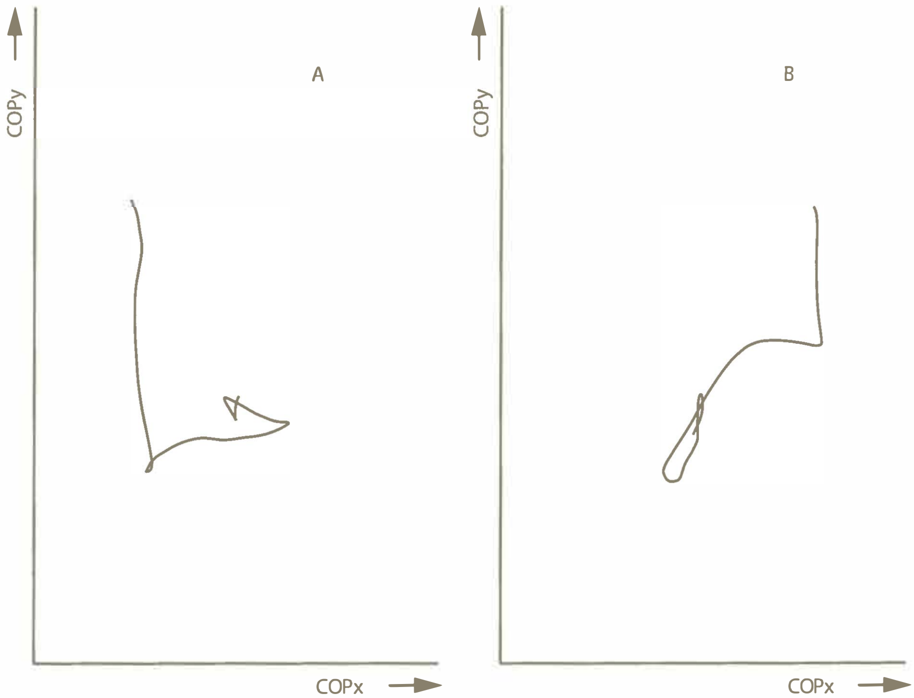


Figure 4.

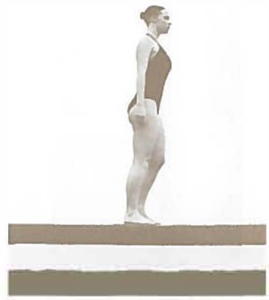
Example of the COP trajectories in a TF subject. A. When leading with the right prosthetic limb and trailing with the left non-affected limb the COPy1 and COPy2 both shifted toward posterior. B. When leading with the left non-affected limb and trailing with the right prosthetic limb the COPy1 was displaced toward the forefoot.

Discussion

The first goal of this study was to determine the limitations in function of the prosthetic limb. For adequate propulsion in gait initiation a posterior COPY displacement and an propulsive FY execution are essential. The stiffness of the prosthetic foot, absent ankle dorsal flexors and deficient sensory feedback resulted in a decreased posterior COPY shift in amputees. The COPY trajectory in amputees mostly differed from AB subjects in the leading non-affected limb condition, in which the COPY was located near the forefoot at the transition to single-limb stance on the trailing prosthetic limb. In the leading prosthetic limb condition a small posterior COPY shift could be achieved in amputees, because in bipedal stance the non-affected trailing limb assisted in the execution of postural adjustments. Tokuno et al.[11] came to the same conclusions concerning the COP shift in amputees, whereas other authors described a similar COP trajectory in AB subjects and both leading limb conditions of amputees.[1, 8, 18]

The reduced propulsive FY in the leading prosthetic limb condition and in the trailing prosthetic limb was caused by the restricted posterior COPY shift and the absence of ankle plantar flexors. The trailing limb normally produces the major part of the propulsive FY in the first step.[6, 19-22] Propulsive FY was predominantly decreased in the trailing prosthetic limb, which corresponded with the absent COPY shift posteriorly in the leading non-affected limb condition. In previous studies a smaller propulsive FY in the prosthetic limb was also seen, most obviously when used as leading limb.[1, 8, 11, 12] As hypothesised, gait initiation velocity was decreased in amputees due to the lower propulsive FY. Velocity at the end of the first step in TF amputees was lower than in TT amputees, especially in the leading non-affected limb condition, which is in accordance with the smaller propulsive FY in TF amputees compared to TT amputees in this condition.

The second aim of this study was to identify adjustment strategies used by amputees in gait initiation. Amputees did not increase propulsive FY in the trailing non-affected limb, but prolonged single-limb stance duration in this limb. In this manner a larger propulsive impulse could be reached in the trailing non-affected limb. An additional explanation for the long period of single-limb stance in the trailing non-affected limb in TF amputees is provided by the properties of the prosthetic knee. A prosthetic knee generally requires a



longer swing phase to reach knee extension at initial contact. It is known from studies in normal walking that swing phase of the prosthetic limb is prolonged in TF amputees.[23, 24]

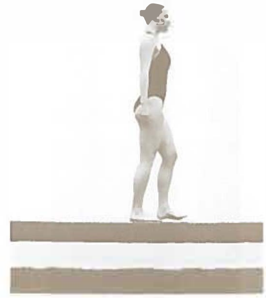
The choice of the leading limb did not influence gait initiation velocity in amputees, which was in agreement with other studies.[8, 18] In our study, a lower gait velocity was expected in the leading non-affected limb condition, since most limitations in propulsion were seen in this condition. The only adjustment strategy to enhance propulsion was found in the trailing non-affected limb. However, in previous studies amputees took more time to load the trailing prosthetic limb and increased bipedal stance duration in the leading non-affected limb condition.[1, 8, 11, 18] The consequently larger propulsive impulse may function as an adjustment strategy in the leading non-affected limb condition and explain why gait initiation velocity was independent of the leading limb.

The reduced balance control in amputees, especially in single-limb stance on the prosthetic limb[25], resulted in the occurrence of several adjustment strategies. The limited posterior COPY shift could function as an adjustment strategy to prevent a large disequilibrium between the COM and the COP.[11] Furthermore, placing the COP in front of the knee contributed to prosthetic knee extension in TF amputees to ensure stability in stance. In TT amputees single-limb stance duration in the trailing prosthetic limb was reduced, which could have served as an adjustment strategy to ease balance control. The type of prosthetic foot may have influenced the duration of single-limb stance as well. A prosthetic foot with a roll-over shape that shifts the COP quickly toward the toes may force an amputee to place the leading non-affected limb on the floor and thus shorten single-limb stance.

Another adjustment strategy that supported balance was an increased limb loading on the non-affected limb. In amputees vertical FZ in the trailing non-affected limb was higher than in the trailing prosthetic limb, which was similar to other studies.[1, 12] From the mediolateral FX and COP data we can conclude that limb loading in favour of the non-affected limb was already present in bipedal stance. More limb loading on the trailing limb in bipedal stance requires less mediolateral FX to shift the COM above the trailing limb in single-limb stance.[26] In amputees the mediolateral FX in the leading non-affected limb condition was increased, suggesting that a large shift of the COM toward the trailing prosthetic limb was needed due to the asymmetric limb loading in bipedal stance. Furthermore, amputees showed a large COPX shift toward the trailing prosthetic limb at the instant of transition to single-limb stance. This increased COPX shift may endanger stability, because the lacking somatosensory input in the prosthetic limb makes shifting the body weight accurately above the trailing prosthetic limb difficult. Studies on quiet bipedal standing in amputees reported a COPX displacement toward the non-affected limb as well.[1, 27-29] In the literature several other explanations for asymmetric limb loading in amputees are described: reduced ankle mobility, stump pain, discomfort of the rigid prosthetic socket, poor hip abductor muscle strength, inadequate sensory information in the prosthetic limb, lack of confidence, or habitual stance.[29-31]

The preference in amputees to lead gait initiation with the prosthetic limb may indicate an adjustment strategy. Previous research did not result in a unanimous conclusion on leading limb preference in amputees.[11, 12] Leading with the prosthetic limb has advantages: the trailing non-affected limb produces most part of the propulsive FY, a posterior COPY shift is achieved in bipedal and single-limb stance, the body weight is already shifted toward the trailing non-affected limb in bipedal stance, and therefore no large increase in mediolateral COPX shift is required. Based on our results, we would advice experienced active amputees to start gait initiation with the prosthetic limb, despite the fact that gait initiation velocity was similar in both leading limb conditions.

The present study contains several limitations. In the TF amputation group the right limb was amputated more often, which may have resulted in a higher leading prosthetic limb preference. In our AB control group a preference for the right limb was seen, whereas in previous gait initiation studies on AB subjects preferences for both limbs were demonstrated.[12, 19, 20] Due to the long period of time between amputation and participation in the study, inquiring subjects after their leg dominance prior the amputation was considered to be unreliable. Another limitation was that only the first step in gait initiation was studied. Analysis of the second step may alter the advice on leading limb preference. When leading with the prosthetic, it may be difficult for TF amputees to ensure knee extension at initial contact due to the short swing phase. Furthermore, data on leading limb preference, temporal variables and joint angles were assessed in different runs than COP and GRF data. Since no major differences were seen in the gait pattern among runs, we assumed it was justified to analyse the data together. Finally, stance width may have affected the mediolateral COPX shift and the FX.[32] We did not standardise stance width among subjects, because we chose to investigate self-selected gait initiation.



Conclusion

The absence of an active flexible ankle joint resulted in limitations in function in the prosthetic limb, which were mostly identical in TF and TT amputees. TF and TT amputees used several adjustment strategies to achieve adequate propulsion and balance control. Amputees should be advised to initiate gait with the prosthetic limb, since fewer limitations in function were found and less adjustment strategies were needed in this condition. Improvement of prosthetic properties to achieve a more active ankle function could facilitate the gait initiation process.

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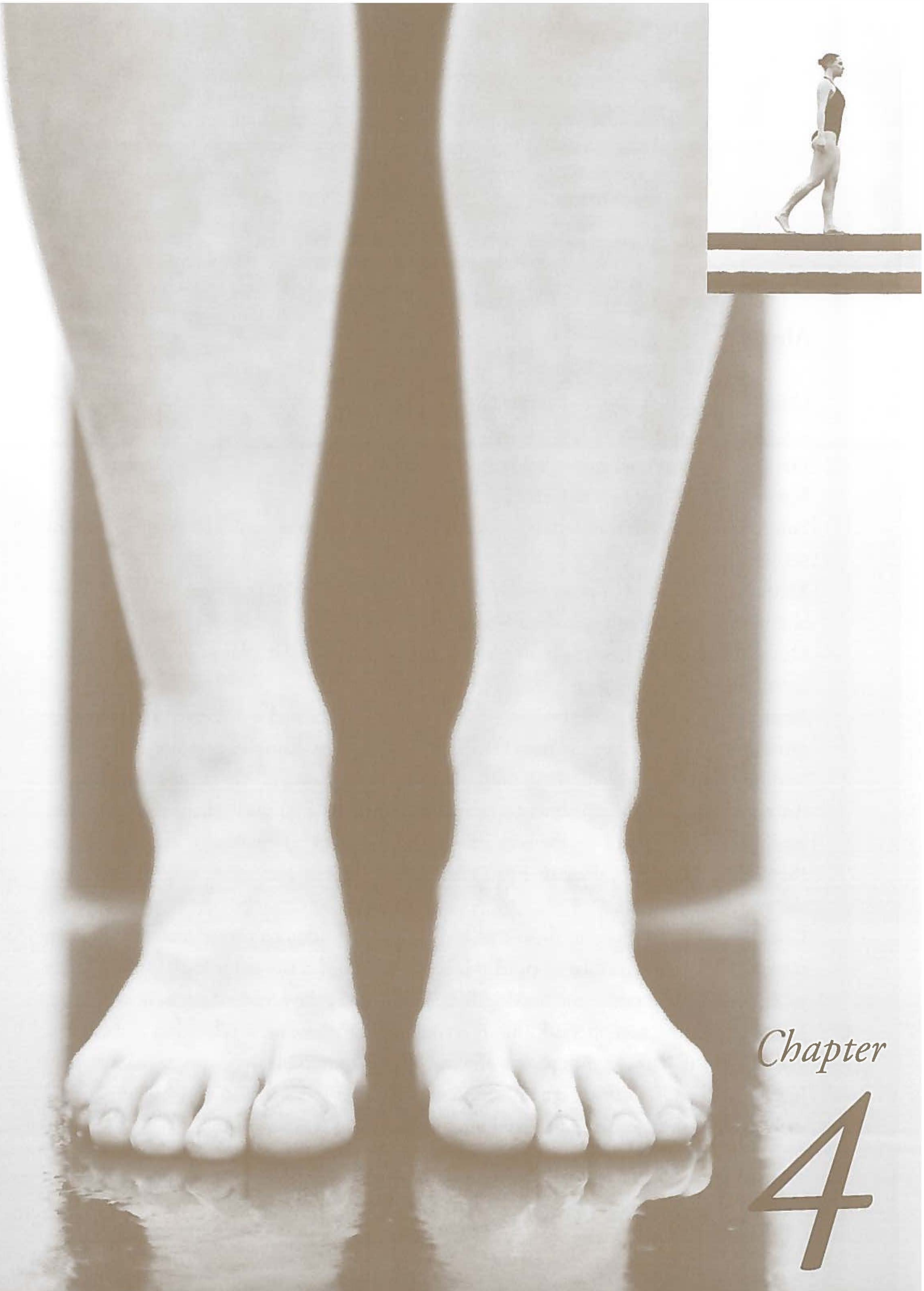

Chapter 4

Gait termination in lower limb amputees

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Abstract

Objective: To study the limitations in function and adjustment strategies of lower limb amputees in gait termination.

Design: Observational cohort study.

Setting: University Medical Center.

Subjects: Unilateral transfemoral and transtibial amputees and able-bodied control subjects

Methods: Subjects performed twelve self-initiated gait termination runs on a force plate and walkway in a motion analysis laboratory.

Outcome measures: Leading limb preference, temporal variables, lower limb joint angles, ground reaction forces and centre of pressure shift.

Results: Compared to able-bodied subjects, amputees showed a decreased peak braking ground reaction force in the prosthetic limb, no anterior centre of pressure shift during leading with the prosthetic limb and an increased mediolateral centre of pressure shift. Amputees used several adjustment strategies to compensate for the limitations in function; leading limb preference for the non-affected limb, longer production of braking force in the non-affected limb, decreased gait termination velocity and more weight bearing on the non-affected limb.

Conclusion: Limitations in functions and adjustment strategies were mainly similar in transfemoral and transtibial amputees. Due to the lack of active ankle function, amputees were not able to increase the braking force and to shift the centre of pressure anteriorly. Leading with the non-affected limb is favourable for adequate deceleration and balance control, but in daily life not always applicable. It is important that amputees are trained in gait termination during rehabilitation and prosthetic design should focus on a more active role of the prosthetic foot and knee.

Introduction

For safe and independent walking, it is necessary that a person is able to adapt the gait pattern to various situations. One example of such an adaptation occurs in the transition from walking to standing, also called gait termination. In everyday life, gait termination is a common movement which is often performed in daily activities.[1] Compared to normal walking, gait termination places a bigger demand on the control of postural stability and requires a complex integration and cooperation in the neuromuscular system.[2-4]

For safe gait termination, forward movement of the body has to be slowed down to achieve a stable upright position.[2,5] In able-bodied (AB) persons the leading limb, which is the limb that stands still first, is mainly responsible for the production of the necessary braking ground reaction force (GRF). Compared to normal walking the braking GRF is increased in the final stance phase.[6-8] A large burst of soleus muscle activity and reduced activation in the tibialis anterior muscle of the leading limb bring the foot flat to the ground.[5,7,9,10] The vasti and gluteus medius muscles are activated, respectively, to extend the knee and to prevent the trunk from bending forward. In the trailing limb the tibialis anterior, biceps femoris and gluteus medius muscles increase activity to bring the body down and backwards with the foot flat to the ground, resulting in a further decrease in forward movement.[9] The muscle activity moves the centre of pressure (COP) anteriorly and keeps the centre of mass (COM) behind the leading limb.[5,9] The position of the COP in front of the COM and the increased braking GRF lead to deceleration of the body.[11] Also stability requirements have to be fulfilled for safe gait termination. In the final bipedal standing position the COM closely coincides with the COP and lies within the base of support.[12,13] The leading limb has the task to create a stable landing placement at the end of the gait termination process.

Transfemoral (TF) and transtibial (TT) lower limb amputees are not able to use an ankle strategy and in TF amputees an active knee function is also absent. In addition, after a lower limb amputation balance control is reduced.[14-16] Due to the loss of nerves, muscles and joints, it is to be expected that gait termination may lead to difficulties in lower limb amputees. No studies on gait termination in amputees have been published so far. Our first objective was to determine which functions were limited in amputees during planned gait termination. We formulated three hypotheses: (1) the braking GRF in the leading prosthetic limb will be decreased due to absence or inefficiency of ankle plantar flexors and knee extensors, (2) the anterior shift of the COP in the prosthetic limb will be reduced as the result of a deficient ankle strategy and (3) the mediolateral COP shift will be increased owing to reduced balance control.

To carry out gait termination in a safe manner, amputees may have to adjust their gait pattern. The second purpose of this study was to assess which adjustment strategies amputees use during gait termination in order to compensate for the limitations in function. We hypothesized four possible adjustment strategies: (1) the production of a larger braking GRF in the non-affected limb will compensate for the reduced braking GRF in the prosthetic limb, (2) gait termination velocity will be reduced so that less braking GRF is needed, (3) a preference for the non-affected limb as leading limb will be seen as a result of the larger braking GRF in this limb and (4) swing phase duration of the non-affected limb will be shortened to minimize single-limb stance duration on the prosthetic limb, which improves stability.

Methods

Subjects

Amputees who were regularly attending the local prosthetics workshop were invited to participate in this study. Subjects with a unilateral TF or TT amputation at least twelve months before inclusion, who used a prosthesis on a daily basis and were able to walk more than 50 meters without walking aids, were included. We recruited a control group of AB subjects through advertisements at the local radio and television, blood bank and hospital. We excluded subjects who suffered from medical conditions that could affect their mobility or balance: neurological, orthopaedic or rheumatic disorders, cognitive problems, significantly impaired vision, reduced sensation of the non-affected leg, or use of antipsychotic drugs, antidepressants or tranquillisers. Amputees with pain or wounds of their amputation limb or fitting problems of the prosthesis were excluded as well.

Prior to the start of the study we obtained approval from the Medical Ethics Committee. Seven TF amputees, twelve TT amputees and ten AB subjects agreed to participate in the study. Before testing, all subjects signed informed consent. The amputees used different

types of prosthetic feet and knees. All TF amputees used free moving prosthetic knees. The subject characteristics are provided in Table 1.

Table 1. Subject characteristics, leading limb preference, gait termination velocity and swing phase duration.

Group		TF (n = 7)	TT (n = 12)	AB (n = 10)
Gender (M / F)		6 / 1	10 / 2	9 / 1
Age (years)		44.0 ± 14.1	49.6 ± 11.6	45.2 ± 9.4
Weight (kg)		81.4 ± 12.4	84.2 ± 8.2	86.5 ± 9.1
Height (cm)		182.6 ± 6.2	180.9 ± 8.5	184.4 ± 6.7
Time (months)		210.7 ± 158.1	207.8 ± 169.4	
Side (R / L)		5 / 2	6 / 6	
Cause		4 trauma, 3 oncology	6 trauma, 2 vascular, 4 oncology	
Prosthetic foot		2 C-walk[1], 2 SACH[1], 3 Multiflex[2]	4 C-walk[1], 3 SACH[1], 1 Greissinger plus[1], 1 Multiflex[2], 2 Quantum[5], 1 S.A.F.E. II[6]	
Prosthetic knee		3 Graph-lite[7], 1 C-leg[1], 1 3R60[1], 1 Total knee[3], 1 SafeLife[4]		
AAS		35.9 ± 26.9	33.8 ± 26.1	
ABC		83.5 ± 15.9 *a	88.4 ± 5.4 *b	98.7 ± 1.0 *a,b
Leading limb pref (%)		42.9 ± 9.5	47.4 ± 6.0	54.2 ± 11.5
Gait termination velocity (ms−1)	LP	0.74 ± 0.14 *a	0.85 ± 0.21 *b	
	LN	0.75 ±0.12 *a	0.89 ± 0.23 *b	1.10 ± 0.28 *a,b
Swing phase duration (s)	LP	0.55 ± 0.07 *a,c†	0.44 ± 0.08 *c	
	TP	0.43 ± 0.06 *a,c†	0.34 ± 0.06 *c	
	LN	0.44 ± 0.09 †	0.39 ± 0.05	0.43 ± 0.04 *a,c
	TN	0.27 ± 0.05 †	0.30 ± 0.06	0.31 ± 0.04

Mean values and standard deviations of age, weight, height, time since amputation, AAS, ABC, gait termination velocity in the leading prosthetic (LP) and in the leading non-affected (LN) limb condition, and swing phase duration in the leading prosthetic (LP), the trailing prosthetic (TP), the leading non-affected (LN) and the trailing non-affected (TN) limb in the TF, TT and AB study groups. Mean values and standard error of leading limb preference for the prosthetic limb in TF and TT amputees and for the right limb in AB subjects. Gender, side and cause of amputation and the used prosthetic feet and knees in absolute numbers. Prosthetic devices by [1] Otto Bock, Duderstadt, Germany, [2] Endolite, Centerville, USA, [3] Össur, Reykjavik, Iceland, [4] Proteval, Valenton, France, [5] Hosmer, Campbell, USA, [6] Foresee, Oakdale, USA, [7] The Lin, Kuala Lumpur, Malaysia. *a Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. *b Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. *c Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences in TF and TT between the limb (condition).

Apparatus

The study was performed in a motion analysis laboratory, which is equipped with an 8.0 m long aluminium walkway and a force plate (Bertec, Columbus, USA) of 0.4 x 0.6 m. We recorded the gait pattern with two video cameras in the coronal and sagittal plane. The frame frequency was 25 Hz. In the walkway runs we collected data on leading limb preference, temporal variables and joint angles. We used six Penny & Giles SG 150 goniometers (Biometrics, Gwent, UK) in which high accuracy and repeatability were demonstrated, to measure the joint angles. The goniometers were placed on the (prosthetic) ankle, (prosthetic) knee and hip of both limbs. We calibrated the goniometers by placing the subject in an erect position with hips and knees in extension and the feet in a plantigrade position. Subjects walked with their own shoes. The soles were provided with flexible aluminium strips at the heel and forefoot. Contact of the strips with the conductive walkway indicated the timing of initial contact and toe-off. The signals of the goniometers and foot contacts were recorded by a portable data acquisition system at a sampling frequency of 800 Hz. (TMSI, Enschede, NL) The runs on the force plate were used for the assessment of the GRF and COP data. Recording, analysing and synchronising of all measurements was performed by using a custom-developed Gait Analysis System (UMCG, Groningen, NL) at 100 Hz.

Procedure

Amputees filled in two questionnaires to determine their activity level and balance confidence, respectively, the modified Amputee Activity Score (AAS)[17, 18] and the Activities-specific Balance Confidence scale (ABC). [19-21] AB subjects only filled out the latter questionnaire. Subjects performed eight runs on the walkway. They were instructed to walk at their self-selected velocity, to stop walking on their own initiative approximately halfway the walkway and to stand still for at least two seconds. The timing and exact

position of stopping were voluntary chosen. No instructions were given on which limb should be used as leading limb.

For the force plate runs, we instructed the subjects to terminate walking by stepping with the leading limb on the force plate, followed by placing the trailing limb next to the leading limb. The subjects performed repeated runs, until the prosthetic and non-affected limb were both used twice as leading limb in an adequate manner. The subjects performed at least three steps prior to the gait termination step to achieve steady state gait.[22-24] Adjustment of the step length in order to hit the force plate was avoided by practising the task in advance to select an appropriate distance from the starting point to the force plate. The subjects were instructed to look at the end of the walkway instead of at the force plate.

Outcome parameters

In amputees we analysed the outcome parameters in two limb conditions: (1) leading with the prosthetic limb and trailing with the non-affected limb and (2) trailing with the prosthetic limb and leading with the non-affected limb. Consequently, data on four lower limbs were obtained: leading prosthetic, trailing prosthetic, leading non-affected, and trailing non-affected. When amputees used the same limb as leading limb in all walkway runs, temporal variables and joint angles of the opposite leading limb could not be determined. In AB subjects we used the mean outcome of the right and left limb in the data analysis to compensate for any asymmetry in the limbs. Similar to amputees, the results for the leading and trailing limb in AB subjects were analysed separately.

We assessed the maximum amplitude of GRF in the vertical (FZ), anteroposterior (FY) and mediolateral (FX) direction. (Figure 1) The first peak GRF (FX,Y,Z1) represented the maximum exerted force by the leading limb in single-limb stance and the second peak GRF (FX,Y,Z2) the maximum produced force in bipedal stance when the trailing limb was placed. To exclude the influence of body weight we expressed GRF as a percentage of body weight (% BW). The trajectory of the COP was described by using four measuring points. (Figure 2) These points were related to the final bipedal stance position at the end of gait termination. Due to the use of a single force plate prior to initial contact of the trailing limb the COP shift under the leading limb was measured, whereas after trailing limb initial contact the resultant COP shift under both limbs was assessed.

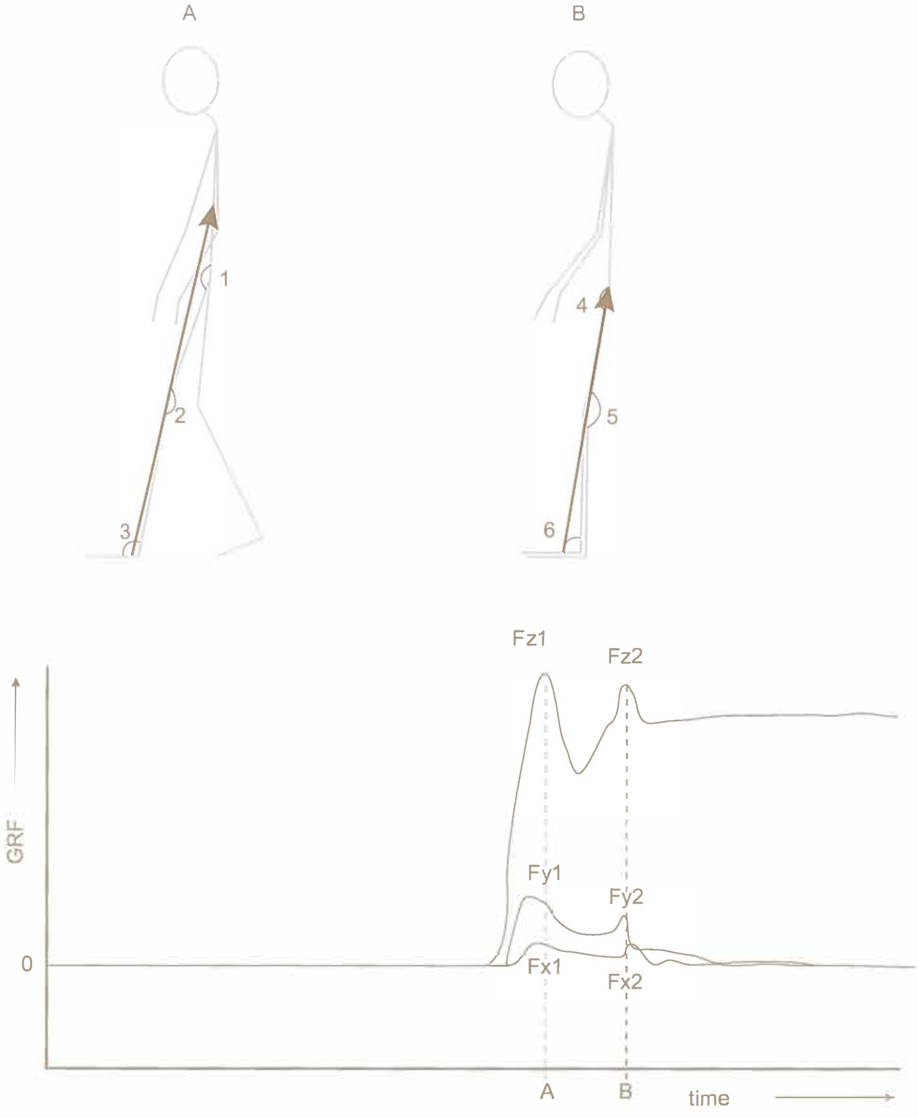


Figure 1.

Stick diagram showing the analysed lower limb joint angles and the resultant GRF and schematic illustration of the analysed peak components of the GRF in an AB subject. Fz1,2 represent the first and second peak in the vertical direction, Fy1,2 in the anteroposterior direction and Fx1,2 in the mediolateral direction. A. The hip, knee and ankle joint angles of the leading limb at the instant of trailing limb toe-off, which coincides with Fz1. B. The hip, knee and ankle joint angles of the trailing limb at the instant of trailing limb initial contact, which coincides with Fz2. The arrows represent the direction and amplitude of the resultant GRF at these events in the gait cycle.

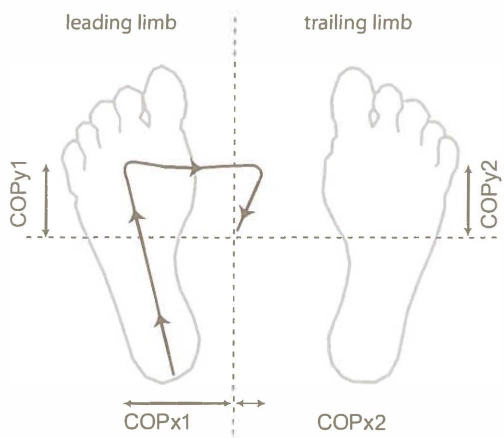


Figure 2.

Schematic representation of the COP trajectory in an AB subject. The COP moves anteriorly and toward the side of the leading limb during single-limb stance. The most anterior and lateral position is reached at midstance. Just prior to initial contact of the trailing limb the COP moves toward the trailing limb side, followed by a small posterior until the final bipedal stance position is reached. The COP trajectory was described by four distances from the final bipedal stance position to: COPy1, the most anterior position on the leading limb side; COPx1, the most lateral position on the leading limb side; COPy2, the most anterior position on the trailing limb side; COPx2, the most lateral position on the trailing limb side.

The leading and trailing limb were determined by using the video images. To obtain a leading limb preference the percentage of prosthetic leading limb runs was determined in amputees, while in AB subjects the percentage of right leading limb runs was scored. The analysed temporal variables were swing phase duration and gait termination velocity in the final stride. Gait termination velocity was assessed at leading limb initial contact and was derived from FY by integration. The joint angles of the hips, knees and ankles were analysed; for the leading limb at the moment of toe-off of the trailing limb and for the trailing limb at the moment of initial contact of the trailing limb. (Figure 1) These events of the gait cycle are critical in gait termination and coincide with the maximum peaks of the FZ.

Statistical analysis

For each limb condition the mean value of the outcome variables was calculated. Normality of the variables within the groups was tested with the Kolmogorov-Smirnov test. Differences in outcome parameters among the three groups were analysed by using an ANOVA followed by post-hoc analysis according to the least-significant difference (LSD) method. Differences in time since amputation, AAS and leading limb preference were only tested between TF and TT amputee groups. The paired t-test was used to analyse the differences between the non-affected and the prosthetic limb within TF and TT study groups. The level of significance was set on $p \leq 0.05$.

Results

Unless otherwise mentioned, only statistically significant differences are presented. Data on activity level, balance confidence, leading limb preference, gait termination velocity and swing phase duration are shown in Table 1. TF and TT demonstrated a lower ABC score than AB did. Activity level in TF and TT was quite similar. Leading limb preference revealed that TF and TT used the non-affected limb more often (but non-significantly) as leading limb than the prosthetic limb. All subjects alternated the leading limb and used both limbs at least once as leading limb in the walkway runs. In TF and TT the gait termination velocity was lower than in AB. Whether gait termination was led with the prosthetic or non-affected limb did not influence gait velocity. In TF the swing phase duration was longer in the prosthetic limb compared to their non-affected limb, and compared to TT and AB.

The joint angles are shown in Figure 3. In TF hip flexion of the leading prosthetic limb and knee flexion on the leading and trailing prosthetic side were decreased compared to TT, to AB, and to the non-affected limb in TF. In TT hip flexion in the leading prosthetic limb was reduced compared to the non-affected limb within the TT group. Knee flexion in the trailing prosthetic limb of TT was lower than in the non-affected limb and compared to AB.

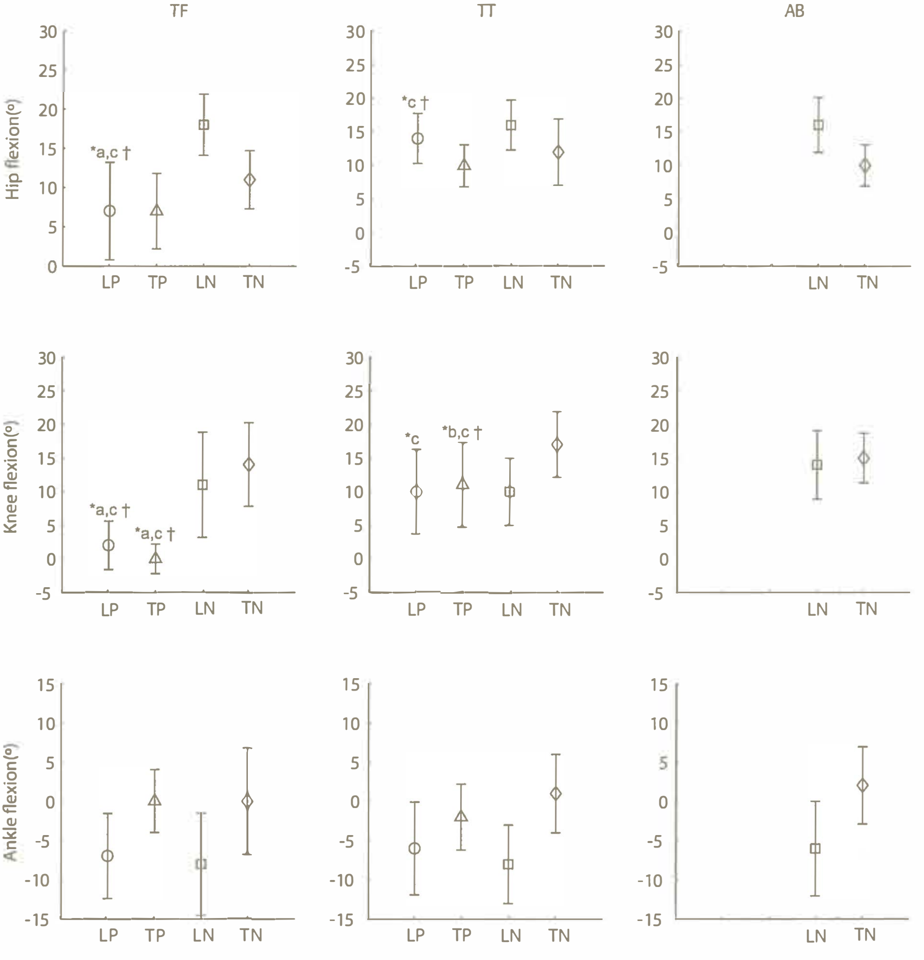


Figure 3.

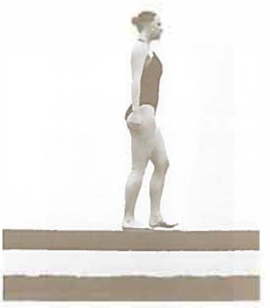
Mean values and standard deviations of hip, knee and ankle joint angles in the leading prosthetic (LP), the trailing prosthetic (TP), the leading non-affected (LN) and the trailing non-affected (TN) limb of TF, TT and AB subjects. Ankle dorsiflexion is positive, ankle plantar flexion negative. *a Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. *b Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. *c Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences between the limb (condition) in TF and TT subjects.

In Figure 4 the results of the GRF are provided. In TF and TT the FZ1 of the leading prosthetic and non-affected limb was reduced compared with the FZ1 in AB. In TT the FZ2 in trailing with the prosthetic limb was also lower than in AB. The FZ2 in trailing with the non-affected limb was larger in TF than in TT. In AB the FZ1 in the leading limb and the FZ2 in the trailing limb were similar, whereas in TF and TT the FZ2 was larger than the FZ1 in both limbs. The FY1 and FY2 in respectively leading and trailing with the prosthetic limb in TF and TT was decreased compared to the FY1 and FY2 in leading and trailing with the non-affected limb in TF and TT and in AB. The FY2 in trailing with the prosthetic limb in TF was smaller than in TT. The leading non-affected limb in TF and TT demonstrated a decreased FY1 compared to AB. In all groups the FY1 exerted by the leading limb was larger than the FY2 produced in the trailing limb. In TF and TT the trailing prosthetic limb showed a decreased FX2 compared to the non-affected limb. In the trailing non-affected limb of TF the FX2 was larger than in AB.

The results of the COP trajectory are presented in Figure 5. In comparison with AB, the COPX1 and COPX2 were increased in TF and TT when leading with the prosthetic limb. In TF the COPX2 in leading with the non-affected limb was also larger than in AB. The COPY1 of the leading prosthetic limb in TF and TT was directed posteriorly, whereas in leading with the non-affected limb in TF, TT and AB the COPY1 was located anteriorly. The posterior COPY1 in the leading prosthetic limb of TT was smaller than in TF. The COPY2 did not show significant differences in both limb conditions. Figure 6 shows a typical example of the COP trajectory of a subject in the TF group.

Discussion

The most important requirements for gait termination are the production of sufficient braking FY and an anterior displacement of COPY.[11] From our study we can conclude that amputees are only able to produce a limited amount of braking FY in the prosthetic limb. The braking FY in the prosthetic limb of TF amputees was reduced by at least 50%, and in TT amputees by approximately 33% compared to AB subjects. In the leading prosthetic limb the COPY shift remained near the heel, whereas in trailing with the prosthetic limb and in AB subjects the COPY was moved anteriorly.



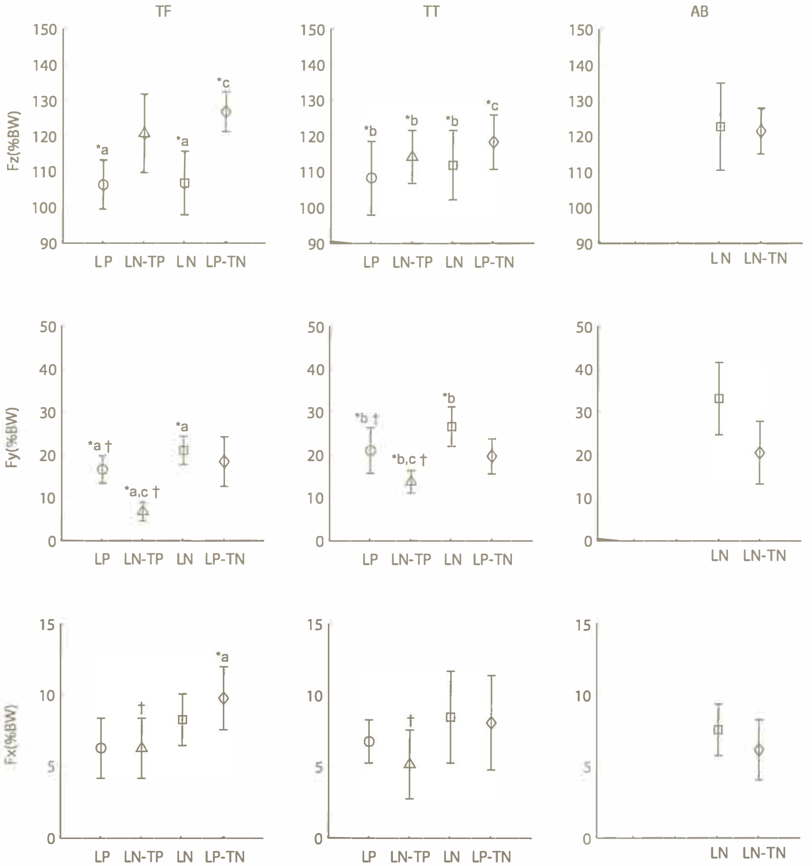


Figure 4.

Mean values and standard deviations of Fz,y,x1 produced in single-limb stance by the leading prosthetic limb (LP) and the leading non-affected limb (LN) of TF, TT and AB subjects, and of Fz,y,x2 produced in bipedal stance in the leading non-affected and trailing prosthetic limb condition (LN-TP) and in the leading prosthetic and trailing non-affected limb condition (LP-TN) of TF and TT subjects and in the leading and trailing non-affected limb condition (LN-TN) of AB subjects. Fz is positive in the upward direction, Fy in the posterior direction and Fx in the trailing limb direction. *a Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. *b Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. *c Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences between the limb (condition) in TF and TT subjects.

The limitations in function of the prosthetic limb can be explained by the deficient lower limb musculature and the different properties of a prosthetic device compared to a non-affected limb. The stiffness of a prosthetic foot impedes a smooth shift of the COPY toward the forefoot. In a prosthetic knee a locking mechanism ensures knee extension during weight bearing in early stance. Knee flexion in the trailing prosthetic limb was reduced in both amputee groups, especially TF amputees, which impedes positioning of the body behind the leading limb and lowering of the body. The reduced weight of and the more proximally located COM in the prosthetic limb compared to a non-affected limb may have contributed to the smaller FY in the prosthetic limb. However, in normal walking no significant effects of inertial prosthetic properties on GRF in amputees are found.[25]

In amputees, adjustment strategies were seen that benefit deceleration. First, in TF amputees the hip and knee in the leading prosthetic limb were more extended. Second, the lower braking FY in the prosthetic limb was compensated by a longer period of force production. Braking FY is mainly executed in single-limb stance duration of the leading limb, which is similar to swing phase duration of the trailing limb. In our study TF amputees stood longer on their non-affected leading limb and were able to increase the braking FY in this way. Third, amputees lowered gait termination velocity, resulting in a decrease in the required braking FY. We allowed the subjects to walk at their own self-selected velocity to pursue a true to nature observation. Consequently, the results may be influenced by gait velocity since joint angles and GRF depend on gait velocity.[26-28] Our hypothesis of a larger production of braking FY in the non-affected limb was not confirmed in this study.

Apart from slowing down the forward movement, the leading limb has to provide stability as well. In TF amputees swing phase duration of the prosthetic limb was prolonged, thus TF amputees spent more time in single-limb stance on the non-affected limb. Literature has shown that amputees mainly experience difficulties in stability in single-limb stance on the prosthetic limb.[14] Consequently, a longer period of single-limb stance in the non-affected limb will not endanger balance control seriously. The larger mediolateral COPX shift in amputees may be the result of decreased balance control. In leading with the prosthetic limb the mediolateral COPX shift was increased in both amputee groups, whereas in the other limb condition COPX shift was only larger in TF amputees. The larger mediolateral COPX shift can also be caused by an increase in stride or stance width.

Mediolateral COPx shift was most clearly seen in leading with the prosthetic limb during single-limb stance, which supports the hypothesis of reduced balance control in amputees being the main problem. Stance width among subjects was not standardised, because we chose to investigate self-selected gait termination.

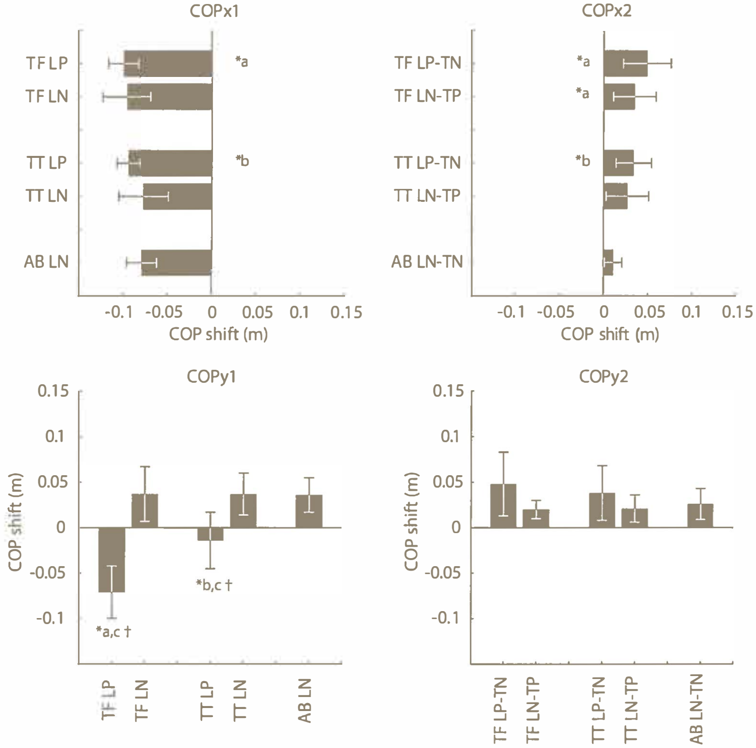


Figure 5.

Mean values and standard deviations of COPx,y1 in single-limb stance of the leading prosthetic limb (LP) and leading non-affected limb (LN) of TF, TT and AB subjects and of COPx,y2 in bipedal stance in the leading prosthetic and trailing non-affected limb condition (LP-TN) and in the leading non-affected and trailing prosthetic limb condition (LN-TP) of TF and TT subjects and in the leading and trailing non-affected limb condition (LN-TN) of AB subjects. COPx is positive in the direction of the leading limb and COPy in the anterior direction. *a Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TF subjects. *b Statistically significant p-values ($p \leq 0.05$) of group differences between AB and TT subjects. *c Statistically significant p-values ($p \leq 0.05$) of group differences between TF and TT subjects. † Statistically significant p-values ($p \leq 0.05$) of within group differences between the limb (condition) in TF and TT subjects.

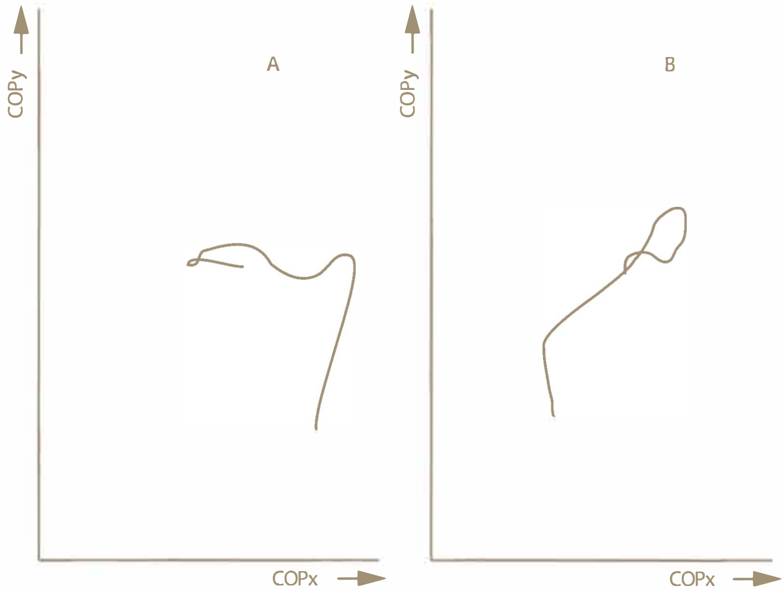


Figure 6.

Example of the COP trajectories in a TF subject. A. When leading with the right non-affected limb and trailing with the left prosthetic limb the COPy1 and COPy2 are both shifted towards the forefoot. B. When leading with the left prosthetic limb and trailing with the right non-affected limb the COPy1 under on the prosthetic limb did not move anteriorly in single-limb stance.

Amputees also used an adjustment strategy to improve balance control. Amputees loaded the non-affected limb more than the prosthetic limb. Mediolateral FX in the trailing non-affected limb was larger than in the prosthetic limb, resulting in a more lateral COPX shift in trailing with the non-affected limb. As soon as the non-affected limb was placed on the ground, amputees loaded their body weight on this limb to enhance stability. The results of the vertical FZ confirm the preference for weight bearing on the non-affected limb in amputees; vertical FZ in the non-affected limb was larger than in the prosthetic limb at the moment of trailing limb initial contact.

In addition, the small preference for the use of the non-affected limb as leading limb in amputees may represent an adjustment strategy. Amputees profit from leading with the non-affected limb, because the braking FY is larger, the COPY moves in front of the COM, and the mediolateral COPX shift is smaller. However, in real life the choice of the leading

limb in self-selected gait termination will often coincide with reaching the destination. When in daily life the stopping location is at an exact position, such as a door, chair, or wall, amputees will terminate gait with the limb that reaches that location first. Another possibility is that amputees adjust their step length prior to reaching the stopping location to emerge with the preferred limb. Amputees only have a choice in leading limb when gait termination occurs at a self-selected place and time. Therefore, it is important to train gait termination during rehabilitation, but specific advice on leading limb preference is of minor importance.

A limitation of the present study was the lack of information on leg dominance. In most subjects the amputation was already performed years ago, and therefore we could not determine the leg dominance prior to the amputation in a reliable way. Another limitation was that only outcome variables in the last step were assessed. Although most deceleration occurs in this step[7, 22, 23], other studies have shown important adjustments in the trailing limb in the step prior to termination, namely a decrease in push-off GRF.[5, 8, 9, 11] Finally, due to technical limitations the data of leading limb preference, temporal variables and joint angles were collected in different runs than the GRF and COP data. Since the walking pattern of the subjects was consistent in all runs, we assumed it was justified to analyse the data together.

Conclusion

AB subjects adjust their gait pattern to gait termination by increasing the braking FY and shifting the COPY anteriorly. In the prosthetic limb of both amputee groups the braking FY is decreased, the mediolateral COPX shift is enlarged, and in leading with the prosthetic limb the COPY is not moved toward the forefoot. Amputees used several adjustment strategies to compensate for the limitations in function. They preferred leading with the non-affected limb, prolonged the production of braking FY in the non-affected limb, decreased the gait termination velocity and loaded more body weight on their non-affected limb. It is important that amputees are trained in a gait termination task during rehabilitation. Leading with the non-affected is favourable for adequate deceleration and balance control, but in daily life not always applicable. In the future, technology that can assist in a more active role of the prosthetic foot and knee may ease gait termination in amputees.

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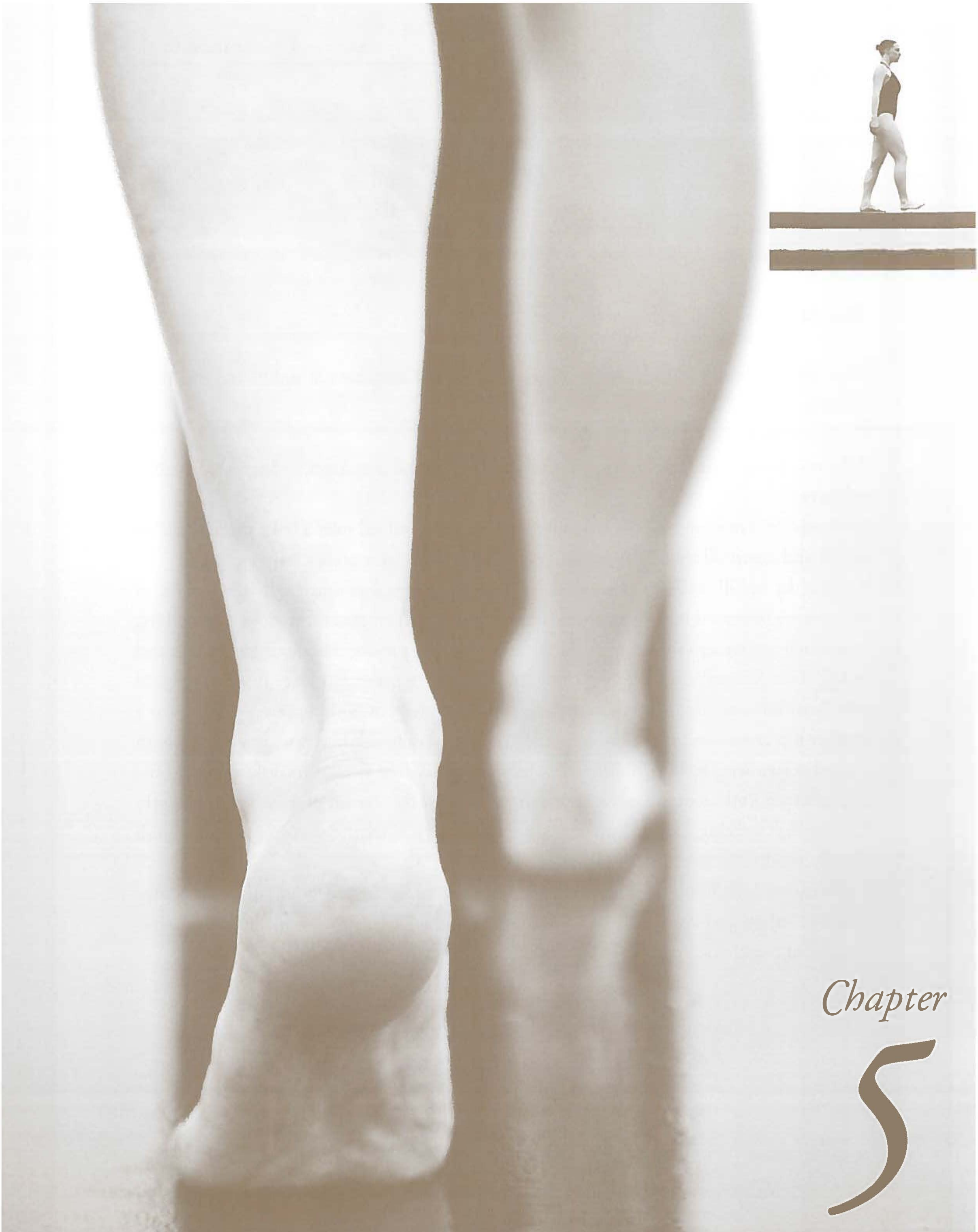

Chapter 5

Uphill and downhill walking in unilateral lower limb amputees

Aline Vrieling, Helco van Keeken, Tanneke Schoppen, Bert Otten, Jan Halbertsma, At Hof, Klaas Postema

Gait & Posture, Volume 28, Issue 2, August 2008, Pages 235-242

Abstract

Objective: To study adjustment strategies in unilateral amputees in uphill and downhill walking.
Design: Observational cohort study.
Subjects: Seven transfemoral, twelve transtibial unilateral amputees and ten able-bodied subjects
Methods: In a motion analysis laboratory the subjects walked over a level surface and an uphill and downhill slope. Gait velocity and lower limb joint angles were measured.
Results: In uphill walking hip and knee flexion at initial contact and hip flexion in swing were increased in the prosthetic limb of transtibial amputees. In downhill walking transtibial amputees showed more knee flexion on the prosthetic side in late stance and swing. Transfemoral amputees were not able to increase prosthetic knee flexion in uphill and downhill walking. An important adjustment strategy in both amputee groups was a smaller hip extension in late stance in uphill and downhill walking, probably related with a shorter step length. In addition, amputees increased knee flexion in early stance in the non-affected limb in uphill walking to compensate for the shorter prosthetic limb length. In downhill walking fewer adjustments were necessary, since the shorter prosthetic limb already resulted in lowering of the body.
Conclusion: Uphill and downhill walking can be trained in rehabilitation, which may improve safety and confidence of amputees. Prosthetic design should focus on better control of prosthetic knee flexion abilities without reducing stability.

Introduction

Since our surroundings are not all at the same level, it is important to be able to walk up and down a sloping surface. Uphill and downhill walking increases the chance of falling due to slipping or losing balance.[1] However, the human locomotor pattern is highly adaptable to changes in gradient.[2-5] Studies performed on able-bodied (AB) subjects have demonstrated adjustment strategies in walking up and down a slope.

In uphill walking AB subjects increase hip and knee flexion and ankle dorsiflexion in swing and at initial contact to provide safe foot clearance and to enable positioning of the foot on a higher surface.[3-8] The increased hip flexion may also be caused by forward bending of the trunk.[2, 3] The increased ankle dorsiflexion in stance is a direct result of the slope gradient.[6] Furthermore, knee flexion is reduced in midstance, which serves to lift the body and eases foot clearance of the other limb.[6] In downhill walking AB persons increase knee flexion from loading response to early swing. The resultant shortening of the limb lowers the body, facilitates initial contact of the other limb on the lowered surface, and reduces the impact force. Moreover, knee flexion assists in rotation of the tibia in the sagittal plane and brings the body forward over the stance foot.[1, 2, 4, 6, 7, 9, 10] Ankle dorsiflexion is increased from late stance until midswing, whereas hip flexion is decreased from midswing to early stance, which results in a pull back of the swing limb, shortening of the step length and easier positioning of the foot on the lower surface.[2, 4, 6, 7, 9]

Amputees may experience limitations in function in slope walking due to the loss of muscles, joint(s) and nerves in the amputated limb. Prosthetic knees and feet possess different properties compared to human joints, and the length of a prosthetic limb is usually shortened. Consequently, transfemoral (TF) and transtibial (TT) amputees may not be able to perform the required adjustment strategies in uphill and downhill walking, which may cause loss of balance. To date, slope walking in amputees has not been studied. The objective of this study was to determine the strategies that amputee use to adjust their gait to uphill and downhill walking.

We hypothesized that TF and TT amputees would not be able to increase flexion in the prosthetic knee and dorsiflexion in the prosthetic foot in swing and early stance in uphill walking. As an adjustment strategy, flexion in swing and early stance of the intact lower limb joint(s) proximal of the amputation level would be larger in amputees than in AB subjects. As an adjustment to the shorter prosthetic limb length, we expected an increased flexion in the non-affected lower limb joints in amputees in swing and early stance, compared to AB subjects. In downhill walking we hypothesized that prosthetic knee flexion in TF amputees would not increase in stance and early swing, and prosthetic ankle dorsiflexion in TF and TT amputees would not increase from late stance until midswing. To compensate for the higher impact force, we expected an increase in non-affected knee flexion in both amputee groups in early stance, compared to AB subjects.

Methods

Subjects

TF and TT amputees were approached via a prosthetics workshop. Inclusion criteria were: a unilateral amputation for at least one year, the use of a prosthesis on a daily base, and the ability to walk more than 50 m without walking aids. An AB control group was recruited through advertisements at the local blood bank, hospital, and television station. Subjects were excluded if they had any medical conditions that could affect their mobility or balance, such as neurological, orthopaedic or rheumatic disorders, or if they had cognitive problems or severe impaired vision. In amputees, reduced sensation of the non-affected limb, wounds or pain at the stump, or fitting problems of the prosthesis were exclusion criteria as well.

Seven TF amputees, twelve TT amputees and ten AB subjects agreed to participate in the study. The Medical Ethics Committee approved the study protocol. All subjects signed informed consent before testing. The amputees used different types of prosthetic feet. TF amputees were provided with a free moveable prosthetic knee. The Amputee Activity Score (AAS) was used to obtain information on the activity level in amputees.[11, 12] A higher score on the AAS represents a higher activity level. The subject characteristics and prosthetic components are presented in Table 1.

Apparatus

The study was performed in a motion analysis laboratory. An upward and downward ramp, both 2.0 m long, were placed halfway an 8.0 m long walkway. The gradient of the slope was 5%, which is advised in building instructions as maximum gradient.[13] The runs

Table 1. Subjects characteristics, prosthetic components and gait velocity.

		TF (n = 7)	TT (n = 12)	AB (n = 10)
Gender (M / F)		6 / 1	10 / 2	9 / 1
Age (years)		44.0 (30-71)	49.6 (27-65)	45.2 (34-63)
Weight (kg)		79.0 (67-97)	84.2 (71-98)	86.5 (72-98)
Height (cm)		182.6 (174-194)	180.9 (165-194)	184.4 (172-192)
Time (months)		210.7 (18-504)	207.8 (23-672)	
Cause		4 trauma, 3 oncology	6 trauma, 4 oncology, 2 vascular	
Prosthetic foot		2 C-walk[1], 2 SACH[1], 3 Multiflex[2]	4 C-walk[1], 3 SACH[1], 1 Greissinger plus[1], 1 Multiflex[2], 2 Quantum[5], 1 S.A.F.E. II[6]	
Prosthetic knee		3 Graph-lite[7], 1 C-leg[1], 1 3R601, 1 Total knee[3], 1 SafeLife[4]		
AAS		35.9 ± 26.9	33.8 ± 26.1	
Gait velocity (ms−1)	Level	1.03 ± 0.19	1.22 ± 0.16	1.34 ± 0.13
	Slope	1.01 ± 0.23	1.21 ± 0.16	1.34 ± 0.14

Mean values and range of age, weight, height, and time since amputation, and mean values and standard deviations of Amputee Activity Scale (AAS) and gait velocity of the TF, TT and AB study groups. Prosthetic devices by [1] Otto Bock, Duderstadt, Germany, [2] Endolite, Centerville, USA, [3] Össur, Reykjavik, Iceland, [4] Proteval, Valenton, France, [5] Hosmer, Campbell, USA, [6] Foresee, Oakdale, USA, [7] The Lin, Kuala Lumpur, Malaysia.

were videotaped with a camera in the sagittal plane, which moved with the subject along the walkway. The recording frequency was 25 Hz. Subjects were provided with six SG 150 Penny & Giles goniometers (Biometrics, Gwent, UK), for which a high accuracy and repeatability were proven. The goniometers were placed bilaterally on the ankle joint (or prosthetic foot), knee joint (or prosthetic knee), and hip joint. Joint angles were measured in the sagittal plane. Prior to testing, the goniometers were zeroed while the subjects stood upright with both limbs straight and ankles in a plantigrade position. The beginning and end of the walkway were fitted with infrared lights, which registered the passing of a subject. Data were recorded by a portable data acquisition system (TMSI, Enschede, NL) and analysed with a custom-developed Gait Analysis System (GAS, UMCG, Groningen, NL). The sampling frequency was 800 Hz, which was low-passed filtered and resampled to 100 Hz. The goniometer data were filtered by a low-pass second-order Butterworth filter with a cut-off of 10 Hz.

Procedure

All subjects were instructed to walk at their self-selected comfortable velocity. Three different walking conditions were performed; level, uphill and downhill walking. The study consisted of four level and four slope walking runs. In slope walking the subjects first walked up and then down the slope.

Outcome parameters

Average gait velocity in level and slope runs was calculated from the length of the 8.0 m walkway divided by the necessary time to walk over this walkway. In slope walking the mean gait velocity of uphill and downhill walking was assessed. The joint angles of the prosthetic and non-affected limbs in amputees were analysed separately. In AB subjects the mean joint angles of the right and left limb were used in the analysis to minimize the influence of asymmetry. In slope walking, the middle stride on the slope trajectory was selected. In level walking three successive strides in the middle of the walkway were analysed. The joints angles of the hip, knee and ankle were assessed at several events in the gait cycle and these are shown in Figure 1.

Statistical analysis

Data were tested with the Kolmogorov-Smirnov test and were normally distributed. Significant differences in subject characteristics were tested with ANOVA. A mixed design ANOVA with repeated measures was performed on gait velocity and joint angles, followed by post-hoc analysis according to the least-significant difference (LSD) method. This analysis allowed adjustment for differences in outcome parameters among study groups that already existed in level walking and detection of differences in outcome parameters that were specific to uphill and downhill walking. The joint angles in both the prosthetic and non-affected limbs of TF and TT amputees were compared with the non-affected limb of the AB study group. Within-subject variables included the outcome parameters in the different walking conditions (level-uphill and level-downhill), and the between-subject factors were the study groups. The level of significance was set in all analyses at $p \leq 0.05$.

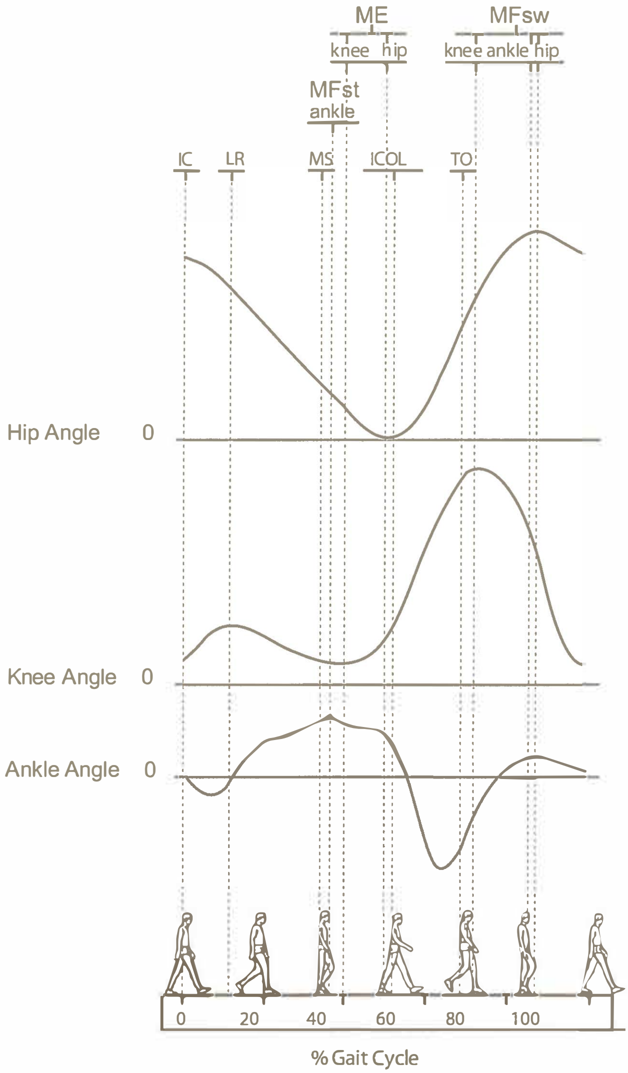

Figure 1.

The analysed angles of the hip, knee and ankle at initial contact (IC), loading response (LR), midstance (MS), maximum dorsiflexion in stance (MFst), maximum extension in stance (ME), initial contact of the opposite limb (ICOL), toe-off (TO) and maximum dorsi(flexion) in swing (MFsw). LR was assessed at maximum knee flexion in stance and MS when the ankle of the swing limb passed the stance limb.

Results

No statistically significant differences were found in the characteristics of the study groups. Gait velocity was similar in level and slope walking within the study groups. Gait velocity was increased in AB compared to TF and TT, and TT walked faster than TF. (Table 1) The lower limb joint angles are shown in Figures 2-4. Only significant interaction effects in joint angles between the study groups in uphill and downhill walking compared to level walking are reported in this section.

Uphill walking - prosthetic limb in amputees

In the hip an interaction effect was shown at IC ($F_{9.8}$, $p < 0.01$), ICOL ($F_{5.5}$, $p = 0.01$) and MFSW ($F_{11.4}$, $p < 0.01$). In uphill walking hip flexion at IC and MFSW increased in AB and TT, which was not seen in TF. Hip extension at ICOL decreased in TT in uphill walking, but not in TF and AB. In the knee an interaction effect was demonstrated at IC ($F_{6.8}$, $p < 0.01$) and MFSW ($F_{4.1}$, $p = 0.03$). Knee flexion at IC in uphill walking was enlarged in AB and TT, but not in TF. In TF knee flexion at MFSW was reduced in uphill walking, which was not demonstrated in AB and TT. The ankle showed no interaction effects.

Uphill walking - non-affected limb in amputees

The hip showed an interaction effect at MEst ($F_{3.8}$, $p = 0.04$). In TF hip extension was decreased at MEst in uphill walking, but not in AB and TT. In the knee an interaction effect was seen at LR ($F_{3.7}$, $p = 0.04$) and TO ($F_{6.4}$, $p < 0.01$). TF and TT increased knee flexion more than AB did at LR in uphill walking. TF increased knee flexion at TO in uphill walking, which was not shown in AB and TT. In the ankle an interaction effect was demonstrated at TO ($F_{5.5}$, $p = 0.01$). TF increased ankle plantar flexion at TO in uphill walking, AB and TT did not.

Downhill walking - prosthetic limb in amputees

In the hip an interaction effect was shown at ICOL ($F_{6.1}$, $p < 0.01$) and MFSW ($F_{3.6}$, $p = 0.04$). Hip extension at ICOL in AB and TF increased slightly in downhill walking, whereas in TT a decrease was shown. In downhill walking hip flexion at MFSW was reduced in TF, but not in AB and TT. The knee showed a significant interaction effect at MEst ($F_{3.7}$, $p = 0.04$) and MFSW ($F_{4.1}$, $p = 0.03$). In downhill walking AB and TT decreased knee extension

at MESt, but this was not seen in TF. In AB and TT knee flexion at MFSW in downhill walking was increased, in TF reduced. In the ankle no interaction effects were found.

Downhill walking - non-affected limb in amputees

No significant interaction effects were demonstrated in this limb.

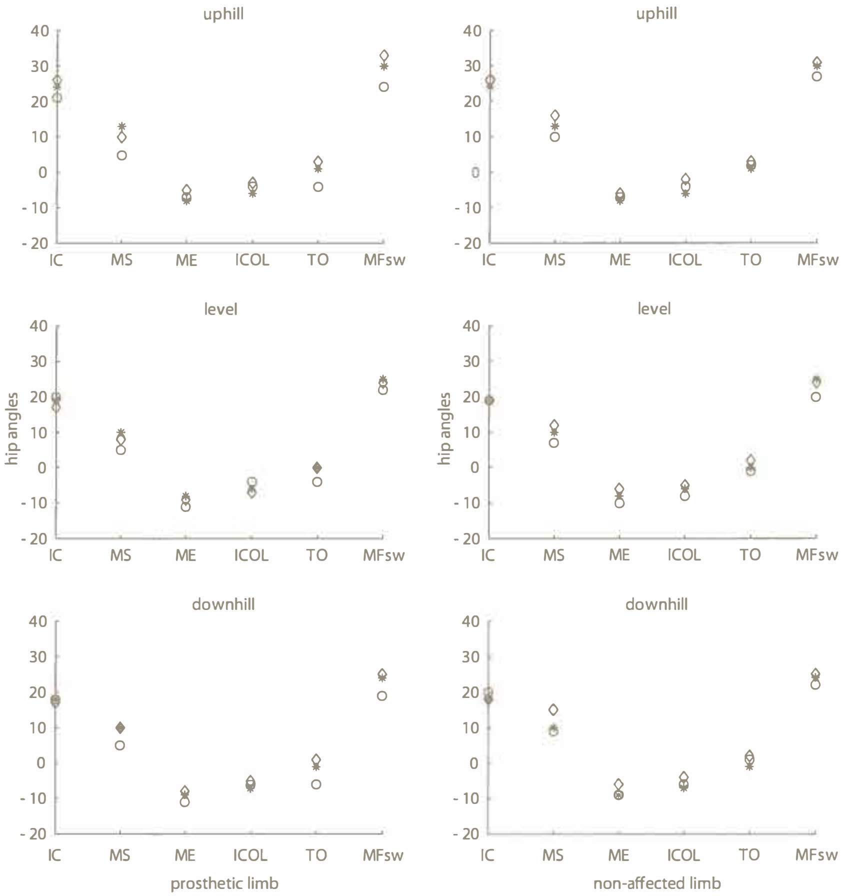

Figure 2.

Mean values of hip angles in the prosthetic and non-affected limb in TF(o), TT(◇) and AB(*) subjects during uphill, level and downhill walking. Hip flexion is positive, extension negative.

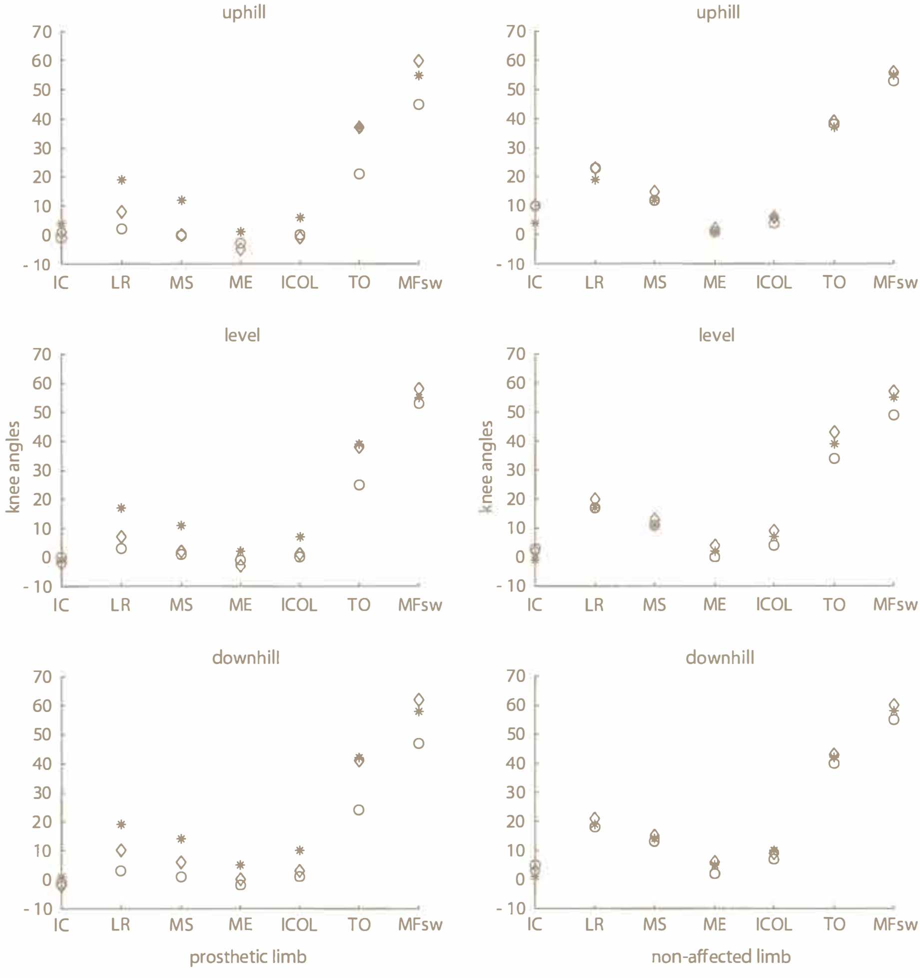

Figure 3.

Mean values of knee angles in the prosthetic and non-affected limb in TF(o), TT(◇) and AB(*) subjects during uphill, level and downhill walking. Knee flexion is positive, extension negative.

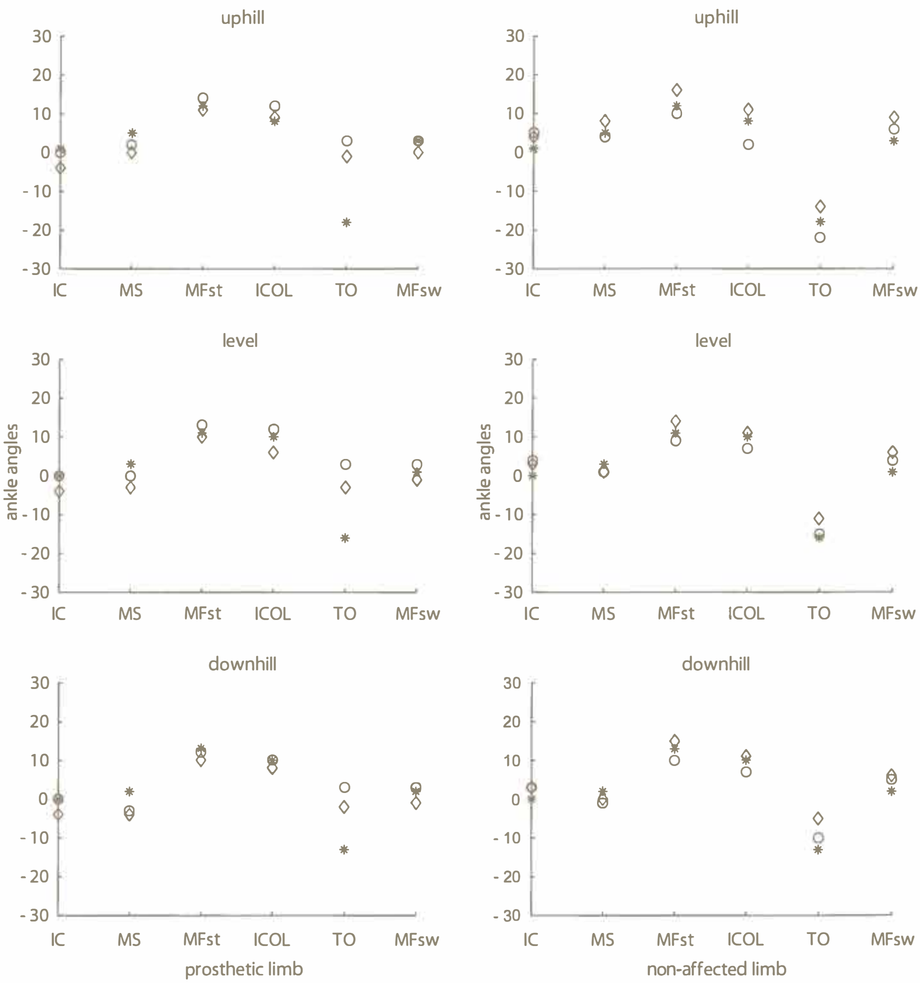

Figure 4.

Mean values of ankle angles in the prosthetic and non-affected limb in TF(o), TT(◊) and AB(*) subjects during uphill, level and downhill walking. Ankle dorsiflexion is positive, plantar flexion. negative.

Discussion

The main goal of this study was to establish how amputees adjust their gait pattern to uphill and downhill walking. Adjustments strategies can involve the prosthetic or the non-affected limb. The adjustment strategies on the prosthetic side of TT amputees were very similar to those in AB subjects. In uphill walking hip and knee flexion at initial contact and hip flexion in swing were increased in the prosthetic limb of TT amputees. In downhill walking TT amputees showed more knee flexion on the prosthetic side in late stance and swing.

As hypothesized, TF amputees were not able to increase prosthetic knee flexion in uphill and downhill walking. For example, maximum swing knee flexion in the prosthetic limb in uphill walking was 8° lower than in level walking. Prosthetic knee flexion in swing relies on activity of the hip flexors and the properties of the prosthetic knee. In stance, prosthetic knee flexion is only possible to a limited extent, depending on the type of prosthetic knee, to prevent unlocking during weight bearing. In this study one subject was fitted with a microprocessor-controlled prosthetic knee joint. Advantages of such a knee joint are the ability to flex in the beginning of stance and to decrease damping in late stance, which respectively contribute to shock absorption in loading response and to knee flexion in swing. The subject provided with a microprocessor-controlled prosthetic knee showed an increased knee flexion in late stance and swing in level, uphill and downhill walking compared to TF amputees with a conventional prosthetic knee, but at loading response no differences were observed.

Since the prosthetic knee offers no possibility to adjust gait to slope walking, TF amputees are required to make use of other strategies. Adjustment strategies can be generated in the hip on the prosthetic side. We hypothesized that TF amputees would increase hip flexion in the prosthetic limb in swing and initial contact to provide safe foot clearance and foot positioning in uphill walking, but this adjustment strategy was not found. In downhill walking TF amputees showed a smaller hip flexion in the prosthetic limb in swing, which could indicate a shorter step length. By reducing the step length, positioning of the prosthetic foot on the lowered surface is eased, because the difference in height is smaller.

In TT amputees the hip on the prosthetic side was used for another adjustment strategy. TT amputees decreased hip extension on the prosthetic side at initial contact of the non-affected limb in uphill and downhill walking. In uphill walking the reduced hip extension in late stance facilitates positioning of the opposite non-affected foot on the higher surface, because the body remains lifted. The reduced hip extension can be caused by a shorter step length, which decreases the height difference that the prosthetic limb has to adjust to. Furthermore, a forward bending of the trunk may add to the decrease in hip extension. In downhill walking the reduced hip extension is most likely explained by a shorter step length. However, reduced hip extension in stance is not beneficial for lowering of the body, which is required in downhill walking.

Uphill and downhill walking did not result in major adjustments in the prosthetic ankle of amputees. Dependent on the stiffness of the prosthetic foot, a passive adaptation to the slope gradient is possible in stance. Thirteen amputees in our study used relatively flexible prosthetic feet, whereas six amputees were provided with more rigid prosthetic feet. Due to the majority of flexible prosthetic feet, amputees could achieve an increase in dorsiflexion that was similar to AB subjects in early stance in uphill walking and in late stance in downhill walking.

Amputees can also use the non-affected limb to carry out adjustment strategies. In uphill walking TF amputees decreased maximum hip extension in the non-affected limb, which led to an easier positioning of the prosthetic foot. The decrease in hip extension in this limb can be interpreted similarly to the prosthetic limb, which we mentioned earlier. A second adjustment strategy that TF amputees applied in the non-affected limb in uphill walking was increasing knee flexion and ankle plantar flexion at toe-off, which may assist in loading of the prosthetic limb. Since more height has to be overcome, it is harder in uphill walking to shift the body weight above the prosthetic limb.

In both amputee groups, knee flexion in the non-affected limb at loading response was increased in uphill walking. Since the shorter length of the prosthetic limb restricted lifting of the body, amputees needed to place the non-affected limb in a more flexed position on the elevated surface. In downhill walking no specific adjustment strategies were seen in the non-affected limb of amputees. We hypothesized that an increase in non-affected knee flexion would compensate for the higher impact force. Shortening of the stance limb to lower the body is an important feature in downhill walking and the more extended position of the prosthetic limb would force the non-affected limb to arrive at the ground from a higher position. However, in amputees the shorter prosthetic length already ensured lowering of the body.

Knee extension in stance results in positive mechanical work in uphill walking, which is essential for moving the body up the slope against gravity. The already straightened position of the prosthetic knee at loading response made a large increase in positive work in TF amputees not possible. A conventional prosthetic knee can only provide a small amount of positive work due to energy storage in the extension spring. The larger knee flexion of the non-affected limb at loading response in amputees increased the amount of positive work in this limb and enabled an adequate lift of the body in amputees. In this way, one adjustment strategy, namely increased knee flexion in the non-affected limb at loading response, compensated for both the shorter length and the limited flexion of the prosthetic limb. In downhill walking knee flexion in late stance creates negative mechanical work, which is necessary for lowering of the body. In TF amputees the increase in flexion in the prosthetic knee was lower than in AB subjects and TT amputees. However, TF amputees were the only study group that showed an increase in the production of negative work on the prosthetic side when walking downhill compared to level walking, because in TF amputees maximum prosthetic knee extension was not decreased in downhill walking.

Since the differences in lower limb joint angles between the conditions and study groups were mostly limited to several degrees, the clinical relevance can be questioned. However, the gradient of the slope was only 5% and walking up and down steeper slopes may give rise to more clinically important results. Increasing the steepness of the slope may result in the usage of other adjustment strategies in amputees and AB subjects. Finally, the study was limited by the different types of prosthetic knees and feet, which may have led to the application of different adjustment strategies. Due to the small sample size we were not able to study the influence of the diverse prosthetic devices on the outcome measures.

Conclusion

An intact knee joint is important for walking up and down a hill safely. TT amputees increased knee flexion in the prosthetic limb to adjust gait to the slope gradient, whereas

in the prosthetic knee of TF amputees this strategy was absent. To improve safety and confidence of amputees during walking up and down a hill, training of these motor tasks in rehabilitation is recommended in order to practise adjustment strategies on slopes of different gradients. Prosthetic knee design should focus on better flexion properties in stance and swing without compromising stability requirements.

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Chapter 6

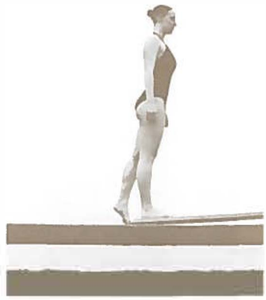
Balance control on a moving platform in unilateral lower limb amputees

Aline Vrieling, Helco van Keeken, Tanneke Schoppen, Bert Otten, Jan Halbertsma, At Hof, Klaas Postema

Gait & Posture, Volume 28, Issue 2, August 2008, Pages 222-228

Abstract

Objective: To study balance control on a moving platform in lower limb amputees.
Design: Observational cohort study.
Subjects: Unilateral transfemoral and transtibial amputees and able-bodied control subjects.
Methods: Balance control on a platform that moved in the anteroposterior direction was tested with eyes open, blindfolded and while performing a dual task.
Outcome measures: Weight bearing symmetry, anteroposterior ground reaction force and centre of pressure shift.
Results: Compared to able-bodied subjects, in amputees the anteroposterior ground reaction force was larger in the prosthetic and non-affected limb, and the centre of pressure displacement was increased in the non-affected limb and decreased in the prosthetic limb. In amputees body weight was loaded more on the non-affected limb. Blindfolding or adding a dual task did not influence the outcome measures importantly.
Conclusion: The results of this study indicate that experienced unilateral amputees with a high activity level compensate for the loss of ankle strategy by increasing movements and loading in the non-affected limb. The ability to cope with balance perturbations is limited in the prosthetic limb. To enable amputees to manage all possible balance disturbances in real life in a safe manner, we recommend to improve muscle strength and control in the non-affected limb and to train complex balance tasks in challenging environments during rehabilitation.



Introduction

Maintenance of balance is necessary during activities in daily life. In able-bodied (AB) individuals the ankle joint and lower leg musculature play an essential role in maintaining balance by appropriately shifting the centre of pressure (COP).[1, 2] Muscle contractions produce a torque around the ankle, which in turn generates changes in the COP and the direction of ground reaction forces (GRF) and modulates the anteroposterior movements of the centre of mass (COM).[2-4] Following lower limb amputation, somatosensory input, muscle activity and joint mobility in the amputated part of the limb are compromised. As a consequence, lower limb amputees are unable to use the same motor strategies for balance control as AB subjects and therefore have to adjust the habitual stance control strategies and develop new strategies.[3, 5, 6]

To date, research concerning balance control in amputees has mainly focused on sway in quiet standing. The results of these studies are contradictory. In several studies the postural sway in individuals with recent or long-standing amputation was increased compared with AB subjects[2, 3, 7-10], whereas other studies found no differences.[11-14] A single force plate was often used for balance measurements and the prosthetic and non-affected limbs were analysed together without taking into account the different properties.[2, 3, 8, 10-12] Studies that analysed the non-affected and prosthetic limbs separately showed a decrease in weight bearing and COP excursions in the prosthetic limb.[9, 15-17] Research has revealed that a good standing balance on the non-affected limb is beneficial for the functional outcome of amputees.[18]

Static balance tests may not be sufficiently challenging to detect essential strategies for maintaining balance in daily activities, since balance control is often required during ambulation.[3, 4, 19] Falls regularly occur when balance control is hindered by an external perturbation.[20] A moving platform is a common method to study perturbations in balance.[1, 21-24] Moving the platform displaces the COP away from the projection of the body's COM on the ground. To regain equilibrium, balance control strategies are used to shift the COM in the same direction as the platform displacement.[1, 25] GRF are used to adjust the movements of the COP and COM to the perturbations. It is known that amputees experience most difficulties in balance control in the anteroposterior direction.[3]

Apart from the motor control system, balance control in daily life is also dependent on the sensory, visual, cognitive and vestibular systems.[13, 25] Humans are able to switch between these balance control systems to compensate for a deficiency in one of the systems or to adjust to the environmental demands. To mimic balance performance in daily life it is important to assess motor skills in combination with other tasks.[19, 26] In this way more subtle differences in balance performance between study groups can be detected.[27]

In previous studies on quiet standing in amputees, balance control was made more difficult by closing the eyes and adding a dual task. Mean COP sway of both limbs and loading on the non-affected limb in amputees clearly increased when the eyes were closed[8-11, 15, 17], whereas in AB subjects only a small [8-10] or no[11] effect was found. This would suggest that in amputees an increased contribution of visual control compensated for the impairment within the somatosensory system. Adding a dual task increased postural sway in amputees more than in AB subjects, implicating that the maintenance of balance was not fully automated in amputees and required conscious control.[2]

In this study we focused on the performance of more complex balance tasks in amputees by moving a platform, depriving vision and adding a dual task. The first aim was to establish the balance control strategies of the prosthetic and non-affected limbs in amputees and of AB subjects during standing on a moving platform. We hypothesised that in amputees the vertical and anteroposterior GRF and the anteroposterior COP displacement would increase in the non-affected limb and decrease in the prosthetic limb compared to AB subjects. The second aim was to study the influence of visual deprivation and an acoustic dual task on balance control strategies. We hypothesized that amputees would increase the anteroposterior GRF and COP displacement in both limbs and would shift the vertical GRF more toward the non-affected limb when vision is deprived or a dual task is added compared to the normal condition.

Methods

Subjects

Amputees were approached via a prosthetics workshop. Inclusion criteria for the amputee study group (AMP) were age over 18 years, a unilateral lower limb amputation at least one year earlier, the use of a prosthesis on a daily basis, and the ability to stand with a prosthesis

without walking aids for at least 30 minutes. An AB control group was recruited through advertisements at the local blood bank, hospital, and television station. Subjects were excluded if they had any medical conditions that could affect their mobility or balance, such as neurological, orthopaedic or rheumatic disorders, the use of antipsychotic drugs, antidepressants or tranquillisers, otitis media, or impaired vision. Additional exclusion criteria for amputees included reduced sensation of the non-affected limb, ulceration or pain at the stump, or fitting problems of the prosthesis.

Eight AMP subjects and nine AB subjects agreed to participate in the study. In the AMP study group three patients with a transfemoral (TF) and five patients with a transtibial (TT) amputation were included. The Medical Ethics Committee approved the study protocol and all subjects signed informed consent before testing. AMP subjects used their own prosthetic limb. The different types of prosthetic feet and knees are detailed in Table 1. To obtain information on functional skills, AMP subjects filled in the modified Amputee Activity Score (AAS), a suitable measure for outpatient amputees with good test-retest reliability and validity.[28, 29] The range of the score is - 70 to + 50 and a higher score represents a higher activity level. Both groups filled out the Activities-Specific Balance Confidence Scale (ABC), which is designed to assess balance confidence when performing activities such as climbing stairs, reaching above the head, and walking on different surfaces. The maximum score is 100. The ABC is shown to be reliable and there is strong support for validity.[30, 31]

Apparatus

Balance measurements were performed on the Computer Assisted Rehabilitation Environment system (CAREN)[32] which consists of a 2.0 m diameter platform that can rotate around three orthogonal axes and translate in three directions along these axes. The platform contains two built-in 0.40 x 0.60 m force plates (Bertec, Columbia, USA) to register GRF with a sampling frequency of 100 Hz. COP data were derived from the GRF and platform moment of force data.

Procedure

Subjects stood erect on the moving platform with their hands alongside their bodies. For reasons of safety subjects were provided with a safety belt that was connected to the

Table 1. Subject characteristics.

	AMP (n = 8)	AB (n = 9)	p-value groups
Gender (M / F)	6 / 2	8 / 1	
Age (years)	51.8 ± 12.7	44.8 ± 9.9	0.239
Weight (kg)	83.3 ± 9.7	85.6 ± 9.1	0.622
Height (m)	1.78 ± 0.09	1.84 ± 0.07	0.142
Level	3 TF, 5 TT		
Time (months)	257.5 ± 195.6		
Cause	5 trauma, 1 vascular, 2 tumour		
Prosthetic foot	3 SACH[1], 2 C-walk[1], 1 Multiflex[2], 1 Greissinger plus[1], 1 Quantum[3]		
Prosthetic knee	1 3R60[1], 1 Total knee[4], 1 Graph-lite[5]		
ABC	84.5 ± 13.0	98.8 ± 1.1	0.017 *
AAS	32.0 ± 31.7		

Mean values and standard deviations of age, weight, height, time since amputation, the AAS and ABC of the AMP and AB study group. Gender, level and cause of amputation, and the used prosthetic feet and knees in absolute numbers. Prosthetic devices by [1] Otto Bock, Duderstadt, Germany, [2] Endolite, Centerville, USA, [3] Hosmer, Campbell, USA, [4] Össur, Reykjavik, Iceland, [5] The Lin, Kuala Lumpur, Malaysia. * Statistically significant p-values ($p \leq 0.05$) of group differences.

ceiling. The feet were placed in a self-selected position, one on each force plate. Subjects were instructed to stand with both feet on the floor as motionless as possible and to maintain balance while the platform swayed for 60 s in the anteroposterior direction. The platform movements were sinusoidal. During the first 15 s the excursions gradually increased to a maximum amplitude of 0.02 m. Maximum platform excursions were executed from the 15th to the 45th s, after which the excursions slowly diminished towards the end of the test. The frequency of the excursions was 1 Hz. The mean anteroposterior velocity was 0.046 ms^{-1}, the maximum velocity 0.13 ms^{-1} and the maximum acceleration 0.79 ms^{-2}. Between the tests a pause of 60 s was allowed.

Subjects were tested in three conditions: (1) normal: single task with eyes open, (2) blindfolded: diving goggles with non-transparent black glasses, and (3) dual task: adding the acoustic Stroop test[33, 34]. In this test the words "high" and "low" were pronounced in

a high or low pitch. Subjects had to name the pitch in which the word was spoken and suppress the tendency to repeat the word they heard. Prior to the balance tests the Stroop test was practiced once. The conditions were presented in random order to avoid learning effects.

Outcome parameters

MATLAB software was used for data analysis. Since we were interested in the period of maximum platform excursions, the first and last 15 s of the balance recordings were excluded and a period of 30 s remained. Balance control was described by three parameters: 1) the weight bearing index (WBI) as a measure for symmetry in body weight distribution, which was calculated from the vertical component of GRF (FZ), 2) the anteroposterior component of GRF (FY), and 3) the anteroposterior COP displacement (COPY). WBI in AMP subjects was the ratio of FZ in the non-affected limb divided by FZ in the prosthetic limb. A ratio score is more often used to quantify limb asymmetries.[35] To calculate WBI in AB subjects the limb with the largest FZ was divided by FZ in the other limb. In the AMP study group the COPY and FY data of the prosthetic and non-affected limbs were analysed separately, whereas in the AB control group the mean of the right and left limbs was used. COPY was defined as the sum of absolute values of the COPY differences, and FY and FZ as the sum of absolute FY and FZ values.

Statistical analysis

The Kolmogorov-Smirnov test showed that all data were normally distributed and Levene's test showed that variances in COPY and FY were equal, whereas in WBI heterogeneity of variance was seen. Differences between the groups were analysed by the independent t-test; in COPY and FY with equal variances, in WBI with unequal variances. Differences between conditions (normal-blindfolded and normal-dual task) within the groups were analysed by the paired t-test. The level of significance was set on $p \leq 0.05$.

Results

Characteristics of the subjects are presented in Table 1. Apart from a higher score on the ABC in AB no statistically significant differences in subject characteristics were found between AMP and AB. The results of WBI, FY and COPY are presented in Tables 2–4. All subjects were able to maintain balance during the tests without taking a step. AMP significantly preferred to bear weight on their non-affected limb in all three conditions. Wbi in amp was significantly more asymmetric than in ab, although in ab weight bearing was not equally divided between the two limbs. In amp 62–63% of the body weight was loaded on the non-affected limb. Fy in the non-affected limb of amp was significantly larger in comparison with ab in all three conditions. Fy in the prosthetic limb of amp was also larger than in ab, but only significantly in the normal condition. In the normal and dual task conditions copy of the non-affected limb in amp was significantly larger than in ab. Copy under the prosthetic limb in amp was lower than in ab in the blindfolded and dual task conditions. A typical example of fy and copy in the prosthetic and non-affected limbs of a subject in the amp group during the normal condition is presented in Figure 1.

Table 2. Weight bearing index.

	AMP (n = 8)	AB (n = 9)	p-value groups
WBI normal	1.65 ± 0.42	1.15 ± 0.14	0.025 *
WBI blindfolded	1.67 ± 0.49	1.17 ± 0.15	0.008 *
p-value normal-blindfolded	0.812	0.465	
WBI dual	1.69 ± 0.49	1.19 ± 0.18	0.010 *
p-value normal-dual task	0.755	0.173	

Mean values and standard deviations of the WBI in the AMP and AB study group. * Statistically significant p-values ($p \leq 0.05$) of group differences.

The only significant effect of condition was demonstrated in ab in which copy in the blindfolded condition was increased compared to the normal condition. In wbi and fy there were no significant differences between the normal, blindfolded and dual task conditions.

Discussion

The first aim of this study was to establish balance control strategies on a moving platform in amputees. In our study, amputees loaded 37–38% of their body weight on their prosthetic limb, whereas studies on quiet standing reported that tt amputees loaded approximately 45% and tf amputees 40% of their body weight on the prosthetic limb.[15,

Table 3. Horizontal ground reaction forces.

	Limb	AMP (n = 8)	AB (n = 9)	p-value groups
Fy normal (% BW)	non-affected	33.9 ± 4.5	23.1 ± 3.3	0.000 *
	prosthetic	30.9 ± 8.7		0.022 *
Fy blindfolded (% BW)	non-affected	36.6 ± 7.8	23.7 ± 4.8	0.001 *
	prosthetic	33.0 ± 13.5		0.065
p-value normal-blindfolded	non-affected	0.119	0.763	
	prosthetic	0.388		
Fy dual (% BW)	non-affected	32.1 ± 9.0	22.1 ± 5.1	0.013 *
	prosthetic	29.7 ± 15.3		0.188
p-value normal-dual task	non-affected	0.435	0.633	
	prosthetic	0.698		

Mean values and standard deviations of Fy of the prosthetic and non-affected limb in the AMP and AB study group. Fy was expressed in % body weight. * Statistically significant p-values ($p \leq 0.05$) of group differences.

Table 4. Centre of pressure differences.

	Limb	AMP (n = 8)	AB (n = 9)	p-value groups
ΔCOP normal (m)	non-affected	3.38 ± 1.69	1.91 ± 0.62	0.027 *
	prosthetic	1.36 ± 0.41		0.053
ΔCOP blindfolded (m)	non-affected	4.28 ± 2.18	2.82 ± 0.87	0.082
	prosthetic	1.39 ± 0.41		0.001 *
p-value normal-blindfolded	non-affected	0.063	0.001 †	
	prosthetic	0.546		
ΔCOP dual (m)	non-affected	3.47 ± 1.67	2.14 ± 0.61	0.043 *
	prosthetic	1.30 ± 0.30		0.003 *
p-value normal-dual task	non-affected	0.689	0.222	
	prosthetic	0.352		

Mean values and standard deviations of COPy of the prosthetic and non-affected limb in the AMP and AB study group. COPy was expressed in m. * Statistically significant p-values ($p \leq 0.05$) of group differences. † Statistically significant p-values ($p \leq 0.05$) of differences between the normal and blindfolded condition within the groups.

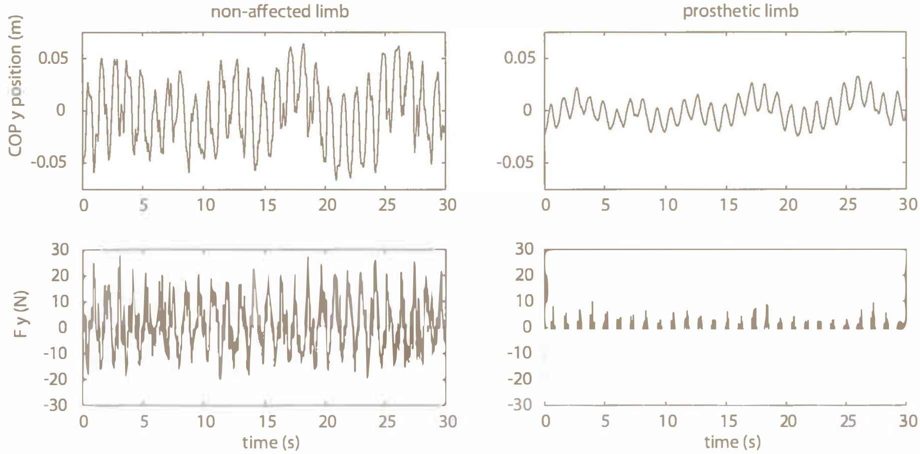


Figure 1.
A typical example of the COPy and Fy in a subject with a TF amputation. COPy and Fy in the non-affected limb are larger than in the prosthetic limb during the 30 s measuring time.

[17, 36] Hence, loading of the non-affected limb seems to increase slightly when balance is perturbed, compared to quiet standing. Various explanations have been suggested for the asymmetric weight bearing strategy in amputees; reduced ankle mobility, stump pain, discomfort due to the rigid prosthetic socket or prosthetic alignment, poor hip abductor muscle strength, inadequate sensory information, lack of confidence, poor balance, or habitual stance.[11, 15, 16] Whereas the advantage of increased weight bearing on the non-affected limb is improved control, the disadvantage is more frequent overloading and arthritis of the non-affected limb.[11, 15, 37]

In the present study, COPY in the prosthetic limb was limited, which can be explained by the absent ankle musculature, the deficient flexibility of the prosthetic foot and the decreased weight bearing on this limb. As an adjustment strategy amputees increased COPY in the non-affected limb, which can be explained by an increased muscle activity in this limb and the trunk. In the normal condition COPY in the non-affected limb of amputees was increased by a factor 2.5 compared to the prosthetic limb, and by a factor 1.8 compared to AB subjects. Earlier studies on quiet standing in amputees showed similar COP results; in the non-affected limb of experienced amputees COP excursions were approximately

twice as large as in the prosthetic limb[15,17]. In subjects with a recent amputation COP velocity in the non-affected limb was 3.5 times larger than in the prosthetic limb at the end of rehabilitation.[2] From this we may conclude that this adjustment strategy of the non-affected limb does not change when the difficulty of the balance task increases from quiet standing to standing on a moving platform.

Despite the inability to compensate at the ankle of the prosthetic limb, a larger FY was found in amputees than in AB subjects. In a study on subjects with somatosensory loss of the lower leg induced by anaesthesia FY was also increased in response to platform displacements.[25] By increasing FY more somatosensory input can be received in the prosthetic limb.[8,15,17] Amputees may have increased FY by using the intact hip musculature in the prosthetic limb. Flexion and extension in the hip shift the COM forward and backward[1,3], and consequently result in a larger FY in both limbs. Muscle activity in the prosthetic limb is required to limit the degrees of freedom of the prosthesis during platform perturbations and to keep the prosthetic knee locked in extension.

The second aim of this study was to determine the effect of visual and conscious control on balance control strategies. In our study only the blindfolded condition had an effect in AB subjects. In amputees balance control was not significantly influenced by the additional task. Several reasons can explain why amputees did not shift to visual or cognitive balance control strategies during standing on a moving platform. The increase of COPY and weight bearing on the non-affected limb in amputees was already large in the normal condition. Using these adjustment strategies more intensively may have endangered stability on the platform in amputees. In contrast, AB subjects were not using their balance control strategies to a full extent in the normal condition. They were therefore able to increase muscle activity in the blindfolded condition, resulting in a larger COPY. Furthermore, managing balance perturbations may have been an entirely automated task, because the subjects were experienced prosthetic users and the platform movements were predictable. The absence of a condition effect may also be explained by the ease of the dual task, the small number of subjects, or an inadequate performance of the dual task.

The present study has a number of limitations. Although clear differences between the study groups were found, the study groups were only small and the AMP group consisted of subjects with different amputation levels. The results can only be considered indicative for experienced lower limb amputees with a normal to high activity level and may not be generalized to all amputees. Since we wished our study to resemble a real life situation, we did not standardise standing position. As a result, foot position and therefore base of support may have been different in subjects, which influenced the outcome parameters. Previous studies reported that AB subjects stood with their feet closer together than amputees[7], and that visual dependency in amputees was reduced when the base of support was wider.[11] Due to technical limitations FY and COPY values were not corrected for inertia of the force plate in relation to the sensors. Because this measuring error was only small and similar in groups and conditions, a significant effect on the results would be unlikely.

In order to mimic a real life situation it is important that a task closely simulates an activity which is difficult for amputees.[4] The moving platform in our experiment would simulate standing in a moving bus. The blindfolded and dual task conditions would simulate standing in a bus in the dark, while having a conversation. However, in a bus the perturbations occur more unexpectedly and with a higher velocity. The present study demonstrated the effect of expected perturbations on balance control. Future research should focus on unexpected balance disturbances.

Conclusion

The results of this study indicate that experienced unilateral amputees compensate for the loss of ankle strategy by increasing movements and loading in the non-affected limb. The ability to cope with balance perturbations is limited in the prosthetic limb. To enable amputees to manage all possible balance disturbances in real life in a safe manner, we recommend to improve muscle strength and control in the non-affected limb and to train complex balance tasks in challenging environments during rehabilitation.

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Chapter 7

Gait adjustments in obstacle crossing, gait initiation and gait termination after a recent lower limb amputation

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Abstract

Objective: To describe the adjustments in gait characteristics of obstacle crossing, gait initiation and gait termination that occur in subjects with a recent lower limb amputation during the rehabilitation process.
Design: Prospective and descriptive study.
Subjects: Fourteen subjects with a recent transfemoral, knee disarticulation or transtibial amputation.
Methods: Subjects stepped over an obstacle and initiated and terminated gait at four different times during the rehabilitation process.
Outcome measures: Success rate, gait velocity and lower limb joint angles in obstacle crossing, centre of pressure shift and peak anteroposterior ground reaction force in gait initiation and termination.
Results: In obstacle crossing amputees increased success rate, gait velocity and swing knee flexion of the prosthetic limb. Knee flexion in transfemoral and knee disarticulation amputees was not sufficient for safe obstacle crossing, which resulted in a circumduction strategy. In gait initiation and termination amputees increased the anteroposterior ground reaction force and the centre of pressure shift in the mediolateral direction in both tasks. Throughout the rehabilitation process the centre of pressure was shifted anteriorly before single-limb stance on the trailing prosthetic limb in gait initiation, whereas in gait termination the centre of pressure in single-limb stance remained posterior when leading with the prosthetic limb.
Conclusion: Subjects with a recent amputation develop adjustment strategies to improve obstacle crossing, gait initiation and gait termination. Innovations in prosthetic design or training methods may ease the learning process of these tasks.

Introduction

During rehabilitation amputees learn to adapt their gait pattern to the prosthesis by training new motor strategies and by adjusting the existing motor strategies. Successful rehabilitation of an amputee does not only include the ability of steady-state level walking, but also the performance of more complex motor tasks such as obstacle crossing, gait initiation and gait termination. Up to now, research into these motor tasks was only performed in experienced prosthetic users. In the prosthetic limb of transtibial (TT) amputees an increase in hip and knee flexion is observed in obstacle crossing, which did not occur in transfemoral (TF) amputees.[1] In the non-affected limb ankle plantar flexion in stance is increased in both amputation levels, which facilitated passing of the obstacle with the prosthetic limb in swing.[1, 2] From studies on gait initiation in TF and TT amputees, it is known that the displacement of the posterior centre of pressure (COP) on the prosthetic side is decreased and the propulsive ground reaction force (GRF) produced by the prosthetic limb is limited.[3-8] A study on gait termination in TF and TT amputees reported a decreased braking GRF in the prosthetic limb and an absent anterior COP shift when leading with the prosthetic limb.[9]

In early rehabilitation amputees are inexperienced in walking with a prosthesis and have not yet developed any adjustment strategies. Through gait and balance training amputees learn to make use of the prosthetic foot and knee properties and to replace the role of the absent muscles by that of the proximal remaining musculature. Up to now, it is not known how and when subjects with a recent amputation develop the adjustment strategies that are necessary to step over an obstacle, to initiate gait and to terminate gait in a safe way. The goal of the current research is to establish the changes in several gait characteristics of obstacle crossing, gait initiation and termination that occur during the rehabilitation process in subjects after a recent lower limb amputation.

Methods

Subjects

Subjects with a recent unilateral TF, knee disarticulation (KD) or TT amputation, who were admitted to a rehabilitation centre, were asked to participate in the study. Subjects with an age between 18 and 80 years and a prognosis of becoming a prosthetic user were included. Subjects with disorders of the locomotor system that influenced mobility and balance, serious and unstable cardiac or pulmonary problems, severe impaired vision, or cognitive impairments were excluded.

All subjects were trained according to a regular rehabilitation program. Specific training of the tested motor tasks was not part of this program. Gait training started by walking with two different types of temporary prosthetic devices. The Pneumatic Post Amputation Mobility aid (PPAM, Ortho-Europe, Abingdon, UK) consists of a metal frame with an inflatable bag and a rocker foot. This device does not allow knee flexion. The Hoensbroek Training Prosthesis (HTP) consists of an adjustable socket and a dynamic prosthetic foot (1D10, Otto Bock, Duderstadt, Germany) and only fits TT amputees. The individually fitted prostheses were composed of diverse types of prosthetic knees and feet. Several walking aids were used during rehabilitation; a rollator (four-wheeled walker, Premis Medical, Woudenberg, NL), crutch(es), and a walking cane. The use of walking aids was gradually reduced. During the rehabilitation process the subjects were regularly seen by a certified prosthetist to control fitting and alignment. In Table 1 information on the prosthetic components and walking aids is presented.

Apparatus

The study was performed in a motion analysis laboratory, which is equipped with an 8 m long walkway and a 0.4 x 0.6 m force plate (Bertec, Columbus, USA). The beginning and end of the walkway were fitted with infrared beams, which recorded the passing of a subject. The trials were videotaped by two video cameras with a sampling frequency of 25 Hz in the coronal and sagittal plane. The joints of the hip and knee on the prosthetic side were provided with flexible twin axis electrogoniometers (SG150, Penny & Giles Biometrics, Gwent, UK). Accuracy in these goniometers is ± 2° and repeatability is 1°, both measured over a range of 90°. The goniometers were zeroed while the subject stood in an upright position with the hip and knee joint in a neutral position and the ankle joint in a plantigrade position. In case of contractures in the hip or knee the most neutral position of that joint was called zero. Goniometer signals were recorded on a portable data acquisition system (TMSI, Enschede, NL) at a sampling frequency of 800 Hz and were filtered with a cut-off at 10 Hz. Coordinates of the COP and the anteroposterior GRF were measured with the force plate at 100 Hz. GRF signals were low-pass filtered with a 6

Table 1. Information on number of subjects, timing of assessments, SIGAM scores, prosthetic components and walking aids at each assessment.

	Group	T1	T2	T3	T4
Number of subjects	High	7	6	4	5
	Low	7	5	6	4
Days after amputation	High	62 ± 38	116 ± 57	186 ± 60	330 ± 83
	Low	92 ± 68	156 ± 81	216 ± 116	359 ± 117
SIGAM*	High	7 B	2 B, 3 C, 1 D	4D	1 C, 2 D, 2 E
	Low	7 B	1 B, 1 C, 2 D, 1 F	1 D, 3 E, 2 F	1 D, 3 F
Prosthetic foot†	High	7 PPAM	2 1D35[1], 2 Multiflex[2], 1 SACH[1], 1 C-walk[1]	2 Multiflex[2], 1 SACH[1], 1 C-walk[1]	1 1D35[1], 2 Multiflex[2], 1 SACH[1], 1 C-walk[1]
	Low	2 HTP, 5 PPAM	1 1D35[1], 2 Multiflex[2], 2 C-walk[1]	1 1D35[1], 2 Multiflex[2], 3 C-walk[1]	1 1D35[1], 1 Multiflex[2], 2 C-walk[1]
Prosthetic knee†	High	7 PPAM	2 3R106[1], 2 Total knee[3], 1 Senior[4], 1 SafeLife[4]	1 Total knee[3], 1 Senior[4], 1 SafeLife[4], 1 locked[5]	1 3R106[1], 1 Total knee[3], 1 Senior[4], 1 SafeLife[4], 1 locked[5]
	Low	2 HTP, 5 PPAM	-	-	-
Walking aid	High	1 rollator, 6 two crutches	6 two crutches	4 one cane	2 one cane, 3 none
	Low	2 rollator, 3 two crutches, 2 one cane	2 two crutches, 1 one cane, 2 none	6 none	4 none

Number of subjects, days after amputation (mean and SD), SIGAM scores, prosthetic components and walking aids of the high and low amputation level study groups at the four assessments (T1-4). * In the SIGAM questionnaire amputees who do not possess their own prosthesis have SIGAM score A, but since the subjects used a temporary prosthetic device during therapy SIGAM score B was most appropriate to represent the mobility level of the subjects at T1. † HTP Hoensbroek Training Prosthesis, PPAM Pneumatic Post Amputation Mobility aid. Prosthetic components by [1] Otto Bock, Duderstadt, Germany, [2] Endolite, Centerville, USA, [3] Össur, Reykjavik, Iceland, [4] Proteval, Valenton, France, [5] Medipro, Bayreuth, Germany.

Hz cut-off. A custom-developed Gait Analysis System (UMCG, Groningen, NL) recorded, synchronised and analysed all measurements with a sampling frequency of 100 Hz.

Procedure

The Medical Ethics Committee approved the study protocol. All subjects signed informed consent before testing. Four assessments were performed; T1 when the subject walked approximately two weeks with a temporary prosthetic device, T2 when the subject walked approximately two weeks with an individually fitted prosthesis, T3 two months after T2, and T4 six months after T2. The mobility level was assessed with the Dutch version of the mobility score developed by the Special Interest Group in Amputee Medicine (SIGAM).[10] Table 1 provides information on the timing of the assessments and the SIGAM scores.

A wooden obstacle of 0.1 m height, 0.02 m thickness and 1.0 m width was placed in the middle of the walkway. (Figure 1) Subjects were instructed to walk at their comfortable gait velocity and to step over the obstacle without touching it. The limb that crossed the obstacle first was defined as the leading limb, the other limb as trailing limb. A physiotherapist walked alongside the amputee and provided assistance if necessary. For safety reasons the obstacle could easily fall over in case of foot contact. To allow passing of a rollator, a supplementary obstacle was designed. If the obstacle fell over due to contact with a walking aid, the trial was repeated. In gait initiation subjects started walking from a bipedal standing position on the force plate on their own initiative. The leading limb stepped of the force plate first, followed by the trailing limb. In gait termination the subjects were instructed to step with the leading limb on the force plate, followed by placing the trailing limb next to the leading limb. If a walking aid touched the force plate, the trial was repeated. Gait velocity, foot position and timing of gait initiation and termination were self-selected. To avoid overloading of the amputated limb at T1 and T2 the number of trials in obstacle crossing was limited to six and in gait initiation and termination to four. At T3 and T4 the number of trials was doubled. In half of the trials the prosthetic limb was used as leading limb, and in the other half the non-affected limb.

Outcome parameters

Outcome parameters in obstacle crossing were success rate, gait velocity and hip and knee joint angles in the prosthetic limb. An obstacle crossing trial was successfully completed when the subject passed the obstacle with both limbs without the obstacle falling over

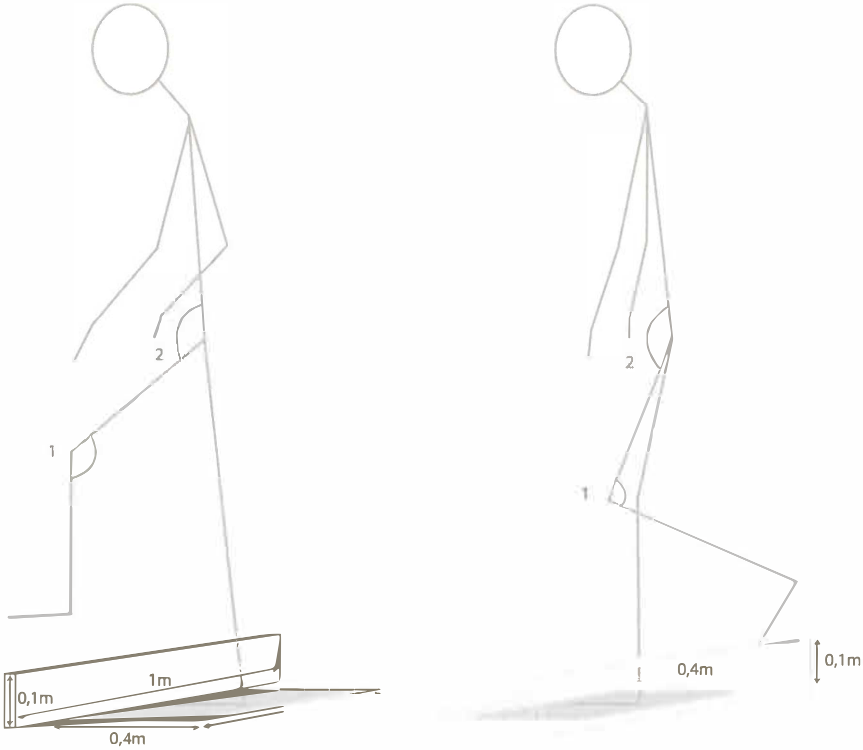

Figure 1.

Stick diagram showing the analysed hip and knee angles in the prosthetic leading and trailing limb. The video images were used to determine the instant that the middle of the foot, either leading or trailing, was above the obstacle. At the instant that the middle of the leading prosthetic foot was exactly above the obstacle (left figure) we assessed the joint angles in the knee (1) and hip (2) of the leading prosthetic swing limb. At the instant that the middle of the trailing prosthetic foot was exactly above the obstacle (right figure) we assessed the joint angles in the knee (1) and hip (2) of the trailing prosthetic swing limb. The knee angle of the prosthetic limb was only analysed in subjects with a free moveable knee and not in subjects using the PPAM aid or locked knee.

and/or the subject needing assistance from the physiotherapist. The percentage of successful obstacle crossing trials was assessed. The average gait velocity was calculated by dividing the length of the walkway by the time that the subject needed to walk over the walkway. The method for measuring the joint angles in the prosthetic limb is demonstrated in Figure 1.

Outcome parameters in gait initiation and termination were the COP shift and the anteroposterior component of the GRF. Outcome parameters were assessed separately for the leading and trailing prosthetic limb and non-affected limb. In single-limb stance the COP shift could be attributed to a limb (leading prosthetic, trailing prosthetic, leading non-affected, or trailing non-affected) and in bipedal stance to a limb sequence (leading prosthetic limb followed by trailing non-affected limb, or leading non-affected limb followed by trailing prosthetic limb). The trajectory of the COP on the force plate was expressed by two measuring points in the mediolateral direction (COPX1,2) and two in the anteroposterior direction (COPY1,2), which is illustrated in Figure 2. We determined the peak amplitudes of the anteroposterior component of the GRF (FY) of the trailing limb in gait initiation and of the leading limb in gait termination, both in single-limb stance.

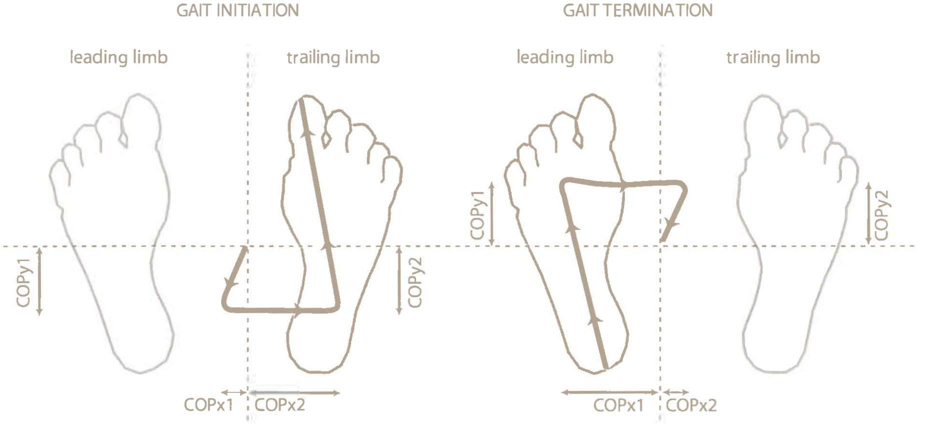

Figure 2.
In gait initiation (left) we assessed the distance from the bipedal starting COP position to the most lateral and posterior COP position on the leading limb side (COPx1 and COPy1) and on the trailing limb side (COPx2 and COPy2). In gait termination (right) we assessed the distance from the final bipedal standing COP position to the most lateral and anterior position on the leading limb side (COPx1 and COPy1) and on the trailing limb side (COPx2 and COPy2). Since we used a single force plate in bipedal stance the resultant COP of the leading and trailing limb were assessed.

Statistical analysis

The outcome parameters were analysed using descriptive statistics. First the means of the outcome parameters per subject were calculated, followed by the group means. In obstacle crossing TF and KD subjects were analysed as one study group (high amputation level) and TT subjects as a separate study group (low amputation level), since knee flexion is of great importance in this task.[1, 2, 11] In gait initiation and gait termination TF, KD and TT subjects were analysed as one study group, since previous research did not show important differences in outcome parameters between the levels of amputation.[6, 9]

Results

Seven TT, four TF and three KD subjects agreed to join the study. The study group included 11 male and 3 female subjects with a mean age of 57.3 (45.8 - 69.4) years. The cause of amputation was vascular disease in twelve subjects, a tumour in one subject and infection in one subject. Not all subjects were able to perform all assessments. After T1 one TT and one TF subject dropped out, respectively due to persistent skin problems of the residual limb and an operation for a hip endoprosthesis on the non-affected side. After T2 one TF subject was excluded because of a vascular bypass operation in the non-affected limb. Two TT subjects could not perform T3 and T4 by reasons of emigration to another country and intermittent claudication in the non-affected limb. One TT subject missed T2 due to a gallbladder operation and one KD subject failed T3 because of an operation for a melanoma in the residual limb. In Table 1 the numbers of subjects per assessment are presented.

Obstacle crossing

Data on success rate are presented in Table 2. In the leading prosthetic limb trials of the high amputation level group success rate at T1 was 94.5%, but decreased to 60.7% at T2. In the trailing prosthetic limb trials of the high amputation level group success rate increased from 72.2% at T1 to 89.8% at T2. From T3 on almost all trials were successfully performed with both the leading and trailing prosthetic limb in the high amputation level group. At T1 success rate of the trailing prosthetic limb in low amputation level group was 82.2% and of the leading prosthetic limb 94.0%. From T2 onwards, the low level amputation group achieved a 100% success rate with either the prosthetic limb leading or trailing over the obstacle. Table 3 contains the results on gait velocity. Gait velocity in the high amputation level group showed a small decrease from 0.43 ms^{-1} to 0.40 ms^{-1} between T1 and T2. After T2 gait velocity in the high amputation level group increased to 0.64 ms^{-1} at T4. Across the four measurements gait velocity in the low amputation level group increased gradually from 0.44 ms^{-1} to 1.0 ms^{-1}. The joint angles of the prosthetic limb are shown in Table 4. In both study groups knee and hip flexion in the leading prosthetic limb were larger than in the trailing prosthetic limb in all measurements. Knee flexion in the leading prosthetic limb in the high amputation level group increased from T2 to T4, but was much smaller than in the low amputation level group. In the trailing prosthetic limb in the high amputation level group hardly any swing knee flexion was measured in all assessments. Swing hip and knee flexion in the leading and trailing prosthetic limb in the low amputation level group were at least twice as large at T4 compared to T1.

Table 2. Success rate in obstacle crossing.

	Level	Limb	T1	T2	T3	T4
Success rate (%)	High	LP	94.5 ± 13.5	60.7 ± 23.0	100.0 ± 0.0	97.6 ± 5.4
		TP	72.2 ± 22.5	89.8 ± 12.3	100.0 ± 0.0	100.0 ± 0.0
	Low	LP	94.0 ± 10.3	100.0 ± 0.0	100.0 ± 0.0	100.0 ± 0.0
		TP	82.2 ± 22.3	100.0 ± 0.0	100.0 ± 0.0	100.0 ± 0.0

Mean values and standard deviation of success rate in leading with the prosthetic (LP) limb and in trailing with the prosthetic (TP) limb of the high and low amputation level study groups at the four assessments (T1-4).

Table 3. Gait velocity in obstacle crossing.

	Level	T1	T2	T3	T4
Gait velocity (ms^{-1})	High	0.43 ± 0.09	0.40 ± 0.08	0.60 ± 0.07	0.64 ± 0.13
	Low	0.44 ± 0.11	0.77 ± 0.15	0.93 ± 0.11	1.00 ± 0.13

Mean values and standard deviation of gait velocity in the high and low amputation level study groups at the four assessments (T1-4).

Table 4. Lower limb joint angles in the prosthetic limb during obstacle crossing.

	Level	Limb	T1	T2	T3	T4
Hip (°)	High	LP	13.6 ± 7.3	27.7 ± 11.1	15.6 ± 8.9	30.3 ± 12.2
		TP	3.6 ± 5.0	19.9 ± 10.6	6.8 ± 11.7	8.7 ± 11.3
	Low	LP	18.1 ± 7.8	32.5 ± 5.1	38.1 ± 9.2	38.2 ± 6.2
		TP	3.7 ± 6.9	13.6 ± 5.6	15.8 ± 3.0	16.1 ± 7.1
Knee (°)*	High	LP	-	14.9 ± 11.6	17.9 ± 16.2	33.8 ± 23.9
		TP	-	4.1 ± 7.4	2.7 ± 9.0	7.5 ± 10.8
	Low	LP	31.3 ± 1.8	51.5 ± 13.2	65.4 ± 18.8	71.1 ± 10.9
		TP	27.0 ± 7.1	47.3 ± 13.1	54.8 ± 15.8	61.6 ± 13.4

Mean values and standard deviation lower limb joint angles in the leading prosthetic (LP) and trailing prosthetic (TP) limb of the high and low amputation level study groups at the four assessments (T1-4). * At T1 the knee angle of the prosthetic limb was only analysed in 2 TT subjects who used a temporary training devise that allowed active knee flexion. At T3 and T4 one subject in the High amputation level group walked with a locked knee and was not included in the analysis.

Gait initiation

The outcome parameters are presented in Table 5. The posterior COPY1 shift in leading with the non-affected limb was larger than in leading with the prosthetic limb in all four assessments. In both leading limb sequences the posterior COPY1 shift at T1 and T2 was comparable and increased from T2 to T4. The COPX1 shift towards the leading non-affected limb increased from T1 to T4, whereas the COPX1 shift towards the leading prosthetic limb increased from T1 to T3. In leading with the prosthetic limb the COPY2 shifted posteriorly and increased slightly from T1 to T4. On the contrary, in leading with the non-affected limb the COPY2 was shifted anteriorly of the bipedal starting position at all assessments and decreased from T1 to T4 (See Figure 3A). The COPX2 shift was larger in the direction of the trailing prosthetic limb than towards the trailing non-affected limb. From T1 to T4 the COPX2 shift showed a slight increase when leading with the prosthetic limb. In leading with the non-affected limb minimal changes in the COPX2 shift were found. In the trailing non-affected limb the FY was higher than in the trailing prosthetic limb. The FY of the trailing non-affected limb was almost similar at T1 (−10.9% BW) and T2 (−10.7% BW) and increased to −19.3% BW at T4. In the trailing prosthetic limb the FY increased from −6.3% BW at T1 to −14.8% BW at T4.

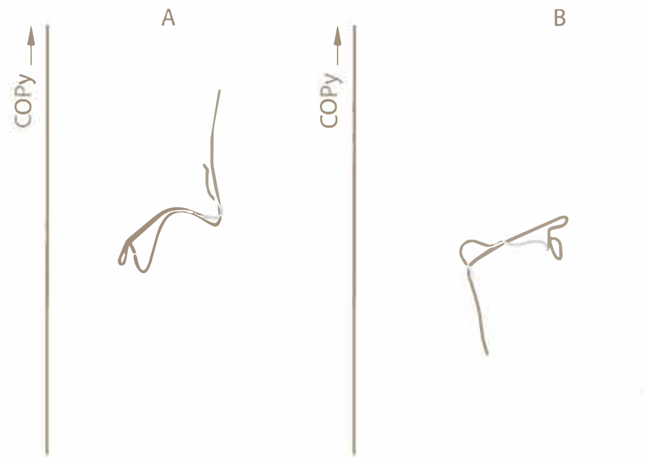

Figure 3.

Representative examples of COP patterns at the first (T1) and fourth assessment (T4). Figure 3A shows the COP pattern of a TF subject when leading gait initiation with the left non-affected limb at T1 (grey line) and at T4 (black line). At the moment of single-limb stance on the trailing prosthetic limb the COPy2 was shifted toward the forefoot. Compared to T1 the forward COPy2 shift was decreased at T4. Figure 3B shows the COP pattern of a TF subject when leading gait termination with the left prosthetic limb at T1 (grey line) and at T4 (black line). In the leading prosthetic limb no anterior COPy1 shift was seen. From T1 to T4 the COPy1 was shifted more towards the heel of the prosthetic limb in stead of towards the forefoot.

Gait termination

The results are presented in Table 6. In the leading prosthetic limb the COPY1 remained posteriorly from the bipedal standing position in all measurements and from T1 to T4 the COPY1 shift toward the heel became larger (See Figure 3B). In contrast, in the leading non-affected limb the COPY1 shift from T1 to T4 was directed anteriorly. The leading prosthetic limb showed minimal changes in the COPX1 shift, whereas the COPX1 shift towards the leading non-affected limb increased from T1 to T4. The anterior COPY2 shift in leading with the prosthetic limb was similar at T1 and T2 and increased after T2. When leading with the non-affected limb the COPY2 shift did not show important changes over time. In both limb sequences the COPX2 shifted more towards the trailing limb at T4 than at T1. The FY was higher in the leading non-affected limb than in the leading prostheticlimb. In the leading prosthetic limb the FY increased from 5.7% BW at T1 to 13.5% BW at T4 and in the leading non-affected limb from 8.5% BW at T1 to 18.2% BW at T4.

Table 5. COP and GRF in gait initiation.

	Limb	T1	T2	T3	T4
COPy1 (cm)	LP-TN	−1.08 ± 0.74	−1.08 ± 1.32	−1.40 ± 1.07	−1.84 ± 1.22
	LN-TP	−2.73 ± 3.10	−2.60 ± 0.91	−3.36 ± 0.86	−3.57 ± 1.70
COPx1 (cm)	LP-TN	0.55 ± 1.05	1.40 ± 0.92	2.73 ± 1.43	2.29 ± 1.43
	LN-TP	0.73 ± 0.74	1.97 ± 2.19	2.03 ± 1.16	2.33 ± 1.10
COPy2 (cm)	LP-TN	−2.21 ± 2.42	−2.96 ± 2.07	−2.84 ± 3.63	−3.00 ± 3.45
	LN-TP	7.03 ± 3.29	3.65 ± 6.48	3.76 ± 2.71	3.18 ± 3.21
COPx2 (cm)	LP-TN	−6.32 ± 2.03	−7.48 ± 1.87	−8.33 ± 1.59	−7.87 ± 1.83
	LN-TP	−11.64 ± 2.83	−11.28 ± 2.71	−12.12 ± 1.77	−10.71 ± 2.20
Fy (% BW)	TN	−10.9 ± 3.5	−10.7 ± 4.5	−15.5 ± 4.3	−19.3 ± 5.1
	TP	−6.3 ± 3.2	−8.0 ± 3.6	−13.1 ± 3.5	−14.8 ± 3.5

Mean values and standard deviation of centre of pressure shift (COP) in the leading prosthetic–trailing non-affected (LP-TN) and in the leading non-affected–trailing prosthetic (LN-TP) limb sequence and peak anteroposterior ground reaction force (Fy) in the trailing non-affected (TN) and trailing prosthetic (TP) limb at the four assessments (T1-4). COPy is positive in the anterior direction and negative in the posterior direction. COPx is positive in the direction of the leading limb and negative in the direction of the trailing limb. Fy is positive in the posterior direction and negative in the anterior direction. COPx and COPy are expressed in cm, Fy in percentage body weight (% BW)

Discussion

From our study we can conclude that TT amputees required less time to perform the necessary adjustment strategies in obstacle crossing than TF and KD amputees did. At the first assessment a number of obstacle contacts was seen in TT amputees, since most TT amputees used the PPAM aid, which did not allow knee flexion. At the second assessment all obstacle crossing trials were performed successfully and knee and hip flexion in the prosthetic limb both showed a large increase compared to the first assessment. When TT amputees were supplied with an individually fitted prosthesis and could use their active knee function, they were all able to step over the obstacle without touching it. In TF and KD amputees the success rate in early rehabilitation depended on which limb crossed the obstacle. When walking with the PPAM aid which does not allow knee flexion, TF and KD

Table 6. COP and GRF in gait termination.

	Limb	T1	T2	T3	T4
COPy1 (cm)	LP-TN	−1.16 ± 4.01	−2.32 ± 4.18	−3.48 ± 3.53	−3.98 ± 2.77
	LN-TP	0.67 ± 3.76	1.62 ± 1.79	1.22 ± 2.78	1.51 ± 3.02
COPx1 (cm)	LP-TN	10.96 ± 2.14	9.60 ± 2.25	10.35 ± 2.10	9.23 ± 2.32
	LN-TP	4.44 ± 1.97	7.36 ± 2.12	7.65 ± 1.80	8.27 ± 1.54
COPy2 (cm)	LP-TN	2.69 ± 2.40	2.54 ± 3.51	3.73 ± 2.41	4.09 ± 2.90
	LN-TP	1.98 ± 1.53	2.53 ± 1.89	2.29 ± 1.16	1.96 ± 1.57
COPx2 (cm)	LP-TN	−1.04 ± 2.02	−2.92 ± 1.87	−4.07 ± 1.98	−4.97 ± 2.85
	LN-TP	−0.77 ± 1.81	−1.22 ± 2.38	−1.73 ± 2.35	−2.16 ± 1.99
Fy (% BW)	LP	5.7 ± 2.1	7.9 ± 2.8	13.0 ± 3.3	13.5 ± 6.0
	LN	8.5 ± 4.4	11.5 ± 3.8	15.6 ± 5.4	18.2 ± 7.0

Mean values and standard deviation of centre of pressure shift (COP) in the leading prosthetic–trailing non-affected (LP-TN) and in the leading non-affected–trailing prosthetic (LN-TP) limb sequence and peak anteroposterior ground reaction force (Fy) in the leading prosthetic (LP) and leading non-affected (LN) limb at the four assessments (T1-4). COPy is positive in the anterior direction and negative in the posterior direction. COPx is positive in the direction of the leading limb and negative in the direction of the trailing limb. Fy is positive in the posterior direction and negative in the anterior direction. COPx and COPy are expressed in cm, Fy in percentage body weight (% BW).

amputees primarily experienced difficulties in swinging the trailing prosthetic limb over the obstacle. The limited hip flexion in the trailing prosthetic limb at the first assessment likely played a role in the low success rate. In addition, the trajectory of the trailing limb can not be guided by vision, whereas during crossing with the leading limb the obstacle can be seen.[2, 12-15] Moreover, in the trailing limb the distance from toe-off to the position of the obstacle is shorter than in the leading limb, which allows limited time to achieve adequate foot clearance in the prosthetic trailing limb or to execute any adjustment strategies.[12, 16]

When TF and KD amputees were provided with an individually fitted prosthesis at the second assessment, the obstacle was most frequently touched by the leading prosthetic limb. At this time in rehabilitation TF and KD amputees were at the beginning of the

learning process of walking with a free moveable prosthetic knee. The small amount of knee flexion in the leading prosthetic limb at the second assessment indicates that TF and KD amputees attempted to make use of the flexion properties of the prosthetic knee joint, but knee flexion was too small to reach sufficient foot clearance in all trials. Failures of the leading prosthetic limb in TF and KD amputees were also caused by touching the obstacle with the prosthetic foot at initial contact. Prosthetic knee stability in early stance is provided by activation of hip extensors, which may have resulted in a backward pull of the prosthetic foot. Moreover, the slow walking velocity may have limited the forward swing of the prosthetic limb as well.

Knee flexion in the prosthetic limb of TF and KD amputees remained small as time since amputation progressed. TF and KD amputees were able to successfully cross the obstacle at the third assessment, approximately six months after the amputation. From previous studies on obstacle crossing we know that able-bodied subjects and subjects with a total knee arthroplasty use adjustments at the hip and pelvis level to pass the swing limb over the obstacle.[1, 17, 18] Although we did not assess hip abduction and elevation in this study, the video images clearly showed that TF and KD amputees used a circumduction strategy in the prosthetic limb to step over the obstacle.

In able-bodied subjects the gait initiation process normally starts with a posterior COPY shift, which creates a forward moment of force.[3, 19-27] In all four assessments of our study on amputees the posterior COPY shift was limited and was already displaced toward the forefoot early in the gait initiation process, namely at the instant of shifting the body weight on the trailing prosthetic limb in single-limb stance. In gait termination able-bodied persons shift the anteroposterior COPY quickly towards the toes after initial contact of the leading limb to produce a backward moment of force.[28-30] In contrast, in the present study the anteroposterior COPY under the leading prosthetic limb in single-limb stance remained near the heel in gait termination from the beginning until the end of the rehabilitation process. Amputees could not shift the anteroposterior COPY in an active manner when standing in single-limb stance on the prosthetic limb, because the prosthetic limb lacks the ability to actively control ankle plantar and dorsiflexion moments.

In both gait initiation and termination from the first to the last assessment amputees showed an increase in the peak propulsive and braking FY in both the prosthetic and non-affected limb. In general, throughout the rehabilitation process amputees increased the COP shift in both directions, resulting in a COP pattern that was more similar to able-bodied subjects. However, in one outcome parameter of gait termination an adjustment was shown that did not enhance the performance of this task. The posterior COPY shift in leading with the prosthetic limb increased from the first to the fourth assessment, whereas an anterior COPY shift in the leading limb would have contributed to deceleration. From the video images we ascertained that some amputees placed the trailing non-affected limb slightly in front of the leading prosthetic limb in gait termination. As a result, the final bipedal standing position moved forward. Because the anteroposterior COPY shift in the leading limb was related to the bipedal standing position, this may in part account for the increase in the posterior COPY shift throughout rehabilitation.

Future research has to be conducted to decide whether amputees can improve obstacle crossing, gait initiation and termination by training methods or innovations in prosthetic components. According to the general concept of motor programming, intensive and task-specific practice is necessary to learn a new motor skill.[7, 31-33] We advise to start training difficult motor tasks early in rehabilitation. Based on our study results, in gait initiation it is advisable to use the prosthetic limb as leading limb, since most difficulties occur in the trailing prosthetic limb. In gait termination the main gait deviations were observed in the leading prosthetic limb, and if possible, leading with the non-affected limb is preferred. For obstacle crossing some basic conditions should be present in the gait pattern, since practising this task too early in the rehabilitation process may result in failures and influence the confidence and motivation negatively, or even lead to falls. Our results indicate that TT amputees could start executing an obstacle crossing task as soon as they are provided with a prosthesis that allows knee flexion, either a temporary prosthesis or an individually fitted prosthesis. For TF and KD amputees we advise to wait until they are able to flex the prosthetic knee and to propel the prosthetic foot forward in swing during level walking. A temporary prosthetic device that allows knee flexion would make early obstacle crossing training possible for this group. Training in TF and KD amputees should focus on increasing prosthetic knee flexion by using hip flexor activity and ground friction, and by maintaining gait velocity when crossing the obstacle.

The studied motor tasks may benefit from innovations in prosthetic design. Microprocessor-controlled prosthetic knees can control swing knee flexion by making adjustments to the damping, and the electronic stance phase control allows for a less stable alignment which eases the initiation of swing phase.[34,35] Both these features could result in more swing knee flexion during obstacle crossing. However, this adaptation is too slow for stepping over a single obstacle because a microprocessor-controlled knee requires at least one step to alter knee flexion.

The roll-over length and shape, the stiffness and the ankle mobility of prosthetic feet may provide opportunities to improve gait initiation and termination. In level walking, it is reported that the use of prosthetic feet with a longer functional forefoot length increases the external dorsiflexion moment in late stance due to the longer moment arm.[36] In gait initiation and termination a longer effective foot length may allow for more COP shift in the anteroposterior direction and result in a higher anteroposterior GRF. However, it remains unclear if amputees are able to use the total effective foot length. A more flexible prosthetic foot would ease sagittal ankle and foot motion, but at the same time compromise the dynamic stability. For example, in gait initiation a more flexible heel would be of assistance in shifting the COP under the heel of the trailing prosthetic limb, but may also lead to losing balance in the posterior direction. Amputees may also benefit from the prescription of single- or multi-axis ankles that permit some ankle dorsiflexion in gait initiation and some ankle plantar flexion in gait termination. In the future adaptive prostheses may be available that could actively generate the necessary ankle motion.

Several limitations that exist in this study have to be considered. In obstacle crossing an increase in gait velocity was observed, which may be responsible for some of the observed differences in joint angles. Research on level walking in amputees reported that gait velocity influences kinematic variables.[37] Furthermore, the walking aids are likely to have affected the data in gait initiation and termination. Walking aids change the base of support and therefore alter the effects of the COP shifts beneath the feet. Initiating or terminating gait with less support of walking aids could increase the freedom of movement and result in more COP shift.

The changes in prosthetic components during the rehabilitation process are likely to have influenced the outcome parameters as well, since prosthetic foot and knee properties influence joint angles, gait velocity, the vertical and anteroposterior GRF and COP pattern.[38-47] The main alteration in prosthetic device occurred when the temporary prosthetic device at the first assessment changed into an individually fitted prosthesis at the second assessment. In obstacle crossing, five TT amputees swapped the PPAM aid with no knee function for a prosthesis in which they could use their own active knee function, and all TF and KD amputees switched to a free moveable prosthetic knee. In gait initiation and termination most amputees used the PPAM aid at the first assessment. The rocker foot can be easily rolled over in the anteroposterior direction due to its limited length and convex shape. From the second assessment on amputees were provided with a prosthetic foot and shoe in which the roll-over shape, ankle motion and stiffness determine the relation between the COP position and the inclination of the prosthetic limb.

As a result of the small number of subjects in each group and the number of drop-outs we were not able to analyse whether gait parameters significantly improved over time. Overall, variability in the outcome parameters was reasonably high, which can be explained by the small and heterogeneous study groups. Subjects with different levels and causes of amputation were collected and the differences in ankle and knee mechanisms, stump length and alignment were not taken into account. In addition, variability was high because each subject only performed a limited number of trials in early rehabilitation.

Conclusion

Subjects with a recent lower limb amputation developed adjustment strategies to improve obstacle crossing, gait initiation and termination. Innovations in prosthetic design or training methods may ease the learning process of these tasks.

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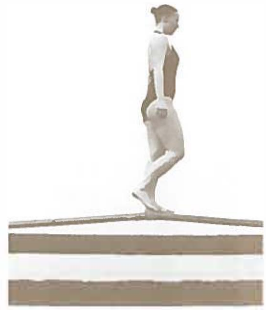

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Chapter

8

General discussion

General discussion

The aim of this thesis was to study how persons with a lower limb prosthesis perform complex balance and gait tasks, and how they adjust to these tasks in the rehabilitation process. Prior to this study, it was unclear which adjustments strategies amputees use to compensate for the functional limitations in difficult motor tasks. In this discussion the most important research findings are summarised. Potential solutions to improve functioning of lower limb amputees in complex walking and balance tasks are considered, from which the clinical implications of the study and recommendations for further research can be derived. The discussion is completed with some final considerations on the content of this thesis.

Limitations in function and adjustment strategies in complex motor tasks

In this thesis we studied the performance of experienced transtibial (TT) and transfemoral (TF) amputees and of able-bodied (AB) subjects in obstacle crossing, gait initiation and termination, walking up and down a ramp and standing on a swaying platform. The absence of an active ankle strategy in the prosthetic limb in TT and TF amputees resulted in a limited anteroposterior displacement of the centre of pressure (COP) and a decreased anteroposterior component of the ground reaction force (GRF) in gait initiation and gait termination. In standing on a moving platform the COP displacement and weight bearing on the prosthetic side were decreased. The deficient knee strategy in TF amputees reduced prosthetic knee flexion in obstacle crossing and uphill walking in swing and in downhill walking in stance. Because of the intact knee function, TT amputees were able to adapt their gait in obstacle crossing, uphill and downhill walking in a similar way as AB subjects.

When performing complex motor tasks, amputees applied several adjustment strategies to compensate for the limitations in function of the prosthetic limb. An adjustment strategy concerning both the prosthetic and non-affected limb was the choice for the leading limb in gait tasks. In obstacle crossing TF amputees preferred leading with the non-affected limb to compensate for the deficient knee strategy, whereas TT amputees chose their prosthetic limb as leading limb. In gait initiation TT and TF amputees favoured to start walking with the prosthetic limb first, whereas in gait termination leading with the non-affected limb was preferred in both amputee groups. Adjustment strategies were also generated in the intact musculature and joint proximal of the amputation level on

the prosthetic side. In obstacle crossing TF amputees placed the prosthetic limb over the obstacle with a circumduction movement to compensate for the limited control of knee flexion.

An important source for the creation of adjustment strategies in amputees was the non-affected limb. In standing on a moving platform the larger COP shift in the non-affected limb adjusted for the lack of control of the prosthetic foot in TT and TF amputees. In standing on a moving platform and in gait initiation and termination, weight bearing on the non-affected limb was increased in TT and TF amputees to relieve the prosthetic limb. In gait initiation and termination the period of anteroposterior GRF production was prolonged in the non-affected limb of TT and TF amputees to compensate for the reduced peak anteroposterior component of the GRF in the prosthetic limb. In obstacle crossing plantar flexion in the non-affected ankle in stance, which is known as vaulting, was larger in TF amputees to ease passing of the prosthetic limb by providing more ground clearance.

To gain insight into the adaptation process of walking with a prosthesis, a group of patients with a recent lower limb amputation was repeatedly tested during rehabilitation while performing obstacle crossing, gait initiation and termination tasks. Apart from TT and TF amputees also persons with a knee disarticulation (KD) were included in this study group. The limitations in ambulatory function were most pronounced in the early rehabilitation period. During the rehabilitation program the amputees trained motor skills and became more familiar with the prosthesis, resulting in fewer limitations in function in the prosthetic limb and showing improved adjustment strategies. In obstacle crossing all amputees touched the obstacle less frequently and increased knee flexion in the prosthetic limb during swing. However, since the increase in prosthetic knee flexion in TF and KD amputees was only small, a circumduction strategy remained necessary. In gait initiation and termination, the anteroposterior GRF in the prosthetic limb increased, whereas regaining sufficient COP shift in the anteroposterior direction remained a problem, especially while standing on the prosthetic limb in single-limb stance.

Improving prosthetic design

Amputees ask for a prosthesis that meets their functional needs to the fullest extent. One option to improve the performance of complex motor tasks in amputees is to decrease the limitations in function of the prosthetic limb. Innovations in prosthetic design play an important role in improving the functionality of a prosthetic limb. Nowadays, prosthetic knee joints are available that are able to sense angular velocities, loading, knee angles and stride rate and adjust swing time and knee flexion by altering the damping.

The motor tasks in this thesis could benefit from a microprocessor-controlled prosthetic knee design. A microprocessor-controlled knee is likely to facilitate obstacle crossing and uphill walking. For both these tasks an increase in knee flexion in swing phase is desired. In our study almost all amputees were supplied with a conventional prosthetic knee, in which the knee axis is aligned posterior to the GRF vector in order to avoid buckling of the knee in stance when weight is applied to the prosthesis. This posterior alignment makes rapid knee flexion during preswing more difficult. A microprocessor-controlled knee can be aligned more anteriorly since knee stability in stance is achieved by an automatically increase in knee joint damping.[1,2] Moreover, swing knee flexion in a microprocessor-controlled knee is eased because the damping is automatically minimized in terminal stance.[2] Studies on level walking in amputees have already reported that the knee flexion angle of a microprocessor-controlled prosthetic knee was very similar to the non-affected limb and to AB subjects.[1,3] Moreover, authors reported a higher self-selected and fast walking velocity while walking with a C-leg (Otto Bock, Duderstadt, Germany) in comparison with a conventional mechanical knee joint.[3-5] When walking over an obstacle course that included walking on a ramp, stairs and uneven terrain and stepping over an obstacle, amputees reached a faster walking velocity when using a C-leg compared to a conventional knee.[4,6,7] A combination of less knee joint damping, a more anterior alignment and a higher gait velocity could result in an increased flexion of a microprocessor-controlled knee during swing.

The stance phase control mechanism in a microprocessor-controlled limb could support balance tasks and downhill walking. Amputees showed less body sway when standing on a moveable platform with a prosthesis supplied with a C-leg compared to a Mauch SNS knee joint (Össur, Reykjavik, Iceland).[8] Downhill walking could be enhanced by the ability of flexion in a microprocessor-controlled knee during stance. One study is available on hill

descent in amputees with a microprocessor-controlled knee. Amputees supplied with a C-leg walked faster downhill with a longer step length than amputees with a mechanical prosthetic knee.[7] Knee angles were not measured in this study. A study on level walking in amputees described some knee flexion in a C-leg at loading response.[8] However, this result was not confirmed in two other studies, in which the C-leg was extended during the whole stance phase.[2, 3] A possible explanation for this inconclusive result could be the tuning of the C-leg. With sufficient training and an adequate tuning TF amputees may be able to walk down a hill with somewhat knee flexion in a microprocessor-controlled prosthetic limb during stance.

Some studies have reported on the use of microprocessor-controlled prosthetic knee mechanisms in difficult motor tasks other than described in this thesis. When a C-leg was used in stair descent in stead of a mechanical hydraulic prosthetic knee, the vertical and anteroposterior GRF in the prosthetic limb were more similar to AB subjects.[9, 10] Maximum knee flexion in stance during stair descent was similar in a C-leg and hydraulic knee joint.[9] Performance of stair descent, measured by errors, technique, velocity and walking aids was better in a group of amputees supplied with a C-leg compared to a conventional mechanical prosthetic knee joint.[4, 7] One study focussed on the performance of stair ascent but found no differences between a C-leg or mechanical knee.[7] In three studies a cognitive demanding task was added to level walking, but all concluded that walking with a microprocessor-controlled limb was not less cognitively demanding than walking with a conventional prosthetic knee.[7, 11, 12]

To establish the benefits of microprocessor-controlled prosthetic knees in functional activities outside of level walking more research is required. The problem remaining is that a microprocessor-controlled knee is not able to alter damping in the knee within one step, which is necessary for stepping over one sole obstacle, gait initiation and termination and walking up and down a short ramp.

Besides the prosthetic knee, also the prosthetic ankle and foot are objects of interest for researchers in the field of prosthetics. Prosthetic foot design may contribute to a better performance of difficult standing and walking tasks in lower limb amputees. Adjustments can be made to the length, stiffness or ankle mobility of prosthetic feet. A longer effective roll-over shape, which is related to the length and stiffness of the keel, may allow for more

COP shift in the anteroposterior direction and result in a higher anteroposterior GRF in gait initiation and termination and while standing on a swaying platform. More range of motion in passive prosthetic feet can be provided by either a single- or multi-axis ankle joint or a less stiff keel. The ability of dorsal flexion in stance during gait initiation and walking uphill allows the foot to stay in contact with the ground for a longer period of time and to prolong the moment arm, resulting in a higher moment of force and impulse. In a similar way, a larger plantar flexion in stance could support gait termination and walking downhill. On the other hand, more ankle range of motion would be at the expense of stability in standing on moving platform, since amputees are not able to control the COP excursions adequately on the prosthetic side. Besides, when the foot is too long and stiff, more muscle power is required to roll over the foot and comfort may be reduced.

Whether these adjustments to prosthetic feet actually result in better performances of the tasks of this thesis needs more research. Studies on level walking provide some support for the positive effects of the above-mentioned adjustments. A study that focused on the arc length of prosthetic feet reported that a longer effective roll-over shape increased the external dorsiflexion moment in late stance on the prosthetic side, produced a smaller first peak vertical GRF in the non-affected limb and resulted in higher gait velocities in TT amputees.[13] In level walking strong correlations were found between the prosthetic ankle range of motion and gait velocity and between the forefoot flexion and the anteroposterior GRF in the non-affected limb in energy storing and releasing (ESAR) feet; increased ankle motion due to a less stiff keel increased gait velocity and a stiffer prosthetic forefoot resulted in a lower first peak of the anteroposterior GRF in the non-affected limb in TT amputees.[14] Adding a multi-axis ankle to a SACH or an ESAR foot produced a significant increase in propulsive impulse in the prosthetic limb in level walking, primarily by increasing the propulsive GRF duration, but also by increasing the propulsive GRF magnitude.[15]

Most of today's commercially available prosthetic feet are completely passive and their mechanical properties remain fixed with walking speed and terrain. Adaptive prosthetic feet that make use of artificial intelligence and actuator and sensor technology possess more advanced options to improve complex walking tasks. Recently, a prosthetic foot has been developed that can automatically control ankle motion (PROPRIO foot, Össur, Reykjavik, Iceland). Evidence-based studies into the gait parameters of such a foot compared to

other prosthetic feet have not been performed yet. A long-term goal in prosthetic design is the development of a prosthetic foot that can actively control joint impedance and generate motor power in order to propel an amputee forward.[17] A prosthetic foot that can actively provide more ankle dorsal flexion in swing may enhance toe clearance, which is necessary in uphill walking and obstacle crossing. Furthermore, during walking up or down a hill the ankle mobility in stance can be adjusted to the slope gradient after the first step. In adaptive prostheses an amputee may indicate their intent to start walking, for example via EMG or a mechanical signal, and the prosthesis could generate or allow an increased amount of dorsiflexion at the ankle. Similarly, during gait termination an advanced prosthesis could sense the amputee's intent to stop walking and create plantar flexion at the ankle.[16]

Another challenge in prosthetic design is improving somatosensory feedback. Recent research shows that low level noise signals can improve the detection and transmission of weak signals within the sensory system via a mechanism called stochastic resonance.[18,19] Balance control in healthy young and elderly subjects, and in patients with diabetes, stroke and a TT amputation improved by applying stochastic resonance stimulation.[18-23] In accordance with this finding, a possible application of stochastic resonance in prosthetic design could be to provide stump sockets with an electrical or mechanical noise signal. A second potential solution to improve somatosensory feedback during gait would be to supply prosthetic feet with a system that presents a vibration stimulus to the residual limb at initial contact and toe-off. In this manner an amputee is aware whether the prosthetic limb is in swing or stance phase. The intended next step in prosthetic innovation is the development of a prosthetic limb with neurosensing, in which the prosthetic device is connected to the human nervous system. In the field of upper limb prosthetics current research focuses on targeted muscle and sensory reinnervation.[24]

Although all these new developments in prosthetic design are promising and potential helpful in improving functionality in amputees, it is important to realize that only prescribing a modern and often expensive prosthetic limb to an amputee is not sufficient. An amputee will only gain a better ambulatory function when he or she is able to apply the extensive functions of the prosthetic limb in daily life. Therefore, a period of rehabilitation training is necessary to teach the amputee to use all of the functional abilities of the new prosthetic knee or foot. Besides, an amputee's ability to walk is only partly dependent on the type of prosthesis. Patient characteristics such as motivation, fear, motor capacity, stump length, muscle strength, comorbidity and age are of major importance as well. In addition, a good fitting of the prosthetic socket is essential for achieving a high level of functioning.

Improving rehabilitation training

Learning new and re-learning old motor skills consumes the largest portion of time in the rehabilitation process. Although general guidelines on amputee walking and balance training are available, evidence-based foundation of specific training methods is scarce. In most rehabilitation centres all amputees are trained according to a similar method, in which the patient is the passive recipient of the treatment, while the therapist is the expert taking the active part. However, people learn in different ways and active learning is nowadays regarded as more successful. Therefore, more attention in rehabilitation could be paid to the cognitive learning strategies of amputees, since they could be a potential important determinant of a successful rehabilitation process. In addition, the methods of learning skills to an amputee differ between rehabilitation centres. Two methods of teaching skills to patients that can be used in amputee rehabilitation are errorless learning or learning by discovery. In errorless learning a patient is prevented from making errors while learning a task, whereas in learning by discovery a patient is encouraged to try to guess or figure out the correct response and learn from any errors made. Sacha van Twillert is currently performing research in the Center for Rehabilitation UMCG on the content of rehabilitation programs. The goal of her study is to investigate whether psychosocial and behavioural interventions in order to give an amputee a more active role in the rehabilitation process would result in better rehabilitation programs.

Present knowledge in rehabilitation emphasizes the need for intensive task-specific, repetitive training in challenging environments with increasing physical demands in order to promote motor learning of a motor task.[25-28] Since difficult motor tasks require more adaptations, it is advisable to pay specific attention to these tasks during the rehabilitation process. We recommend to start early in the rehabilitation process with training of complex balance and walking tasks to accelerate the process of adjusting and renewing of motor programs in the central nervous system. Furthermore, the trained tasks should closely resemble situations or activities an amputee encounters in real life. In early rehabilitation the safety of amputees is of main importance and instructions by and

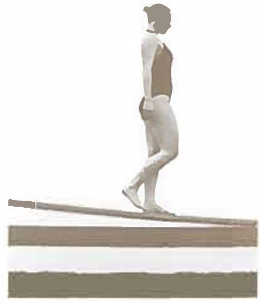
assistance of a therapist are required then. When amputees posses the basic motor skills of walking with a prosthesis, they could start learning by discovery; perform a difficult motor task and experience how to adjust the gait pattern. For this purpose a rehabilitation centre could be supplied with a track that contains obstacles of different sizes, slopes of various gradients, and uneven surfaces. An alternative approach is to train an amputee outside of the rehabilitation centre, for example in the street, in busy traffic, or on rough surfaces. These training techniques will enhance the ability of amputees to manoeuvre safely in all environments, including distracting environments.

Another more futuristic option to improve training of complex motor tasks by simulating real life situations is the use of virtual reality. Virtual reality is a computer-generated world in which the user can experience and interact with a virtual environment that resembles the real world. Sensory information, such as vision and proprioception, and real time feedback on the performance of tasks can be integrated in a virtual reality setting.[27, 29-31] The advantages of virtual reality are extensive; it is safe, can be applied early in the rehabilitation process, requires little space, can be carried out with minimal supervision, and enhances motivation and enjoyment.[27, 31-38] In rehabilitation medicine, virtual reality can be used as an assessment or exercise environment in which patients can be exposed to complex situations.[27, 32] Up to now, some evidence is reported on the positive effects of virtual reality on rehabilitation outcome. Training that uses virtual reality technology improved upper and lower limb function and obstacle crossing in subjects with a stroke[32, 36, 39-44], gait in subjects with multiple sclerosis[45], and balance confidence and ADL function in subjects with traumatic brain injury.[46, 47] Promising is that the subjects were able to transfer the virtual training to real world tasks and activities in daily living.[40, 42, 48] On the contrary, other studies have created doubts on the usability of virtual reality, because outcome parameters that reflect balance control in walking and standing deteriorated in a virtual reality environment.[49-52] In addition, from our own experience with the CAREN system we know that synchronising perturbations on a platform with events occurring on the video screen is technically difficult.

In reference to our study, virtual reality could be used to create complex environments in which amputees can exercise balance and gait tasks. While standing on a moveable platform, perturbations can be presented to which amputees can anticipate and respond, because the perturbations are in real time shown on a video screen or head-mounted display. For example, standing in an overloaded bus that drives through busy traffic can be simulated. To practise gait tasks, virtual reality can be accompanied with walking on a treadmill. With this method stepping over virtual obstacles of different magnitudes can be trained. When foot contact with an obstacle occurs, the amputee can be warned by visual, tactile or auditory feedback and has the ability to instantly adjust the stepping strategy. A tactile feedback method that corrects the movement immediately would have the preference. Feedback techniques have previously been used in training environments to improve amputee rehabilitation. In TT amputees evidence was found that auditory feedback of weight bearing information in early balance training improved rehabilitation outcome.[53] In addition, supplying visual or auditory feedback of foot pressure improved gait performance.[21] Another research reported that when amputees were stimulated to walk symmetrically by real time visual feedback, they were able to decrease energy consumption.[54]

Moreover, the adjustment strategies that are necessary to successfully perform the complex motor tasks can be used to improve rehabilitation training. The learning process of applying these adjustments strategies may be accelerated or strengthened by specific training. Since most adjustment strategies are executed by the non-affected limb, training of muscle strength and coordination of this limb is essential in rehabilitation. This training is especially of relevance for vascular amputees in which the musculature of both limbs is often weakened, since the amputation is often preceded by a long period of immobility and followed by a long period of convalescence, for example due to delayed wound healing. Although adjustment strategies are functional to fulfil a task, the application of some of these strategies also poses a risk to amputees. Previous research reported that amputees are at increased risk of developing knee pain and osteoarthritis in the non-affected limb as the result of increased limb loading and gait abnormalities.[55-57] Particularly in vascular amputees, in whom the non-affected limb is affected by diabetes or arteriosclerosis as well, continuous overloading of this limb is harmful.

For a good evaluation of rehabilitation training, it is important to register the gait pattern of amputees during the rehabilitation process. A simple and widespread available method is to record video images. When several recordings are made during the rehabilitation period, the process of adapting to the prosthetic limb can be registered. The rehabilitation physician and physiotherapist can use the video images to improve gait tasks in amputees.



Watching the recordings together with the amputee is a method of augmented feedback in order to inform, instruct and motivate the amputee. Besides, examining the videos in slow motion can provide details on the gait pattern that can not be seen in real time walking. Performing an extensive gait analysis in an advanced gait laboratory on every amputee is unpractical, expensive and time-consuming. Only in the case of persisting gait abnormalities or stump problems without a satisfying explanation, we would recommend performing a gait analysis in which temporal, kinematic, and kinetic data are assessed.

Final considerations

The composition of the prosthesis is of major importance for the gait pattern of amputees. Previous studies have shown that the type of prosthetic foot influences ankle mobility, energy expenditure, gait symmetry, muscle activation patterns, gait velocity and loading of the lower limbs[58-65], whereas the type of prosthetic knee affects gait velocity, stride time, swing phase duration, knee kinematics and energy expenditure.[66-68] We assume that the mechanical properties of the prosthetic devices used in our study have influenced the results. Due to the small sample size, statistical analysis of the effect of prosthetic characteristics on the performance of complex motor tasks could not be performed. Differences in suspension and alignment were not taken into account in the present study and the same applies to patient characteristics. Future research could provide an answer to the question which patient and prosthetic features have a positive effect on performing complex motor tasks. At the time of writing, at the Center for Rehabilitation of the UMCG research is conducted by Carolin Curtze on the interaction between prosthetic foot properties and the motor capacity of persons with a transtibial amputation. More knowledge about the influence of prosthetic properties on gait-related activities in amputees with different levels of motor capacity could facilitate the prescription of the optimal prosthetic components.

An ongoing point of discussion in amputee rehabilitation is whether a symmetric gait pattern is the goal that has to be achieved. In AB persons a symmetrical gait pattern is considered to be energy efficient and less effortful, but in amputees the recovery of gait parameters toward normal may not be synonymous with functional recovery. Instead, due to structural alterations in the nervous and locomotor system after a lower limb amputation, asymmetrical adjustments in the gait pattern are necessary for safe and independent walking. Whereas these adjustments are abnormal for AB persons, they may be necessary to improve the walking ability in amputees. Since the adjustment strategies also affect the gait parameters in the non-affected limb in amputees, this limb can no longer be considered as a normal limb. On the other hand, from an amputee's perspective a symmetrical gait pattern is often desired for cosmetic purposes, since most amputees prefer a walking pattern that disguises the prosthetic limb.

In gait analysis studies in amputees an important question is what kind of reference values should be used for comparison.[69] In research, including this study, comparisons are often made between amputees and AB subjects, or between the prosthetic and non-affected side and gait parameters in the prosthetic limb are considered normal when they are similar to those in the non-affected limb of either amputees or AB subjects. In order to determine functional recovery of gait in amputees, it would be preferable to compare an amputee with another amputee who possesses similar physical and prosthetic features. A solution could be to create a database with information on gait parameters of amputees which can be used as reference values. This would provide an opportunity to match every amputee with a comparable group of amputees from the database whose gait parameters are considered optimal or representative for the "average" amputee. The gait parameters could be categorized in K-levels (see Table 1) and in a more extended version also other patient characteristics and prosthetic components could be added to the database.

Table 1. K-levels

K0	The patient does not have the ability or potential to ambulate or transfer safely with or without assistance and a prosthesis does not enhance his/her quality of life or mobility.
K1	The patient has the ability or potential to use a prosthesis for transfers or ambulation on level surfaces at fixed cadence. Typical of the limited and unlimited household ambulator.
K2	The patient has the ability or potential for ambulation with the ability to traverse low level environmental barriers such as curbs, stairs, or uneven surfaces. Typical of the limited community ambulator.
K3	The patient has the ability or potential for ambulation with variable cadence. Typical of the community ambulator who has the ability to traverse most environmental barriers and may have vocational, therapeutic, or exercise activity that demands prosthetic utilization beyond simple locomotion.
K4	The patient has the ability or potential for prosthetic ambulation that exceeds basic ambulation skills, exhibiting high impact, stress, or energy levels. Typical of the prosthetic demands of the child, active adult, or athlete.

The findings of this study can not be generalized to all amputees. The population of this study consisted of a relatively high number of amputations due to trauma or tumour and consisted of subjects of young age with a normal to high functional level. This amputee group was selected because they could carry out the tasks in a safe way and were most likely to encounter difficult gait situations in daily life. However, in Europe over 85% of the amputations are performed on persons over 60 years, in whom vascular disease with or without diabetes is the most frequent reason for amputation.[53, 70] The functional capacity and mobility in elderly and vascular amputees is often limited due to intermittent claudiation, osteoarthritis, diabetic neuropathy, and age-related degeneration of the visual, vestibular and cognitive system.[53] These amputees may not be able to perform the required adjustment strategies in complex motor tasks or to use the advanced functions of modern prostheses. Instead, they may avoid difficult motor tasks in daily life, for example by walking around a high obstacle. Because the group of elderly vascular amputees will increase in the upcoming decennia, it is important to adjust the environment in which the amputee lives. The demands on elderly amputees can be limited for example by removing height differences at homes, in shops and on streets, which likely avoids falls and supports independence.

General conclusion

Safe and independent locomotion with a prosthetic limb requires the ability to perform motor tasks in complex circumstances and environments. In this thesis it is shown that persons with a unilateral lower limb amputation experienced several limitations in function in complex balance and gait tasks. They were able to execute complex motor tasks in a safe manner by using adjustment strategies in the prosthetic and non-affected limb. This knowledge is helpful for innovations in prosthetic design and training methods and may contribute to an improved performance of complex motor tasks in amputees. Future research has to be conducted to decide whether amputees can improve the performance of complex motor tasks by training methods or innovations in prosthetic components.

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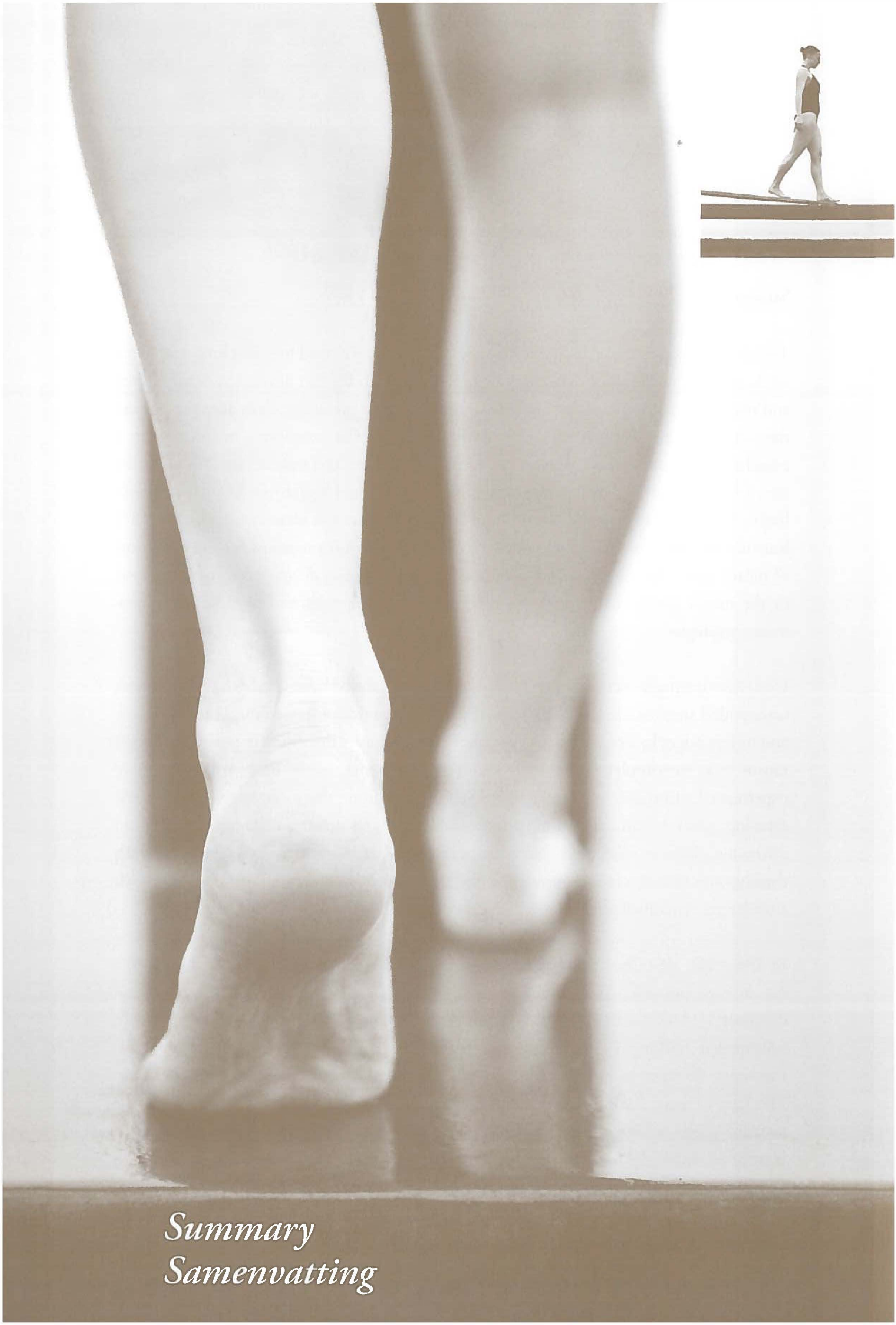


Summary
Samenvatting

Summary

This study describes the adjustment strategies that were developed by lower limb amputees while performing complex balance and gait tasks. Due to the loss of somatosensory input and the absence of musculature and one or more joints, a prosthetic limb does not possess the same functional properties as a non-affected limb. The limitations in function in a prosthetic limb cause difficulties in the control of balance and movements. An amputee uses adjustment strategies to compensate for the functional limitations of the prosthetic limb. Adaptation of balance control and movement patterns after an amputation is a learning process. The central nervous system contains motor programs for the execution of motor tasks. After a lower limb amputation the prosthetic device has to be integrated in the motor cortex by changing the existing motor strategies and by developing new motor strategies.

Until now the majority of research on motor tasks performed by amputees has focussed on unimpeded steady-state level walking and on balance control in standing. However, safe and independent locomotion with a prosthetic limb requires the ability to perform difficult motor tasks in complex circumstances and environments. In the first part of the study experienced prosthesis users were tested while performing the following tasks: obstacle crossing, gait initiation and termination, walking up and down a slope and standing on a swaying platform. This study has been performed in a motion analysis laboratory with three groups of subjects: a group with a unilateral transfemoral amputation, a group with a unilateral transtibial amputation and a control group with able-bodied subjects.

In the study described in chapter 2 the subjects were tested while stepping over a 0.1 m high obstacle. Able-bodied subjects and subjects with a transtibial amputation demonstrated an increase in knee flexion when stepping over an obstacle compared to unimpeded walking ("knee strategy"). The subjects with a transfemoral amputation showed a decrease in knee flexion, despite the technical possibilities to flex the knee. The lack of knee flexion in transfemoral amputees was compensated by bringing the prosthetic limb with a circumduction movement over the obstacle and by an increase in plantar flexion of the non-affected ankle in stance phase at the instant the prosthetic limb was above the obstacle. Due to the absent active knee function transfemoral amputees

preferred to lead obstacle crossing with the non-affected limb, whereas subjects with a transtibial amputation chose to step over the obstacle with the prosthetic limb first.

Chapter 3 and 4 report on gait initiation and gait termination respectively. In both tasks the absence of an active and flexible ankle joint resulted in adjustment strategies that were very similar for transtibial and transfemoral amputees. During gait initiation able-bodied subjects were able to shift the centre of pressure backwards and to enlarge the propulsive component of the ground reaction force. Compared to able-bodied subjects, the propulsive ground reaction force and the posterior centre of pressure shift in the prosthetic limb were decreased, especially when the prosthetic limb was used as trailing limb. The gait initiation velocity was lower for all amputees. Applied adjustment strategies by the transtibial and transfemoral amputees were more limb loading on the non-affected limb and prolonging the period of propulsive force production in the non-affected limb. Furthermore, amputees favoured to initiate gait with the prosthetic limb, since this resulted in less limitations in function and fewer adjustment strategies were required to perform the task successfully.

During gait termination the able-bodied subjects showed an increase in the braking component of the ground reaction force and a shift of the centre of pressure anteriorly at initial contact of the leading limb. The opposite was observed for amputees, where the braking ground reaction force in the prosthetic limb was decreased and an anterior centre of pressure shift was absent in leading with the prosthetic limb. The amputee subjects used several adjustment strategies: leading limb preference for the non-affected limb, longer production of braking force in the non-affected limb, more weight bearing on the non-affected limb and a decreased gait termination velocity. Leading gait termination with the non-affected limb may be positive for adequate deceleration in amputees, but is not always applicable in daily life.

In chapter 5 the subjects were tested while walking up and down a 5% gradient slope. In uphill walking, hip and knee flexion at initial contact and hip flexion in swing were increased for the prosthetic limb of subjects with a transtibial amputation compared to able-bodied subjects. In downhill walking, transtibial amputees showed more knee flexion on the prosthetic side in late stance and swing phase. For subjects with a transfemoral amputation no increase in prosthetic knee flexion was observed in uphill and downhill walking. In accordance with obstacle crossing, the transfemoral amputees did not use all the functional possibilities of the prosthetic knee during swing phase when walking on a slope. In stance phase prosthetic knee flexion is only possible to a limited extent, dependent on the type of prosthetic knee and the alignment, to prevent unlocking during weight bearing.

Chapter 6 describes how subjects maintained balance during standing on a platform that moved in the anteroposterior direction in three conditions: eyes open, blindfolded and while performing a dual task. Compared to able-bodied subjects, the total generated anteroposterior ground reaction force was larger in both the prosthetic and non-affected limb of subjects with an amputation. For this group the centre of pressure displacement was increased in the non-affected limb and decreased in the prosthetic limb. Furthermore, amputees loaded more body weight on the non-affected limb. Blindfolding or adding a dual task did not influence the results significantly. Subjects with an amputation compensated for the loss of active ankle function by increasing the movements and loading in the non-affected limb.

The second part of this thesis focussed on the learning process of applying the necessary adjustment strategies in complex motor tasks. In chapter 7 a prospective study is presented in which we describe the adjustments in gait characteristics occurring after a lower limb amputation. A group of subjects with a recent transtibial, transfemoral or knee-disarticulation amputation was repeatedly tested during and after the rehabilitation period while performing obstacle crossing, gait initiation and termination tasks. The study showed that the subjects developed several adjustment strategies to execute the tasks. With progression of time since the amputation, the obstacle was touched less frequently and knee flexion in the prosthetic limb increased. However, for subjects with a transfemoral amputation and knee disarticulation flexion in the prosthetic knee was not sufficient to clear the obstacle and a circumduction movement remained necessary. This was similar to the strategies of the experienced prosthesis users. In gait initiation the centre of pressure was shifted anteriorly prior to single-limb stance on the trailing prosthetic limb during all measurements and the propulsive component of the anteroposterior ground reaction force in the prosthetic limb increased from the first to the fourth measurement. In gait termination the centre of pressure was displaced posteriorly during loading of the leading

prosthetic limb and the braking component of the anteroposterior ground reaction force in the prosthetic limb increased gradually.

Chapter 8 contains the main conclusions of this thesis and recommendations to optimize the performance of complex motor tasks for amputees. Improving the prosthetic knee and ankle components could reduce the limitations in functions in a prosthetic limb. Possible solutions are the use of microprocessor-controlled prosthetic knee mechanisms and adaptive prosthetic feet, adjustments to the length, stiffness or ankle mobility of prosthetic feet and improving somatosensory feedback to the stump. To accelerate the process of adjusting and renewing of motor programs in the nervous system our advice is to start early in the rehabilitation process with the training of complex balance and walking tasks. Amputees could be trained with a method called learning by discovery; perform a difficult motor task and experience how to adjust the gait pattern. Also the use of virtual realty provides opportunities for better rehabilitation programs. Future research should decide whether and to what extent persons with a lower limb prosthesis benefit from adjustments in prosthetic design and training programs when performing complex motor tasks.

Samenvatting.

Deze studie beschrijft de aanpassingsstrategieën die patiënten met een beenamputatie hebben ontwikkeld voor de uitvoering van complexe loop- en balanstaken. Door het verlies aan sensorische informatie en de afwezigheid van spieren en één of meer gewrichten bezit een prothesebeen niet dezelfde functionele eigenschappen als een niet-aangedaan been. De afgenomen functionele mogelijkheden van een prothesebeen veroorzaken moeilijkheden in de controle van balans en bewegingen. Een persoon met een beenamputatie gebruikt aanpassingsstrategieën om te compenseren voor de functionele beperkingen van het prothesebeen. Het aanpassen van de balanshandhaving en het bewegingspatroon na een beenamputatie is een leerproces. Het centrale zenuwstelsel bevat motorische programma's voor het uitvoeren van motorische taken. Na een amputatie van een been moet het prothesebeen worden geïntegreerd in de motorische cortex door bestaande motorische strategieën aan te passen of nieuwe strategieën aan te leren.

De meeste studies op het gebied van motorische vaardigheden van amputatiepatiënten hebben zich tot nu toe gericht op het lopen over een vlakke ondergrond en op balans in stand. Het veilig en zelfstandig lopen met een prothesebeen vereist echter de vaardigheid om moeilijke motorische taken uit te voeren in complexe omgevingen en situaties. In het eerste deel van het onderzoek zijn ervaren protheselopers getest tijdens het uitvoeren van de volgende taken: stappen over een obstakel, starten en stoppen met lopen, een helling op- en aflopen en staan op een bewegend balansplatform. Dit onderzoek werd uitgevoerd in een laboratorium voor houdings- en bewegingsanalyse bij drie groepen proefpersonen: een groep met een unilaterale transfemorale amputatie, een groep met een unilaterale transtibiale amputatie en een controlegroep met gezonde proefpersonen.

In de studie beschreven in hoofdstuk 2 werden de proefpersonen getest tijdens het stappen over een obstakel van 0.1 m hoog. De gezonde proefpersonen en de proefpersonen met een transtibiale amputatie lieten een toename van de knieflexie zien bij het stappen over een obstakel vergeleken met normaal lopen ("kniestrategie"). De proefpersonen met een transfemorale amputatie vertoonden juist een afname in knieflexie, ondanks de technische mogelijkheden van de protheseknie om te flecteren. Het gebrek aan knieflexie van de proefpersonen met een transfemorale amputatie werd gecompenseerd door het prothesebeen met een circumductiebeweging over het obstakel te brengen en door een

toename in de plantairflexie van de niet-aangedane enkel in de standfase op het moment dat het prothesebeen zich boven het obstakel bevond. Door de afwezigheid van een actieve kniefunctie prefereerden de transfemorale amputatiepatiënten om eerst met het niet-aangedane been over het obstakel te stappen, terwijl de transtibiale amputatiepatiënten bij voorkeur eerst met prothesebeen over het obstakel stapten.

Hoofdstuk 3 en 4 beschrijven respectievelijk het starten en stoppen met lopen. In beide taken leidde de afwezigheid van een actief en flexibel enkelgewricht tot aanpassingsstrategieën die vrijwel identiek waren bij transfemorale en transtibiale amputatiepatiënten. Bij het starten met lopen waren de gezonde proefpersonen in staat om het aangrijpingspunt van het lichaamsgewicht naar achteren te verplaatsen en om de propulsieve component van de grondreactiekracht te vergroten. In vergelijking met gezonde proefpersonen was de propulsieve grondreactiekracht en de achterwaartse verplaatsing van het aangrijpingspunt van de grondreactiekracht van het prothesebeen afgenomen, met name wanneer het prothesebeen als volgbeen fungeerde. De snelheid waarmee het lopen gestart werd was lager bij proefpersonen met een beenprothese. Het plaatsen van meer lichaamsgewicht op het niet-aangedane been en het gedurende langere tijd produceren van een propulsieve grondreactiekracht in het niet-aangedane been waren aanpassingsstrategieën die door de proefpersonen werden toegepast. Tevens gaven de amputatiepatiënten er de voorkeur aan om eerst met het prothesebeen uit te stappen, omdat dit resulteerde in minder functionele beperkingen en er minder aanpassingsstrategieën nodig waren om de taak succesvol uit te voeren.

Bij het stoppen met lopen vertoonden de gezonde proefpersonen een toename in de afremmende component van de grondreactiekracht en een voorwaartse verplaatsing van het aangrijpingspunt van deze kracht tijdens voetcontact met het leidende been. In tegenstelling hiermee was bij de proefpersonen met een amputatie de afremmende grondreactiekracht in het prothesebeen afgenomen en was de voorwaartse verplaatsing van het aangrijpingspunt van de grondreactiekracht afwezig wanneer het prothesebeen gebruikt werd als leidend been. De proefpersonen met een amputatie gebruikten diverse aanpassingsstrategieën: het niet-aangedane been was bij voorkeur het leidende been, in het niet-aangedane been werd gedurende een langere periode een remmende grondreactiekracht waargenomen, het lichaamsgewicht werd voor een groter deel op het niet-aangedane been geplaatst en de snelheid waarmee gestopt werd was lager.

Eerst stoppen met het niet-aangedane been kan bij patiënten met een beenamputatie een adequate snelheidsvermindering ondersteunen, maar is niet altijd toepasbaar in het dagelijks leven.

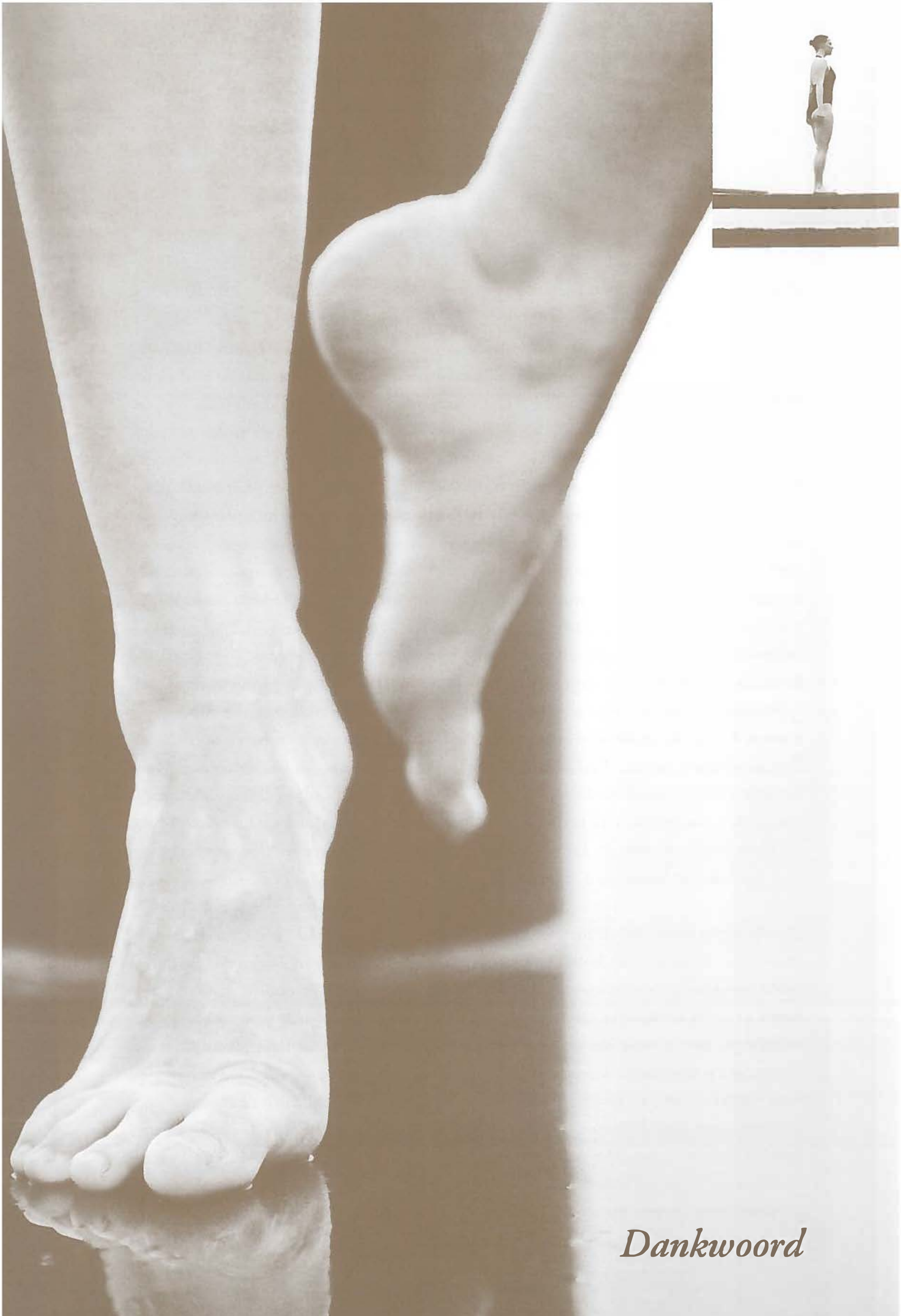
In hoofdstuk 5 werden de proefpersonen getest tijdens het op- en aflopen van een helling met een hellingsgradiënt van 5%. Bij het omhoog lopen lieten de proefpersonen met een transtibiale amputatie in vergelijking met gezonde proefpersonen in het prothesebeen een toegenomen flexie zien van de heup in de zwaaifase en van de heup en knie bij voetcontact. Bij het aflopen van een helling vertoonden deze proefpersonen meer knieflexie in het prothesebeen in de late standfase en in de zwaaifase. Bij proefpersonen met een transfemorale amputatie werd zowel tijdens het omhoog als het omlaag lopen geen toename van de knieflexie van het prothesebeen waargenomen. In overeenstemming met het stappen over een obstakel gebruikten de amputatiepatiënten tijdens het lopen over de helling niet alle functionele mogelijkheden van de protheseknie in de zwaaifase. In de standfase is knieflexie maar beperkt mogelijk, afhankelijk van het soort knie en de uitlijning, om ontgrendeling van de protheseknie tijdens het belasten te voorkomen.

Hoofdstuk 6 beschrijft hoe de proefpersonen hun balans handhaafden tijdens het staan op een platform dat bewoog in voorachterwaartse richting onder drie condities: ogen geopend, geblindeerd en met een aandachtvragende taak. In vergelijking met gezonde proefpersonen was bij de proefpersonen met een amputatie de in totaal gegenereerde voorachterwaartse grondreactiekracht in het prothesebeen en in het niet-aangedane been groter. Bovendien was de verplaatsing van het aangrijpingspunt van de grondreactiekracht bij deze groep toegenomen in het niet-aangedane been en afgenomen in het prothesebeen. Daarnaast plaatsten de amputatiepatiënten hun lichaamsgewicht meer op het niet-aangedane been. Blinderen of het toevoegen van een aandachtvragende taak had geen duidelijke invloed op de resultaten. De amputatiepatiënten compenseerden het verlies aan actieve enkelfunctie door de bewegingen van en de belasting op het niet-aangedane been te vergroten.

Het tweede deel van het onderzoek richtte zich op het leerproces van het toepassen van de benodigde aanpassingsstrategieën voor het uitvoeren van complexe motorische taken. In hoofdstuk 7 wordt een prospectief onderzoek gepresenteerd waarin we de aanpassingen van het looppatroon beschrijven die optraden na een beenamputatie. Een

groep proefpersonen met een recente transtibiale of transfemorale amputatie of knie-exarticulatie werden meermaals getest tijdens en na de revalidatie terwijl ze stapten over een obstakel en startten en stopten met lopen. Uit het onderzoek bleek dat de proefpersonen diverse aanpassingsstrategieën ontwikkelden om de taken uit te voeren. Naar mate de amputatie langer geleden verricht was, werd het obstakel minder vaak geraakt en nam de knieflexie in het prothesebeen toe. De knieflexie bij proefpersonen met een transfemorale amputatie en knie-exarticulatie was echter te beperkt om voldoende voetheffing te krijgen en een circumductiebeweging bleef noodzakelijk in de zwaaifase. Dit was vergelijkbaar met de strategieën van ervaren protheselopers. Tijdens het starten met lopen was het aangrijpingspunt van de grondreactiekracht bij alle metingen naar voren verplaatst wanneer het prothesebeen als volgbeen fungeerde en nam de grootte van de propulsieve component van de grondreactiekracht in het prothesebeen toe van de eerste tot de vierde meting. Tijdens het stoppen met lopen was het aangrijpingspunt van de grondreactiekracht naar achteren verplaatst tijdens het belasten van het leidende prothesebeen en nam de grootte van de remmende component van de grondreactiekracht in het prothesebeen geleidelijk toe.

Hoofdstuk 8 bevat de belangrijkste bevindingen van dit proefschrift en aanbevelingen om de uitvoering van complexe motorische taken door amputatiepatiënten te optimaliseren. Het verbeteren van de prothese-eigenschappen kan resulteren in meer functionele mogelijkheden van een prothesebeen. Hierbij kan gedacht worden aan het gebruik van microprocessor-gestuurde protheseknieën, adapterende prothesevoeten, aanpassingen aan de stijfheid, lengte en enkelmobiliteit van de prothesevoet en het verbeteren van de somatosensorische feedback naar de stomp. Om het proces van aanpassen en vernieuwen van motorische programma's in het zenuwstelsel te bevorderen, adviseren we om vroegtijdig in het revalidatieproces te starten met het trainen van complexe balans- en looptaken. Amputatiepatiënten kunnen getraind worden volgende een methode "learning by discovery", waarbij ze een taak uitvoeren en zelf ervaren hoe ze het looppatroon moeten aanpassen. Ook het gebruik van virtual reality biedt mogelijkheden om revalidatieprogramma's te verbeteren. Toekomstig onderzoek moet uitwijzen of en in welke mate patiënten met een beenprothese tijdens het uitvoeren van complexe motorische taken profiteren van aanpassingen aan het protheseontwerp en trainingprogramma's.



Dankwoord

Dankwoord

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