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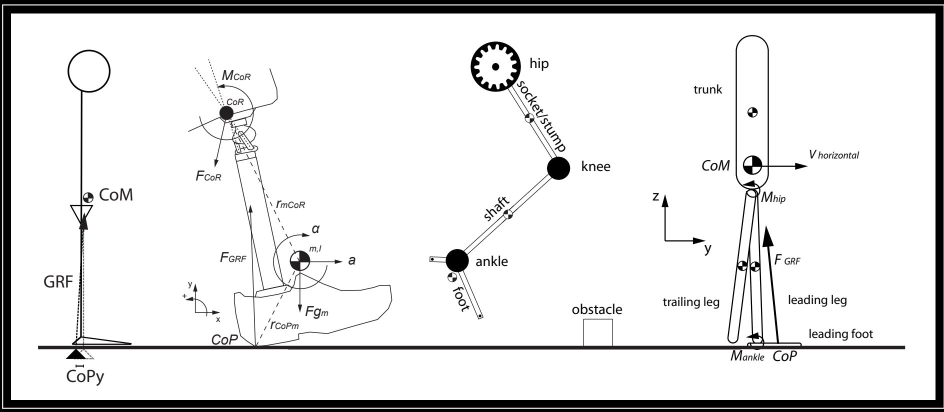
Predictions on how not to walk symmetrically with a mechanical prosthetic limb



PhD thesis

Helco G. van Keeken

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Predictions on how not to walk symmetrically with a mechanical prosthetic limb

The studies presented here have been conducted at
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Predictions on
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- **Controlling horizontal deceleration during gait termination in transfemoral amputees; measurements and simulations.**
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  Med Eng Phys. 2012.
- **Principles of obstacle avoidance with a transfemoral prosthetic limb.**
  van Keeken HG, Vrieling AH, Hof AL, Postema K, Otten B.
  Med Eng Phys. 2012.
- **Stabilizing moments of force on a prosthetic knee during stance in the first steps after gait initiation.**
  van Keeken HG, Vrieling AH, Hof AL, Postema K, Otten B.
  Med Eng Phys. 2012.
- **Controlling propulsive forces in gait initiation in transfemoral amputees.**
  van Keeken HG, Vrieling AH, Hof AL, Halbertsma JP, Schoppen T, Postema K, Otten B.
  J Biomech Eng. 2008.

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Contents

Chapter 1

General Introduction

1.1 Introduction

In our modern society, high impact decisions in numerous fields, varying from the tangible field of biology [1] to the more abstract financial field [2], are made based on mathematical models that predict the consequences of these decisions in the near and far future. Also in the field of rehabilitation medicine, which has a long tradition of good craftsmanship, the mathematical models become more and more important [3].

Traditionally, in clinical settings, analysis of temporal and spatial parameters [4;5;6;7;8] and functional assessments such as the timed walk test [9] and the timed up-and-go (TUG) test [10], are used to assess (limited) aspects of basic functional capacities. Also, subjective approaches such as questionnaires and functional rating scales [11] are used to assess the functional capacities, the quality of life and health status [12].
Observations done in clinical settings can be used to form (new) theories about functional ability and prosthetic performance. These theories can be represented by mathematical models. The advantage of these models is that they describe and explain the behaviour and results observed in a straight forward, quantitative and precise way. The outcome of these models can be used to gain new insights into the rehabilitation process, which contributes to the improvement of the theories. Also, these insights can be used to improve patients' functional ability and prosthetic performance in a more practical way. Therefore, these mathematical models are an asset to the field of rehabilitation medicine. Biomechanical analysis can offer additional information about functional ability and prosthetic performance [13;14;15], based on motion, force and muscle activity data. Inverse dynamics mathematical models can be used to calculate the joint forces, moments and powers, which can not be measured directly. The output of these models helps to understand adaptations in motor strategies and energy transfer mechanisms, that occur with respect to the integration of the prosthesis into the dynamic system of the individual [16]. Also, forward dynamics mathematical models allow researchers to inspect systematically the consequences of certain input parameters, without the interference of compensations strategies, which would occur in human subjects. The findings and insight gained with these models can help professionals working in this field to improve the functional ability of transfemoral (TF) prosthetic limb users by making the current rehabilitation programs, from early gait re-education to movement refinement and performance optimization, better and to create new and better prosthetic limbs (figure 1.1).

1.2 Prosthetic limb gait phenomena in transfemoral amputees observed in the real world

Maintaining sufficient balance is essential during gait, which can be seen as a process of continuous falling according to the German philosopher Hans Vaihinger (1852 - 1933) [17].

'Gehen ist ein reguliertes Fallen: mit jedem Schritt fällt der Mensch auf eine Seite durch Veränderung seines Gleichgewichtes und sucht den Fall durch Vorsetzung des anderen Fusses zu hemmen; auf dem antagonistischen Spiel solcher Funktionen beruht nicht bloß das Gehen,

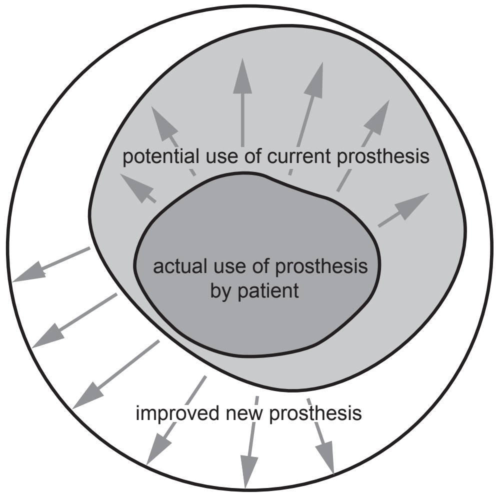


Figure 1.1: Improve the functional ability of transfemoral (TF) prosthetic limb users by providing insights and knowledge that contribute to the development of improved rehabilitation programs and prosthetic limbs.

sondern beruhen auch sonstige organische Bewegungen.' *

This continuous falling, which is an antagonistic interaction between the two lower limbs, can be divided in four specific stages of prosthetic gait: gait initiation, weight bearing, prosthetic limb swing and gait termination. During these stages the prosthetic limb user has to control the inequality between the two limbs that take part in the teamwork, which is necessary during gait, to control the motions of the center of mass (CoM) by changing the orientation of the ground reaction force (GRF) and its origin, the center of pressure (CoP) under the prosthetic foot.

1.2.1 Gait initiation

Gait initiation is a task that challenges the balance control system by forcing an individual from a state of stable balance to an unstable motion during walking [18;19;20;21]. Gait initiation demands a complex integration of neural mechanisms, muscle activity and biomechanical forces [22;23;24]. In persons with an amputation of the lower extremity, gait initiation may cause difficulties, because it is not possible to use an active ankle strategy [25] and the reduced sensory input system in the prosthetic limb. The lack of ankle strategy, which normally contributes to the posterior displacement of the CoP at gait initiation and thereby creating a forward momentum [21;26] off the CoM, has to be compensated for with other strategies [27;28]. The lack of propulsive force during the end of the stance phase of the trailing prosthetic limb due to the absence of the calf muscles also influences the amputees' performance [25;29;30;31].

*Walking is a regulated falling: with every step a human being falls to one side by changing his equilibrium and tries to inhibit the falling by placing the other foot forward: the antagonistic interaction during such functions is not only the base for walking, but also for other organic based movements.

1.2.2 Weight bearing

The preference of experienced TF amputees to initiate gate with their prosthetic limb leading, indicates that they have implicit knowledge of the active control possibilities in their sound ankle, which they use to gain forward velocity [27]. Because of these active control possibilities it seems advisable to initiate gait with the prosthetic limb leading. When considering the first step after gait initiation, in which the leading prosthetic limb becomes the stance limb and is used for weight bearing, the leading limb has to be placed in such a manner that sufficient knee stability is reached when loading the limb. The magnitude and orientation of the GRF under the prosthetic foot is determined by the angle at which the limb is placed, the internal moment of force around the hip joint, the angular velocities and gravitational forces on the body segments. When this GRF generates an external moment of force around the knee joint that remains within the limits of the knees stability, the knee will not buckle and stable stance will be achieved. In contrast to experienced prosthetic limb users, inexperienced patients are taught to initiate gait with their sound limb leading in the initial stage of therapy in our rehabilitation facility. This strategy ensures a stabilizing external extension moment on the prosthetic stance limb during gait initiation and minimizes the risk of falling during the first step, as the sound limb, with more control possibilities, becomes the stance limb. Consequently, in the second step the prosthetic limb becomes the stance limb again, with the same need to stabilize the knee.

1.2.3 Prosthetic limb swing

To make these steps, the prosthetic limb has to be swung forward, while the sound stance limb establishes a base of support that appropriately maintains stability to avoid slipping or falling. To move the limb forward in a safe and precise manner, the TF amputees have to take into account not only the properties of the prosthetic limb, but also the environment in which they are walking, especially during obstacle avoidance, which is a common problem during daily living activity [32]. During a very complex combination of movement strategies, the swing limb must clear an obstacle successfully to avoid tripping [33;34]. The applied joint moments of the swing limb and the obstacle-foot distance during stance determine the clearance achieved during obstacle avoidance [35]. Active flexion of the knee, as seen in able-bodied (AB) subjects and transtibial (TT) amputees [36;37;38;39;40], is not possible with a TF prosthetic limb. TT amputees increase swing hip elevation and hip and knee flexion as a function of obstacle height during obstacle avoidance. An increase of the knee flexion on the prosthesis side is achieved by modulating the moment of force at the hip, not at the knee, as seen on the amputee's sound side [41;42]. In addition, the stance limb hip flexion, knee flexion and (on the sound side) ankle plantarflexion increase slightly with increased obstacle height, but the stance limb hip elevation does not. The lack of a knee strategy in TF amputees is compensated for by circumduction at the hip on the prosthesis side and by plantar flexion on the sound side [43]. However, the extension strategy also has disadvantages. Not only does it make the prosthetic limb more visible, but also changes in the gait cycles are necessary when accelerating and decelerating the prosthetic limb in a lateral direction. Therefore, more degrees of freedom must be controlled. Additional free space is necessary for the clearance as the foot moves farther outward.

1.2.4 Gait termination

Successful gait termination with a TF prosthetic limb requires indirect control over this device with limited degrees of freedom. Gait termination studies in AB subjects show that several strategies are used to reduce the forward motion of the CoM [44;28]. By placing the leading limb on the ground in front of the body, a CoP under the foot is formed. The GRF originating from this CoP is used to decelerate the CoM. Also, by decreasing the push-off force with the trailing limb the forward motion is reduced [45;46;47;48]. During gait termination, the leading limb is for the most part responsible for the production of the necessary braking force [47]. As a result of the absence of active control in the ankle joint, a prosthetic limb produces less braking ground reaction force under the leading prosthetic limb, compared to the force under the sound limb in a sound limb leading situation [44]. Studies in prosthetic limb users show that the motion of the CoP is directly related to the stiffness of the prosthetic ankle, the orientation of the shaft, the position of the foot and the type of foot that is used [49;50].

1.3 Aim of the thesis

In the current thesis 'Model and measurement studies on stages of prosthetic gait.', we use several two dimensional (2D) inverse and forward dynamics mathematical models to investigate principles of TF amputee prosthetic gait. For every aforementioned stage of gait a mathematical model is designed which allows us to conceptually analyze phenomena observed in the real world (figure 1.2), based on the ideas of Dym [51].

'In an elementary picture of the scientific method (see figure 1.2), we identify a "real world" and a "conceptual world". The external world is the one we call real; here we observe various phenomena and behaviors, whether natural in origin or produced by artifacts. The conceptual world is the world of the mind - where we live when we try to understand what is going on in that real, external world. The conceptual world can be viewed as having three stages: observation, modeling, and prediction.
In the observation part of the scientific method we measure what is happening in the real world. Here we gather empirical evidence and "facts on the ground". Observations may be direct, as when we use our senses, or indirect, in which case some measurements are taken to indicate through some other reading that an event has taken place. For example, we often know a chemical reaction has taken place only by measuring the product of that reaction.
In this elementary view of how science is done, the modeling part is concerned with analyzing the above observations for one of (at least) three reasons. These rationales are about developing: models that describe the behavior or results observed; models that explain why that behavior and results occurred as they did; or models that allow us to predict future behaviors or results that are as yet unseen or unmeasured.
In the prediction part of the scientific method we exercise our models to tell us what will happen in a yet-to-be-conducted experiment or in an anticipated set of events in the real world. These predictions are then followed by observations that serve either to validate the model or to suggest reasons that the model is inadequate.
The last point clearly points to the looping, iterative structure apparent in figure 1.2. It also suggests that modeling is central to all of the conceptual phases in the elementary model of the scientific method. We build models and use them to predict events that can confirm or

deny the models. In addition, we can also improve our gathering of empirical data when we use a model to obtain guidance about where to look.'

The outcome of the mathematical models are used to make predictions about certain choices in strategies that can be made when walking with a prosthesis. In the current thesis, data of TF amputees and AB subjects using a kneewalker prosthetic device [52] are used to validate the models, which are all checked for (internal) consistency, conservation of energy and unrealistic values [51]. The four stages that were studied are described in separate chapters in this thesis.
The outcomes and insights gained from these studies are used to predict how TF amputees can compensate for the limitations in the active control of the CoP position during gait initiation and termination, and the absence of active knee extension control during weight bearing on the prosthetic limb. Also, these findings are used to predict which strategies contribute to successful weight bearing and prosthetic limb forward swing during obstacle avoidance. They provide insights into what should be taken into account during obstacle avoidance. These models can be of interest when forming new ideas about how to improve prosthetic limbs and how to improve rehabilitation programs.

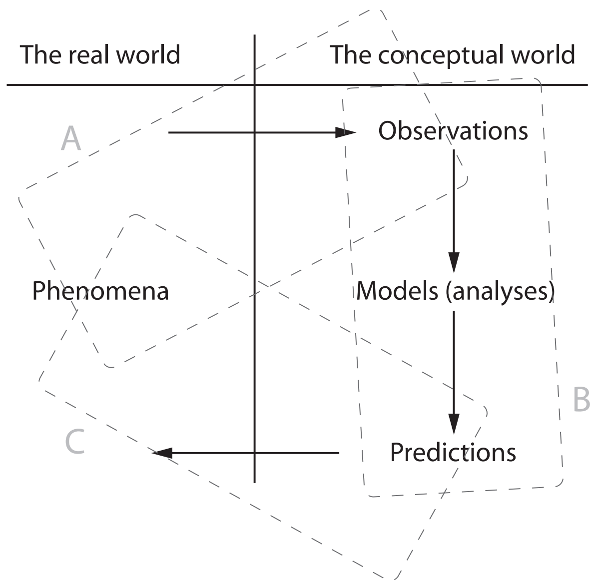


Figure 1.2: An elementary depiction of the *scientific method* that shows how our conceptual models of the world are related to observations made within that real world [53] (original figure from Dym, 2004).
A: Phenomena and observations, described in the thesis of dr. A.H. Vrieling (2009).
B: Observations, models and predictions, described in the current thesis.
C: Predictions verified in the real world.

1.4 Background of the thesis

This thesis is part of the project 'Postural control after lower limb amputation; changes in body representation and the recovery in postural control'. The project is the result of a collaboration between the Center for Rehabilitation Medicine of the University Medical Center Groningen, the Netherlands and the Center for Human Movement Sciences of the University Medical Center Groningen, University of Groningen, the Netherlands.

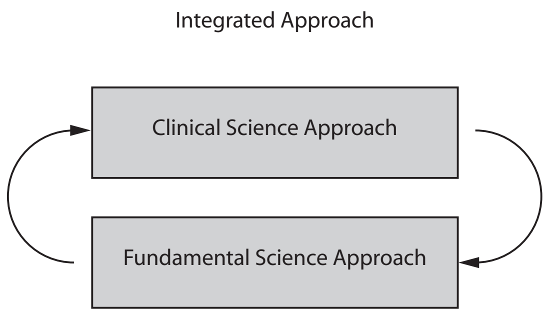


Figure 1.3: The integrated approach unites two types of research: from a clinical approach and from an fundamental science approach. Findings from the clinical science approach are tested in the fundamental science approach and visa versa.

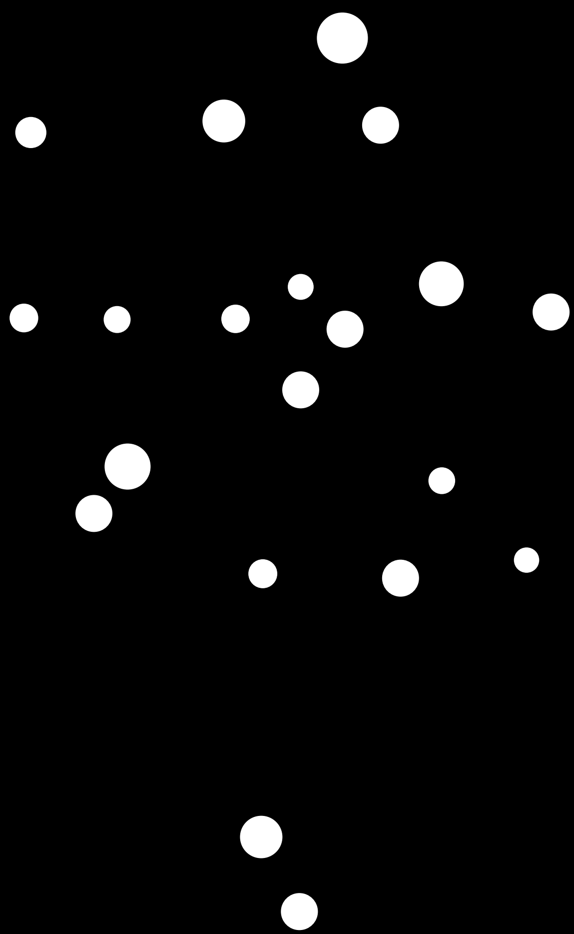
This integrated approach unites two types of research: research from a clinical science approach and research from an fundamental sciences approach (figure 1.3). The clinical research was performed by medical specialist for rehabilitation Aline H. Vrieling [8]. Her thesis (figure 1.2A) formed the base for the current thesis (figure 1.2B). Many of the findings that were reported in her thesis are studied from a biomechanics perspective in the second part of this project. Parts of the datasets that were reported in her thesis are also used in the current thesis.

1.5 Outline

In the current thesis, data of TF subjects, data of AB subjects with a kneewalker prosthetic device and 2D inverse and forward dynamics mathematical models are used to investigate the influence of prosthetic limb properties and the interaction with the floor and obstacles. In chapter 2, we describe how the position of the CoP, the CoM and orientation of the GRF are used for gait initiation. Chapter 3 focuses on loading weight and weight bearing on the prosthetic limb and producing an adequate external moment of force on the knee. Moving the prosthetic limb and making use of ground friction to get foot clearance are described in chapter 4. Chapter 5 handles the gait termination process and the consequences of prosthetic limb properties. In chapter 6 we discuss the findings, predictions, implications, limitations and future direction of research and expectations.

Chapter 2

Controlling propulsive forces; gait initiation in transfemoral amputees.



Abstract

During prosthetic gait initiation transfemoral amputees control the spatial and temporal parameters which modulate the propulsive forces, the positions of the center of pressure and the center of mass. Whether their sound limb or prosthetic limb is leading, the transfemoral amputees reach the same end velocity. We wondered how the center of mass velocity build up is influenced by the differences in propulsive components in the limbs and how the trajectory of the center of pressure differs from the center of pressure trajectory in healthy subjects. Seven transfemoral subjects and eight able-bodied subjects were tested on a force plate and on an eight meter long walkway. On the force plate, they initiated gait two times with their sound limb leading and two times with their prosthetic limb leading. Force data were used to calculate the center of mass velocity curves in horizontal and vertical directions. Gait initiated on the walkway was used to determine the limb preference. We hypothesized that because of the differences in propulsive components, the motions of the center of pressure and the center of mass have to be different, as ankle muscles are used to help generate horizontal ground reaction force components. Also, due to the absence of active ankle function in the prosthetic limb, the vertical center of mass velocity during gait initiation may be different when leading with the prosthetic limb compared to when leading with the sound limb. The data showed that whether the transfemoral subjects initiated gait with their prosthetic limb or with their sound limb, their horizontal end velocity was equal. The subjects compensated the loss of propulsive force under the prosthesis with the sound limb, both when the prosthetic limb was leading and when the sound limb was leading. In the vertical center of mass velocity a tendency for differences between the two conditions was found. When initiating gait with the sound limb, the downward vertical center of mass velocity at the end of the gait initiation was higher compared to when leading with the prosthetic limb. Our subjects used a gait initiation strategy that depended mainly on the active ankle function of the sound limb; therefore they changed the relative durations of the gait initiation anticipatory postural adjustment phase and the step execution phase. Both limbs were controlled in one single system of gait propulsion.
The shape of the center of pressure trajectories, the applied forces and the center of mass velocity curves are described in this chapter.

ground reaction forces, prosthetic gait initiation, motor strategy, center of pressure, center of mass, velocity

2.1 Introduction

Gait initiation is a task that challenges the balance control system by forcing an individual from a state of stable balance to an unstable posture during walking[18;19;20;21]. Gait initiation demands a complex integration of neural mechanisms, muscle activity and biomechanical forces[22;23;24].

In persons with an amputation of the lower extremity, gait initiation may cause difficulties, because of the disability to use an active ankle strategy[25] and the reduced sensory input system in the prosthetic limb. The lack of ankle strategy, which normally contributes to the posterior displacement of the center of pressure (CoP) at gait initiation and thereby creating a forward momentum[21;26] on the center of mass (CoM) (figure 2.1), has to be compensated for with other strategies. The lack of propulsive force during the end of the stance phase of the trailing prosthetic limb due to the absence of the calf muscles also influences the amputees' performance[25;29;30;31].

Gait initiation can be divided in an anticipatory postural adjustment (Apa) phase and a step execution (Exe) phase. During the Apa phase postural adjustments are made. These adjustments are invariably proportional to the focal gait initiation movement in the Exe phase in which the leading limb is moved forward. The postural adjustments are an integral part of the planning of the movement. The adjustments consist of muscle activation which take in account the coming change in posture and assist the movement by creating a horizontal distance between the CoP and the CoM[54;55;56;57].

After an amputation, a reorganization of motor strategies in gait initiation takes place. To regulate the speed of progression during gait initiation, the amputees control the spatial and temporal parameters of the propulsive forces. When gait is initiated with the sound limb, amputees control the intensity of the propulsive forces during the Apa phase and the Exe phase. In contrast, when gait is initiated with the prosthetic limb, the modulation of the horizontal CoM (CoMy) velocity results mainly from the propulsive forces generated during the Exe phase[58].

In a study of Michel and Chong (2004), the CoMy end velocity (VmHor) at heel contact of the leading limb, was not significantly different in the prosthetic limb compared to the sound limb. This finding may imply that the subjects treat their different limbs as a functional unit, resulting in the same VmHor when initiating gait.

Another study of Michel and Do (2002) showed that the absence of ankle and knee muscles did not affect the CoMy velocity within amputee subjects. The average VmHor of amputees is lower than the average VmHor of healthy subjects. Michel and Chong stated that the absence of ankle and knee muscles did not affect the CoMy velocity. However, a study by Kerrigan et al. (2000) implied that the vertical CoM (CoMz) velocity is influenced by the absence of the ankle musculature. In the study by Kerrigan et al. the relevance of heel rise in the reduction of CoMz vertical displacement was shown. Heel rise during the push off phase in the gait cycle results in an elongation of the CoP and CoM distance and therefore prevents the CoMz moving in a downwards direction during gait initiation. A study by Nolan and Kerrigan (2003) showed that toe-standing gait initiation influences the anterior - posterior CoP (CoPy) trajectory. Less posterior CoPy translation was seen when initiating gait in toe-standing position compared to heel-toe standing gait initiation. To compensate for this decrease CoPy translation to posterior, the subjects used a different muscle activation pattern, resulting in a delayed forward translation of the CoPy and therefore creating a greater forward momentum. Kim et al. (2003) showed in a laboratory setting in which a cadaveric leg was mounted on a foot and ankle joint simulator, that the trajectory of the

CoPy was influenced by the angle of the tibia and the foot, and the extrinsic ankle muscles. When applying an amount of 5 kg of muscle loading on the calf muscle with the tibia in upright position, a CoPy displacement in anterior direction of maximal 0.008 meter was reached. Kim et al. indicated that the study may be interpreted as simulation of a bipedal stance with a small amount of postural sway.

A prosthetic limb lacks the active adjustment of the ankle dorsiflexion and plantarflexion angle; most of the artificial ankles are passive systems. During gait initiation the CoPy moves forward as a result of an increasing torque on the ankle caused by the forward rotation of the tibia. The CoPy under the passive prosthetic ankle can not be moved to posterior during the gait initiation and therefore does not contribute to increasing forward momentum during the gait initiation in amputees. The range of this CoPy motion increases with increasing stiffness of the prosthetic ankle.

During gait initiation a prosthesis user has to deal with both an active sound ankle and a passive prosthetic ankle. It is not known how prosthesis users have incorporated the CoPy motion and the disability to actively control the trajectory in their gait initiation process under the prosthetic limb and the sound limb.

We hypothesized that because of the absence of the active ankle function in the prosthetic limb a relative smaller range of motion of the CoPy is shown within the prosthesis users during gait initiation when standing on the prosthetic limb compared to standing on the sound limb with an active ankle function. This trajectory is not actively influenced and therefore the CoPy moves in a trajectory which is related to the CoMy and tibia orientation. Still, posterior positioning of the CoPy at the beginning of gait initiation, as seen in healthy subjects, is necessary for gait initiation. Therefore we hypothesized that this posterior positioning of the CoPy is only possible when the sound limb is still in contact with the ground. Furthermore, due to the absence of active ankle function in the prosthetic limb, the CoMz velocity during gait initiation and the CoMz end velocity (VmVert) may be different when leading with the prosthetic limb compared to leading with the sound limb.

Finally, based on the findings of Michel and Chong, we expected that although there are differences in the two leading limb conditions, our subjects should be able to initiate gait with the prosthetic limb and the sound limb. They use a strategy in which their two limbs operate as a functional unit, resulting in the same CoM end velocity when initiating gait in the two conditions. However, because of the differences in the two conditions, we expect that our subjects prefer to use the prosthetic limb as leading limb during voluntarily gait initiation. In addition, observations during rehabilitation therapy show that transfemoral (TF) amputee subjects initiate gait with the prosthetic limb more often. During quiet stance at ease the TF subjects tend to stand mainly on their sound limb. This posture makes initiating gait with the prosthetic limb more likely, because the body weight is already shifted towards the stance limb.

The aim of this study is to identify modifications in the CoP and CoM movement control strategies in TF amputees in the anticipatory postural adjustment (Apa) phase and the step execution (Exe) phase during gait initiation. These data can be of interest to improve prosthetic knees and feet, to improve rehabilitation programs and to understand how the central nervous system adjusts to impairments caused by absence of active muscular control of joints.

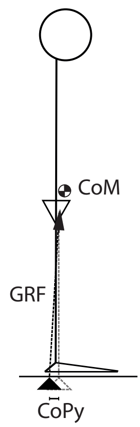


Figure 2.1: Posterior displacement of the center of pressure (CoP) creating a forward momentum.

2.2 Methods

2.2.1 Subjects

For this study, amputee subjects were recruited by a prosthetics workshop with clients in the three northern provinces of the Netherlands. Inclusion criteria for the amputee group were: a TF amputation because of trauma or oncology for at least one year, daily use of a prosthesis and the ability to walk with the prosthesis more than 50 m without walking aids. Amputees were excluded if they had any medical conditions affecting their mobility or balance, like neurological, orthopedic or rheumatic disorders, otitis media, cognitive problems, severely impaired vision, reduced sensibility of the sound limb, or the use of antipsychotic drugs, antidepressants or tranquilizers. Furthermore, amputee subjects with pain or wounds of the stump, and fitting problems of the prosthesis were excluded. A matched control group of able bodied (AB) subjects was also selected. They were recruited via advertisements at the local blood bank, hospital, television and radio station. AB subjects were included when they did not suffer from lack fitness during common daily activities. Exclusion criteria for AB subjects were orthopedic or neurological disorders, otitis media, reduced sensibility in the lower limbs and the use of antipsychotic drugs, antidepressants or tranquilizers.
The study group consisted of seven TF amputees and eight AB subjects. The medical ethics committee of the University Medical Center Groningen approved the study protocol. All subjects signed an informed consent before testing. The characteristics of the subjects are shown in table 2.1. The groups showed no statistically significant differences in gender, age, height and weight (in amputees with prosthesis). The TF group contained five right-sided and two left-sided amputees. The TF amputees used different types of prosthetic knees, all supplied with a free movable knee: Teh Lin (3), C-leg (1), Total knee (1), Otto Bock 3R60 (1) and Proteval (1). The following prosthetic feet were used by the subjects: C-walk (2), dynamic SACH (2) and Endolite (3). The subjects walked with their own shoes.

Table 2.1: Patient characteristics of the TF subjects and the AB subjects. Mean values and standard deviations of age, weight, height, time since amputation, and side of amputation. Gender and side of amputation are provided in absolute numbers. No statistically significant differences were found ($p \leq 0.05$).

Group	TF ($n = 7$)	AB ($n = 8$)
Gender (male / female)	6/1	7/1
Age (years)	44.0 ± 14.1	46.4 ± 9.7
Weight (kg)	81.4 ± 12.4	85.0 ± 10.1
Height (m)	1.82 ± 0.06	1.84 ± 0.07
Time since amputation (months)	210.7 ± 158.1	-
Side amputation (right / left)	5 / 2	-

2.2.2 Apparatus

The study was performed at the Motion Analysis Laboratory of the Center for Rehabilitation of the University Medical Center Groningen. A Bertec force plate of 0.40 x 0.60 m was used to collect the ground reaction forces (GRF) in three directions and the position of the CoP in two directions. The forces were sampled at 100 Hz. The gait initiation was recorded with two video cameras: one scanning the coronal plane, the other the sagittal plane. The video sampling frequency was 25 Hz. Recording and analysis of the force measurements and video registration was done with a custom-developed Gait Analysis System (GAS) based on LabView software. An eight meter long walkway was used to assess the leading limb preference.

2.2.3 Procedure

Force plate trials were used to obtain data on the GRF, moments, and the CoP. The data were obtained in vertical (GRFz; CoPz), anteroposterior (GRFy; CoPy) and mediolateral (GRFx; CoPx) direction. The CoM (CoMz; CoMy; CoMx) end velocity (VmVert; VmHor) and duration of the Apa phase and the Exe phase were calculated from the GRF data (figure 2.2; figure 2.3).

The start of the Apa phase was defined as the moment in time when GRFy was larger than one percent of the body weight in Newton. The end of the Apa phase and start of the Exe phase was defined as the moment in time the CoPx reached the highest velocity when transitioning from the leading limb towards the trailing limb side just before foot off. The end of the Exe phase was defined as the moment in time that GRFz was at its maximum, before the leading swinging limb touches the floor (heel strike).

The measurement started with quiet standing at ease in a bipedal standing position and ended after the subject walked at least two steps away from the force plate. The position of the feet on the force plate was self-selected. The TF subjects performed two trials with the sound limb as leading limb and two trials with the prosthetic limb as leading limb. In the AB group the temporal and spatial parameter values of the right and left limb was assessed and averaged, and used in the data analysis. This method was chosen to minimize the influence of asymmetry between the limbs. In this way we analyzed two conditions in

the AB group: the sound leading limb and the sound trailing limb.
The leading limb preference was determined from video images of four trials, in which the subjects had to walk over an eight meters long walkway. They had to stop and start walking halfway the walkway. The leading limb was self-selected. To obtain a leading limb preference score in the amputee group, the amount of trials in which the prosthetic and sound limb were used as leading limb was expressed as a percentage of the total number of trials. In the AB group the number of right and left leading limb trials was expressed as a percentage as well.

2.2.4 Outcome Parameters

The forces and moments were used to calculate the final outcome parameters, similar to Michel and Chong (2004). The CoM acceleration vector was directly calculated from Newton's second law: $m * \vec{a} = \vec{W} + \vec{F}$, where m is the subject's mass, $\vec{a}$ is the CoM acceleration, $\vec{W}$ is the subject's gravity vector and $\vec{F}$ is the GRF vector. The instantaneous CoM velocity was obtained by integration of the acceleration.
The duration of the two phases (dApa, dExe), the slope of the CoMy velocity (sApa, sExe), the CoMy velocity at Apa phase - Exe phase transition before foot off (VFO), the gain of CoMy velocity during step execution (G), the duration of the gait initiation (tVm), and the VmHor were calculated. Also the VmVert was calculated at the end of the Exe phase.

2.2.5 Statistical Analysis

For each subject, individual means of the trials for the leading and trailing (prosthetic and sound) limb were calculated. Leading limb preference was investigated by a one-group t-test. A paired t-test was performed in which the leading limb preference score of the TF and AB groups was compared to zero. The level of significance was set to $p \leq 0.05$.

2.3 Results

The results of the outcome parameters are presented in table 2.2. The leading limb preferences are presented in table 2.3. Schematic diagrams of gait initiation characteristics of the TF and AB subjects are presented in figure 2.2. This figure contains the CoPy trajectory, forces and velocities and a stick figure representation of the gait initiation. The stick figure representation is divided in 5 sub phases, the start of the Apa phase, the middle of the Apa phase, the Apa phase - Exe phase transition, the middle of the Exe phase and the end of the Exe phase. The CoPy position during these sub phases is represented by a triangle. The stick figures are based on video images.
The trajectory of CoPy, the force curves (GRFy; GRFz), the CoM velocity curves, and the Apa phase duration and Exe phase duration from typical AB subjects and typical TF subjects during gait initiation are presented in figure 2.3.

Table 2.2 shows that some significant differences were found between and within the TF and AB groups in the properties of the phases, dApa, SApa, VFO, dExe and G. Within the TF group no significant differences were found for the VmHor and tVm when leading with the sound limb or with the prosthetic limb. During gait initiation, the TF group reached a lower VmHor than the AB group. A tendency was found for differences in VmVert for the two different leading limb conditions in the TF group ($p = 0.06$). The VmVert when leading with the sound limb was higher in downward direction compared to leading with the prosthetic limb. The tVm needed to for gait initiation was not longer in the TF group compared to the AB group. The force curves and CoM velocity curves had characteristic signatures in both the TF group and the AB group. The subjects who deviate from these characteristic signatures are described at the end of the results section.

Table 2.2: Calculated gait initiation phases for AB subjects and TF subjects and their characteristics: the duration of the two phases (dApa, dExe) and the slope of the CoMy velocity (sApa, sExe), the CoMy velocity at Apa phase - Exe phase transition before foot off (VFO), the gain of the CoMy velocity during step execution (G), the duration of the gait initiation (tVm), and the CoMy end velocity (VmHor) calculated similar to Michel and Chong (2004), supplemented with the vertical end velocity (VmVert). The end velocities were calculated at the end of the Exe phase.

Calculated Item	AB Sound limb	TF Sound limb	TF Prosthetic limb
dApa (sec)	0.46 ±0.11	0.68 ±0.19 †	0.23 ±0.25 †*
SApa (m/s^2)	0.51. ±0.10	0.52. ±0.9	0.19 ±0.10 †*
VFO (m/s)	0.23 ±0.06	0.35 ±0.10 †	0.06 ±0.08 †*
dExe (sec)	0.51 ±0.04	0.35 ±0.04 †	0.72 ±0.12 †*
SExe (m/s^2)	1.54 ±0.32	1.02 ±0.24 †	0.94 ±0.24 †
G (m/s)	0.77 ±0.11	0.36 ±0.10 †	0.66 ±0.12 *
tVm (sec)	0.97 ±0.10	1.03 ±0.17	0.95 ±0.18
VmHor (m/s)	1.00 ±0.16	0.71 ±0.16 †	0.71 ±0.15 †
VmVert (m/s)	-0.13 ±0.04	-0.16 ±0.06	-0.09 ±0.05

Statistically significant P-values ($p \leq 0.05$) of differences between the TF group and the AB group are marked with †.
Statistically significant P-values ($p \leq 0.05$) of differences between the TF - sound limb leading group and the TF - prosthetic limb leading group are marked with *.

Table 2.3: Leading limb preference score for the prosthetic (P) and the sound (N) limb in TF subjects and for the right (R) and left (L) limb in AB subjects in gait initiation.

Group	limb	Leading limb preference (%)
TF (n=7)	P	71.4 ± 39
	N	28.6 ± 39
AB (n=8)	R	47.2 ± 23
	L	52.8 ± 23

No significant differences were found.

2.3.1 TF, sound limb leading

Figure 2.2 and figure 2.3a show that during the relative long Apa phase, the CoPy moves to posterior in the first part of the Apa phase and to anterior in the second part. At the end of the Apa phase a translation of the CoPy to posterior is seen. This posterior translation continues at the beginning of the Exe phase, but changes in anterior direction shortly afterwards. During the posterior translation at the end of the Apa phase, a drop in GRFy is seen. Also the GRFz drops. At the same time the CoMz velocity increases in upward direction. This increase of the CoMz velocity is preceded and followed by a decrease of the CoM velocity, resulting in a downward - upward - downward motion of the CoM. Both the CoM velocity and GRFy increase at the end of the Exe phase. The CoMy velocity increases during the total gait initiation. These typical curves were found in five out of seven TF subjects. In two TF subjects a double peak in the GRFy was seen at the end of the Apa phase.

2.3.2 TF, prosthetic limb leading

Figure 2.2 and figure 2.3b show that during a relative short Apa phase and during the first half of the Exe phase the CoPy moves in a posterior direction. During the last part of the Exe phase an anterior translation of the CoPy is seen. Compared to TF, sound limb leading, similar GRFz curves are found. The CoMz velocity curve differs from the TF, sound limb leading, group. No decrease in CoMz velocity and therefore no downward motion were found before the Apa phase - Exe phase transition. Also the GRFy curve differs from the GRFy curves in TF, sound limb leading. A continuous increasing of the GRFy is seen. The increase of the GRFy diminishes halfway the Exe phase. Combined to this diminution in the GRFy, a stationary position of the CoPy was found during the first part of this diminution. The second part was accompanied by a motion of the CoPy in anterior direction. These curves were found in six of the seven TF subjects. The curves of the deviating subject are described at the end of this section.

2.3.3 AB, sound limb leading

Figure 2.2 and figure 2.3c show that in the AB group, the dApa and dExe are almost equal. During gait initiation the CoPy moves to posterior during most of the Apa phase. At the end of the Apa phase the CoPy moves slightly to anterior. At the beginning of the Exe phase the CoPy moves to posterior again. These translation shifts were seen in at least three of four trials per subject. The rest of the Exe phase the CoPy moves to anterior again. The GRFz curves are quite similar to the curves of the TF group in both conditions. The GRFy curve also shows resemblances with the TF group, when leading with the prosthetics limb. The only difference is the more pronounced bump at the beginning of the Exe phase. The CoMz velocity curve shows no upward velocity. These curves were found in seven out of eight subjects. The curves of the deviating subject are described at the end of this section.

2.3.4 Leading limb preference

The TF subjects showed a tendency towards a preference for the prosthetic limb as the leading limb. The prosthetic limb was the leading limb in 71.4 percent of the total trials.

2.3.5 Deviators

Figure 2.4 and figure 2.5 show the data of two deviating subjects, who initiated gait differently as the majority above. Both subjects, one from the TF, prosthetic limb leading, group and one from the AB group, initiated gait while using a toe-standing strategy in the Exe phase. Before the Apa phase - Exe phase transition the most posterior position of the CoPy was reached. During the transition the CoPy translates in anterior direction. The vertical velocity increases and moves the CoM in upward direction. A decrease in GRFy was found. The GRFz curves and the CoMy velocity curves were similar to the curves of the typical subjects above.

2.4 Discussion

The aim of this study was to identify modifications in CoP and CoM movement control strategies in TF prosthesis users in the Apa phase and the Exe phase during gait initiation. In this section we discuss the CoPy motions, the applied forces and the resulting CoM velocities separately to get more insight in gait initiation strategies used by the TF subjects.

2.4.1 CoP

As stated earlier in the introduction, the lack of ankle strategy, which contributes to the CoPy displacement in posterior direction at gait initiation and thereby creating a forward momentum[21;26] on the CoM has to be compensated with other strategies. Our data show that TF subjects adopt a strategy which is mainly depending on the sound limb enabling them to move their CoPy in posterior direction while initiating gait.
The absence of active ankle function in the prosthetic limb is compensated with the active ankle function in the sound limb. When the TF subjects are leading with their prosthetic limb, the Apa phase is relative short. The prosthetic ankle does not contribute to keeping the CoPy in posterior position, while there is a need to terminate the swing phase before the next stance on the prosthesis. During the first part of the relative long Exe phase the trailing sound limb keeps the CoPy in posterior position. When the TF subjects are leading with the sound limb the TF subjects change the relative durations of the Apa phase and Exe phase, making the Apa phase longer at the cost of the Exe phase, shifting the transition more towards the end of gait initiation. This timing shift enables them to produce sufficient propulsive GRF by positioning the CoPy in posterior position. This shift is based on the use of active ankle function of the sound limb as long as possible while still being able to move the leading sound limb forward during a short swing phase.

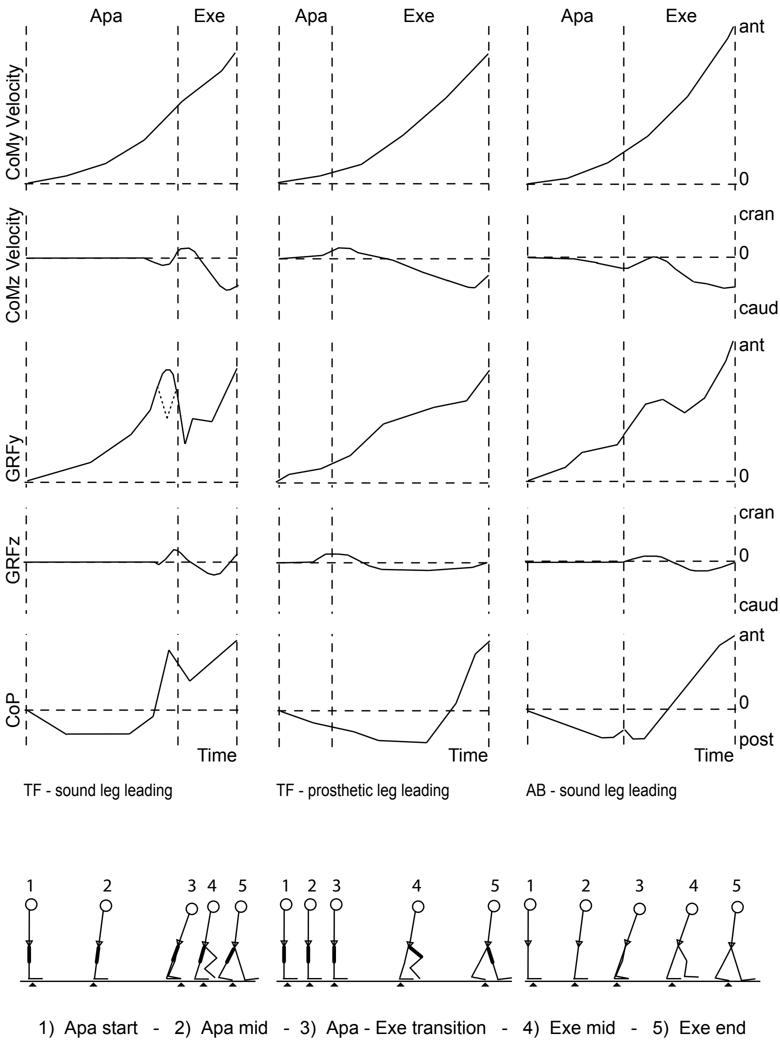


Figure 2.2: Schematic overview of typical CoMy and CoMz velocities, the forces (GRFy; GRFz) and the CoPy motion during gait initiation of TF subjects leading with their sound limb or the prosthetic limb and AB subjects. The stick figures show five sub phases of gait initiation: 1) the start of the Apa phase, 2) the middle of the Apa phase, 3) the Apa phase - Exe phase transition, 4) the middle of the Exe phase and 5) the end of the Exe phase. The CoP position during these sub phases is represented by a triangle. The stick figures are based on video images. Two typical GRFy curves were found In the TF - sound limb leading group. These curves are represented by the solid and the dotted line.

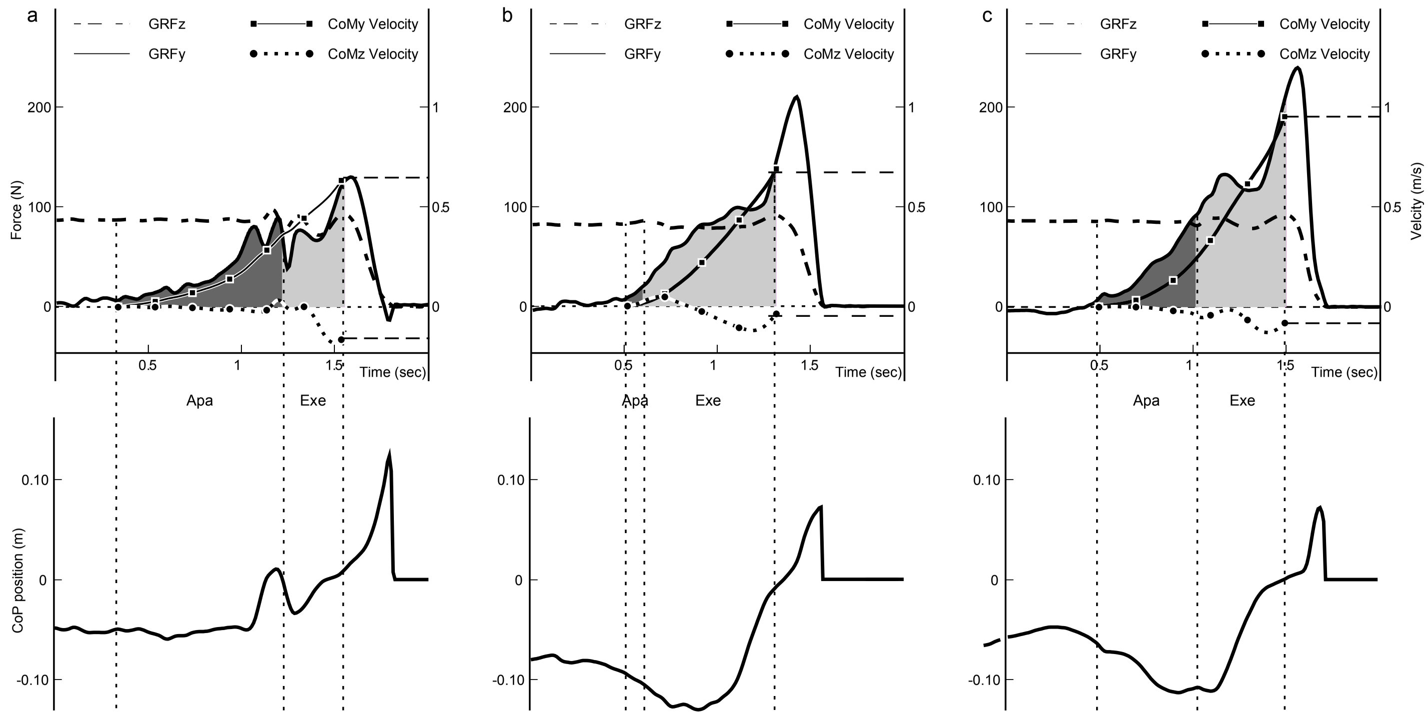


Figure 2.3: Typical time courses of GRFz and GRFy, CoPy motion and the resulting velocities of typical TF subjects (a: sound limb leading; b: prosthesis limb leading) and c: AB subjects). The time course is divided in an Apa phase and an Exe phase. The start of the Apa phase is defined as the moment in time that GRFy is bigger than one percent of the body weight in Newton. The start of the Exe phase is defined as the moment in time the CoPx reaches the highest velocity when moving from the leading limb towards the trailing limb (not in the figure). The end of the Exe phase is defined as the as the moment in time that GRFz is at its maximum, before the leading swinging limb touches the floor (heel strike). The end velocities at the moment the leading limb touches the floor are marked with the dashed lines. To fit GRFz in the figure a scaling factor of 0.1 is used.

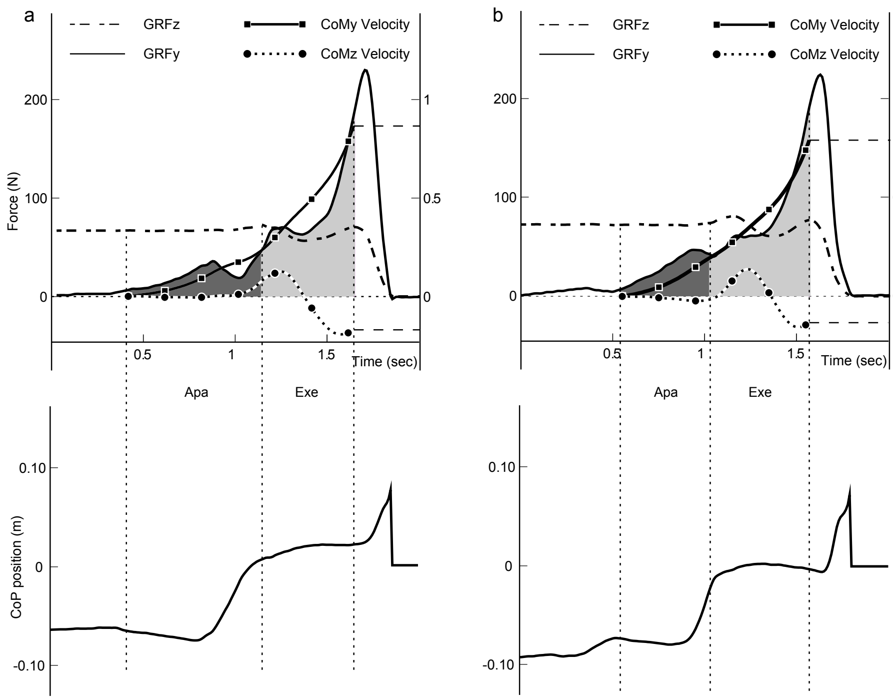


Figure 2.4: Deviator from TF amputee group - prosthetic limb (a) leading and AB group(b). The time course of GRFz and GRFy, CoPy motion and the resulting velocities of the deviating subjects, divided in an Apa phase and an Exe phase. To fit GRFz in the figure a scaling factor of 0.1 is used.

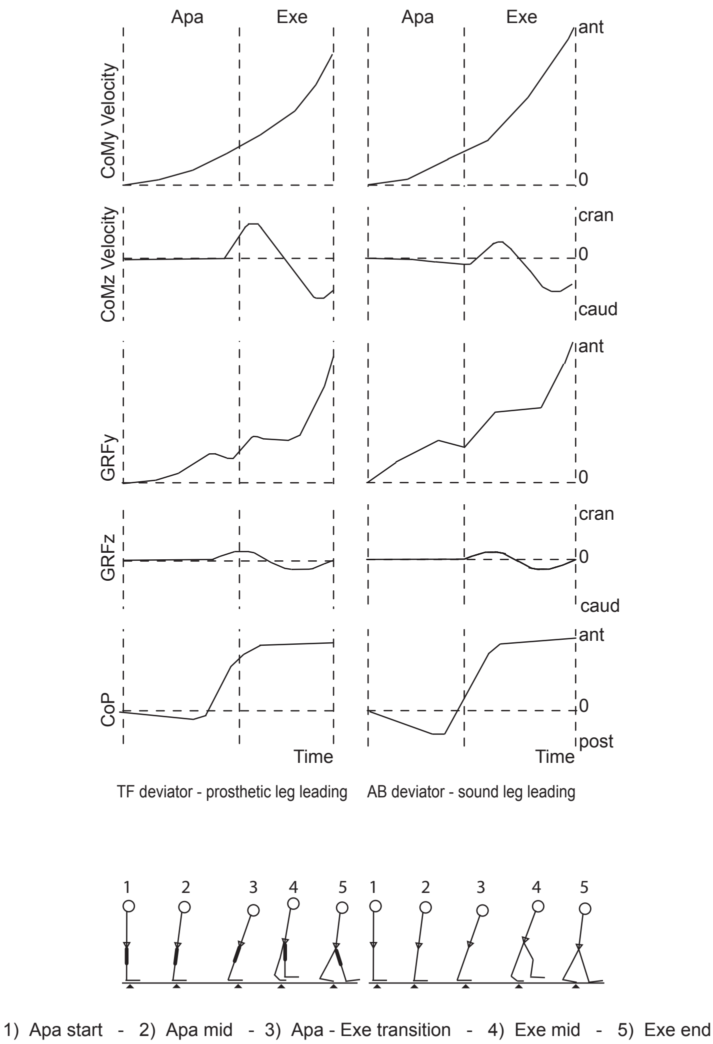


Figure 2.5: Schematic overview of typical CoMy and CoMz velocities, the forces (GRFy; GRFz) and the CoPy motion during gait initiation of TF amputee deviators leading with their prosthetic limb and AB deviators. The stick figures show five sub phases of gait initiation: 1) the start of the Apa phase, 2) the middle of the Apa phase, 3) the Apa phase - Exe phase transition, 4) the middle of the Exe phase and 5) the end of the Exe phase. The CoP position during these sub phases is represented by a triangle. The stick figures are based on video images.

The dependence on the sound limb active ankle function can be seen quite clearly during the Apa phase - Exe phase transition in the TF - sound limb leading group. Just before the transition, the CoPy moves to anterior as a result of active plantar flexion muscle activity and tibia rotation of the sound limb. At the moment that the leading sound limb leaves the ground, the CoPy moves to posterior. This motion of the CoPy is the result of repositioning the CoPy under the trailing prosthetic foot. A similar CoPy motion can be seen in the AB group, when repositioning the CoPy under the trailing sound foot. In the TF - prosthetic limb leading condition this CoPy motion is not present, because the trailing sound limb is controlling the CoPy motion.
The two deviators differ from the strategy above, due to their preference to use a toe-standing strategy in the Exe phase when initiating gait. Already in the Apa phase the CoPy moves in anterior direction. This anterior motion is caused by the preference to stand on the toes while swinging the leading limb forward.
Analysis of the CoP of the TF subject shows not only that the CoPy moves forward, but also that the CoPx does not move towards the sound limb during the first part of the Apa phase. That means that both the sound foot and the prosthetic foot are equally responsible for the CoPy movement. Therefore no active ankle plantar flexion is applied with the sound limb during this first phase. The CoPy motion is only possible with a prosthetic ankle, which is loose enough to allow the subject to fall forward, but is stiff enough to imitate the CoPy movement from under the sound foot. Analysis of the video images of the TF subject show that the heel of the prosthetic foot is lifted of the group during the end of the Apa phase. At the end of the Apa phase the CoPx moves towards the sound limb, because of the plantar flexion of the sound ankle which is necessary for the toe standing. Analysis of the video images of the TF subject on the walkway also showed that the subject used a toe-walking strategy during gait. A possible reason for this strategy is that the TF deviator tried to gain more clearance for the leading prosthetic limb during the swing phase.
Analysis of the video images of the AB subject showed that during gait, the subject did not use the toe walking strategy. Only during gait initiation the toe-standing strategy was used in the Exe phase. A possible reason for this strategy is that the AB subject tried to gain more clearance for the leading limb.

2.4.2 Forces

The GRFz curves that were applied by our groups did not differ much between the groups. The shapes of the curves were quite similar, although the position of the maximum and minimum of the GRFz curves are located differently in gait initiation. In the TF group, the maximum is around the Apa phase - Exe phase transition, while in the AB group the maximum is after the Apa phase - Exe phase transition.
The GRFy curves showed greater differences, especially when comparing the TF - sound limb leading with the TF - prosthetic limb leading and the AB group.
A drop in the GRFy curves is seen at the moment the sound limb leaves the ground during the Apa phase - Exe phase transition in the TF - sound limb leading group. Again, this drop shows that the prosthetic user uses a strategy in which the sound limb has an important role in gait initiation. Another difference which was found is the shape of the GRFy curve after the Apa phase - Exe phase transition in the TF - prosthetic limb leading group and the AB group. The TF group had a less evident increase and decrease of GRFy, compared

to the AB subjects, although both groups had a sound active ankle function in their trailing limb. An explanation for this difference can be found in the acceleration and deceleration of the swing limb.
In the AB group the swing limb is the sound limb, which has considerable higher inertia than the swinging prosthetic limb in the TF group. Also, the sound limb has to be moved forward in a relative shorter time. Since the CoPy shows a comparable profile, it must the inertia of the swing limb and the duration of the swing phase that causes this difference.
The first of the double peak which is found in half of the TF subjects - sound limb leading group, coincides with the latest moment in time the center of pressure is in the posterior position. The second peak coincides with the final push off of the sound limb. The valley in between can be explained by loss in force due to the anterior displacement of the center of pressure, coming closer to a position under the CoM. In the AB group and the group with the other TF subjects these two peaks are fused, since the remaining sound limb is able to keep the center of pressure in a posterior position even when the leading limb is making a roll over movement.
The force curves of the two deviators showed a drop in the applied forces compared to the other subjects. This drop is partly the result of the CoPy shift in anterior direction.

2.4.3 CoM

The CoMy velocity curves of the groups are all alike. The curvatures show a non-linear increase in velocity. A small difference can be seen in the curve of the TF - sound limb leading group compared to the other groups. The curve shows a small velocity change at Apa phase - Exe phase transition, as a result of the diminished GRFy as the sound limb leaves the ground. After that, the velocity increases again.
On the other hand, the vertical velocities show a clear difference. Although the shape of the curves is quite similar, the maximum velocity which can be found in the Exe phase of the TF group shows a positive (upward) velocity, while in the AB group the maximum stays below zero, showing no upward velocity. In other words, the TF group uses CoM displacement in upward direction during gait initiation. When the TF subject is leading with the prosthetic limb this CoM motion can be the consequence of several actions: 1) raising of the swing limb, 2) plantar flexion of the trailing stance ankle, and 3) elevation of the swing hip. All these motions provide extra clearance in case the prosthetic limb is the swing limb. When the subject is leading with the sound limb, the clearance issue is not the biggest concern. However, a positive CoMz velocity is found during the Exe phase. In the TF - sound limb leading group the Apa phase - Exe phase transition occurs more at the end of the gait initiation and is therefore combined with a relative short Exe phase. The TF subject has to move his sound limb upwards and forwards relatively quickly. These fast motions also result in a more pronounced CoM displacement.
The relative short Exe phase is the result of a passive ankle function of the prosthetic limb. During the relative long Exe phase when leading with the prosthetic limb the subjects make use of the active ankle function of the sound trailing limb. This strategy enables them to 1) gain more time for the swing limb, 2) maintain more height with their CoM and 3) end with less negative CoMz velocity at the end of the gait initiation.
The CoMz velocity curves from the two deviators showed relative higher velocities compared to the other subjects. A big upward velocity was found as a result of the toe-standing

preference of the subjects. In the CoMy velocity a drop in the curve was found. This drop in velocity is the result of the CoPy shift in anterior direction and the decrease in GRFy.

2.4.4 Overall

Similar to the findings of Michel and Chong (2004), the data show that although the duration of the gait initiation is the same in both the TF group and the AB group, the VmHor is lower in the TF group whether the subjects are leading with their sound limb, or with their prosthetic limb. These differences in VmHor may be caused by the absence of active ankle function in the prosthetic limb.

Although the VmHor and tVm are the same for the two leading limb conditions in the TF group, a tendency is found for the VmVert to be different between the conditions. When the TF group is leading with the sound limb, the VmVert at the end of the Exe phase in downward direction is relatively higher compared to the condition in which the TF subjects are leading with the prosthetic limb. This relatively higher VmVert is probably the result of the absence of an active ankle function in the trailing prosthetic limb. During the Exe phase the TF subjects act more like an inverted pendulum without an active ankle function when leading with the sound limb and trailing with the prosthetic limb.

Michel and Chong (2004) stated that the absence of ankle and knee muscles does not affect the CoMy velocity. The study by Kerrigan et al. (2000) implies however that the CoMz velocity is influenced by the absence of the ankle musculature. Although we did not find a significant difference in VmVert, but only a tendency, we agree with Kerrigan et al.. Not only there was a tendency of a VmVert difference, also the motion of the CoPy was clearly affected by the absence of an active ankle function in the prosthetic limb. The subjects adopt a strategy in which they depend on the functioning of the healthy ankle.

Our subjects preferred to initiate gait with their prosthetic limb on the walkway. From a gait initiation view point possible reasons for that are: 1) they are already standing with most of their weight on the sound limb before they start the gait initiation. Therefore it is easier to start the gait initiation without the necessary weight shift which occurs at the Apa phase - Exe phase transition in AB subjects who have distributed their weight evenly over both their limbs; 2) the Apa phase - Exe phase transition takes place at the beginning of the gait initiation; their VFO is still low, which might give them the impression that the risk of falling is less; 3) moving their CoM in upward direction is easier when standing still straight up; and 4) that's the way they were taught to initiate gait during rehabilitation.

When considering the next step after gait initiation, there is another reason to initiate gait with the prosthetic limb. The choice of the leading limb in gait initiation also has consequences for the next step in which the leading swing limb becomes the stance limb. Gait initiation with the sound limb in the TF groups gives the penalty of a higher downward CoMz velocity just before heel strike, which needs to be compensated by the lifting action of the sound limb when on the ground again. The prosthetic limb without the active ankle function can not contribute actively to regaining CoM height.

When initiating gait with the prosthetic limb, a lower downward CoMz velocity needs less catching action of the prosthetic limb, due to the possibilities the active ankle function of the trailing sound limb offers.

We hypothesized that because of the absence of the active ankle function in the prosthetic limb a relative smaller range of motion of the CoPy is shown within the prosthesis users

during gait initiation when standing on the prosthetic limb compared to standing on the sound limb. Also, we hypothesized that this posterior positioning of the CoPy is only possible as long as the sound limb is still in contact with the ground.
We found that the range of motion of the CoPy is mainly influenced by the sound limb. The figures show that when the sound limb is in contact with the ground a clear towards posterior - towards anterior translation of the CoPy is seen.
Due to the lack of the active ankle function in the prosthetic limb, our subjects compensated the disability to move the CoPy and generate the GRF with the active ankle function of the sound limb. They used their two limbs as a functional unit resulting in the same VmHor, whether they were leading with their prosthetic limb or their sound limb. It seems that as long as there are no adequate active prosthetic ankles, more symmetry within a TF subject during gait initiation is impossible.
We also hypothesized that because of the absence of active ankle function in the prosthetic limb, the CoMz velocity during gait initiation may be different when leading with the prosthetic limb compared to leading with the sound limb. We found a tendency in vertical end velocities. When leading with the sound limb a bigger downward VmVert was found. Also, the composition of the velocity curves during the gait initiation showed differences.
The finding in this paper can be useful for professionals who work with amputee patients. The information can help the professionals to advance their instructions about gait initiation for new amputees that just learn to walk with a prosthesis or for amputees who have trouble initiating gait. The gait initiation instructions are based on gait initiation strategies from experienced prosthetic users who have advanced their technique during common daily activity.
From a biomechanical perspective, the results of experienced prosthesis users show that it makes no difference whether gait is initiated with the sound limb leading or with the prosthetic limb leading. In both cases the same forward end velocity is reached in the same amount of time. Therefore, patients can be trained with their preferred limb leading. When a patient wants to initiate gait with the prosthetic limb, the patient should be instructed to lift his prosthetic feet of the ground as early as possible and then move the prosthetic limb forwards, while keeping the pressure under the heel of the sound foot. When the prosthetic limb is moved forward sufficiently, the CoP will move forward automatically under the sound foot. When the heel of the prosthetic limb is almost on the ground, the sound foot starts with the push off. This strategy is clinically known as the way patients tend to initiate gate. When the patient prefers to initiate gait with the sound limb, the patient should be instructed to lean forward with both feet on the ground and the hips and limbs straight. This motion will create forward velocity. As soon as the patient feels sufficient pressure under the sound forefoot, the sound limb has to be moved forward, while accepting weight on the prosthetic limb.

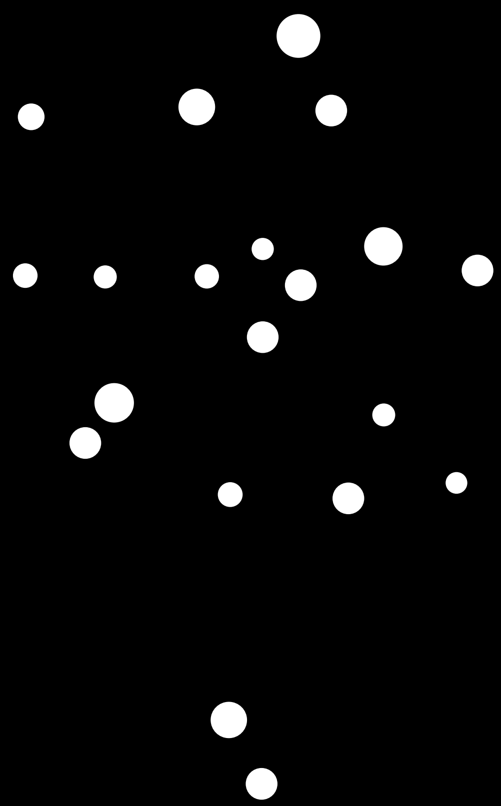
2.5 Conclusions

During gait initiation TF amputees generate anterior - posterior and vertical forces that take them from a state of stable dynamic balance to an unstable single supported posture during walking. The VmHor within the TF subjects is the same, whether the TF subjects are leading with the prosthetic limb or the sound limb. Also, the duration of the gait initiation

is the same, irrespective of which limb is chosen to be the leading limb. A tendency was found for differences in VmVert, depending on which limb was trailing.
TF subjects make use of a gait initiation strategy in which they control their two limbs as a functional unit. This strategy has a great dependency on the sound limb with the active ankle function. The Apa phase - Exe phase transition shift strategy is adapted to whether the sound limb is leading or trailing. The sound limbs active ankle function is used to move the CoPy, to manipulate the forces and to produce the end velocities.

Chapter 3

Stabilizing moments of force on a prosthetic knee during stance in the first steps after gait initiation.



Abstract

In this study, the occurrences of stabilizing and destabilizing external moments of force on a prosthetic knee during stance, in the first steps after gait initiation, in inexperienced users were investigated. Primary aim was to identify the differences in the external moments during gait initiation with the sound limb leading and the prosthetic limb leading. A prosthetic limb simulator device, with a flexible knee, was used to test able-bodied subjects, with no walking aid experience. Inverse dynamics calculations were performed to calculate the external moments. The subjects learned to control the prosthetic limb within 100 steps, without walking aids, evoking similar patterns of external moments of force during the steps after the gait initiation, either with their sound limb loading or prosthetic limb leading. Critical phases in which a sudden flexion of the knee can occur were found just after heelstrike and just before toe off, in which the external moment of force was close to the internal moment produced by a knee extension aiding spring in the opposite direction.

transfemoral prosthetic limb, gait initiation, ground reaction force, inverse dynamics, leading limb

3.1 Introduction

Amputee subjects who learn to walk with a prosthetic limb with an artificial knee, perform poorly during the initial gait training, hence the use of parallel bars, supported by therapists and other safety measures. In the weeks that follow, subjects develop adjustment strategies to improve obstacle crossing, gait initiation and gait termination [28]. It is suggested that the most significant gait adaptations occurred following receipt of a functional prosthesis. Research does not show a clear benefit in gait patterns at discharge following use of generic prosthetic devices (early walking aids with limited functionality) during the initial rehabilitation process [62]. Therefore, it is of value to study the gait initiation strategies, with fully functional prosthetic limbs, in inexperienced prosthetic limb users in the early phase of motor learning. In the current study we investigated the stabilizing external extension moments on the prosthetic knee in the first steps after gait initiation with the sound limb or prosthetic limb leading in inexperienced prosthetic limb users.

The preference of experienced transfemoral (TF) amputee subjects to initiate gate with their prosthetic limb leading, indicates that they have implicit knowledge of the active control possibilities in their sound ankle, which they use to gain forward velocity [63;27]. Because of these active control possibilities it seems advisable to initiate gait with the prosthetic limb leading. When considering the first step after gait initiation, in which the leading prosthetic limb becomes the stance limb, the leading limb has to be placed in such a manner that sufficient knee stability is reached when loading the limb. The ground reaction force (GRF) under the prosthetic foot results from the angle at which the limb is placed, the internal moment of force around the hip joint and gravitational forces on the body segments. When this GRF generates an external moment of force around the knee joint that remains within the limits of the knees stability, the knee will not buckle and stable stance will be achieved.

In contrast to experienced prosthetic limb users, inexperienced patients are taught to initiate gait with their sound limb leading in the initial stage of therapy in our rehabilitation facility. This strategy ensures a stabilizing external extension moment on the prosthetic stance limb during gait initiation and minimizes the risk of falling during the first step, as the sound limb, with more control possibilities, becomes the stance limb. Consequently, in the second step the prosthetic limb becomes the stance limb again, with the same need to stabilize the knee.

Based on differences in step length and velocity of the prosthetic limb in the steps after the gait initiation with either the prosthetic limb or the sound limb leading, we expect different ground reaction forces under the prosthetic foot. These forces generate external flexion or extension knee moments which may stabilize or destabilize the prosthetic knee. During the swing phase knee flexion is necessary for ground clearance. At the end of the swing phase, an internal hip extension moment can be applied to extend the prosthetic knee, using the inertial properties of the lower part of the prosthetic limb. This internal hip extension moment also contributes to ground reaction forces, which contribute to the stabilization of the knee during stance.

In this study, we investigated whether an inexperienced prosthetic limb user with only limited training (100 steps) is able to initiate gait and walk two steps without walking aids on a prosthetic limb with a flexible knee, without the occurrence of a sudden flexion of that knee during stance, either in a sound limb leading condition, in which the prosthetic limb is used in the second step, or in a prosthetic limb leading condition, in which the prosthetic limb is used in the first step. To ensure that our subjects had no experience with the use of a prosthetic limb, or related walking aids, and to control for learning period and comorbidity, we used able-bodied (AB) subjects. The AB subjects used a kneewalker prosthetic limb, on which AB subjects use the same compensation strategies as inexperienced prosthetic limb users, and that kinematic and kinetic analysis results are similar to gait analysis from people with TF amputations [52]. The flexible knee is equipped with an extension aiding spring, which can deliver an internal extension moment. We wondered if the evoked GRF delivers an external flexion moment that is close to the internal extension moment.

3.2 Methods

3.2.1 Subjects

Eleven inexperienced naive AB subjects (7:4 (m:f); 28 y (±3); 75.7 kg (±8.4); 1.85 m (±0.07)) wearing a kneewalker prosthetic limb (figure 3.1) with no impairments of walking volunteered to participate in the study. The healthy subjects were recruited via advertisement on a local university bulletin board. They had no known neurological or orthopedic complaints or diseases. Informed consent was obtained from all subjects before testing.

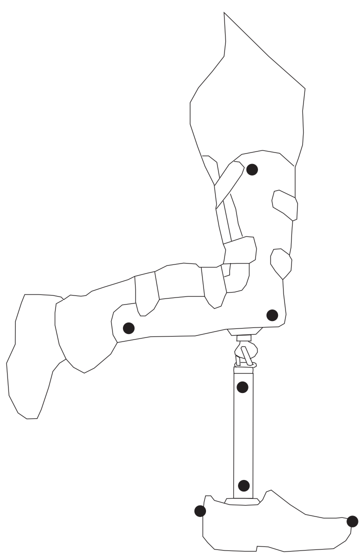


Figure 3.1: Kneewalker prosthetic limb for able-bodied subjects. The black dots indicate the optical marker positions.

3.2.2 Apparatus

We used a PRIMAS 3D motion capture camera system, a Bertec force plate and a kneewalker prosthetic limb for AB subjects. The 3D motion capture camera system consists of six infrared cameras recording at 100 Hz. Seven retroreflective markers, positioned on the socket (3: upper leg, knee joint, lower leg), the shaft (2: proximal, distal) and the foot (2: heel, toe) (figure 3.1) were used to record the motion of the prosthetic limb. The GRF and center of pressure (CoP) under the prosthetic foot were recorded at 100 Hz with the Bertec force plate. The marker data and force plate data were rotated around the vertical (y) axis and projected in the sagittal plane through the artificial knee joint, enabling us to calculate the external flexion and extension moment on the knee. The motion and force data were filtered with a second order 5 Hz low pass zero time-lag Butterworth filter and processed in MATLAB with custom made software for the 2D inverse dynamics calculations. The outcome parameters were analyzed with SPSS.
The kneewalker prosthetic limb is a prosthesis for AB subjects [52] which consists of an Otto Bock Habermann modular four bar linkage knee joint (3R36), an Otto Bock dynamic foot with toes (1D10, size 26) and a shoe (size 43/9, toe-heel length 0.30 m) (figure 3.1). The artificial knee is equipped with an extension aiding spring. This spring has two main functions. Firstly, the spring supports the forward motion of the foot and shaft at the end of the swing phase, reducing the swing time. Secondly, the spring enables a prosthetic limb user to raise the prosthetic limb forward against gravity without flexion of the knee, assumed that the motion is not performed with high accelerations. This second feature provides a prosthetic limb user control over the passive knee when positioning the prosthetic foot for the stance phase at low speed. By making use of the extension spring the prosthetic knee remains locked in full extension. The spring generates an internal moment between 45 and 0 degrees flexion. The magnitude of the moment is inversely related to the amount of flexion, decreasing down to 0 Nm at 45 degrees flexion. Hyperextension of the prosthetic knee is prevented by a mechanical stop, i.e. a very high stiffness. The spring produces a maximal internal extension moment of 12.4 Nm in full extension. The length of the shaft can be adjusted to match the contralateral leg length. The mass of the knee-shaft-socket system is 2.08 kg. The prosthetic ankle-foot system of the prosthetic limb is relatively stiff. The upper leg socket of the kneewalker is constructed in such a way that the prosthetic limb is connected to the upper and lower leg, which is fixed in 90 degrees flexion at the knee joint. Because of this construction, the AB subjects are able to put weight on the kneewalker via their knee and the socket/leg connection. In this way the prosthesis can be used in a comparable manner as a prosthesis for knee-exarticulation amputees. All subjects used the same shoe under the prosthetic foot. The heel-toe length was 0.3 m.

3.2.3 Procedure

Before the measurements the subjects were allowed to walk with the kneewalker prosthetic limb without walking aids. The subjects were not allowed to make more than 100 steps. No other instructions were given. All subjects were informed and they experienced that the kneewalker was equipped with a flexible knee that not only can flex during the swing phase, but also flexes in loaded condition when used inadequately.
Before gait initiation measurements started, the subjects had to balance on the kneewalker

with little support from the contralateral foot for at least three seconds to determine a midstance position per subject, based on the angles of the joints and segments. After this measurement all subjects were tested in two conditions. In the first condition subjects initiated gait with their sound limb leading, placing their prosthetic limb on the force plate in the second step. In the second condition subjects initiated gait with their prosthetic limb leading, placing the prosthetic limb on the force plate in the first step. After gait initiation the subjects had to continue walking two steps at their own preferred speed. The subjects were randomly distributed in terms of which condition was tested first. After five successful trials (foot on force plate, no sudden flexion of the knee during stance) in one condition the subjects initiated gait with the opposite limb in the other condition. Only successful trials of gait initiation were used for the analysis. The subjects did not use walking aids during gait initiation.

3.2.4 Outcome parameters

The forces and markers on the prosthetic limb in both leading limb conditions were used to determine 1) the ground reaction force in horizontal (F_x) and vertical (F_y) direction, 2) the position of the CoP over time under the prosthetic foot (CoP_x), 3) the segment angle of the shank, 4) the joint angle of the knee, 5) the velocity of the heel just before heel strike, 6) the external moment (M) on the center of rotation (CoR) of the knee (figure 3.2), and 7) the horizontal (a_x) and vertical (a_y) linear acceleration and angular (α) acceleration of the knee-shank-foot system. For the analysis of the data we made a distinction between subjects that used knee flexion during the swing phase with the prosthetic limb and subjects that kept their prosthetic knee in extension during the swing phase.

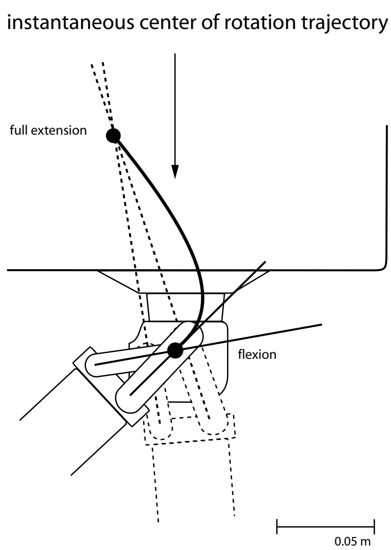


Figure 3.2: The position and trajectory of the center of rotation of the four bar linkage knee joint in the kneewalker prosthetic limb as a result of position, orientation and length of the grounded links.

The external moment on the center of rotation of the knee (M_{CoR}) was calculated based on static and dynamic components in 2D inverse dynamics. The position of the CoR of the modelled four bar linkage knee joint was calculated by the intersection of the lines of the two grounded links (figure 3.3).The static components are formed by the moment arm of the knee-shank-foot system (r_{mCoR}) to the center of rotation, the static moment of the ground reaction force (F_{GRF}) and the moment arm (r_{CoPm}) on the mass of the system (m), and the gravitational force (Fg_m) on the system. The dynamic components are the moments needed to produce the angular (α) and linear (a) accelerations of the system with its inertial properties (I). For the inverse dynamics calculations we modeled the knee and shank, and the foot as two slender rods. The inertia of the knee-shank-foot system of the kneewalker was calculated based on the properties of the components and was on average 0.05 kg m^2 in relation to the CoM of the system depending on the length of the shank.

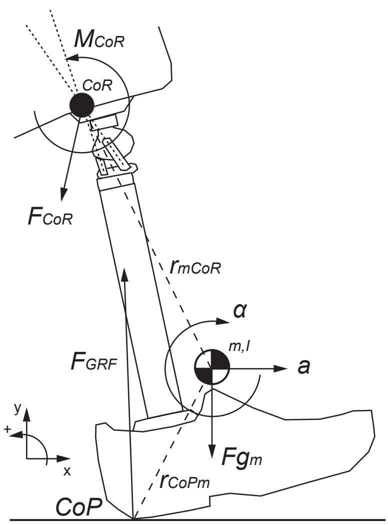


Figure 3.3: External moment on the center of rotation of the knee (M_{CoR}), consisting of static and dynamic components. The static components are formed by the moment arm of the knee-shank-foot system (r_{mCoR}) to the center of rotation, the static moment of the ground reaction force (F_{GRF}) and the moment arm (r_{CoPm}) on knee-shank-foot the system (m), and the gravitational force (Fg_m) on the system. The dynamic components are the moments needed to produce the angular (α) and linear (a) accelerations of the system with its inertial properties (I).

3.2.5 Statistical analysis

All outcome parameters were distributed normally. Paired T tests for differences between conditions were used for the analysis of outcome parameters. To determine differences between two subject groups that used or did not use knee flexion during swing subjects we used independent samples T tests between groups. The level of significance was set to $p<0.05$.

3.3 Results

One out of the eleven subjects needed one extra trial in the sound limb leading condition to produced the five successful trials per leading limb condition, because of a sudden flexion of the knee.

Similar external moment production patterns were found between subjects. The patterns within the subjects were very consistent in the two conditions (figure 3.4). The data showed no significant intra-individual differences in the temporal values of the gait cycle stance phase between the two conditions. On average the total duration of the foot contact was 1.1 s (±0.2). Midstance was reached at 56% (±4) of the stance phase after heel strike. Critical knee flexion moment phases, which were close to the 12.4 Nm produced by the extension aiding spring, were found at 9% (±4) of the stance phase in the sound limb leading condition and at 13% (±8) of the stance phase in the prosthetic limb leading condition, and at 93% (±6) of the stance phase for both leading limb conditions.

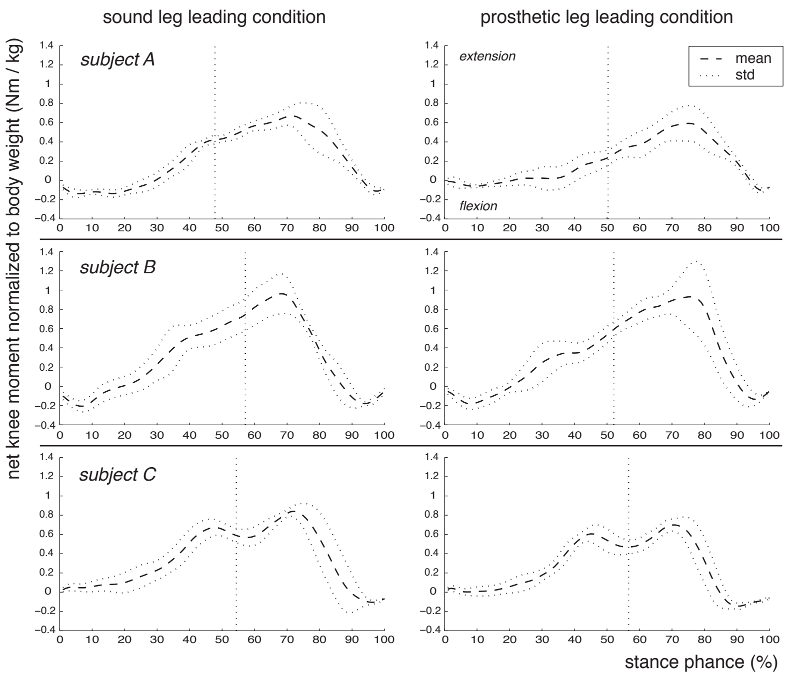


Figure 3.4: Net moments of force on the knee normalized to body weight during stance produced by three inexperienced prosthetic limb users in the sound limb and prosthetic limb leading condition. Several different external moment production patterns were found between subjects in the two conditions. The patterns within the subjects were very consistent in the two conditions. Subject A used a knee extension strategy at the end of the stance phase, subjects B and C used a knee flexion strategy. Notice the decrease in flexion moment at the end of the stance phase as a result of unlocking the knee. The vertical dotted line represents midstance.

Table 3.1: Outcome parameters of prosthetic limb motion and external moments of force on the prosthetic limb during gait initiation in sound limb leading (second step) and prosthetic limb leading (first step) conditions ($n = 11$). Negative moments (Nm) indicate that a flexion moment on the knee is applied. Negative angles indicate that the shank is rotated with the distal end in front of the proximal end. An angle of 0 radians indicates that the shank is in vertical position. Negative velocities indicate that the motions are downwards (vertical z-axis) and backwards (horizontal y-axis).

Outcome parameters prosthetic limb side	Sound limb leading condition (2nd step)	prosthetic limb leading condition (1st step)
External knee moment at heel strike	-2.5 Nm (±2.8)$*^1$	0.2 Nm (±2.0)
Horizontal velocity of the heel just before heel strike	0.13 m/s (±0.11)$*^2$	0.08 m/s (±0.07)
Vertical velocity of the heel just before heel strike	-0.02 m/s (±0.05)	-0.04 m/s (±0.02)
Shank angle at heel strike	-14.90 degrees (±2.86)$*^2$	-16.62 degrees (±4.58)
Minimal external moment on the knee in the first phase	-10.4 Nm (±3.6)	-8.9 Nm (±2.8)
External knee moment at midstance	46.1 Nm (±10.0)$*^1$	32.8 Nm (±9.7)
Minimal external moment on the knee in the second phase [a]	KF: -14.9 Nm (±0.49)	KF: -13.7 Nm (±0.97)
	KE: -11.9 Nm (±1.9)	KE: -10.6 Nm (±2.7)

[a] Knee Flexion group (KF; 3 subjects) and Knee Extension group (KE; 4 subjects). Four subjects were excluded for this second phase analysis because they made use of both knee strategies in the two conditions. No statistical analysis was performed in this second phase analysis because of the small group sizes.
$*^1$ Significant difference ($p<0.01$) between two leading limb conditions.
$*^2$ Significant difference ($p<0.05$) between two leading limb conditions.

Table 3.1 shows the average outcome parameters. Significant differences between the two conditions were found in the external knee moments at heel strike and at midstance. At heel strike a larger flexion moment on the prosthetic knee was found in the sound limb leading condition. In midstance a larger extension moment on the knee was found in the sound limb leading condition. Significant differences were also found in the horizontal velocity of the prosthetic foot just before heel strike and the shank angle at heel strike. The velocity of the foot was higher in the sound limb leading condition, while the angle of the shank with the vertical was smaller, compared to the prosthetic limb leading condition. No significant differences were found in the vertical velocity and the knee moment before and after midstance.

In the swing phase, three groups could be distinguished. Three subjects used knee flexion strategy when swinging their foot forward (figure 3.5). Four subjects used a knee extension and circumduction strategy when moving their foot forward for the next step. The external knee moments used in these two groups differed. The remaining four subjects in the last group used both strategies, with a preference for the knee flexion strategy. One of the four subjects used the knee flexion strategy three times out of the five trials. The other three subjects used the knee flexion strategy four times out of five times. The data from the last group were excluded from the analysis.

Table 3.2 shows the averaged linear and angular acceleration outcome parameters of the knee-shank-foot system of the kneewalker prosthetic limb during stance phase. A significant difference was found in the linear horizontal acceleration. In the sound limb leading condition the minimal horizontal acceleration was higher compared to the prosthetic limb leading condition.

Table 3.2: Mean and standard deviation of the minimum and maximum values of the angular (α) and the horizontal ($\overline{a}_y$) and vertical ($\overline{a}_x$) linear accelerations of the CoM of the knee-shank-foot system of the kneewalker prosthetic limb ($m = 2.1\ kg;\ I = 0.05\ kg\ m^2$) during the stance phase.

Accelerations of the knee-shank-foot system	Sound limb leading condition (2nd step)		prosthetic limb leading condition (1st step)	
	Min	Max	Min	Max
$\overline{a}_x\ (m/s^2)$	-1.9 (± 1.4)$*$	5.5 (± 0.74)	-1.1 (± 0.75)	5.1 (± 0.80)
$\overline{a}_y\ (m/s^2)$	-0.67 (± 0.43)	1.37 (± 0.50)	-0.78 (± 0.43)	1.31 (± 0.34)
$\overline{\alpha}\ (rad/s^2)$	-12.2 (± 3.6)	6.0 (± 1.7)	-10.6 (± 3.3)	5.5 (± 2.0)

$*$ Significant difference (p<0.01) between two leading limb conditions.

Figure 3.5 shows the graphs of a subject during the stance phase after gait initiation. A flexion moment (-13 Nm) on the knee is seen after midstance at 1.00 s, resulting in flexion of the knee. This moment results from the horizontal and vertical forces (Fx and Fy) and their point of application ($CoPx$) under the prosthetic foot. Figure 3.6 shows that the external moment on the center of rotation of the knee (M_{CoR}) is mainly influenced by the ground reaction force in the static moments. When the knee starts flexing at the end of the stance phase (1.00 s), the influence of the dynamic moments can be seen when the ground reaction force is reduced substantially.

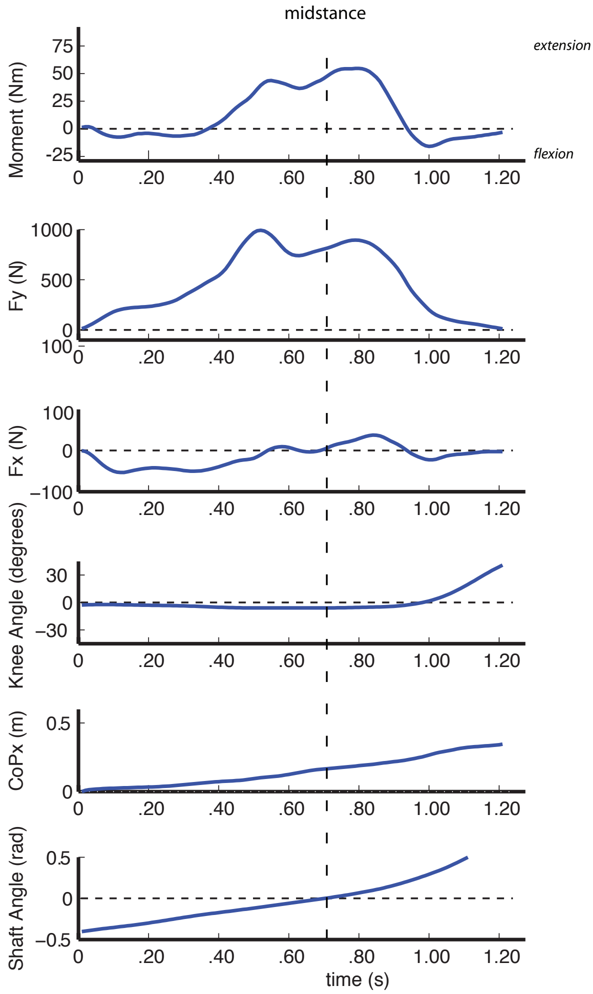


Figure 3.5: An example of a subject with the kneewalker prosthetic limb leading, using the knee flexion strategy at 1.00 s after the beginning of the stance phase at 0 s. The vertical dashed line represents midstance.

3.4 Discussion

In this study, we investigated whether an inexperienced prosthetic limb user with only limited training (100 steps) is able to initiate gait and walk two steps without walking aids on a prosthetic limb with a flexible knee, without the occurrence of a sudden flexion of that knee during stance, either with their sound limb leading or their prosthetic limb leading.

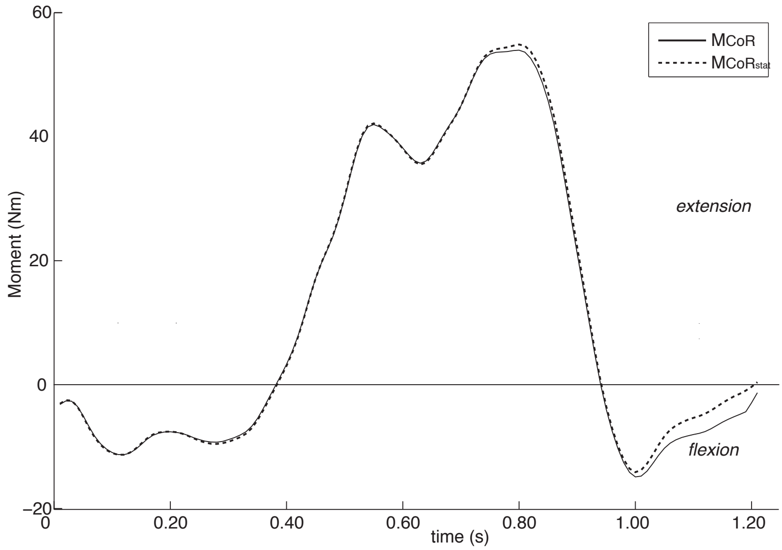


Figure 3.6: An example of the external moment on the center of rotation of the knee (M_{CoR}) consisting of static ($M_{CoR_{stat}}$) and dynamic ($M_{CoR_{dyn}}$) moments during the stance phase after gait initiation with the prosthetic limb leading ($M_{CoR} = M_{CoR_{stat}} + M_{CoR_{dyn}}$). The subject uses a knee flexion strategy during the swing phase. This flexion strategy starts at the end of the stance phase, at 1.00 s after the heel strike at 0 s.

The flexible knee is equipped with an extension aiding spring, which can deliver an internal extension moment. We wondered if the evoked GRF delivers an external flexion moment that is close to the internal extension moment.

We used inexperienced young AB subjects instead of recently amputated TF subjects, to ensure that our subjects had no experience with the use of a prosthetic limb and because of the absence of experience and comorbidity in this group, which otherwise could influence the outcome of the experiments. Indeed physical differences and age differences between the AB subjects and TF amputee subjects can be expected, but a study by Lemaire et al. (2000) has shown that the AB subjects have to learn to deal with mechanical properties of a body extension in a similar way as inexperienced prosthetic users. Therefore we believed it was justified to use AB subjects instead of TF amputee subjects to get first insights in gait initiation strategies in inexperienced prosthetic limb users.

In order to be able to walk with the kneewalker prosthetic limb, with conventional mechanical components, the subjects have to learn to control external moments on the passive prosthetic limb. These moments are influenced by the angle in which the limb was placed on the ground, the applied internal moments by the hip muscles and gravity, and the mechanical properties of the artificial knee.

The results of our experiment showed no clinical differences between the two leading limb conditions. All the subjects were able to prevent a sudden flexion of the flexible passive prosthetic knee, with the exception of one subject who needed one extra trial in the sound limb leading condition to have five successful trials. The duration of the stance phase and

the patterns of the external moments on the CoR of the knee that were evoked were similar in both conditions.
The only clear differences between the two conditions could be found just before the beginning of the stance phase. The horizontal velocity and the angle of the shank differed. These differences had to be dealt with by the subjects. Remarkably, these differences had no consequences for the patterns of the external moments during the stance phase after the heel strike.
The results from our study show that although different velocities of the prosthetic limb were found, the patterns of the external moments applied on the prosthetic knee in the steps after gait initiation did not differ between the two leading limb conditions.

We found two moments during stance phase in which a sudden flexion of the knee could occur. The first flexion moment occurred at the beginning of the stance phase, just after heel strike in both conditions. This flexion moment was remarkably close to the critical moment produced by the extension aiding spring. It is remarkable since only a shift of 1 mm of the CoP position backwards would create an unstable situation. We identified this as a critical phase of the stance phase after gait initiation. This phase was accomplished by a diminished loading rate, compared to normal loading patters, which has also been described as an awareness effect to prior slip experiences[64], showing a cautious loading pattern. In this phase the CoP is positioned relatively more backward under the heel of the prosthetic foot. When the CoP is moved forward, less flexion moment is produced. Notice the relation in figure 3.5 between the shank angle and CoPx position during gait initiation. This relation is the result of the stiff prosthetic ankle properties. The CoP moves forward as a result of an increased external moment on the ankle caused by the rotation of the shaft relative to the foot. The range of this CoP motion increases with increasing stiffness of the prosthetic ankle. It seems that using a prosthetic foot with a relatively flexible heel or rocker will limit the hazard of a sudden flexion of the knee, as the CoP is moved forward more quickly. The outcome from this study applies on the design of this specific prosthetic limb with a stiff ankle-foot. Therefore some considerations have to be taken into account when applying these finding.
The second flexion moment in which the external moment on the prosthetic knee was close to the moment of the extension aiding spring was found just before toe off. In this part of the stance phase some subjects evoked a moment that was used to flex the knee for the upcoming swing phase. The other subjects evoked less external moment on the knee, resulting in maintaining the prosthetic knee in extension during the upcoming swing phase.

The external moments on the knee are mainly the result of the static moments produced by the GRF. Figure 3.6 shows that the total moment on the center of rotation of the knee (M_{CoR}) is only influenced by the dynamic moments ($M_{CoR_{dyn}}$) when the prosthetic limb is flexing at the end of the stance phase. The GRF is then reduced substantially (figure 3.5). The relative small contribution of the dynamic moments can be calculated with the dynamic moment equations. The relative small mass and inertia of the knee-shank-socket system and the relative small linear and angular acceleration have minor contributions to the total moment that is produced.

The small differences in knee moments that were found between the two conditions (1-3 Nm, table 3.1) could be the result of the measurement techniques. The static external moment equations showed that when applying a GRF of 800 N at a moment arm of 0.015 m a moment of 12 Nm is produced. An error of +0.001 m in moment arm length, as a result of false marker position information, leads to an increase of 0.8 Nm.

In our study we were interested in the clinical consequences of the leading limb choice. The results of our experiment showed no clinically differences between the two leading limb conditions. On a more detailed level, there might be some differences, which can be of interest. We did not investigate the moment patterns in detail, but selected discrete points. For more in depth knowledge further investigation of the moment patterns is essential.

Studies have shown that the prosthesis properties influence the way TF prosthetic limb user can apply GRF and its point of application on the prosthetic limb. Many property variables can be adjusted to reach optimal performance. These variables are the stump and socket connection, the alignment, and the components that are used, for example the knee joint, the extension aiding spring, the ankle joint, and the foot and shoe [65;66;67;68;69;70;71;72;73]. Although these variables influence the performance, studies also showed that despite relatively large changes in prosthetic alignment [74] and with mechanical differences [75;76;77], amputee subjects could walk well and without consistent differences in amputee gait. It is suggested that the subjects might adjust gait patterns to the characteristics of the device, blurring actual differences in design [78].
Although our kneewalker might not have been optimally aligned, the findings of these previous studies indicate that our findings have relevance for getting insight in gait initiation strategies, with fully functional prosthetic limbs, in inexperienced prosthetic limb users.

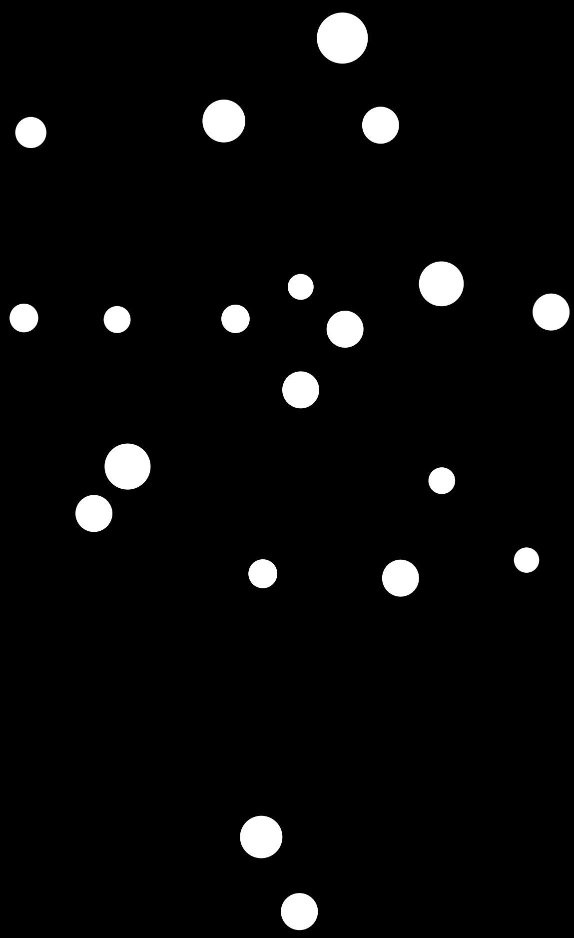
Although the subjects had many variables to vary, all subjects evoked similar patterns of external moments of force under the prosthetic limb in the two steps after gait initiation. The subjects evoked moments that were remarkably close to maximum moment produced by the extension aiding spring to prevent a sudden flexion of the prosthetic knee joint. The subjects had learned to control the prosthetic limb in both conditions within 100 steps of gait, without producing disproportionally big moments of force. This finding suggests that there is no clinical relevance, in relation to the external moments on the knee, to initiate gait with either the sound limb leading or the prosthetic limb leading. Therefore we think that preference of leading limb should not be focus of attention during therapy.

3.5 Conclusions

The experiment showed that the AB subjects learned to control external moments on the kneewalker prosthetic limb in a very efficient manner within 100 steps of gait. In both the prosthetic limb and sound limb leading condition the subjects evoked patterns of external moments of force that were similar. The external moments that were evoked were mainly the result of GRF components evoked during the stance phase. The inertial properties of the prosthetic limb during stance had only limited influence on the external moment on the knee. The critical phases in which a sudden flexion of the knee can occur were found just after heel strike and just before toe off. Based on the control strategy that the subjects used during the stance phase on the prosthetic limb after gait initiation, it makes no difference which limb is used as the leading limb during gait initiation to avoid sudden knee flexion.

Chapter 4

Principles of obstacle avoidance with a transfemoral prosthetic limb.



Abstract

In this study, conditions that enable a prosthetic knee flexion strategy in transfemoral amputee subjects during obstacle avoidance were investigated. This study explored the hip torque principle and the static ground principle as object avoidance strategies. A prosthetic limb simulator device was used to study the influence of applied hip torques and static ground friction on the prosthetic foot trajectory. Inverse dynamics was used to calculate the energy produced by the hip joint. A two-dimensional forward dynamics model was used to investigate the relation between the obstacle-foot distance and the necessary hip torques utilized during obstacle avoidance. The study showed that a prosthetic knee flexion strategy was facilitated by the use of ground friction and by larger active hip torques. This strategy required more energy produced by the hip compared to a knee extension strategy. We conclude that when amputees maintain sufficient distance between the distal tip of the foot and the obstacle during stance, they produce sufficiently high, yet feasible, hip torques and use static ground friction, the amputees satisfy the conditions to enable stepping over an obstacle using a knee flexion strategy.

transfemoral prosthetic limb, obstacle avoidance, hip torques, ground friction, knee flexion, computer simulations

4.1 Introduction

Stepping safely over obstacles is a common daily living activity [32]. During obstacle avoidance, the stance limb must establish a base of support that appropriately maintains stability to avoid slipping or falling. The swing limb must clear the obstacle successfully to avoid tripping [33;34]. The applied joint moments of the swing limb and the obstacle-foot distance during stance determine the clearance achieved during obstacle avoidance [35]. Flexion of the knee in the swing limb is the most important motor strategy used by able-bodied (AB) subjects for foot clearance. This knee flexion is achieved by an increase in the force of the knee flexors [36;37;38;39;40] and through kinetic coupling by the hip flexors. The amputation of a lower limb results in a deficiency in sensory input and an absence of muscles and joint(s). A person with a lower limb prosthesis must adapt to a mechanical device to become functionally independent again [79]. Transtibial (TT) amputee subjects increase swing hip elevation and hip and knee flexion as a function of obstacle height during obstacle avoidance. An increase of the knee flexion on the prosthesis side is achieved by modulating the work performed at the hip, not at the knee, as seen on the amputee's sound side [41;42]. In addition, the stance limb hip flexion, knee flexion and (on the sound side) ankle plantarflexion increase slightly with increased obstacle height, but the stance limb hip elevation does not. Hill et al. (1997) concluded that modulations of the stance limb served to position the pelvis further back from the obstacle as the height of the obstacle increased.

Transfemoral (TF) amputee subjects make use of adjustment strategies to compensate for the loss of muscles and sensory input in their prosthetic limb during obstacle avoidance, and they learn to cope with bilaterally delayed and decreased obstacle avoidance responses in both limbs [80]. Vrieling et al. (2007; 2009) found that the prosthetic knee flexion during obstacle avoidance of transfemoral amputee subjects was reduced in comparison with unimpeded walking and compared to TT amputees and able-bodied subjects. The lack of knee strategy in TF amputee subjects is compensated for by circumduction at the hip on the prosthesis side and by plantar flexion on the sound side [43]. These results suggest that TF amputee subjects use an extension strategy: their knee is fixed in extension, which is combined with hip abduction and exorotation. This strategy has an advantage over the knee flexion strategy. The extended prosthetic knee eases the transition from swing to stance. However, the extension strategy also has disadvantages. Not only does it reveal the use of a prosthetic limb, but also changes in the gait cycles are necessary when accelerating and decelerating the prosthetic limb in a lateral direction. Therefore, more degrees of freedom must be controlled. Additional free space is necessary for the clearance as the foot moves farther outward. Possible reasons for choosing the extension strategy over the flexion strategy are 1) a reduced gait velocity of the TF amputee subjects, which impedes the initiation of the pendulum motion of the prosthetic limb or 2) not being able to produce a sufficient flexion moment at the hip joint. To reduce the number of falls of amputees, Vrieling et al. suggested that it is important to train amputees in 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. Although TF amputee gait and obstacle avoidance include many out of plane actions, including trunk sway [81;82;83] and the previously reported circumduction strategy [43;28], we believe it is of interest to study the possibility of crossing an obstacle with a knee flexion strategy as this strategy reduces the changes in gait cycles and masks the use of a prosthetic limb. In theory, crossing an obstacle with an upper leg prosthesis, which is confined to the sagittal plane, can be executed in four ways, where the prosthetic limb is either leading or trailing with the artificial knee joint either in flexion

(flexion strategy) or fixed in extension and combined with hip abduction and exorotation (extension strategy).
In the present study, we asked how the hip torques and static ground friction [84] contributed to a flexion strategy in obstacle avoidance and what the costs are of this strategy. We hypothesized that to move a prosthetic foot using a knee SSexion strategy over an obstacle that is close by, it is preferable to use ground friction and large hip torques. This combination helps to achieve height with less forward motion compared to a combination of small hip torques and without the use of static ground friction. To achieve sufficient obstacle-foot clearance during the knee flexion strategy, the TF amputee subject must overcome the extension spring force that keeps the prosthetic knee in extension. The applied hip torque and the static ground friction on the prosthetic foot can help overcome the extension spring force. Consequently, the following knee flexion lowers the moment of inertia of the prosthetic limb by bringing the foot and the lower leg shaft closer to the hip joint. These changes may be useful when stepping over an obstacle. In the current study, we limited the modeling to the sagittal plane, as crossing an obstacle with a flexed upper leg prosthesis is confined to the sagittal plane.
In the first part of this study, we experimentally investigated the relationship among the static ground friction on a prosthetic foot, a wide range of hip torques and the trajectory of the prosthetic foot. The temporal (duration) and spatial (forward motion) data, the inverse dynamics (energy produced by the hip, the hip torques and the mean angular velocities of the upper leg) and the statistical relationships among fast or slow hip flexion to move the foot 0.1 m upward, with and without static ground friction conditions, were investigated. In the second part of this study, we used a two-dimensional forward dynamics model to investigate obstacle avoidance for which we focused on a) the influence of a constant hip torque on the first part of the prosthetic foot trajectory, with and without the use of static ground friction and b) the relation between the obstacle-foot distance and the associated necessary time varying hip torques in the sagittal plane while stepping over an obstacle. Testing for these discrete parameters in human subjects, without the interference of compensation strategies, was not feasible; therefore, it was decided to approach this problem in a theoretical way. The outcome and insights we gained from this study can be used to understand why TF amputee subjects prefer to use the knee extension strategy during obstacle avoidance and to provide insights into what we should take into account when teaching a knee flexion strategy during obstacle avoidance to TF amputee subjects who have a prosthetic limb.

4.2 Methods & Results

Informed consent was obtained from all subjects before testing.

4.2.1 Part I - Measurements

In first part of this study, we investigated the relationships among static ground friction, hip torques and the trajectory of the prosthetic foot.

Hip torques

A wide range of hip torques driving a prosthetic limb was produced by four naive AB male subjects (mean 30 y (SD 7); mean 80 kg (SD 7.3); mean 1.87 m (SD 0.08)), with no previous experience using a prosthetic limb. A kneewalker transfemoral prosthetic simulator was used[52;85]. To obtain a high degree of equivalence for the comparison, we used a kneewalker prosthetic limb that was relatively short compared to the length of the sound limb, with the same properties, alignment settings and segments length for all four subjects. To make contact with the ground, the leg length difference between the sound leg and the prosthetic leg was compensated for by flexing the sound stance limb. The kneewalker prosthetic limb consisted of an Otto Bock Habermann modular four-bar linkage knee joint (3R36), an Otto Bock dynamic foot with toes (1D10, size 26) and a shoe (size 43 / 9, toe-heel length 0.30 m) (figure 4.1). The artificial knee was equipped with an extension spring. The spring served two main functions. First, the spring supported the forward motion of the foot and the shaft at the end of the swing phase. Second, the spring enabled the prosthetic limb user to raise the prosthetic limb forward against gravity without flexion of the knee, assuming that the motion is not performed at a high acceleration. This second feature provided a prosthetic limb user control over the passive knee when positioning the prosthetic foot for the stance phase. When using the extension spring, the prosthetic knee remains locked in full extension. The spring generates a moment between 45 and 0 degrees of flexion. The amount of moment is inversely related to the amount of flexion, which decreases to 0 Nm at a 45 degree flexion. Hyperextension of the prosthetic knee is prevented by a mechanical stop, i.e., a very high stiffness. The spring produces a maximal moment of 12.4 Nm in full extension. The length of the shaft can be adjusted to match the contralateral leg length. The mass of the knee-shaft-socket system is 2.08 kg. The prosthetic ankle-foot system of the prosthetic leg is relatively stiff. The upper leg socket of the kneewalker prosthetic limb is constructed in such a way that the prosthetic limb is connected to the upper and lower leg, which is fixed at a 90 degree flexion at the knee joint. Because of this construction, the AB subjects are able to put weight on the kneewalker via their knee and the socket/leg connection, so that the prosthesis can be used in a comparable manner to prostheses for knee-exarticulation amputees. All subjects used the same shoe under the prosthetic foot. The heel-toe length was 0.3 m.

Measurement

We used a PRIMAS 3D motion capture camera system, Bertec force plates and the kneewalker. The 3D motion capture camera system consisted of 6 infrared cameras that recorded at 100 Hz. A total of 8 retroreflective markers were positioned on the spina illiaca posterior superior (1), the socket (3: upper leg, knee joint and lower leg), the shaft (2: proximal and distal) and the foot (2: heel and toe) (figure 4.1). The motion data were filtered using a third-order 5 Hz low-pass zero time-lag Butterworth filter. The three-dimensional forces and the center of pressure position were sampled at 100 Hz using Bertec force plates.

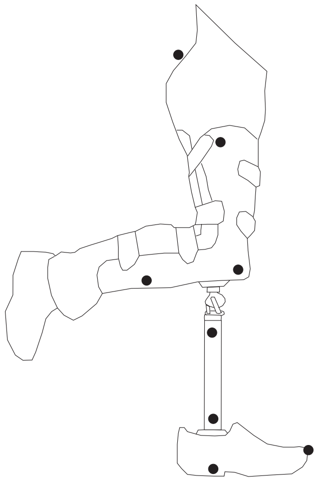


Figure 4.1: Kneewalker prosthesis for able-bodied subjects.
The black dots indicate the position of the retroreflective markers.

Procedure

The subjects were instructed to stand at ease on the force plates. Subsequently, they had to flex their hip at the prosthesis side at four speeds, which varied from slow to very fast, while standing on their sound limb. The subjects performed this task in two conditions: with and without ground friction on the prosthetic foot, i.e., with the foot on the floor and the foot above the floor. Under the ground friction condition, no instructions were given regarding how much body weight should be placed on the prosthetic limb. The subjects were allowed a maximum of five trials to produce the four different speeds. The subject decided which trial was excluded if he or she used five trials. Only the first part of the motion, from the beginning of motion, until the lowest point of the foot (either the heel or toe) reached a height of 0.10 m above the starting position, was used for the analysis. During the movement, the displacement of the pelvis had to be kept at a minimum to limit the compensation strategies. The data from the force plates were used to verify whether the swing foot was in contact with the ground in the two conditions. The data from the optical marker system were used to verify that pelvis displacement had a standard deviation that was less than 0.04 m in any direction. Only the trials that met these criteria were selected. The selected trials were grouped according to conditions with and without ground friction. The two conditions were then subdivided into fast and slow trials, which were based on the time it took to move the foot upward until the lowest point of the foot reached a height of 0.10 m above the starting position. The trials were divided at the median of that duration (0.36 s).

Outcome parameters

The outcome parameters were the percentage of trials that resulted in flexion of the knee, the duration of the trials, the forward motion of the lowest point of the foot, the energy produced by the hip, the hip torque, the mean angular velocity of the upper leg during the trial,

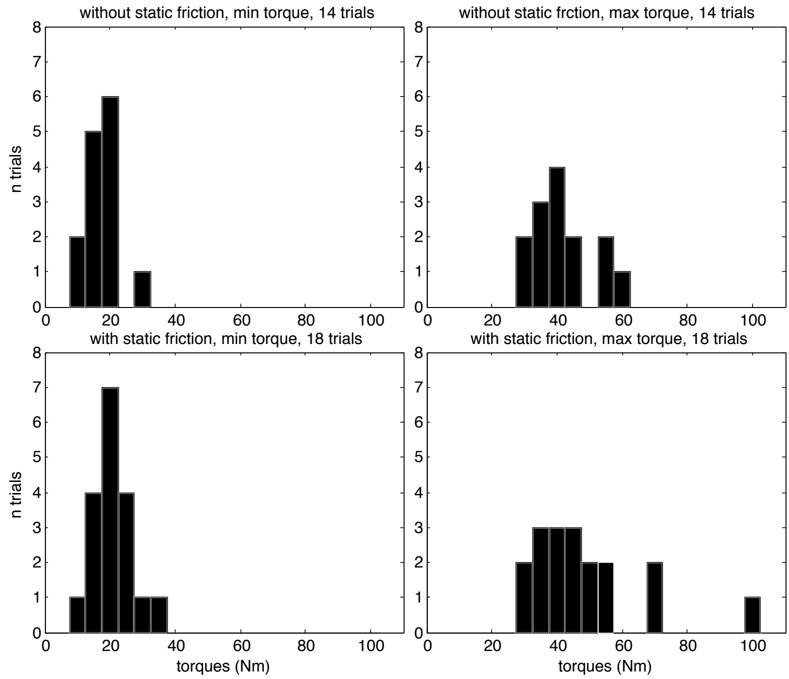


Figure 4.2: Minimal (left panels) and maximal torques (right panels) produced during the trials in the conditions with (lower panels) or without static ground friction (upper panels).

and whether the motion resulted in flexion or extension of the knee. The motion-captured kinematic data and the force plate data were used in a planar inverse dynamics model. The model was based on a Newton-Euler approach with constraint equations, previously presented by Otten (2003), which consisted of three linked elements with a 4 bar linkage knee to calculate the energy, the torques and the mean angular velocity in the sagittal plane. A non-parametric method, i.e., the Kruskal-Wallis test, was used to determine significant differences ($p<0.05$) between the conditions.

Results

Figure 4.2 shows the minimal and maximal torques of the wide range of hip torques that were used in the conditions with and without ground friction. Some trials were excluded because of too much hip motion or ground contact. For the without ground friction condition, 14 trials met the inclusion criteria; for the with static ground friction condition, 18 trials met the criteria. The average hip position standard deviation of the selected trials was 0.02 m in both conditions. Figure 4.3 shows the trajectories of the toe of the kneewalker prosthetic limb during fast and slow hip flexion in the conditions with and without static ground friction. The foot moved upward until the lowest point of the foot reached a height of 0.10 m above the starting position. An overview of the temporal (duration) and spatial (forward motion) data, the inverse dynamics (energy produced by the hip, the hip torques and the mean angular velocities of the upper leg) results, the statistical relationship among fast

Table 4.1: Overview of the temporal (duration) and spatial (forward motion) data and the inverse dynamics (energy produced by the hip and the mean angular velocities of the upper leg) results of the 4 conditions (fast or slow hip flexion, with and without static ground friction) under which the foot was raised foot 0.10 m.

Group	Forward motion (m)	Energy (J)	Duration (s)	Angular velocity mean (rad/s)	Flexion (% of trials)
Without Static Friction, Slow	0.36 (0.08)†	12.86 (2.40)⋆†	0.59 (0.20) ⋆†	0.77 (0.34)⋆†	38
Without Static Friction, Fast	0.22 (0.05)	29.31 (6.15)⋆	0.32 (0.03)⋆	2.04 (0.26)⋆	100
With Static Friction, Slow	0.21 (0.07)	21.36 (4.14)	0.48 (0.09)‡	1.57 (0.39)‡	100
With Static Friction, Fast	0.14 (0.05)†	29.12 (7.86)†	0.29 (0.04)†‡	2.43 (0.42)†‡	100

Significant differences between the pairs of conditions (Group, first column) are marked with ⋆, † and ‡ ($p<0.05$).

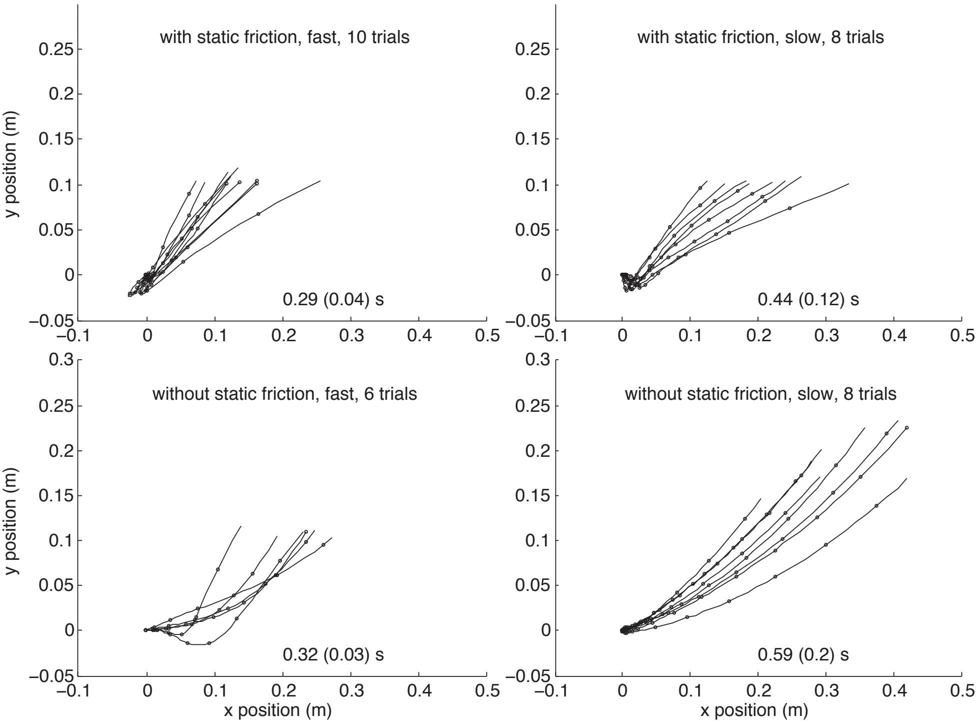


Figure 4.3: Trajectories of the toe of the kneewalker prosthetic limb during fast and slow hip flexion in the conditions with and without static ground friction. The foot was raised upward, until the lowest point of the foot (either the heel or toe) reached a height of 0.10 m above starting position. The numbers at the bottom the graphs represent the average duration and standard deviation in seconds of the motions. Notice that the trajectories of the slow movement without static ground friction are larger because the heel is the lowest point of the foot.

and slow hip flexion in the conditions with and without static ground friction, for the foot travelling 0.10 m upward, and the result of the motion (flexion or extension of the knee) are presented in table 4.1.
This experiment shows that by making use of more energy and static ground friction, the foot moved a smaller distance forward (0.20 m) compared to when less energy is produced and no static ground friction is used. Making use of these two factors shortened the upwards motion duration significantly and increased the angular velocity of the upper leg. The lack of ground friction with fast upper leg motions made no difference in the energy produced nor the duration of the motion, but it almost doubled the forward motion of the foot. In all of the fast trials and the slow trials that made use of static ground friction, flexion of the prosthetic knee was found. In 5 out of the 8 slow trials that did not make use of the static ground friction, the knee was maintained in an extended position.

4.2.2 Part II - Simulations

In the second part of this study, we used the two-dimensional forward dynamics model to investigate obstacle avoidance in a theoretical way, in which we focused on a) the influence of a constantly applied hip torque on the first part of the prosthetic foot trajectory with and without the use of static ground friction for 0.2 s and b) the relationship between the obstacle-foot distance and the associated necessary time to vary hip torques in the sagittal plane when stepping over an obstacle.

Model

The forward dynamics model is a planar system of three linked elements based on a Newton-Euler approach with constraint equations, as presented by Otten (2003). The model consists of an upper leg, a lower leg and a foot with joint torques and forces (figure 4.4). We adjusted the model to simulate the dynamics of a prosthesis that was connected to an upper leg stump. The settings of the model were in the range of the kneewalker prosthetic limb. The stump and socket connection was modeled as a single rigid body. For the sake of simplicity, the model utilized frictionless single axis joints. Values were selected based on numerical stability. The ranges of the ankle joint and knee joint were limited with linear counter torque springs and dampers. The counter torque springs are related to the joint angles. The dampers in the joints damp the motion of the joint based on the joint velocity. The knee counter torque spring, which limits the flexion and extension range of motion of the knee, provided a maximal stiffness of 5000 Nm rad^{-1}. The free range of the knee joint was 100 to 0 degrees (fl / ext). The extension damper factor of the knee was set to 10 Nm s rad^{-1}. The flexion damper factor was set to 1 Nm s rad^{-1}. The ankle counter torque spring provided a maximal counter torque of 1000 Nm rad^{-1}. The free range of the ankle joint was -1 to +1 degrees (plantar fl / dorsal fl; almost stiff ankle). The ankle damper factor was set to 4 Nm s rad^{-1}. The hip joint moved with a constant forward velocity. The torques on the hip joint and the translations of the moveable point in the sagittal plane were variables in the model of which the influences can be studied. The hip joint was the origin of the first element in the chain. Translation and rotation of a parent element effected the connected child element. The proximal (heel) and distal (toe) part of the foot were used as possible contact-points

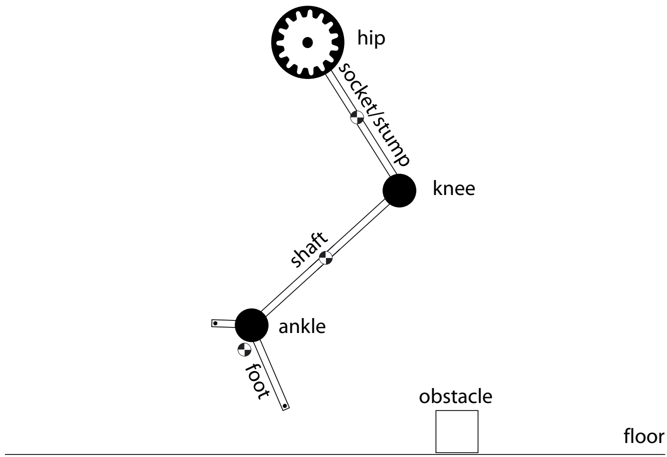


Figure 4.4: A two-dimensional planar system of four linked elements with joint torques and forces in a floor and obstacle environment.

with the external environment. The ground reaction forces, which the model takes into account, were formed in these points during collision. The external environment consisted of a floor and an obstacle of 0.10 m high and 0.10 m wide. The socket and shaft elements were modeled as slender rods. The foot was modeled as a triangle. The masses and lengths of the kneewalker prosthetic limb were set as constants in the model (socket: 8 kg, 0.45 m; shaft 3 kg, 0.53 m; foot 1 kg, 0.31 m; horizontal heel-toe length 0.24 m; foot sole-ankle height 0.07 m). There was no weight support implemented. Other constants in the model included the stiffness of the floor and the obstacle (10^5 N/m). The positions of the elements and their angles, velocities and angular velocities were calculated during the simulation using the Euler integration method with an integration step size of 0.001 s. An extension assist spring, as found in modern artificial knee joints, was added to the model. The modeled spring was active in the range of 15 to 0 degrees flexion. The spring provided a counter torque of 0 Nm at 15 degrees of flexion and increased linearly to 25 Nm at 0 degrees of flexion.

Simulation

Computer simulations ($\Delta T = 0.001$ s) were performed using Matlab (The MathWorks, Inc; Version 7, R14).

Procedure

The model was used to investigate a) the influence of the applied hip torques and the static ground friction on the foot trajectory and b) the relationship between the obstacle-foot distance and the necessary torques for obstacle clearance.

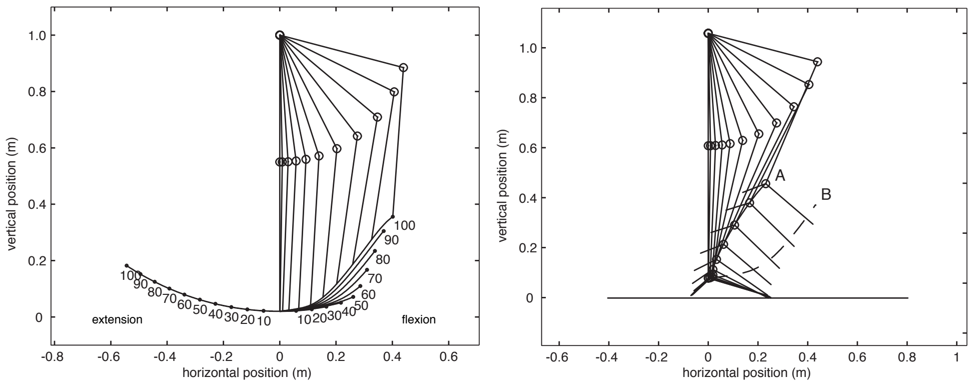


Figure 4.5: Left panel: The influence of various hip flexion and extension torques applied to the model. The numbers in the graph represent the applied hip torques and the end position of the ankle after 0.2 s of simulation. The stick figures represent the end positions at the various values of the hip moment (flexion). Although the mass and inertial properties of the foot were taken into account, the foot is not shown in the figure for clarity reasons.
Right panel: Prosthetic limb trajectory of the model with (A) and without (B, dashed line) static ground friction (hip torque: 100 Nm; duration: 0.2 s).

Part IIa The influence of the applied hip torques was investigated using a simulation in which the prosthesis hung with the hip joint fixed in the two-dimensional space (figure 4.5-left). The prosthetic limb was positioned high above the ground. The foot did not touch the ground. In the initial state, the socket / stump element and the shaft element were hanging vertically. Torques ranging from 0 Nm to 100 Nm in the flexion and extension direction were applied to the hip joint to create flexion and extension moments on the hip. These torques were constant during the simulation. The influence of static ground friction on prosthetic foot trajectory was investigated using a situation in which the prosthetic limb hung vertically with the foot in contact with the ground. The angular velocities of the segments were zero at the beginning of the simulations. We applied the highest flexion torque (100 Nm), as used in the previous simulation, on the hip. The duration of the simulations was 0.2 s (figure 4.5-right).

Part IIb The model was also used to study the relation between the obstacle-foot distance and the needed torques for foot clearance. This object was placed at several distances in front of the toe of the prosthetic foot, ranging from 0.1 m to 0.7 m, in steps of 0.1 m. Hip torque profiles, which vary during a step and are needed to steer the prosthetic limb over an obstacle placed at a specific distance, were searched for in a sequence of trials with a simulated annealing algorithm. For every distance, the model was set with initial parameters. The forward horizontal velocity of the hip was set to 0.9 m s^{-1} constant velocity (unimpeded TF gait velocity: 1.0 m s^{-1} [43;28]). The vertical hip velocity was set to zero. The angular velocities of the segments were zero at the start of the simulations. The distance the model had to cover was set to 1.4 m. The start and end positions of the prosthesis and the desired minimal clearance (0.05 m) from the prosthetic foot over the obstacle (height: 0.10 m;

depth: 0.10 m) were set as objectives for the model. Using a hip torque profile as the input argument, the simulated annealing algorithm [87] searched for the lowest error value, which was the outcome of the obstacle avoidance with the prosthetic limb simulation during a sequence of 3000 trials per distance. The error value to be minimized by the algorithm was based on the knee angle (in rad) and the distance between the foot and the floor in the final position (in m), the clearance of the foot over the object (in m), the number of contacts made with the ground and the obstacle (discrete number), the length of the hip, the knee and foot trajectories (in m) and the total distance traveled (in m). These error values were unweighted sums. Therefore, the error values of the knee angle and the number of ground contacts had a great deal of influence on the final outcome. When searching, the hip torque profile that produced the lowest error value during the previous simulation trials was used as an input for the next trial. This torque profile input was adjusted based on random values, whose magnitudes decreased over the trials, resulting in a new error value. At the beginning of the trials, the torque profile changed more than at the end of the trials, due to the decrease in the magnitude of the random values. This procedure helped to identify an optimal local minimum, and therefore the optimal hip torque profile. The profile that was used the input for the algorithm was the result of a cubic spline-based interpolation over time of five adjustable torque values. These five torque values were on a fixed time interval between the start and the end of the simulation. The values between these discrete points were interpolated, which resulted in a smooth cubic spline curve over time. The range of the possible hip torques was limited to the maximal flexion and extension hip torques that were produced by our subjects and in accordance with literature ranging from 0 and 100 Nm [88].

Results

The simulations showed that the applied hip torques and the static ground friction influenced the foot trajectory. The torque profiles needed for stepping over an obstacle were based on these factors.

Part IIa The forward dynamics simulations with the model showed that when the applied torque changes, the resulting trajectory of the foot also changes. When a hip extension torque is applied to a prosthesis that is hanging freely, the foot follows the trajectory of a pendulum. The knee joint locks the prosthesis in extension, therefore 'converting' the double pendulum into a single pendulum. Larger torque values result in a longer trajectory of the foot, that is caused by the higher velocity of the foot in the same time interval. When a flexion torque is applied the trajectory of the foot is quite different. As long as the torque is small, the trajectory is still similar to the trajectory of a pendulum. However, when a larger torque is applied, the trajectory changes. The curves in figure 4.5-left show that when large torques are applied, the foot gains not only more forward velocity but also for a more upward velocity. In figure 4.5-right, the influence of the static ground friction is visualized. When the prosthetic foot makes contact with the ground, the trajectory of the prosthetic foot with the same torque applied differs from the trajectory when no contact is made. Due to static ground friction, the tip of the foot (toe) becomes the point of rotation of the lower leg for the first period of the simulation. The horizontal forward trajectory distance of the foot after losing contact with the ground is less compared to the situation without static

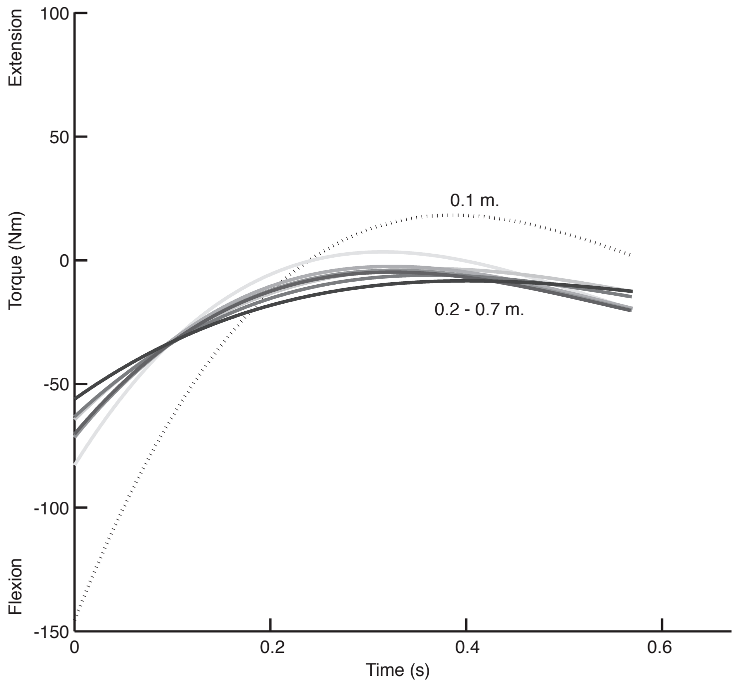


Figure 4.6: Profiles of the hip torques that are needed for stepping over an obstacle while using static ground friction as a result of a simulated annealing algorithm. The numbers in the graph represent the distance between the foot and the obstacle at the beginning of the simulation. When the object is too close to the foot (0.1 m) unrealistic flexion moments are required (dashed line).

ground friction. However, the vertical upward trajectory distance is larger.

Part IIb The trajectories that were found with the simulated annealing algorithm corresponded to the imposed criteria. The simulations of the model show that the torque profiles (figure 4.6), that are needed for stepping over an obstacle positioned at different distances from the initial starting position from the prosthetic foot are comparable for almost all distances, except when the object is very close to the distal tip of the foot (<0.10 m). An obstacle-foot distance of 0.2 m was sufficient to step over the obstacle with normal human hip torques (range: 56 to 83 Nm) (figure 4.6). At very close distances contact with the obstacle can only be avoided when torques that are outside the range of the torques produced by our subjects are applied. A starting torque of more than 100 Nm is needed to steer the prosthetic limb in flexion over the obstacle (figure 4.6, line 0.10 m, starting at 150 Nm). The clearance over the object deviated on average by 0.02 m (SD 0.02) from the imposed 0.05 m. This relatively large standard deviation was caused by the trials in which the object was close to the foot (<0.60 m).

4.3 Discussion

We hypothesized that to move a prosthetic foot using a knee SSexion strategy over an obstacle that is close by, it is preferable to use ground friction and large hip torques. This combination helps to achieve height with less forward motion compared to a combination of small hip torques and without the use of static ground friction. The current study confirmed the above hypothesis. Both the experiment and the model showed that the generated trajectory is the result of the duration and the amount of the applied hip torques and the use of static ground friction. When the foot is not too close to the obstacle (>0.15 m) it should be possible for TF amputee subjects to generate the adequate knee flexion motion with the available hip torques and the use of static ground friction. Of course, it should be taken into account that this specific finding is only valid for this specific prosthetic limb. Other studies have shown that changes in the stiffness and damping of prosthetic joints [89], weight (distribution) [90;91], the composition of the knee joint [67;68], gait velocity and the shape of the socket [43] influence the trajectory of the foot or the needed hip torques. Although the results of this study have to be verified by future studies of other prosthetic limbs and in TF amputees with comorbidity, some explanations for the choices made in obstacle avoidance strategies can be reasoned based on the hip torque and static ground friction principles that were identified. Earlier results have suggested that during obstacle avoidance, TF amputee subjects use a knee extension strategy with an externally rotated and abducted limb [43;28]. A possible explanation for this choice of strategy is that the distance between the prosthetic foot and the object are too small for safe clearance during obstacle avoidance. In that case, the subjects should be taught to use the prosthetic limb as the leading limb during the obstacle clearance, which would result in a sufficient distance between the foot and the object [43]. Another possible explanation for the choice of this strategy might be the amount of energy that is necessary to employ the strategy. According to our study, a flexion strategy that is seen in the fast hip motion trials or when static ground friction is used demands more hip energy compared the slow motion without using static ground friction condition, in which the knee remains extended in most of the trials. The needed energy may be the reason for this choice, although one could image that standing on one limb for a longer period, moving the prosthetic limb in a direction perpendicular to the direction their walking and losing forward velocity is more energy demanding than the fast motion and making use of static ground friction. We were not able to verify this with our data. The third possible explanation is that making use of static ground friction or the application of large hip torques results in increased upper leg angle velocity, which may feel less controlled with no active control over the passive prosthetic knee joint. Learning to cope with a flexing passive knee joint can only be possible when the subject is able to fit the prosthetic limb perfectly into his body scheme and when the properties of the limb are suited to the needs of the user. If the subject is not able to fit the prosthetic limb into his body scheme, he or she will learn that when he or she uses small hip torques, the prosthetic limb remains in an extended knee joint position. An extension spring in the knee enables the patient to slowly lift the limb over an obstacle, without flexion of the knee, which results in controlled clearance over the obstacle. In the knee extension strategy the static ground friction on the prosthetic foot is not used during obstacle avoidance. The last possible explanation for not using a knee flexion strategy when the prosthetic limb is the trailing limb is that the TF amputee subject has no visual control over the prosthetic foot, to compensate the lack of mechanoreceptors in the knee, as long as it is behind or under the body. Therefore, the TF amputee subject does not know if foot clearance is adequate. A study with AB subjects

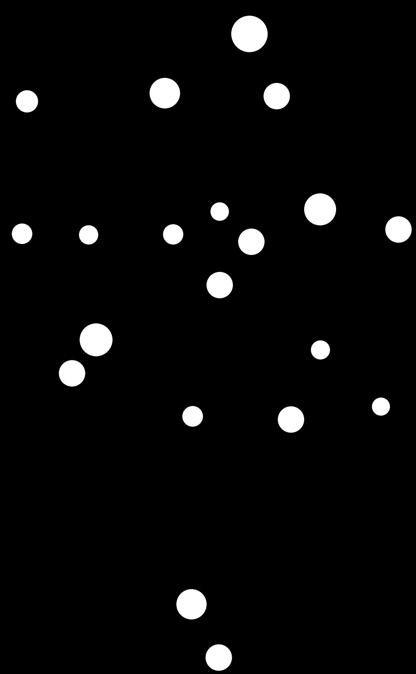
showed that limb elevation over an obstacle was increased for a greater safety margin after removing vision[92]. Leading limb control is modifiable on-line when vision is available during obstacle avoidance. In contrast, trail limb control is based primarily on feedforward visual information and on-line kinesthetic sensory output. The extension strategy enables the TF subject with a diminished on-line kinesthetic sensory output to have more visual control when the limb is trailing.
There are a few limitations to this study. The group of subjects was small in number. This group was used to investigate the foot trajectory of the prosthetic limb when applying hip torques and using static ground friction on the prosthetic foot. We were not interested in the performances of the AB subjects, only the trajectory of the prosthetic foot driven by the subjects. The number of trials made by the subjects was sufficient to test our hypothesis. It can also be argued that the AB subjects using the kneewalker prosthetic limb cannot be compared with TF amputees. However, Lemaire et al. reported that AB subjects use the same compensation strategies as inexperienced prosthetic limb users and that kinematic and kinetic analyses results were similar to gait analysis of people with TF amputations[52], which makes the use of a kneewalker prosthetic limb valid in our study. A limitation in the theoretical part of this study is that we used a constant hip forward velocity when steering the prosthesis over the obstacle in the simulations. Prosthetic limb users do not use constant velocities. Complex combinations of accelerations and decelerations are applied on the hip while walking with a prosthesis[65] and during obstacle avoidance[41;42;43;28]. These hip accelerations do not only influence the motion of the prosthetic components, but they also have a direct influence on foot clearance. If a TF amputee subject elevates or tilts his or her hip or pelvis 0.01 m, the foot will also gain 0.01 m of foot clearance. This hip elevation or tilt strategy is very often used during obstacle avoidance; AB subjects achieve approximately 22% of toe clearance by utilizing hip elevation[93], but this factor was not taken into account in this study. We expect that hip elevation or tilt strategy will contribute to the reduction of the needed hip torques for obstacle avoidance. However, the model shows that this strategy is not necessary in obstacle avoidance when the correct hip torques and static ground friction are used. Note that the minimum and maximum torques found in these simulations deviate slightly from human hip torques. As we did not include a complete muscle model, the simulated profiles show a sudden onset of torque at the beginning of the movement, whereas human torque profiles normally show a slow onset for the first 100 ms[94]. Another limitation in this simulation is that we used a simulated annealing algorithm, which is a stochastic procedure. Although this algorithm does not give a set of unique solutions, there is sufficient convergence of the torque profiles to be useful for testing our hypothesis (figure 4.6). The last limitation is that the models in this study were two-dimensional models, which limits motion to the sagittal plane. To study the influence of non-planar obstacle avoidance solutions, which may necessitate knee extension strategies and hip joint motion with abduction and exorotation, three-dimensional models should be constructed. As obstacle avoidance is confined to the sagittal plane, we feel that a two-dimensional model is acceptable for an analysis of normal swing phase kinematics and for gaining insights into the influence of hip torques and ground friction on the trajectory of the prosthetic foot. The model we used for the forward dynamics simulation consisted of a single axis knee joint for parsimonious reasons. Although this knee joint is still used in prostheses, this knee joint is not representative for all modern prosthetic knee joints. Most modern prosthetic knee joints consist of at least four bar linkage knee joints or are computer controlled. These types of knee joints influence the trajectory of the foot[67;68;95]. Nevertheless, to gain insight into the prosthetics dynamics, the single axis knee joint satisfies our requirements.

4.4 Conclusions

The study showed that the trajectory of the prosthetic foot is influenced by both the hip torques and the static ground friction on the prosthetic foot. Based on our findings, we report that the following factors contribute to a successful clearance using a knee flexion strategy during obstacle avoidance by TF amputee subjects: 1) maintaining a sufficient distance between the foot and the obstacle at the start of the swing phase, 2) producing sufficient hip torques, and 3) making use of the static ground friction on the prosthetic foot.

Chapter 5

Controlling horizontal deceleration during gait termination in transfemoral amputees; measurements and simulations.



Abstract

In this study we investigated how leading limb angles combined with active ankle moments of a sound ankle or passive stiffness of a prosthetic ankle, influence the center of mass velocity during the single limb support phase in gait termination. Also, we studied how the trailing limb velocity influences the center of mass velocity during this phase. We analyzed force plate data from a group of experienced transfermoral amputee subjects using a prosthetic limb, and the outcome from a two dimensional mathematical forward dynamics model. We found that when leading with the sound limb, the subjects came almost to a full stop in the single limb support phase, without the use of the prosthetic limb. When leading with the prosthetic limb, the center of mass deceleration was less in a relatively short single limb support phase, with a fast forward swing of the trailing sound limb. Slowing down the heavier trailing sound limb, compared to the prosthetic limb, results in a relatively larger braking force at the end of the swing phase. The simulations showed that only narrow ranges of leading limb angle and ankle moments could be used to achieve the same center of mass velocities with the mathematical model as the average start and end velocities of the prosthetic limb user. We conclude that users of prosthetic limbs have a narrow range of options for the dynamics variables to achieve a target center of mass velocity. The lack of active control in the passive prosthetic ankle prevents the transfermoral amputee subjects from producing sufficient braking force when terminating gait with the prosthetic limb leading, forcing the subjects to use both limbs as a functional unit, in which the sound limb is mostly responsible for the gait termination.

transfemoral prosthetic limb, gait termination, ground reaction force, inverse dynamics, forward dynamics

5.1 Introduction

Successful gait termination with a transfemoral (TF) prosthetic limb requires indirect control over a device with limited degrees of freedom. With fewer muscles compared to able-bodied individuals, TF amputees have to be able to control the prosthetic limb by making use of the limb properties and the environment in which the limb is used.
Gait termination studies in able-bodied subjects show that several strategies are used to reduce the forward motion of the center of mass (CoM). By placing the leading limb on the ground in front of the body, a center of pressure (CoP) under the foot is formed. The ground reaction force (GRF) originating from this CoP is used to decelerate the CoM. Also, by decreasing the push-off with the trailing limb the forward motion is reduced.[45;46;47;48].
During gait termination, the leading limb is for the most part responsible for the production of the necessary braking force[47].
Studies in prosthetic limb users show that the motion of the CoP is directly related to the stiffness of the prosthetic ankle, the orientation of the limb, the position of the foot and shaft and the type of foot that is used[49;50]. As a result of the absence of active control in the ankle joint, a prosthetic limb produces less braking ground reaction force under the leading prosthetic limb in anterior-posterior direction, compared to the force under the sound limb in a sound limb leading situation[44]. To compensate for the limitations in the prosthetic foot and ankle, the leading prosthetic limb can be placed under a different angle with the vertical at initial contact, to change the position of the CoP, and therefore influence the CoM velocity. (The leading limb angle is defined as the angle between that limb and the vertical.) When making a larger leading limb angle, which results in a more forward positioned foot, the CoM decelerates more as a result of the more backward orientation of the GRF, compared to when making a smaller leading limb angle.

In the current study, we investigated how combinations of leading limb angles and internal active ankle moments of the sound ankle or passive stiffness of the prosthetic ankle influence the CoM velocity during the single limb support phase of the gait termination. Also, we considered if the trailing limb motion influences the CoM velocity during gait termination. Similar to gait initiation[63], we expect that when TF amputee subjects terminate gait with their prosthetic limb leading, the duration of the single limb support phase on the prosthetic limb is shorter compared to a sound limb leading condition. This strategy enables the TF amputee subjects to make use of the active muscle control possibilities in the trailing sound limb as quickly as possible. However, it may have its effect on the CoM velocity, as the fast forward acceleration and deceleration of the trailing sound limb toward the final stance position influences the direction of the GRF.

In our study, we used a two dimensional mathematical forward dynamics model based on Newton Euler and constraint equations and the gait termination data from a group of experienced TF amputee subjects using a prosthetic limb[44]. We divided the gait termination process in a single and double limb support phase, in both a prosthetic limb leading condition and a sound limb leading condition. We compared the CoP position, the GRF and the CoM Velocity of the TF amputee subjects during the single limb support phase with the outcome of the forward dynamics model. The model consisted of a leading limb, either a sound limb or prosthetic limb, with an ankle joint, a trunk and a trailing limb. The two limbs were connected to the trunk via hip joints. The model enabled us to inspect systematically the whole range of possible combinations of leading limb angles, active ankle moments of a sound ankle or passive stiffness of a prosthetic ankle, and trailing limb accelerations. Testing for these parameters in human subjects, without the interference of compensation strategies was not feasible; therefore it was decided to approach this problem in a theoretical way. The outcome and insights we gained from this study can be used to understand how TF amputees can compensate for the limitations in the active control of the CoP position during gait termination by using different leading limb angles.

5.2 Methods

5.2.1 Subjects

For this study, TF amputee subjects were recruited by a prosthetics workshop with clients in the three northern provinces of the Netherlands. Inclusion criteria for the amputee group were having a TF amputation for at least one year, daily use of a prosthetic limb and the ability to walk with the prosthesis more than 50m without walking aids. Amputees were excluded if they had any medical condition affecting their mobility or balance, like neurological, orthopedic or rheumatic disorders, otitis media, cognitive problems, severely impaired vision, reduced sensibility of the sound limb, or the use of antipsychotic drugs, antidepressants or tranquilizers. Furthermore, amputee subjects with pain or wounds to the stump, and fitting problems of the prosthesis were excluded.
The study group consisted of seven TF amputee subjects (6/1 (m/f, absolute numbers); mean 44.0 y (SD 14.1); mean 81.4 kg (SD 12.4); mean 1.83 m (SD 0.06); time since amputation mean 210.7 months (SD 158.1); side of amputation 5/2 (l/r, absolute numbers)). The TF amputee subjects used different types of prosthetic knees, all supplied with a free movable knee: Teh Lin (3), computerized C-leg (1), Total knee (1), Otto Bock 3R60 (1) and Proteval (1). The following prosthetic feet were used by the subjects: C-walk (2), dynamic SACH (2) and Endolite (3). The subjects walked with their own shoes.

5.2.2 Apparatus

The measurements were performed at the Motion Analysis Laboratory of the Center for Rehabilitation of the University Medical Center Groningen. A force plate of 0.40 x 0.60 m was used to measure the three dimensional forces and moments, which were used to calculate the position of the CoP in the horizontal plane. The force plate data were sampled at 100 Hz. The gait termination was recorded with two video cameras: one recording the coronal plane, the other the sagittal plane. The video frame rate was 25 Hz. Recording, synchronization and analysis of the force measurements and video registration was done with a custom-developed Gait Analysis System (GAS) based on LabView software. Synchronization was done by the hardware, using the same clock pulse.

5.2.3 Procedure

The subjects were instructed 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 limb and the sound 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 [96;97;98]. Adjustment of the step length in order to hit the force plate was avoided by practicing 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.
The data were obtained in vertical (GRFz; CoPz), anterior-posterior (GRFy (breaking force); CoPy) and mediolateral (GRFx; CoPx) direction. The CoM (CoMz; CoMy; CoMx) start velocity (CoM Velocity) and duration of the single limb support phase (leading limb foot on the ground) and the double limb support phase were calculated from the force plate data (figure 6.6).
The gait termination process was divided in two phases. The start of the single limb support phase was defined as the moment the leading limb was on the ground, the trailing foot was off the ground and the maximal GRFz was reached. The video images were used to determine if the trailing foot was off the ground. The moment of the transition from the single limb support phase to the double limb support phase was defined as the moment the CoPx reached the highest velocity when moving from under the single limb stance foot a point between the two feet. The end of the double limb support phase was defined as the moment the GRFy was reduced to zero.

5.2.4 Outcome Parameters

The force plate data of the group of TF amputee subjects were used to calculate the final outcome parameters. The CoM acceleration vector was directly calculated from Newton's second law: $m * \vec{a} = \vec{W} + \vec{F}$, where m is the subject's mass, $\vec{a}$ is the CoM acceleration, $\vec{W}$ is the subject's gravity vector and $\vec{F}$ is the GRF vector. The instantaneous CoM velocity was obtained by integration of the acceleration.
The CoM velocity at the start of the single limb support phase (CoM Velocity_{1ft}) and the double limb support phase (CoM Velocity_{2ft}), the duration of gait termination process, the duration of the two phases (Duration_{1ft}, Duration_{2ft}) and the impulses used in the two phases (Impulse_{1ft}, Impulse_{2ft}) were used to determine the global differences in the sound limb and prosthetic limb leading conditions. The trajectories of the CoMy velocity, the GRF and the CoPy position during the one foot phase were compared to the outcome of simulations with the mathematical model.

5.2.5 Statistical Analysis

For each subject, individual means of the outcome parameters over the two trials for the leading and the two trials for the trailing (prosthetic and sound) limb condition were calculated. A paired two sample student's t-test was used to determine the global differences between the two conditions. The level of significance was set to $p \leq 0.05$.

5.2.6 Mathematical Model

A two dimensional forward dynamics model was used to inspect the whole range of possible combinations, that may enable TF amputee subjects to achieve the equal CoM velocities at the end of the single limb support phase during gait termination when varying the leading limb angle and the internal active ankle moment of the sound ankle or passive stiffness of the prosthetic ankle (figure 5.1). The four elements model is based on Newton Euler constrained equations with forward dynamics [86]. We used Euler integration for the simulation steps ($\Delta t = 0.0001$ s). Some additions were made to simulate spring and damper limited joints and contact points with the external world.
The model consisted of a leading foot (length: 0.3 m, mass: 2kg; the proximal and distal end point can form contact points with the external world), with an ankle joint, and limb (length: 1.0 m, mass: 15 kg or 5 kg (respectively the sound limb and prosthetic limb)), a trunk (length: 1.0 m, mass: 48 kg) and a trailing limb (length: 1.0 m, mass: 15 kg or 5 kg) . The two limbs were connected to the trunk via hip joints. The limbs could be set with the inertial properties of a prosthetic limb or a sound limb. The elements were modeled as slender rods. Joints were simulated as frictionless pin point joints. Elements were constrained by joint forces based on equating the acceleration of element endpoints. Internal and external forces were passed on via the joint elements. Hip and ankle muscles were simulated as torque engines in the joints.
Input variables in the model were (1) the leading limb angle at initial contact, (2) the trailing limb motion during single limb support phase, and (3) the ankle stiffness or internal active moment of force of the ankle of respectively the leading prosthetic limb or the leading sound limb. The initial horizontal velocity of the CoM at the start of the gait termination was set to 0.7 m/s. Mixed dynamics, forward dynamics with some constraint or prescribed movement [86], were used to constrain the trunk element to the vertical of the external world by using hip moments. The known angular acceleration of the trunk was set to zero, keeping the trunk in the upright position. The velocity of the CoM was the result of the forces acting on the model, the inertial properties of the model and the internal moments produced by the model. A minimum jerk bell-shaped velocity curve [99] was defined for the angular velocity of the trailing limb. The angular acceleration derived from this angular velocity curve was scaled to the range the trailing limb was allowed to move forward. When the scale value was 1, the trailing limb ended hanging vertically next to the leading limb at the end of the single limb support phase. When it was 0, the trailing limb did not move forward. When the leading limb was set as a prosthetic limb, the passive prosthetic ankle was limited in its range of motion by a spring (C_{spring}) and damper (C_{damper}) system, creating a counter torque (T_c) based on the joint angle (θ) and joint angle velocity ($\dot{\theta}$) when the ankle angle exceeds its limits (Equation 5.1).

$$T_c = C_{spring} * (\theta_{limit} - \theta) - C_{damper} * \dot{\theta} \tag{5.1}$$

in which $C_{spring} = 100\ Nm$ and $C_{damper} = 25\ Ns\ m$.
When the leading limb was set as a sound limb, the internal active ankle moment (T_{ankle}) was set to be the known in the external moments when the leading foot was in contact with the ground. The sound ankle was set to produce a constant internal plantar flexion moment of 54 Nm.

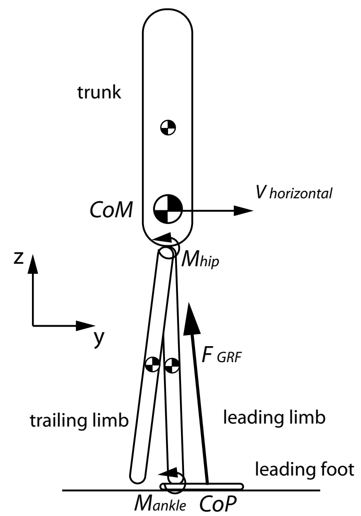


Figure 5.1: Gait Termination Model consisting of a leading limb and foot with ankle joint, a trunk and a trailing limb.

Ground reaction forces ($\vec{F}_{GRF}$) were formed when the leading foot made contact with the floor of the external world. Ground reaction forces were calculated based on continuously checking for the intrusion of the heel and toe ($\vec{X}_{foot_{y,z}}$) into the floor ($\vec{X}_{floor_{y,z}}$), which has a modeled stiffness ($C_{stiffness}$) and damping ($C_{damping}$) (Equation 5.2).

$$\vec{F}_{GRF} = C_{stiffness} * (\vec{X}_{floor_{y,z}} - (\vec{X}_{foot_{y,z}})) - C_{damping} * \frac{\Delta(\vec{X}_{floor_{y,z}} - \vec{X}_{foot_{y,z}})}{\Delta t} \quad (5.2)$$

in which $C_{stiffness} = 1 * 10^5\ N\ m^{-1}$ and $C_{damping} = 5 * 10^3\ Ns\ m$.

The mathematical model was used to study the influence of the leading limb angle at initial contact, the ankle properties (passive stiffness or internal active ankle moments) of the leading limb, and the trailing limb motion during foot contact with the leading limb on the anterior-posterior CoM velocity. Combinations of settings were investigated to study the influence of the ankle and limb properties on the end velocity. Table 5.1 shows the ranges of the initial settings.
The mathematical model was validated by examining the conservation of energy and comparing the modeled external forces with measured ground reaction forces of the TF amputee subjects. The mixed dynamics moments of force constraining the trunk element were checked against unrealistic values by inspecting and comparing the calculated values with known maximal joint moments of force in healthy subjects[100].

Table 5.1: Ranges of the initial settings of the leading limb angle, ankle stiffness or aktive moment of force and trailing limb motion in the gait termination simulation.

Condition	Sound limb leading	Prosthetic limb leading
Leading limb angle	0.13 - 0.23 rad 7.45 - 13.18°	0.18 - 0.28 rad 10.31 - 16.04°
Ankle stiffness	-	0 - 1 [a]
Internal active ankle moment	19 - 69 Nm	-
Trailing limb motion	0 - 1[b]	0 - 1[b]

[a] 1: maximal stiffness (= 100 N m^{-1}).
[b] 1: trailing limb is next to the leading limb at the end of the simulation, hanging vertically.

Table 5.2: Gait termination: Averages and standard deviation of the velocity of the CoM (CoM Velocity) at the start ($t = 0$) of the single limb support phase ($1ft$) and double limb support phase ($2ft$), the gait termination duration (Duration; $total = 1ft+2ft$) and the braking impulse (Impulse) in the sound limb leading and prosthetic limb leading conditions (n=6).

Condition	Sound limb leading	Prosthetic limb leading
CoM Velocity$_{1ft(t=0)}$(m/s)	0.72(SD 0.12)	0.69(SD 0.15)
CoM Velocity$_{2ft(t=0)}$(m/s)	0.14(SD 0.06)*	0.33(SD 0.08)
Duration$_{total}$(s)	1.04(SD 0.41)	1.02(SD 0.31)
Duration$_{1ft}$(s)	0.58(SD 0.15)*	0.32(SD 0.08)
Duration$_{2ft}$(s)	0.46(SD 0.43)	0.70(SD 0.29)
Impulse$_{total}$(N s)	56.1(SD 12.1)	54.5(SD 14.1)
Impulse$_{1ft}$(N s)	48.7(SD 15.6)*	29.6(SD 8.1)
Impulse$_{2ft}$(N s)	7.3(SD 6.6)*	24.9(SD 9.7)

Significant differences ($p<0.05$) between the two leading limb conditions are marked with *.

5.3 Results

5.3.1 Subjects

All subjects, except one, were able to come to a full stop in one gait termination cycle (both the leading and trailing limb placed on the ground). Although the CoMy velocity of the subject that could not come to a full stop, was heavily reduced to almost zero, one small step with his leading limb was made extra to come to the complete full stop in both conditions. We could not find a convincing biomechanical reason for this extra step. We excluded the data from this subject from the analyses.

No significant differences were found in the velocity at the beginning of the gait termination process in the two conditions (table 5.2). Also, the overall duration and the impulse used to come to a full stop were almost equal. However, significant differences were found when the gait termination process was divided in two phases. When the gait was terminated with the sound limb leading, the gait termination was almost completely executed during the single limb support phase (figure 6.6A). 85 percent (SD 14) of the total impulse was generated in the first part of the process. The velocity at the end of the single limb support phase was reduced to 20 percent. On average, the duration of the single limb support phase and the double limb support phase were almost equal, but a large standard deviation was found for the double limb support phase (figure 6.6A,B). Although CoMy velocity was already heavily reduced for most subjects the moment they placed the trailing limb on the ground, the double limb support phase took quite long to come to a complete stop (figure 6.6B).
When terminating with the prosthetic limb, the average single limb support phase took 33 percent (SD 8) of the total process time. During this phase 55 percent (SD 10) of the total brake impulse was generated (figure 6.6D).
The data shows that our subjects use the CoP and the GRF to come to a complete stop with their CoM. Positive GRFy results in braking of the subjects.

CoP In the sound limb leading condition our subjects moved the CoP to anterior under the foot in the first part of the single limb support phase. During the double limb support phase, almost no CoP motion was found (figure 6.6A,B).
When leading with the prosthetic limb, the CoP moved slowly to anterior during the single limb support phase. When the trailing sound foot was placed on the ground the CoP continued moving forward, while transitioning the weight from the leading prosthetic limb toward the sound limb (figure 6.6C,D).

GRF In the sound limb leading condition the GRF decreased as presented in figure 6.6B in 5 subjects. In one subject the GRFy decreased and increased after the start of the single limb support phase (figure 6.6A).
In the prosthetic leading conditions the GRFy also decreased after the start of the single limb support phase, but was still higher at the end of the single limb support phase. Compared to the sound limb leading condition a clear increase was found around the start of the double

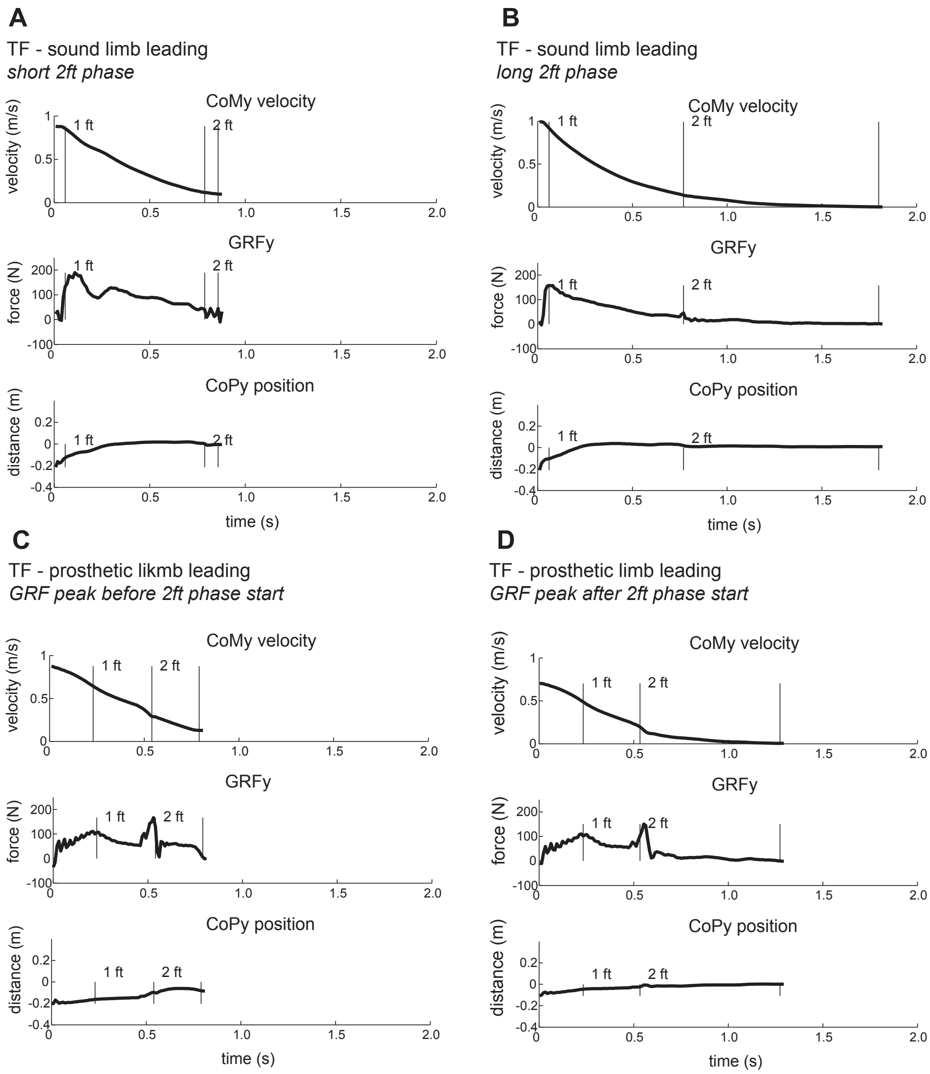


Figure 5.2: Exemplary gait termination by TF amputee subjects in sound limb and prosthetic limb leading conditions, divided in single limb support phase (1ft) and double limb support phase (2ft). Vertical lines represent the start and end of a phase.
A: Sound limb leading, with a short double limb support phase and a deviating decrease and increase in the GRF just after the single limb support phase start. 1 subject produced this GRF pattern.
B: Sound limb leading, with a long double limb support phase, although the CoMy velocity is already heavily reduced at the start of the double limb support phase. 5 subjects produced similar GRF patterns.
C: Prosthetic limb leading, with GRFy peak after start double limb support phase. 4 subjects produced similar GRF patterns.
D: Prosthetic limb leading, with GRFy peak before start double limb support phase. 2 subjects produced similar GRF patterns.
CoPy: 0 = center of force plate; positive value indicates anterior position of the CoP (toward toe of foot).

limb support phase in the GRFy in four of the six subjects. This increase was found not only before but also after the weight shifting from leading limb to trailing limb (figure 6.6D).

CoM The curved trajectory of the CoMy velocity in single limb support phase of the sound limb leading condition differed from the relatively more linear trajectory in the prosthetic limb condition. The CoMy velocity showed an larger decrease around the start of the double limb support phase (figure 6.6C,D).

5.3.2 Mathematical Model

The CoMy velocity, the position of the CoP and the GRFy in the mathematical model showed similarities with the measurements (figure 5.3).

Initial Parameters

The average velocity of the TF amputee subjects at the beginning of the single limb support phase (0.7 m/s) was used as an initial parameter for the mathematical model in both the sound limb leading condition and prosthetic limb leading condition. The goal was to achieve end velocities with the model, which were similar to the average end velocity of the prosthetic limb group. In the sound limb leading condition, the average end velocity was 0.14 m/s. In the prosthetic limb leading condition the average end velocity was 0.33 m/s. These end velocities had to be reached in the same time as the subjects used to reach these end velocities. The average duration in the sound limb leading condition was 0.58 s and 0.32 s in the prosthetic limb leading condition. To achieve the end velocity in the sound limb leading condition, we set the leading limb angle of the model at initial contact to 0.18 rad (10.31°). The sound ankle was set to produce a constant internal plantar flexion moment of 54 Nm. In the prosthetic limb leading condition, we set the leading limb angle of the model at initial contact to 0.21 rad (12.03°). The prosthetic ankle of the model was set as a flexible ankle. In both conditions, the velocity of the trailing limb was set in such a manner that the limb would be next to the stance limb at the end of the single limb support phase. When simulating with the initial parameters, a clear decrease in the CoMy velocity in the sound limb leading condition was found. In the prosthetic limb leading condition a plateau was found in the CoMy velocity halfway the single limb support phase. This plateau was accompanied by a clear increase in the GRFy. The CoPy in the sound limb condition showed a gradual translation toward the toe, while in the prosthetic limb leading condition the CoPy rapidly shifted from one position to another after the foot was flat on the ground.

Combinations of settings

Figure 5.4 shows that only small areas can be found in which the calculated end velocities of the model match the average end velocity of the group of TF amputee subjects. When chosen inadequate, the wrong combination of sound ankle moment and limb angle results in a CoM end velocity that differs substantially from the desired end velocity, which was achieved by the TF amputee subjects. The ankle stiffness appears to have limited influence on the CoM end velocity. The trailing limb motion seems only relevant for the CoM end velocity when the combination of the sound ankle moment or ankle stiffness and the limb angle result in a CoM end velocity which is close to the average end velocity of the TF amputee subjects. This can be seen around the small areas in which the exact end velocity is found.

5.4 Discussion and Conclusion

Although it seemed that no differences in the sound limb leading condition and prosthetic limb leading condition were found in the global outcome values of the overall process (CoM velocity, duration and impulse), clear differences were found when the process was divided in the single limb support phase and double limb support phase. When leading with the sound limb, the subjects came almost to a full stop in the single limb support phase, without the use of the prosthetic limb. When leading with the prosthetic limb, the CoM deceleration was less in a relatively short single limb support phase, with a fast forward swing of the trailing sound limb. One possible reason for limiting the single limb support phase can be given: the active control possibilites in the sound limb are necessary for the gait termination. For succesful gait termination adequate CoP positioning and an associated backwards GRF are necessary. Anterior motion of the CoP during the single limb support phase was found in both leading limb conditions. When the leading sound limb was in contact with the ground, with the foot flat on the ground, the CoP motion was the result of an internal active ankle plantar flexion moment. When the leading prosthetic limb was in contact with the ground the limited CoP motion under the prosthetic foot was the result of a passive coupling between the foot and the changing limb angle during the gait termination. The lack of active control in the passive prosthetic ankle prevents the TF amputee subjects from producing sufficient braking force when terminating gait with the prosthetic limb leading. A more forward placed CoP produces a more backward GRF, but in the TF amputee, CoP can not be moved actively forward using muscular control. To compensate for this shortcoming, the leading prosthetic foot can be placed more forward, resulting in a larger GRFy, which contributes to the production of the braking force. However, the simulations show that CoM velocity is very susceptible to foot placement. This finding shows that using this strategy for accurate control is not very likely. Only narrow ranges of values could be used to achieve the same CoM velocities with the mathematical model as the average start and end velocities of the prosthetic limb user. These outcomes imply that a prosthetic limb user does not have much freedom choosing the leading limb angle, the trailing limb motion and the ankle moment. Based on the simulations, it seems that TF amputee subjects have to make use of sound limb depending strategies during gait termination. Instead of the leading limb being responsible for the necessary braking force as seen in healthy subjects, it is the sound limb, either as the leading limb or trailing limb, which is mostly responsible for the necessary braking force compensating for the shortcomings in the prosthetic limb. Also, it should be taken into account that not every limb angle is possible, as the CoP position and the resulting GRF are essential for passive knee extension. Small changes in CoP placement

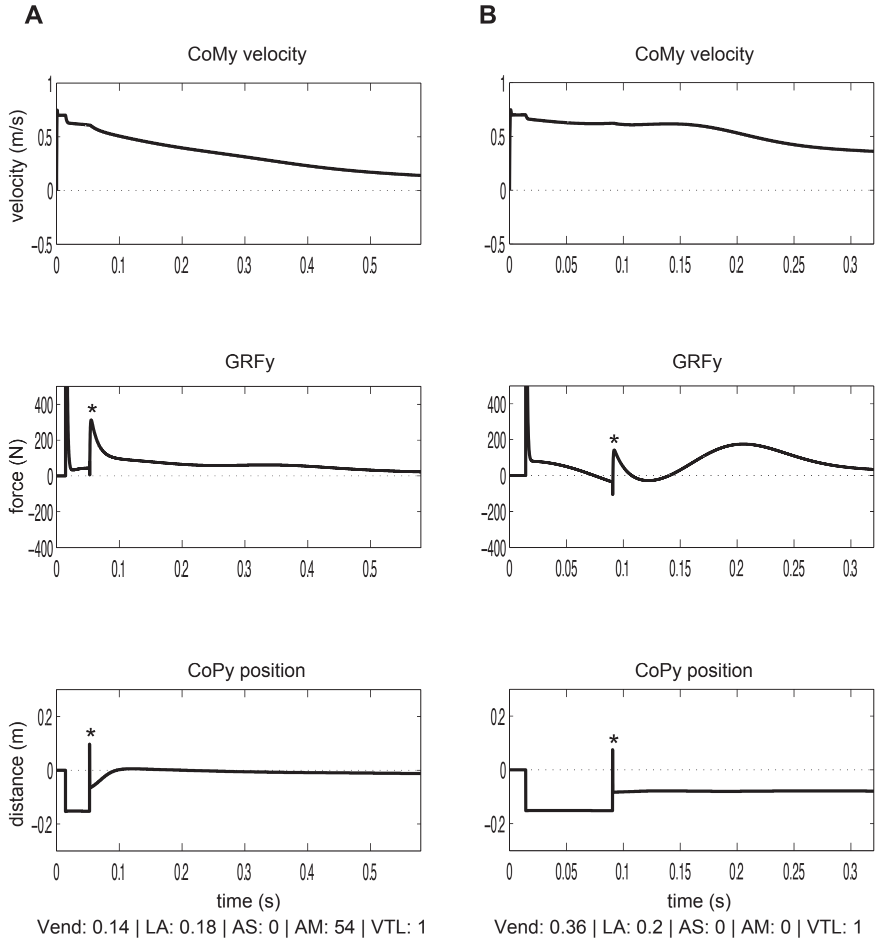


Figure 5.3: Results of the gait termination model during the single limb support phase, either leading with the sound limb (A) or with the prosthetic limb (B). Initial parameters were used to achieve initial velocities and end velocities that were similar to the average velocities of the TF amputee subjects.
Vend: end velocity (m/s); LA: leading limb angle (rad); AS: ankle stiffness (graduated scale from 0 to 1; 0: max. flexible, 1: max. stiff); AM: internal active ankle moment (Nm); VTL: Trailing limb motion (graduated scale from 0 to 1; 0: no motion, 1: inversely related to leading limb)
The sudden changes of the CoPy position and GRF at the transition of the heel contact to the foot flat on the ground, marked with ⋆, are artifacts of the mathematically modeled floor stiffness and elasticity, and the flat shoe sole, causing a quick alternation between heel and toe contact.
CoPy: 0 = center of modelled force plate; positive value indicates anterior position of the CoP (toward toe of foot).

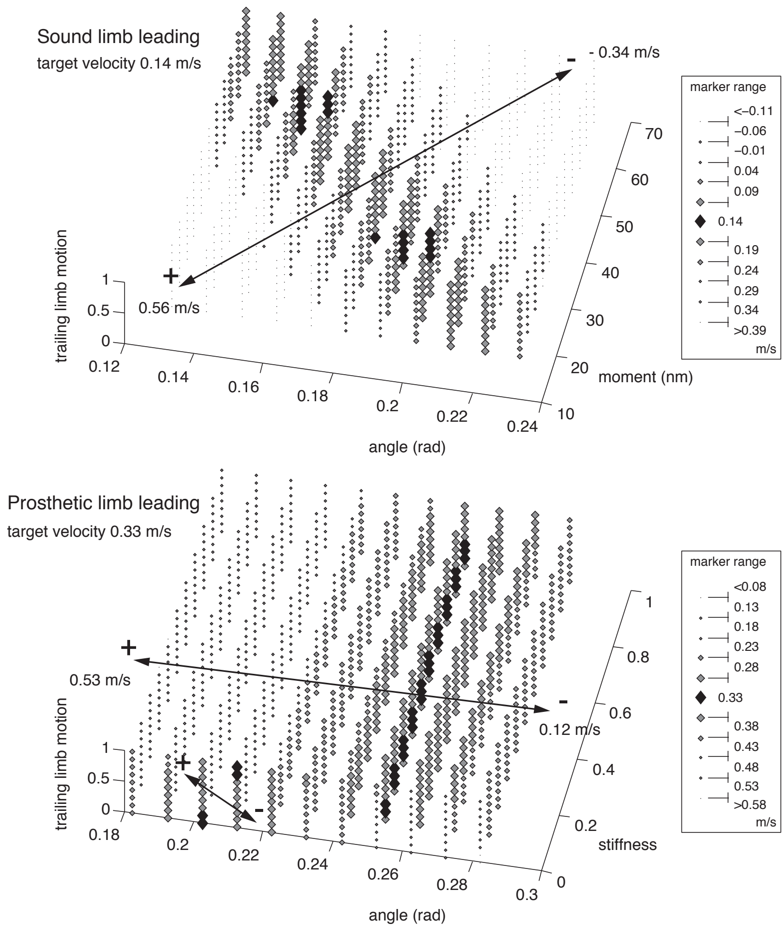


Figure 5.4: Results of a gait termination model with several different parameters; Relation between the influence of the leading limb angle (rad), sound ankle moment (Nm) or prosthetic ankle stiffness (graduated scale from 0 to 1; 0: max. flexible, 1: max. stiff) and trailing limb motion (graduated scale from 0 to 1; 0: no motion, 1: inversely related to leading limb) and the anterior-posterior CoM velocity at the end of the gait termination. The calculated end velocities of the model that are equal to the average end velocity of TF amputee subjects are visualized with black markers. The size of the markers indicates the magnitude of the correspondence with the average end velocity of the TF amputee subjects. The arrows show the direction of the change in velocity. The numbers at the end of the arrows show the range of the end velocity. Negative velocity values represent motion in the backward direction. The initial anterior-posterior CoM velocity is set to 0.7 m/s, the end velocity to 0.14 m/s in the sound limb leading condition and 0.33 m/s in the prosthetic limb leading condition, and the duration of the single limb support phase is set to 0.58 s in the sound limb leading condition and 0.32 s in the prosthetic limb leading condition.

may result in buckling of the knee, depending on the position of the instantaneous knee axis. This sound limb depending strategy has some consequences for the gait termination process. In the short single limb support phase when the prosthetic limb is leading, the relatively heavy sound trailing limb has to be moved forward. This influences the way the subjects can use the braking force to reduce the CoMy velocity. Slowing down the trailing sound limb velocity at the end of its swing phase by producing adequate moments of force with both hips results in a relatively larger braking force, compared to when slowing down the relatively light prosthetic limb in the sound limb leading condition. This relation explains the linear CoMy velocity decrease in the single limb support phase in the prosthetic limb leading condition (figure 6.6C,D).
It seems that the way the trailing sound limb is slowed down is critical, but only when the leading prosthetic limb angle is chosen correctly and in accordance with the stiffness of the prosthetic ankle as can be seen in figure 5.4. When the ankle is very stiff and the limb angle is large, the CoPy remains under the heel. Although the large limb angle would contribute to a higher braking force, the position of the CoP reduces this gain, since it remains under the heel for quite some time. When using a flexible ankle and the same limb angle, the CoPy moves forward at the beginning of the stance phase resulting in a larger braking force. For gait termination it seems that a flexible ankle or heel setting would be preferred as this setting increases braking force. In contrast to gait termination, during gait this setting is counter productive, because CoMy velocity is lost during early stance phase, hindering forward progression. The settings of the prosthetic ankle or heel should be different for gait termination than for gait. Therefore, a well-balanced choice should be made. The choice for the settings depends on the subjectÕs needs.

Some remarks and considerations should be given when interpreting the outcome of the current study. In this study, we were merely interested in specific underlying principals of gait termination. Since no exact numerical correspondences were to be expected, we decided that the models anthropometrics had to match those of a human being roughly. For simplicity reasons we chose not to add moveable knee joints. Moveable knee joints have some benefits during gait termination. A moveable knee joint has its added value in the foot clearance process and can be used to decrease the impact of the GRF during heel strike. We realize that these knee joints influence the CoM velocity. Already we showed that the orientation of the GRF and the weight of the trailing limb are of importance for CoM velocity changes. Similar to this, the inertia of the limb, influenced by the orientation and position of the upper and lower part of the limb, has its effect on the CoM velocity changes. The absences of the knee joints might explain some of the minor differences between the results of our model and the measured data of the TF amputee subjects. Although we did not use moveable knee joints, it is our opinion that the current findings are relevant. Most of the prosthetic knees remain in full extension during the gait termination process, especially in the prosthetic leading limb condition, but also in the prosthetic limb trailing condition. In this last condition, many times the limb does not flex as a result of the low angular velocity and the minimal use of the GRF which is not sufficient to create a flexion moment around the knee joint.
Also for the sake of simplicity, we did not add a trailing foot. Adding a foot would probably enabled us to reproduce the quick braking force alteration around the beginning of the double limb support phase. Based on our observations during the measurements and the consequences of the definition of the single limb support phase to double limb support phase transition, we assume that the quick alteration is the result of the landing of the trailing foot and the moment in which the weight transition to the sound limb occurs.

In one subject the braking force decreased and increased at the beginning of the single limb support phase as a result of plantar flexion of the stance ankle joint of the sound limb. This plantar flexion was used to increase limb length to reach toe clearance of the trailing prosthetic limb at the beginning of its swing phase. Obviously, this phenomenon was not found in the mathematical model because the plantar flexion strategy was not necessary as we did not include a trailing foot.
Because of the absence of the moveable knees and the trailing foot, we were not able to study their influence on the CoM velocity. It is likely that TF amputees make use of these possibilities to influence their CoM velocity. Similar to this, it should be taken into account that also the trunk influences the CoM velocity, and that TF amputees probably use their trunk to influence the CoM velocity as well. We observed variation of the trunk position when analyzing the video images. Four out of the six subjects showed a more backward trunk position in the prosthetic limb leading condition. This strategy results is a more backward positioned CoM, relative to the CoP position, which contributes to the deceleration of the CoM. In our mathematical model, the trunk is controlled to stay in vertical orientation using adequate hip torques. A more advanced model should be used to study the influence of these items. The current model is used to study the relation between the leading limb angles, the internal active ankle moments of the sound ankle or passive stiffness of the prosthetic ankle, the trailing limb motion and the CoM velocity during the single limb support phase of the gait termination.

During gait termination TF amputee subjects can compensate for the absence of voluntary muscle activity and joints in the prosthetic limb by influencing the CoP position and the GRF under their prosthetic limb, and with their sound limb. Patients have to be trained in such a manner that they learn how to produce the optimum of braking force with both of their limbs used as a functional unit: using an adequate limb angle on the prosthetic side and using the capacities of the sound ankle. Because the duration of the total gait termination process and the produced braking impulse do not differ when leading with the prosthetic limb or with the sound limb, it seems that there is no reason to enforce a preferred leading limb during gait termination. However, when terminating gait with the sound limb leading, the TF amputee subjects are able to reduce their CoMy velocity more quickly during the first part of the process. This gait termination strategy is the preferred strategy in the experienced TF amputee subjects group.

Chapter 6

General discussion

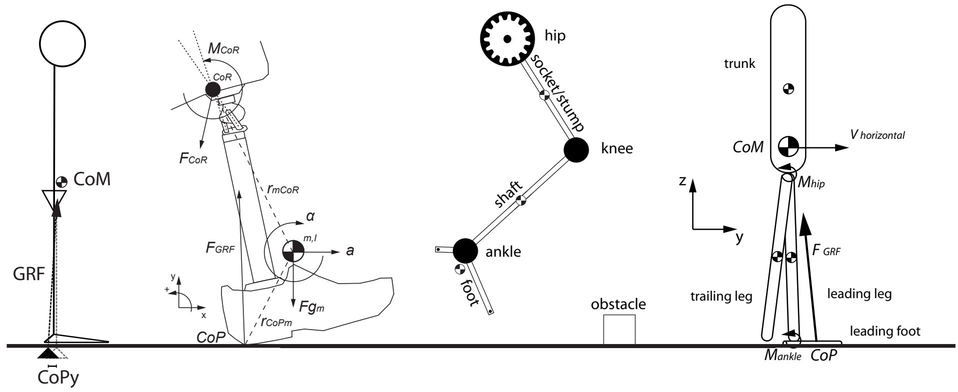


Figure 6.1: Four models that were used in this thesis to study specific stages during gait: gait initiation, weight bearing, prosthetic limb swing and gait termination.

6.1 Introduction

In the current thesis 'Model and measurement studies on stages of prosthetic gait.', we reported predictions, outcomes and insights gained from studies in which we investigated how transfemoral (TF) amputees compensate for the limitations in the active control of the orientation of the ground reaction force (GRF) and the position of its origin, the center of pressure (CoP), during gait initiation and termination, and the absence of active knee extension control during weight bearing on the prosthetic limb. Also, we reported which strategies contribute to successful prosthetic limb forward swing during obstacle avoidance and what should be taken into account during obstacle avoidance.
We used mathematical models to investigate principles of TF amputee prosthetic gait in four specific stages, namely gait initiation, weight bearing, prosthetic limb swing and gait termination (figure 6.1). For every stage, a mathematic model was designed which allowed us to analyze in a conceptual world phenomena we have observed in the real world [8]. Data of TF amputee and able-bodied (AB) subjects using a knee walker prosthetic device were used to validate the models that were all checked for (internal) consistency, conservation of energy and realistic parameters. The four stages were described in separate chapters in this thesis.
The findings were used to increase our insights and contributed to the development of our theory on TF prosthetic gait. The predictions of the models changed our thoughts on certain choices in strategies that are made when using prosthetic limbs. These models can be of interest when forming new ideas about how to improve prosthetic knees and feet and how to improve rehabilitation programs.

6.2 Predictions on four stages of gait

Measurement data and mathematical models were used to investigate principles of TF amputee prosthetic gait in gait initiation, weight bearing, prosthetic limb swing and gait termination. These four stages were described in separate chapters in this thesis. Based on the predictions, outcomes and findings, we found strong indications that confirm the idea that during gait TF amputee mainly depend on the sound limb to accelerate and decelerate their body, by influencing CoP and GRF and changing timing of actions[27;44;28;63;101]. Also, we can confirm the idea that TF amputee can use the GRF to extend their prosthetic knee during stance when bearing body weight on the prosthetic limb[102], and, on the other hand, TF amputee can also use GRF forces and fast hip motions to overcome the knee extension aiding spring[43;103]. Important to note is that TF amputee appear to have to learn to take into account the environment in which they are functioning and learn to combine body segment properties and dynamics, which enables them to enlarge their (limited) leeway during difficult tasks.

6.2.1 Gait initiation

In chapter 2 we investigated how TF amputee control the spatial and temporal parameters which modulate the propulsive forces, the positions of the CoP and the CoM during prosthetic gait initiation. Force measurement data were used to calculate the CoM velocity curves in horizontal and vertical direction. Our forward dynamics model predicted that whether the TF subjects initiate gait with their prosthetic limb or with their sound limb, their horizontal end velocity is equal. The subjects compensated the loss of propulsive force under the prosthesis with the sound limb, both when the prosthetic limb is leading and when the sound limb is leading. In the vertical CoM velocity a tendency for differences between the two conditions was found. When initiating gait with the sound limb, the downward vertical CoM velocity at the end of the gait initiation was higher compared to when leading with the prosthetic limb. We concluded that our subjects used a gait initiation strategy which depended mainly on the active ankle function of the sound limb and therefore they changed the relative durations of the gait initiation anticipatory postural adjustment phase and the step execution phase.

6.2.2 Weight bearing

In chapter 3, we focused on the occurrences of stabilizing and destabilizing external moments of force on a prosthetic knee during stance, in the first steps after gait initiation, in inexperienced users. Primary aim was to identify the differences in the external moments during gait initiation with the sound limb leading and the prosthetic limb leading. A prosthetic limb simulator device with a flexible knee was used to test AB subjects, with no walking aid experience. Inverse dynamics calculations were preformed to calculate the external moments. The subjects learned to control the prosthetic limb within 100 steps, without walking aids, evoking similar patterns of external moments of force during the steps after the gait initiation, either with their sound limb loading or prosthetic limb leading. Our inverse dynamics model predicted that critical phases in which a sudden flexion of the knee can occur were found just after heelstrike and just before toe off, in which the external moment of force is close to the maximum internal moment produced by a knee extension aiding spring in the opposite direction. We conclude that from a safety perspective it does not matter which limb is the leading limb during gait initiation.

6.2.3 Prosthetic limb swing

In chapter 4, conditions that enable a prosthetic knee flexion strategy in TF amputee during obstacle avoidance were investigated. This study explored the hip torque principle and the static ground principle as object avoidance strategies. A prosthetic limb simulator device for AB subjects was used to study the influence of applied hip torques and static ground friction on the prosthetic foot trajectory. Inverse dynamics was used to calculate the energy produced by the hip joint. A two-dimensional forward dynamics model was used to investigate the relation between obstacle-foot distance and the necessary hip torques utilized during obstacle avoidance. Our inverse and forwards dynamics model predicted that a prosthetic knee flexion strategy is facilitated by the use of ground friction and by larger active hip torques. This strategy requires more energy produced by the hip compared to a knee extension strategy. We concluded that when an TF amputee maintains sufficient distance between the distal tip of the foot and the obstacle during stance, he or she produces sufficiently high, yet feasible, hip torques and uses static ground friction, the TF amputee satisfies the conditions to enable stepping over an obstacle using a knee flexion strategy.

6.2.4 Gait termination

In chapter 5 we investigated how leading limb angles combined with active ankle moments of a sound ankle or passive stiffness of a prosthetic ankle, influence the CoM velocity during the single limb support phase in gait termination. Also, we studied how the trailing limb acceleration influences the CoM acceleration during this phase. We analyzed force plate data from a group of experienced TF amputee using a prosthetic limb, and the outcome from a two dimensional mathematical forward dynamics model. We found that when leading with the sound limb, the subjects came almost to a full stop in the single limb support phase, without the use of the prosthetic limb. When leading with the prosthetic limb, the CoM deceleration was less in a relatively short single limb support phase, with a fast forward swing of the trailing sound limb. Slowing down the heavier trailing sound limb, compared to the prosthetic limb, results in a relatively larger braking force at the end of the swing phase. Our inverse and forwards model predicted that only narrow ranges of leading limb angle and ankle moments could be used to achieve the same CoM velocities with the mathematical model as the average start and end velocities of the prosthetic limb user. We conclude that users of prosthetic limbs have a narrow range of options for the dynamics variables to achieve a target CoM velocity. The lack of active control in the passive prosthetic ankle prevents the TF amputee from producing sufficient braking force when terminating gait with the prosthetic limb leading, forcing the subjects to use both limbs as a functional unit, in which the sound limb is mostly responsible for the gait termination.

6.3 Clinical consequences

The predictions, outcomes and insights gained from our model and measurement studies contributed to the development of our theory about asymmetry, funcional ability and learning. Based on our findings, we concluded that it is impossible to walk symmetrically with a mechanical prosthetic limb, unless additional efforts are made to compensate for the shortcomings in the prosthetic limb. We expect that improving functional ability, instead of minimizing asymmetry, will contribute to the improvement of the patients satisfaction. According to the principles of Discovery Learning and learning as a function of attentional focus, improving the functional ability can best be achieved by training in environments which enable the TF amputee to find individual optimal performance patterns for complex motor skills.

6.3.1 Asymmetry and functional ability

For high functioning individuals with lower-limb amputation, gait deviation does not significantly correlate to patient satisfaction[104]. Kark and Simmons (2011) suggested that improving self-percieved functional ability and attitudes toward the prosthesis, rather than minimizing gait deviation, will improve patient satisfaction. To improve functional ability we have to understand the principles of prosthetic limb use and compensation strategies. Studies by Vrieling et al. (2009) showed that 'an important source for the creation of adjustment strategies in amputees was the non-affected limb'. This last finding is also confirmed by some of the studies in the current thesis. The observed real world phenomena and predictions of our models of the conceptual world show that this sound limb dependency is an important strategy, which contributes to functional ability.

Both experienced and recently TF amputated subjects, tested during gait initiation and termination and during obstacle crossing, in a motion analysis laboratory[8], showed a decrease in CoP motion under the prosthetic limb, with more weight bearing of the non affected limb. During gait initiation they were able to produce less forward velocity of the CoM during gait initiation compared to AB subjects. The TF amputee showed less braking force under the prosthetic limb during gait termination compared to the sound limb. This shortcoming when using the prosthetic limb, has consequences for the gait initiation and termination strategy. During gait initiation, subjects preferred to lead with the prosthetic limb. During gait termination, subjects preferred to lead with the sound limb. These findings suggest that the subjects prefer to use the sound limb to produce the necessary impulse. However, our gait termination model study showed that not only the leading limb influenced the magnitude of the impulse during the single limb stance phase, but also the trailing limb accelerations influenced the impulse. This finding suggest that also the moving trailing limb, and probably other moving segments, can be used in a CoM velocity control strategy. We found a relation between the mass of the trailing limb and the magnitude of the impulse. The accelerations of a relatively heavier trailing sound limb influenced the ground reaction force more than a lighter prosthetic limb. This asymmetrical mass distribution within the TF amputee and the consequences for the outcome of limb motions contribute to our theory that the prosthetic limb user has to control the inequality between the two limbs that take part in the teamwork and the idea that symmetry can only be achieved when additional efforts are made to compensate for this mass inequality.

Not only during gait initiation and termination the TF amputee has to make additional effort to compensate for the inequality when pursuing symmetry, but also during obstacle crossing. Vrieling et al. (2009) showed that when crossing an obstacle, TF subjects preferred to lead with the sound limb. Their trailing prosthetic limb showed a decrease in flexion compared to a sound knee, which was compensated with a circumduction movement and an increase in the plantar flexion of the sound ankle in the stance phase. According to our model study it should be feasible to counteract this asymmetrical motion of the prosthetic limb by using ground friction and fast hip motions. When using this strategy, TF subjects should be able to cross the obstacle with a flexed knee. The disadvantage of this strategy is that it costs more energy compared to the knee extension strategy.

Based on the findings in the laboratory and the model studies, which show that compensation strategies mainly depend on the active control possibilities of the sound limb and that more energy is needed when trying to counteract the asymmetrical motions, it seems an unattainable goal to pursue full symmetry during prosthetic gait. We expect, in accordance with Kark and Simmons, that improving functional ability will probably contribute to the improvement of the patients satisfaction. Please notice that we removed the words 'self-percieved' and 'attitude towards the prosthesis' from the sentence used by Kark and Simmons, and added 'probably' as these parameters were not included in the current thesis, which focusses on specific elements of functional ability during gait from a biomechanical perspective.

6.3.2 Learning

Traditionally, TF amputee who learn to walk with a prosthetic limb with an artificial knee, perform poorly during the initial gait training, hence the use of parallel bars, support by therapists and other safety measures. Our studies showed that AB subjects learned within 100 steps how to use the prosthetic limb in a safe way, provided that they were paying attention to how they placed their device on the floor. Their training was only limited, no therapists were involved and the only safety measure provided was the wall next to the subjects. To learn the subjects how to use the prosthesis, we created an environment in which our subjects were allowed to move around and to do and use whatever they thought was necessary. Although it seemed a rather unorganized environment, we made sure that there were sufficient objects that could be of help to understand how the prosthetic device behaves when in use. We placed some balls (to kick) and boxes (to step on and off) in a room that challenged the subjects. This training concept was based on the principles of Discovery Learning[105] and learning as a function of attentional focus[106]. Discovery Learning takes place in problem solving situations where the subjects draw on their own experience and prior knowledge and is a method of instruction through which subjects interact with their environment by exploring and manipulating objects, wrestling with questions and controversies, or performing experiments. Using the objects, we directed the subjects' attention to the effects of their movements (external focus), in contrast to attention to the movement itself (internal focus)[106]. The self-initiated noisy training sessions featured a variety of between-exercises differences, which would help the subjects to learn motor skills that were adapted better to their own physical needs and skills and enabled them to find individual optimal performance patterns for given complex motor skills. Even though an occasional fall occurred, based on the short learning time (100 steps) of our subjects, we feel that this type of training and the concepts of the underlying theories have relevance for the development of improved rehabilitation programs.

In this Discovery Learning concept, rehabilitation professionals working with TF amputee

do have an important role. They have to deploy their practical and theoretical skills to create safe environments in which patients are offered the opportunity to explore all the possibilities of their prosthetic limbs without physical assistance. In these environments, TF amputee are allowed to make errors within certain limits [107]. Domingo and Ferris (2009) reported that physical assistance can hinder motor learning of walking balance, as it does not allow for error detection and correction. Only for more difficult tasks assistance appears less detrimental. Of course, it should be taken into account that freezing situations, in which the TF amputee reduces his Degrees of Freedom during skill acquisition as described by Bernstein (1967) in too difficult tasks must be limited, as it might narrow the range of possibilites.
When creating these environments, therapists should take into account that TF amputees have to use the aforementioned strategies to compensate shortcomings [31]. Therefore, it seems advisable that the therapists create situations in which their patients experience the benefits of initiating gait with the prosthetic limb leading and terminating gait with their sound limb leading. Also, patients should experience that by using ground friction, fast hip motions and maintaining distance between the obstacle and the prosthetic foot during obstacle crossing it is possible to create foot clearance over an obstacle.

6.4 Future research

Although now it might seem possible that we can make predictions about certain phases of gait, which can eventually be used to improve the functional ability of prosthetic limb users, one more important step has to be made in this research process. If we want to improve the functional ability, we first have to see if we can improve the prosthetic limb design and the rehabilitation programs. Our models can contribute to this process, although the validity and usability of our models in both the conceptual and real world have to be verified further by experiments with TF amputees or with AB subjects using a prosthetic limb simulator device in various conditions, with various prosthetic components and properties in various environments (figure 1.2C, chapter 1.3).

Some of the predictions that have to be verified and confirmed, or some of the questions that arise from these models, are:

- Is it possible to learn how to use the fast hip motion strategy, the GRF strategy and the leading limb strategy and does it have benefits in the real world to learn how to use these strategies?
- How does trunk sway influences the four stages of gait?
- How do our current models relate to the modern microprocessor controlled lower limb prosthetics?

6.4.1 Strategies

Is it possible to learn how to use the fast hip motion strategy, the GRF strategy and the leading limb strategy and does it have benefits to learn how to use these strategies? Based

on the available relevant data and physical, mechanical and physiological principles, our models predicted that these strategies should be feasible. It is our expectation that, on the condition that a suitable rehabilitation program can be developed, TF amputees are able to make motions similar to the motions of the model.
A possible limiting factor that might interfere with adopting these strategies is the high demands that are placed on the already heavily used hip muscles. An increased absorption and energy generation by the muscles that control the hip joint of the amputated limb is necessary to compensate for the absence of the lower leg muscles during gait. This main compensatory strategy is seen in unilateral transtibial amputees during gait [16]. It should be taken into account that these high demands might lead to overuse of these muscles.

A remark that has to be made here is that some parts of our models were verified and compared to the data of AB subjects using a knee walker prosthetic device. Hofstad et al. [80] reported bilaterally delayed and reduced responses in persons with a lower limb prosthesis. This finding reflects a basic reorganization within the central nervous system aimed at providing synchronized activity in both lower limbs, even though the peripheral deficit involves only one limb. In our AB subjects using the knee walker this reorganization probably did not occur, as their training was only limited to 100 steps on the device. This of course should be taking into account when verifying and validating the predictions of our models. On the other hand, these 100 steps were sufficient to teach the AB subjects how to use the prosthetic limb in a safe way, provided that they were paying attention to how they placed their device on the floor. Although literature is not conclusive on the effects of early weight bearing on stump healing, volume reduction and functional outcome [109], Kozáková et al. (2010) reported that faster prosthetic fitting and gait training decreases asymmetry from body weight distribution between both the limbs. Perhaps, faster prosthetic fitting influences the reorganization within the central nervous system. The subjects in the study by Hofstad (2009) were amputated more that 5 year before they were studied. Until recently immediate prosthetic fitting was not common, and perhaps these subjects were not trained with an immediate prosthesis fitting protocol. In that case, it would be interesting to repeat the study by Hofstad with TF amputees who were trained in an immediate prosthesis fitting program, and investigate how this influences the central nervous system.

6.4.2 Trunk sway

How does trunk sway influence the four stages of gait? Our models predicted that changes in all body segments influence the outcome of the models. For simplicity reasons, the motions of the trunk in our models were kept to a minimum. Observations revealed that TF amputees use a lot of trunk motion during gait, not only in the forward-backward direction, but also in the left-right direction. We also noticed that the TF amputee tended to hang over with their trunk toward their prosthetic side during the left-right trunk sway(figure 6.2). In common clinical practice this lateral trunk motion toward the transfemoral prosthetic limb (hang over strategy) is attributed to hip muscle weakness and deficient walking agility [81;111]. However, we saw the same motion in AB subjects when using a knee walker prosthetic device. Because these motion did not occurred during normal gait, we doubted whether the lateral trunk motion toward the transfemoral prosthetic limb can actually be attributed to hip muscle weakness and deficient walking agility. While exploring our models, we rationalized that three other possible reasons for this lateral trunk motion can be given. Our first explanation for this lateral trunk motion, which might be too simple and is probably incorrect, is that lateral trunk motion strategy is used to compensate asymmetrical contact times during gait. Two other explanations for the lateral trunk motion, that seem to be more valid, are the power strategy and balance strategy.

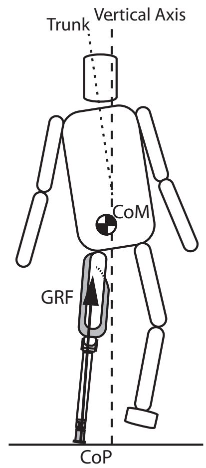


Figure 6.2: Trunk rotation in TF subjects

Asymmetrical contact times

A 2D mathematical model[82] we created to investigate the lateral trunk motion showed relations between gait velocity, CoM position, CoP position and contact times in transfemoral amputees (figure 6.3). The model predicted that a hang over strategy leads to symmetrical contact times. An inverted pendulum model demonstrated that the contact time during

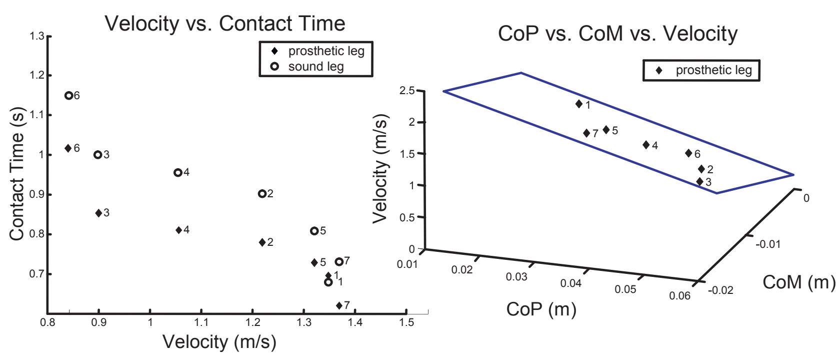


Figure 6.3: Relations between CoM, CoP, velocity and contact time in 7 TF subjects during gait. Notice the asymmetrical contact time between sound limb and prosthetic limb stance.

prosthetic gait is shorter on the prosthetic side, when the prosthetic limb is placed more outward, compared to the sound side. The higher acceleration $\vec{a}$ of the CoM toward the contralateral side is the result of the distance d between the CoP position under the stance foot and the projected CoM (CoM′) position (figure 6.4). The 2D three elements Newton

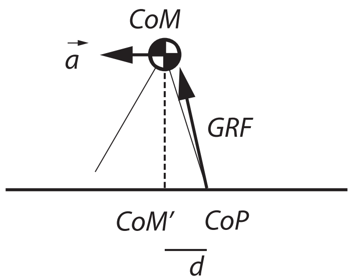


Figure 6.4: Inverted pendulum CoM / CoP relation (front view). Note that the contact time is depending on gait width.

Euler constrained mixed dynamics model (figure 6.5) predicted that trunk sway toward the prosthetic side enlarges the contact time, resulting in more symmetrical contact times (figure 6.6). The predictions made by the model contribute to the theory that TF amputees have a tendency to walk with equal contact times.

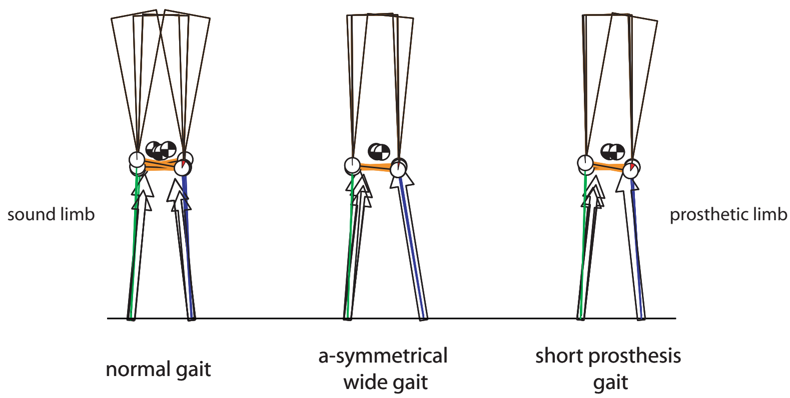


Figure 6.5: Model compensating gait 1(front view). For the corresponding contact time see graphs in figure 6.5.

As stated in the introduction, this first explanation for this lateral trunk motion seems to be too simple and is probably incorrect. If the trunk motion is used to compensate the asymmetrical contact times, then why does the amputee not place his prosthetic limb more inward? The explanation, which is based on the predictions of a mathematical model, and the question which derives from this prediction, should be further investigated in both the real and the conceptual world.

Power and balance strategy

If the asymmetrical contact time explanation would be tested on TF amputees in the real world, by asking the amputees to place their foot more inward, we would probably hear from the amputee that placing the limb more inward, makes him feel less stable. This idea

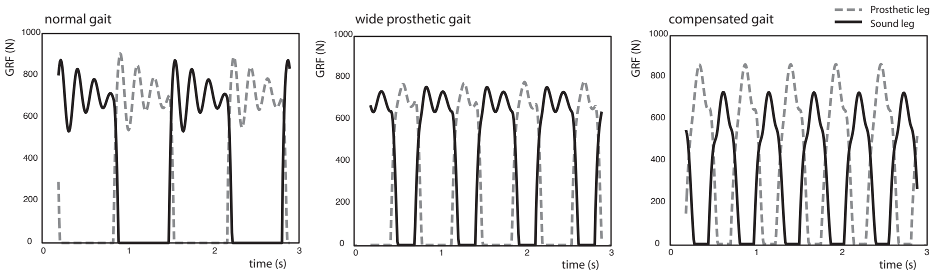


Figure 6.6: Contact time during gait of 2D 3 segment model presented in figure 6.5.

leads to two other explanations for the lateral trunk motion in TF amputees during walking. These explanations derive from a power hypothesis and a balance hypothesis. In these hypothesis, the lateral trunk motion is related to the absence of an active ankle function in the prosthetic limb. The lateral trunk motion is used to increase the GRF to compensate the loss of power in the ankle, or is used to control the position of the CoP under the prosthetic foot to compensate the loss of active balance control in the ankle.

These hypotheses are currently investigated by van Hal et al. in the SPRINT (smart mobility devices with improved patient prosthesis interaction)[112] project, an Innovative Medical Devices Initiative NL, from the Healthy Ageing Network Northern Netherlands (HANNN), which is commissioned by ZonMw, Den Haag. SPRINT contribues to the plans of the high-tech health farm by developing new rehabilitation techniques and devices that restore patient mobility and shift intramural rehabilitation to extramural care. SPRINT includes a unique multi-disciplinary combination of fundamental researchers, applied researchers, health care institutes and industries. This makes it possible to cover the entire chain of innovation, from fundamental research on mobility to market introduction of products. This part of the SPRINT project focuses on the idea that due to loss of active plantar and dorsal flexion of the ankle, TF amputees compensate for the lack of power during push off by actively counterrotating their trunk at the end of the stance phase from their prosthetic limb side towards the sound limb, which increases the GRF. The lack of active balance control at the prosthetic ankle can also be compensated with trunk motions, which are used to maintain the CoM within the base of support area. By placing the prosthetic limb more outward, resulting in a wider gait (figure 6.4), and keeping the CoM$'$ well within the base of this wide support area, a prosthetic limb user can create a safety margin, that will prevent falling over the prosthetic limb in the lateral direction, especially in challenging environments. The end goal of this project is to create a new prosthetic limb that mechanically supports the user optimizing his energy production while maintaining active balance on the prosthetic limb during gait.

6.4.3 Microprocessor controlled lower limb prosthetics

How do our current models relate to the modern microprocessor controlled lower limb prosthetics?

Although we did not simulate microprocessor controlled lower limbs (MCLL), which consist of adaptive microprocessor controlled knees (MPKs) and/or ankles, the current developments in this relatively new area are too important to ignore in this part of the thesis. Since our models showed that the position and motion of the CoP and the orientation of the GRF are very important for both gait and balance with a conventional prosthetic limb, the advantages of the MCLL over the conventional prosthetic limbs indicate that these limbs are of significance in that perspective. Powered MCLL can offer the possibility to steer the motion and position of the CoP and the orientation of the GRF, provided that adequate algoritms algorithms are used. The use of these limbs should be further investigated, amongst others with mathematical models, to improve their functionality which depends on the right choice of hardware and appropriate software algorithms.

Being able to detect events and cycles is essential for mathematical models that are used to control adaptive prosthetic limbs. The algorithms we developed in MATLAB (MathWorks®) were able to detect robustly single events during post processing. The detection algorithms were based on several assumptions and contained many conditions and criteria that had to be met. We were able to detect these events, because we knew, for example, that the force plate data contained a single step in a certain direction. Unfortunately, the current algorithms are not useful in situations in which the gait direction is uncertain, or in which the direction of gait would change during the step. To detect an event in these situations, the algorithms have to be modified by adding more conditions, making them more robust. To detect single events in real time in a large variety of possible motions, not only modifications of our algorithms are necessary. The relatively slow MATLAB programming environment is not suitable for real time single event detection. Therefore, developments are made, and parts of the algorithms are now used in other software environments in which real time evaluation of events is possible.

As stated before, the current developments in the modern microprocessor controlled lower limb prosthetics are too important to ignore. Therefore, some of our thoughts about adaptability, necessities and prosthetic limb control are described below.

Adaptability

Nederhand et al[113] reported that a higher stiffness of a conventional mechanical prosthetic ankle results in better dynamic balance control. In contrast to this finding, Fey et al[114] found that decreasing foot stiffness can increase prosthesis range of motion, mid-stance energy storage and late-stance energy return. However, they also reported that the net contributions to forward propulsion and swing initiation may be limited as additional muscle activity to provide body support becomes necessary. Soares et al[16] stated that rigid feet lead to a fast step from foot strike to toe off, which causes not only changes in the behavior of the prosthetic limb during the stance phase, but also in the sound limb during the swing phase. According to their review, dynamic feet produce different behavior, with increased symmetry between the prosthetic limb and sound limb during the stance and swing phases. This relates to the elasticity of these feet, which gives rise to a more harmonic transition between foot strike and toe-off during the stance phase, since they provide greater range of motion for the prosthetic ankle. Based on these studies and the contradictions found, it appears that it would be preferable to have prosthetic limbs that are adaptive to the environments in which they are used and are able to change their properties on the fly based

on the requirements the users impose.
It seems that MCLL enable the patients to stand and walk symmetrically and improve the functional ability of the TF amputee[115;116;117;118;119;95], which is in line with the wishes of the TF amputee. Ulger et al[120] showed that when TF amputee used a hydraulic knee joint energy consumption decreased, subjects' satisfaction increased and gait was near normal compared to when using their old mechanical knee joint. Fradet et al[121] reported that their findings suggest that the adapted mode of a microprocessor-controlled prosthetic ankle leads to more physiologic kinematics and kinetics in the lower limbs and reported that patients felt safer. Studies investigating powered MCLL reported that using these devices influenced the motions of the prosthetic limb[95], resulting in a more symmetrical motion[117], significantly improving function and balance[116;119] and producing several kinematical characteristics comparable to healthy walking[115] compared to when using the conventional mechanical devices.

Necessities

MCLL, that not all operate in the same manner, ultimately seek to mimic the human anatomical control system (true-to-life system), by incorporating sensor input, processing, output actuation, and feedback input features[122] and accommodate for environmental factors. The prosthetic devices have to combine sufficient dynamic balance control, with adequate energy storage and return capacities and still contribute to propulsion. Devices differ in the ability to accommodate for the various environmental factors and in the extent to which accommodation can be achieved. The resultant output of the device incorporates resistive and/or powered actuation strategies into each move. These conditions require some necessities, which can only be fulfilled by microprocessor controlled combinations of complex mechanical components, hydraulic components and powered components.
One of the necessities is that powered MCLL must have sufficient energy at their disposal to move the limbs. Up until now, the technology is limited by the size and the weight required for a motor and batteries in the prosthesis to provide sufficient net energy. In a recent study, Sup et al.[115] reported that they developed a powered transferoral prosthetic knee and ankle that can provide a range of 12.2 km of level walking and 9.2 km of 5 degrees upslope walking. In terms of steps, these numbers translate to a range of 11.000 to 13.800 steps on a single charge of the 115 Wh battery (weight: 700 g). Healthy individuals who take more than 10.000 steps per day are classified as 'active', are 'highly active' if they make more than 12.500 steps per day[123]. Sup et al. reported that the average power consumption for level walking at self-selected speed of their prosthetic limb was 50 W, with an average net energy delivery by the ankle of 12 J per stride. By comparison, the ankle joint of a similar healthy subject would provide approximately 16 J net energy per stride[124;125]. Sup et al. state that an increased energy delivery with increasing slope correlates with the increased energy requirements of upslope walking, which in turn correspond to increasing the potential energy of the body center of mass as the user walks up the slope. With the information about the subject's mass, the battery energy, the walking distance and the corresponding slope angles provided in the article, and assuming an electrical motor efficiency of 75%, we calculated that the powered prosthetic limb produced less than 15% of the energy that was needed to increase the potential energy. Of course, this is more than what can be produced with a passive prosthesis, which has no ability to deliver net energy, but not as much as what a sound limb would produce. Since the subject was able to walk upslope, we assume that the other 85% is provided by the sound limb and the hip on the prosthetic limb side. Therefore,

we conclude that even with a powered prosthetic limb, there is sound limb dependency. Another necessity is that a robust real time control loop with multiple controllers should be established. MCLL use several strategies to control the limbs. Computational intrinsic control uses sensor information on ambulation, cadence and environment. This form of control has to be combined with interactive extrinsic control, that uses EMG sensors, pattern recognition systems and cortical or peripheral nerve sensors. These two have to be connected to human subjects, who have their own control mechanisms, that use not only mechanical cues but also visual and auditorial cues that are picked up before the information by the MCLL is picked up. Based on our experiences, it seems difficult to establish a robust real time control loop with multiple controllers. Our subjects showed behaviour that we interpret as 'want to be in control'. This feeling of being in control is only possible if the amputees gain extended physiological proprioception [126], similar to a baseball player who has a sense of where the sweet-spot [127] of a 'static' bat is. With 'adaptable' prosthetics where the position and stiffness of the joint changes, there may be limited association of joint position of the prosthesis as it adapts to the environment. To increase confidence in spatial orientation of the prosthesis correlated with the ambulated environments, it seems that very intelligent control, combined with proper biomechanical movement of the prosthesis, is essential.

Control

We agree with Martin et al. [122] that control strategies of the computational intrinsic control or interactive extrinsic control input methods are arguably the most difficult technical barrier for the next generation of prostheses. The movement of the limbs requires precise accommodation for a wide variance of factors, and the ability of the prosthesis to 'think, respond, and react' to environmental changes based on the limited number of sensory and neural inputs is challenging. What makes prosthetics increasingly difficult, when compared with purely robotic systems, is the coupling of man and machine. Although robotic devices have been able to achieve relatively natural bipedal gait, the human factor adds great complexity to the developmental process.
As long as the motions are cyclic, and the deviations from this cyclic motion are limited, the sufficiently fast algorithms, which are the core of the control methods, are able to adjust the properties of the prosthetic limb, to counteract the consequences of these deviations. To counteract deviation, not only real time sensory information is used by the algorithms, but also information from several cycles before the current state. This combination of information is necessary to prevent undesired oscillation between the control of the prosthetic limb and the control of the TF amputee [128]. The consequence is that sudden changes in the current environment are not immediately counteracted. For example, it takes at least one step walking upslope before a computer controlled prosthetic ankle is adjusted to a more dorsal flexion position [115], which eases the upslope walking. Fortunately, although the prosthetic ankle is not adjusted at the first step on the slope, the algorithms seem to be adequate during daily life. Also, for known, not suddenly changing, single events, for example sitting down and standing up [117], appropriate algoritms appear to be available by recognizing patterns in prosthesis sensor data in real time, without the need for instrumentation of the sound-side leg [129]. The Össur® Power Knee™ assists in hese motions by accelerating and decelerating the body's mass, mimicking concentric quadriceps function in a body weighted condition. Data are initially used to train models, which classify the patterns as standing, sitting, or walking [129]. Trained models are subsequently used to infer the user's intent in real time.

For cyclic motions and the known, not suddenly changing, single events appropriate algorithms seem to be available. To our opinion, the biggest challenge now is to develop algorithms that assist TF amputee during unexpected single events, for example stumbling, being pushed or a quick turn at the end of the walkway, in real time. Detection of these events and responding with adequate reactions is very difficult, and seems only possible if multi sensor information is used to reduce false alarm rates [130]. Compensatory stepping strategies should be further investigated, as the control of control of volitional and compensatory limb movements differs in some fundamental ways. Also, visual attention studies should be performed to investigate gaze behaviour in during unfamiliar and complex situations. Information about sudden changes in movements and suddenly redirected gaze will probably have to be part of the algorithms needed, as balance-recovery reaction is typically modulated on the basis of visuospatial environmental information [131].
There is still much work to be done in this area. To our opinion the current state of knowledge in this area is best expressed by the title of the paper by Zhang et al. (2011), mostly because of the first word: *'Towards design of a stumble detection system for artificial legs'*...

6.5 General thoughts

In contrast to Dym (2004), Kent and Franklyn-Miller (2011) suggest that it would be of benefit to develop a uniform, robust modelling strategy for both research and clinical rehabilitation practice. They reported that variation in modelling techniques limits the utility of findings reported in the literature. I do not agree with the idea of a uniform and robust modelling strategy. To my opinion specialised, custom-developed mathematical models can be used to model phenomena at various levels, and may serve different goals. Occam's razer principle states that among competing hypothesis, the one that makes the fewest assumptions should be selected, meaning that one should proceed to simpler theories until simplicity can be traded for greater explanatory power. The simplest available theory need not to be the most accurate. Simple models, which are less accurate, can be used to investigate principles, whereas more complex models, which are more accurate, try to predict a more realistic outcome. The downside of these latter models is that many input parameters and unclear algorithms are required to make realistic predictions, which makes the model less convincing compared to the simple models. As uniform models require multi parameters settings that make the relations within the model unclear, both the research and clinical rehabilitation practice may benefit of a nonuniform modelling strategy, in which specialised, custom developed models are used, that are dedicated and tested to simulate and predict only limited aspects of the observed phenomena. This strategy makes it is easier to study and understand the relation between the identified and included parameters and the predictions made (figure 6.7). Clear relations improve the power of the model. The identified parameters and the clear relations will benefit the clinical practice, provided that the model is wel described.

I strongly feel that making acceptable and valid assertions is only possible if the predictions of a model based on (known) variables and parameters, and of which the assumptions are known, are tested according to the methodological modelling principles as described by Dym (2004). Knowing what is assumed improves the power of the model. Without knowing this,

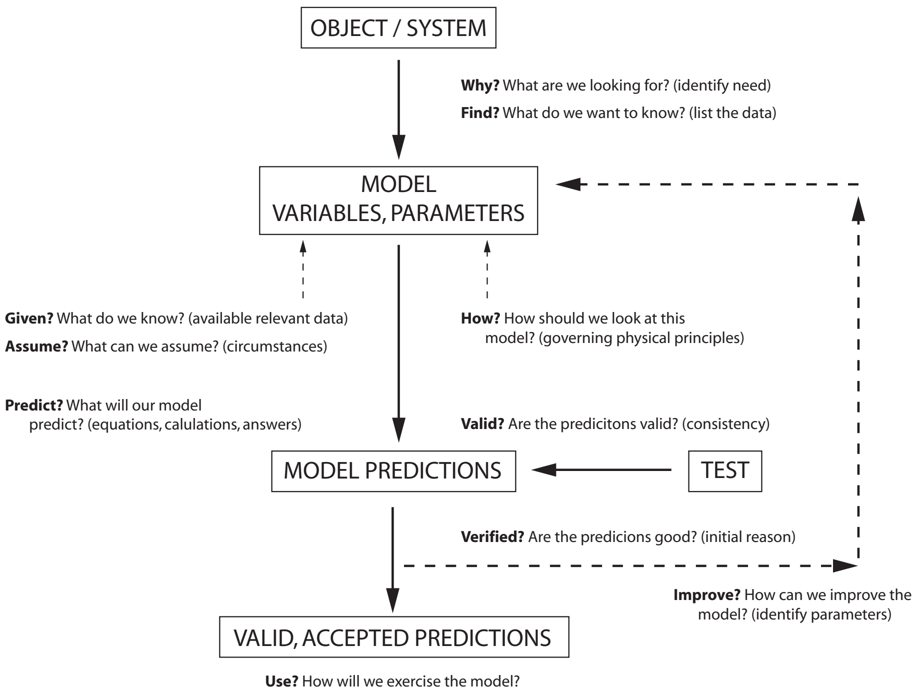


Figure 6.7: A first-order view of *mathematical modeling* that shows how the questions asked in a principled approach to building a model relate to the development of that model (original figure from Dym, 2004), inspired by Carson and Cobelli [132].

the model becomes a black box, of which the circumstances that apply cannot be identified. In our models we made several assumptions that can be debatable. Our models simulate situations in a thin slice of the 3D world, an euphemism for 2D modelling, which therefore has consequences for the predictions made by the models. Compared to our simplified 2D models, with a limited number of body segments and pinpoint joints, 3D models are far more complex, with more parameters and other algoritms. We are aware that the predictions we made with our 2D models might be different when using 3D models, and know that the predictions only apply in one plane. Knowing this, and being aware of it, helps us to understand better what the predictions mean for the strategies chosen by the TF amputees in the real world.
The whole process of methodological modelling is captured by the questions presented in figure 6.7. This flowchart is not an algorithm for building a good mathematical model, but the questions are key to problem formulation generally. The methodological modelling principles show that the predictions of the model have to be validated and verified. In this thesis the validation and verification of the models was done not only using data of TF subjects and AB subjects using a kneewalker in the conceptual world, but also with a physical mechanical Meccano® model as the outcome of one of the mathematical models was a counterintuitive prediction.
When we studied what happened when a prosthetic limb with a flexible knee was moved forward by using a hip flexion moment, we found a remarkable result with the mathematical model. We hypothesized that the foot should move forward when applying hip torques because the whole limb swings forward. However, because of the flexible knee and the inertial properties, the foot moves relatively more in upward direction than in forward direction when applying large hip torques and not using ground friction. To verify this unexpected, counterintuitive prediction we used the mechanical model, to confirm these findings (figure 6.8).

This in upward direction moving of the foot becomes more clear with a heavier foot. Our limb swing model shows the influence of the mass of the foot within only a few minutes of simulation time (figure 6.9). This example demonstrates that mathematical models allow us to gain first insights into changes in movement strategies and prosthetic design. Especially designing, creating and testing of prosthetic limb components can be a costly and lengthy process. Considering this, it is well worth to start this process with exploratory modeling studies.

Although there are data of human subjects in this thesis, the focus of this thesis is to a large extent on the mathematical models. For people who feel that because of the use of these models instead of real patients, of whom we know use many compensation strategies, this thesis is of lesser value, I have the following consideration.
A funny incident that happened may illustrate that working with mathematical models, which are simplified representations of the real world, is not so far off from working with human subjects. Although these mathematical models consist of numerous equations and even more values, these models can show very human-like behaviour. During our search for the optimal trajectory of a prosthetic limb over an obstacle, based on the error values described in chapter 3, I used several computers that were assigned to the calculate all the simulated steps that were necessary to find the solution. This solution would produce a trajectory of the foot over the obstacle, in which the forefoot moved only a few centimeters over the top of the frontside of the obstacle. A fraction of a second later the heel had to

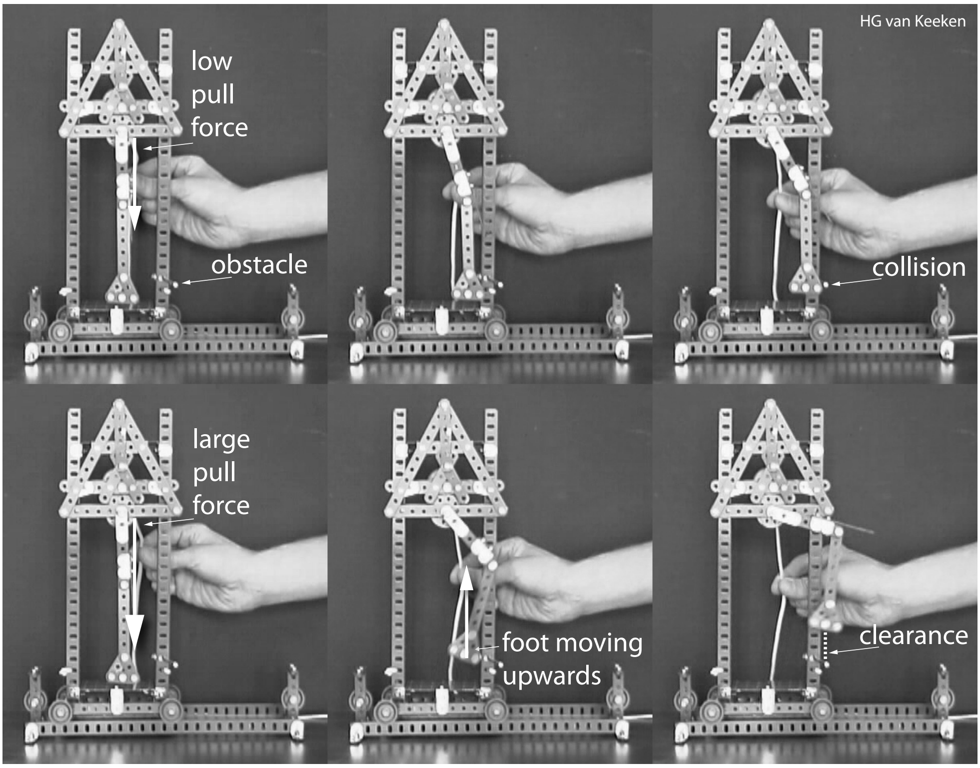


Figure 6.8: Mechanical model used to confirm that the foot moves relatively more in upward direction than in forward direction when applying large hip torques (by pulling the rope with the hand) and not using ground friction (lower image sequence) compared to a low hip torques condition (upper image sequence). This strategy can be used to avoid a collision with an obstacle.

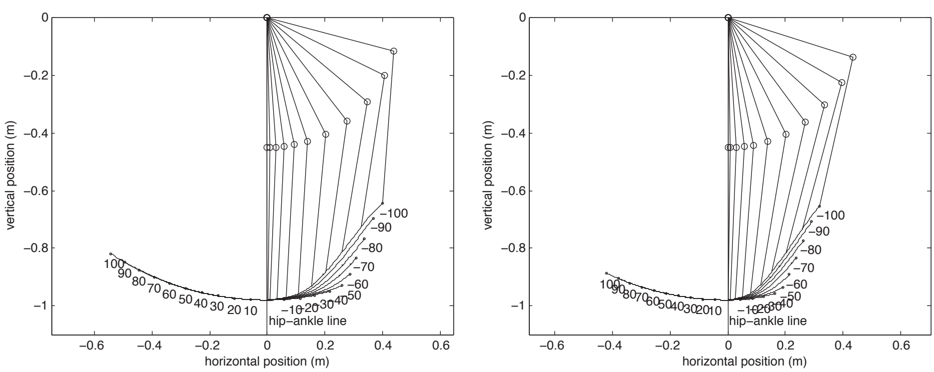


Figure 6.9: Example of the influence of the mass of the foot. The foot in the right panel is twice as heavy as the foot in the left panel.

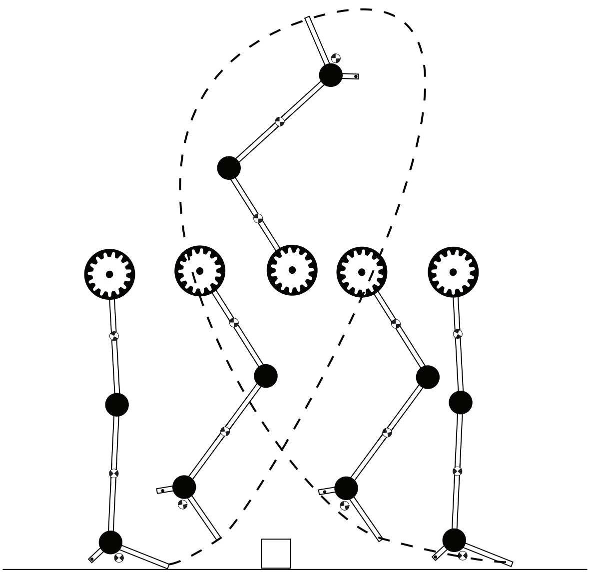


Figure 6.10: Result of 2.5 days simulation with a model that crosses an obstacle based on a optimization algorithm.

pass the top of the backside of the obstacle at a similar distance. Of course the foot was not allowed to collide with the obstacle and the energy consumption should be as little as possible. This procedure, in which all the computers communicated via a server to share their best result, so that the next simulation trial of all the computers would be based on the best solution found until then, took about 2.5 days. I used the weekend so that I could use the computers of my colleagues. When the best solution in a very diverse mathematical landscape was found, the central server sent an e-mail to me, with the input parameters for the model. I entered these parameters in my simulation software at home, on a sunday afternoon, and waited for the outcome, which of course would be perfectly suited to the demands and criteria that were given. Please, have a look at figure 6.10 and imagine my surprise . . .

What have I learned from all of this? The obvious lesson learned was that I had to restrict the motions of the hip. The lesser obvious lesson I learned was that models behave like patients. Analogous to some patients, without the proper instruction, models will come up with solutions to a problem that can be very surprising, without being illogical from a certain perspective, comparable to compensation strategies. Many times I have seen this happening during my work as a physiotherapist. Patients who were asked to avoid an obstacle that was positioned in front of them on the ground decided to walk around it, instead of over it when the environment allowed it. From a certain, probably very functional, perspective they did it exactly as instructed, but not as intended. As a researcher and therapist, I dare to say that models can show very human like behaviour. Your instructions and the learning environment in which the patient moves are key in successful training, teaching and learning!

6.6 General conclusion

The outcomes and insights gained from these studies can be used to predict how TF amputees can compensate for the absence of active control of the prosthetic limb. This absence of active control has consequences for the positioning of the CoP during gait initiation and termination, the active knee extension control during weight bearing on the prosthetic limb and the prosthetic limb swing during the swing phase. The observed real world phenomena and predictions of our models of the conceptual world show that the sound limb dependency is an important strategy during gait initiation and termination, which contributes to patients' functional ability. Also, these findings can be used to predict which strategies contribute to successful prosthetic limb forward swing during obstacle avoidance and what should be taken into account during obstacle avoidance. These models can be of interest when forming new ideas on how to improve prosthetic knees and feet and how to improve rehabilitation programs. The findings were used to increase our insights and contributed to the development of our theory on TF prosthetic gait. Based on the findings in the laboratory, which show that compensation strategies mainly depend on the dexterity of the sound limb, it seems that it is an unattainable goal to pursue symmetry during prosthetic gait.

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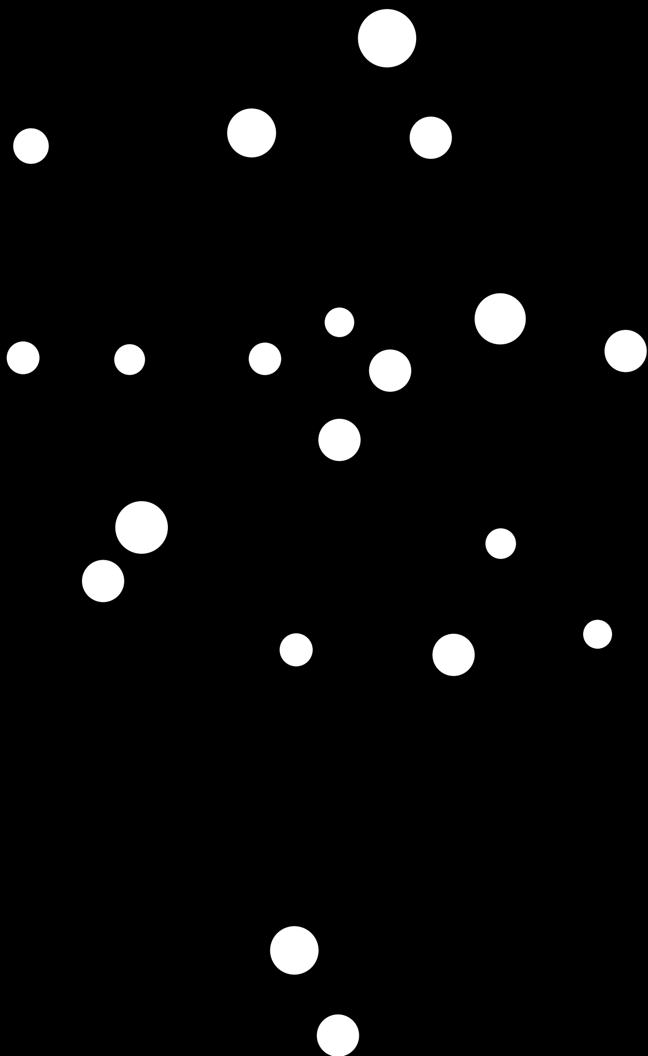

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Summary

Summary

Introduction - In the current thesis, entitled 'Model and measurement studies on stages of prosthetic gait', we used several two dimensional inverse and forward dynamics mathematical models to investigate principles of transfemoral amputee prosthetic gait. For four specific stages of prosthetic gait, namely gait initiation, weight bearing, prosthetic limb swing and gait termination, mathematical models were designed which allowed us to conceptually analyze phenomena observed in the real world. The outcome of the models were used to make predictions about certain choices in strategies that can be made when using prosthetic limbs. Data of transfemoral amputees and able-bodied subjects using a kneewalker prosthetic device were used to validate the models, which were all checked for (internal) consistency, conservation of energy and unrealistic values. The four stages that were studied are described in separate chapters in this thesis.

Gait initiation - During prosthetic gait initiation transfemoral amputees control the spatial and temporal parameters which modulate the propulsive forces, the positions of the center of pressure and the center of mass. Whether their sound limb or prosthetic limb is leading, the transfemoral amputees reach the same end velocity. We wondered how the center of mass velocity build up is influenced by the differences in propulsive components in the limbs and how the trajectory of the center of pressure differs from the center of pressure trajectory in healthy subjects. Seven transfemoral subjects and eight able-bodied subjects were tested on a force plate and on an eight meter long walkway. On the force plate, they initiated gait two times with their sound limb leading and two times with their prosthetic limb leading. Force data were used to calculate the center of mass velocity curves in horizontal and vertical directions. Gait initiated on the walkway was used to determine the limb preference. We hypothesized that because of the differences in propulsive components, the motions of the center of pressure and the center of mass have to be different, as ankle muscles are used to help generate horizontal ground reaction force components. Also, due to the absence of active ankle function in the prosthetic limb, the vertical center of mass velocity during gait initiation may be different when leading with the prosthetic limb compared to when leading with the sound limb. The data showed that whether the transfemoral subjects initiated gait with their prosthetic limb or with their sound limb, their horizontal end velocity was equal. The subjects compensated the loss of propulsive force under the prosthesis with the sound limb, both when the prosthetic limb was leading and when the sound limb was leading. In the vertical center of mass velocity a tendency for differences between the two conditions was found. When initiating gait with the sound limb, the downward vertical center of mass velocity at the end of the gait initiation was higher compared to when leading with the prosthetic limb. Our subjects used a gait initiation strategy that depended mainly on the active ankle function of the sound limb; therefore they changed the relative durations of the gait initiation anticipatory postural adjustment phase and the step execution phase. Both limbs were controlled in one single system of gait propulsion.
The shape of the center of pressure trajectories, the applied forces and the center of mass velocity curves are described in this chapter.

Weight bearing - In this study, the occurrences of stabilizing and destabilizing external moments of force on a prosthetic knee during stance, in the first steps after gait initiation, in inexperienced users were investigated. Primary aim was to identify the differences in the external moments during gait initiation with the sound limb leading and the prosthetic

limb leading. A prosthetic limb simulator device, with a flexible knee, was used to test able-bodied subjects, with no walking aid experience. Inverse dynamics calculations were performed to calculate the external moments. The subjects learned to control the prosthetic limb within 100 steps, without walking aids, evoking similar patterns of external moments of force during the steps after the gait initiation, either with their sound limb loading or prosthetic limb leading. Critical phases in which a sudden flexion of the knee can occur were found just after heelstrike and just before toe off, in which the external moment of force was close to the internal moment produced by a knee extension aiding spring in the opposite direction.

Prosthetic limb swing - In this study, conditions that enable a prosthetic knee flexion strategy in transfemoral amputee subjects during obstacle avoidance were investigated. This study explored the hip torque principle and the static ground principle as object avoidance strategies. A prosthetic limb simulator device was used to study the influence of applied hip torques and static ground friction on the prosthetic foot trajectory. Inverse dynamics was used to calculate the energy produced by the hip joint. A two-dimensional forward dynamics model was used to investigate the relation between the obstacle-foot distance and the necessary hip torques utilized during obstacle avoidance. The study showed that a prosthetic knee flexion strategy was facilitated by the use of ground friction and by larger active hip torques. This strategy required more energy produced by the hip compared to a knee extension strategy. We conclude that when amputees maintain sufficient distance between the distal tip of the foot and the obstacle during stance, they produce sufficiently high, yet feasible, hip torques and use static ground friction, the amputees satisfy the conditions to enable stepping over an obstacle using a knee flexion strategy.

Gait termination - In this study we investigated how leading limb angles combined with active ankle moments of a sound ankle or passive stiffness of a prosthetic ankle, influence the center of mass velocity during the single limb support phase in gait termination. Also, we studied how the trailing limb velocity influences the center of mass velocity during this phase. We analyzed force plate data from a group of experienced transfermoral amputee subjects using a prosthetic limb, and the outcome from a two dimensional mathematical forward dynamics model. We found that when leading with the sound limb, the subjects came almost to a full stop in the single limb support phase, without the use of the prosthetic limb. When leading with the prosthetic limb, the center of mass deceleration was less in a relatively short single limb support phase, with a fast forward swing of the trailing sound limb. Slowing down the heavier trailing sound limb, compared to the prosthetic limb, results in a relatively larger braking force at the end of the swing phase. The simulations showed that only narrow ranges of leading limb angle and ankle moments could be used to achieve the same center of mass velocities with the mathematical model as the average start and end velocities of the prosthetic limb user. We conclude that users of prosthetic limbs have a narrow range of options for the dynamics variables to achieve a target center of mass velocity. The lack of active control in the passive prosthetic ankle prevents the transfermoral amputee subjects from producing sufficient braking force when terminating gait with the prosthetic limb leading, forcing the subjects to use both limbs as a functional unit, in which the sound limb is mostly responsible for the gait termination.

Discussion and conclusion - In contrast to other researchers, who suggest that it would be of benefit to develop a uniform, robust modelling strategy for both research and clinical rehabilitation practice, it is my opinion that specialised, custom-developed mathematical models can be used to model phenomena that occur in the real world. Based on *Occam's*

razer principle, we used models that were relatively simple with only limited assumptions, to investigate the principles of transfemoral amputee prosthetic gait.
It should be taken into account that when we want to use these models to make predictions about the four phases of gait in for example the clinical setting, one more important step has to be made in the research process. The validity and usability of our models in both the conceptual and real world have to be verified further by (learning) experiments with transfemoral amputees or with able-bodied subjects using a prosthetic limb simulator device in various conditions, with various (microprocessor controlled) prosthetic components and properties in various environments.

The predictions, outcomes and insights gained from our model and measurement studies contributed to the development of our theory about asymmetry, funcional ability and learning, and also formed our ideas about trunk motions and microprocessor controlled limb limbs prosthetics. Based on our findings, we concluded that it is impossible to walk symmetrically with a mechanical prosthetic limb, unless additional efforts are made to compensate for the shortcomings in the prosthetic limb. We expect that improving functional ability, instead of minimizing asymmetry, will contribute to the improvement of the patients satisfaction. According to the principles of Discovery Learning and learning as a function of attentional focus, improving the functional ability can best be achieved by training in environments which enable the transfemoral amputee to find individual optimal performance patterns for complex motor skills.

This thesis is part of a series of theses[8;133;134] *resulting from the project 'Postural control after lower limb amputation; changes in body representation and the recovery in postural control'. The project is the result of a collaboration between the Center for Rehabilitation Medicine of the University Medical Center Groningen, the Netherlands and the Center for Human Movement Sciences of the University Medical Center Groningen, University of Groningen, the Netherlands.*
This integrated approach unites two types of research: research from a clinical science approach and research from an fundamental sciences approach. The clinical research was performed by medical specialist for rehabilitation Aline H. Vrieling. Her thesis (2009) formed the base for the current thesis. Many of the findings that were reported in her thesis were studied from a biomechanics perspective in this second part of the project. Parts of the datasets that were reported in her thesis were also used in the current thesis.

Samenvatting

Samenvatting

Introductie - Voor het proefschrift 'Model and measurement studies on stages of prosthetic gait' hebben we vier specifieke stadia van het lopen met een bovenbeenprothese onderzocht. Biomechanische analyses hebben inzicht gegeven in de principes van het starten met lopen, het gewicht plaatsen op het prothesebeen, het naar voren bewegen van het prothesebeen en het stoppen met lopen. Verschillende twee-dimensionale invers en voorwaarts dynamische wiskundige modellen, in combinatie met bewegingsdata, maakten het mogelijk om fenomenen uit de dagelijkse situatie te analyseren in een conceptuele wereld. De uitkomsten van de modellen zijn gebruikt om voorspellingen te doen over de keuzes die prothesedragers maken tijdens het lopen. Naast de uitkomsten van deze modellen zijn voor de verschillende studies ook kinetische en kinematische data van proefpersonen met en zonder transfemorale amputie geanalyseerd. Deze laatste groep maakte gebruik van een prothesebeen dat onder hun knie werd bevestigd. De data zijn gebruikt voor het onderzoeken van de loopbewegingen met het prothesebeen, alsook voor het valideren van onze modellen. In het proefschrift is iedere onderzochte fase in een eigen hoofdstuk beschreven.

Starten met lopen - Tijdens het starten met lopen gebruiken prothesedragers zowel het gezonde been als het het prothesebeen om in beweging te komen. Het verschil in voortstuwende krachten in deze twee benen en de positie van de aangrijpingspunten beïnvloeden de wijze waarop het lichaamsmassamiddelpunt versnelt. Opvallend genoeg maakt het niet uit of het gezonde been of het prothesebeen als eerste naar voren wordt geplaatst. In beide gevallen is na de eerste stap dezelfde voorwaartse snelheid bereikt. Wij vroegen ons af hoe prothesedragers deze snelheidstoename realiseren en welke verschillen er zijn tussen de spatiële en temporele parameters in beide condities. Om daar achter te komen zijn krachtplaatdata en video-opnamen gebruikt om de snelheid van het lichaams-massamiddelpunt in zowel horizontale als verticale richting te berekenen. Onze conclusie was dat het gebrek aan voortstuwende kracht in het prothesebeen werd gecompenseerd door het gezonde been. Hierbij viel op dat de verticale snelheid, in tegenstelling tot de horizontale snelheid, wel verschillend was in de twee condities. Wanneer werd gestart met het gezonde been als eerste naar voren, was de verticale snelheid van het lichaams-massamiddelpunt hoger. Dit werd veroorzaakt door de valbeweging die gemaakt wordt op het gestrekte prothesebeen. Ook bleek dat de prothesedragers voor het genereren van de voorwaartse snelheid met name gebruik maakten van de actieve mogelijkheden in de gezonde enkel. Hiervoor pasten zij de duur van de twee- en éénbenige fase aan. Tijdens het starten met lopen, met het prothesebeen als eerste naar voren, plaatsten zij het prothesebeen al in een vroeg stadium naar voren, in tegenstelling tot wanneer het gezonde been als eerste naar voren werd bewogen. In dat laatste geval werd de verplaatsing zo lang mogelijk uitgesteld en werd het been pas aan het eind van het starten snel naar voren bewogen. Hierdoor kon lang van de actieve enkelfunctie in het gezonde been gebruik worden gemaakt. De prothesedragers lieten een voorkeur zien voor de strategie waarbij het prothesebeen als eerste naar voren werd geplaatst.

Gewicht plaatsen op het prothesebeen - Tijdens de eerste stappen na het starten met lopen ontstaan buigende en strekkende momenten op het kniegewricht van het prothesebeen. Prothesedragers hebben een voorkeur voor het starten met het prothesebeen als eerste naar voren; daarom onderzochten wij de momenten op de knieprothese tijdens het starten met lopen met het prothesebeen en met het gezonde been als eerste naar voren. Een invers dynamisch model is gebruikt om de momenten op een prothesebeen, met een flexibele knie en strekveer, te analyseren. Dit prothesebeen werd door gezonde proefpersonen onder hun knie gedragen. Deze beginnende gezonde prothesebeengebruikers leerden binnen 100 stappen om hun gewicht op het prothesebeen te plaatsen en daadwerkelijk stappen te maken. De berekeningen lieten zien dat de kritische momenten met name optraden aan het begin en vlak voor het einde van de standfase. In deze fasen creëerden de gebruikers interne momenten die nagenoeg gelijk waren aan het moment van de strekveer in de tegenovergestelde richting. Opvallend was dat de momenten op de protheseknie nauwelijks verschilden tijdens de eerste stappen na het starten met lopen met het prothesebeen of met het gezonde been als eerste naar voren.

Het prothesebeen naar voren bewegen - Het adequaat naar voren bewegen van een prothesebeen is belangrijk tijdens zowel het lopen als het passeren van obstakels. Door prothesedragers wordt veelal een strategie gehanteerd waarbij het prothesebeen gestrekt en naar buiten gedraaid over een obstakel heen wordt bewogen. Deze strategie verraadt onmiddellijk de aanwezigheid van een prothesebeen. Een flexiestrategie, waarbij de knie wordt gebogen, komt meer overeen met een looppatroon zoals we dat zien bij gezonde mensen. Een voordeel van deze laatste strategie is dat het lopen nauwelijks onderbroken wordt tijdens het stappen over een obstakel. In deze studie onderzochten wij hoe prothesedragers grondwrijving en heupmomenten kunnen gebruiken tijdens het passeren van een object met een al dan niet gebogen prothesebeen. Met een invers dynamisch model zijn bewegingen van gezonde proefpersonen die een prothesebeen onder hun knie droegen geanalyseerd. Daarnaast hebben we een voorwaarts dynamisch model gebruikt om de relatie tussen heupmomenten, grondwrijving en prothesebeenbewegingen te voorspellen. Dit model is ook gebruikt om te bepalen wat de minimale afstand tussen de voorvoet en een obstakel kan zijn indien de proefpersoon met een gebogen prothesebeen over het obstakel heen wil stappen. Onze conclusie was dat de kniebuiging tijdens de zwaaifase groter werd indien de prothesedrager grotere heupmomenten produceert en gebruik maakt van de grondwrijving. Helaas kost dat wel meer energie, maar daardoor konden deze prothesedragers tot op een afstand van ongeveer twintig centimeter veilig over een tien centimeter hoog obstakel stappen met een gebogen knie.

Stoppen met lopen - Om tot stilstand te kunnen komen, gebruiken prothesedragers verschillende strategieën. Uit deze studie bleek dat de eigenschappen van de enkel en de hoek waaronder het voorste been werd geplaatst, van invloed waren op de snelheid van het lichaamsmassamiddelpunt gedurende de éénbenige fase van het stil gaan staan. Daarnaast bleek dat deze snelheid werd beïnvloed door de versnellingen en vertragingen van het achterste volgbeen. Wanneer prothesedragers stopten met hun gezonde been als eerste naar voren, kwamen ze nagenoeg tot volledige stilstand tijdens de éénbenige fase, zonder gebruik te maken van het prothesebeen. Wanneer het prothesebeen als eerste naar voren werd gezet, was de vertraging minder en werd het gezonde been snel bijgezet om toch tot stilstand te komen. Het gewicht van het relatief zwaardere gezonde been had daarbij een positieve invloed op de vertraging. De simulaties met een voorwaarts dynamisch model lieten zien dat het vertragen van het lichaamsmassamiddelpunt over een beperkt bereik mogelijk is met de door ons onderzochte combinatie van strategieën. Waarschijnlijk worden ook bewegingen van andere lichaamsdelen gebruikt om tot stilstand te komen. Onze conclusie was namelijk dat slechts een beperkt aantal combinaties van beenhoek, enkelmoment en achterste zwaaibeengebruik de gewenste eindsnelheid opleverden.

Discussie en conclusie - De verschillende twee-dimensionale invers en voorwaarts dynamische wiskundige modellen hebben ons geholpen inzicht te verkrijgen in de vier specifieke

stadia van lopen. Hiervoor zijn, gebaseerd op het principe van '*Occam's razer*', relatief simpele modellen met een beperkt aantal aannames gebruikt. In tegenstelling tot andere onderzoekers, die voorstellen om gebruik te maken van een uniforme en robuuste modelleerstrategie voor zowel onderzoek als de revalidatiepraktijk, ben ik van mening dat in onderzoekssituaties specifiek ontwikkelde wiskundige modellen gebruikt dienen te worden voor het modelleren van fenomenen die optreden in het dagelijks leven. Door het aantal aannames te beperken en duidelijke relaties te modelleren, kunnen fenomenen met grotere overtuigingskracht worden beschreven en beter worden begrepen.
Voordat onze modellen daadwerkelijk in bijvoorbeeld de klinische setting gebruikt kunnen worden, moet nog één belangrijke stap in het onderzoeksproject worden gemaakt. De validiteit en de bruikbaarheid van de modellen moeten in zowel de werkelijke wereld als in de conceptuele wereld verder worden onderzocht. Een extra serie (leer-)experimenten met transfemoraal geamputeerden en gezonde proefpersonen, die gebruik maken van een prothesebeen onder hun knie, zal moeten plaatsvinden. Deze experimenten moeten gedaan worden onder verschillende condities, in verschillende omgevingen en met verschillende (computergestuurde) prothese onderdelen.

De voorspellingen, uitkomsten en inzichten die uit onze model- en meetstudies naar voren zijn gekomen, dragen bij aan onze theorie over asymmetrie, functionele vaardigheid en leren. Daarnaast dragen ze bij aan onze ideeën over rompbewegingen en computergestuurde prothesen. Gebaseerd op de onderzoeksbevindingen, concluderen we dat het onmogelijk is om symmetrisch te lopen met een mechanisch prothesebeen, tenzij extra inspanningen worden verricht om de tekortkomingen in het prothesebeen te compenseren. Wij verwachten dat het verbeteren van de functionele vaardigheid, in plaats van het verminderen van asymmetrie, zal leiden tot een verbetering van de mate van tevredenheid bij de prothesedragers.

Dit proefschrift is onderdeel van een serie proefschriften[8;133;134] *die voortkomt uit het project 'Postural control after lower limb amputation; changes in body representation and the recovery in postural control'. Het project is het resultaat van een samenwerking tussen het Centrum voor Revalidatie van het Universitair Medisch Centrum Groningen en het Centrum voor Bewegingswetenschappen van het Universitair Medisch Centrum Groningen, Rijksuniversiteit Groningen, Nederland.*
Deze geïntegreerde aanpak verenigt twee typen van onderzoek: onderzoek vanuit de klinische benadering en onderzoek vanuit een fundamentele benadering. Het klinisch onderzoek in het eerste deel van het project was uitgevoerd door Aline H. Vrieling, medisch specialiste revalidatie. Haar proefschrift (2009) kent een sterke relatie met het huidige proefschrift. Veel van de bevindingen die gerapporteerd zijn in haar proefschrift, zijn in het tweede deel van het project onderzocht vanuit een biomechanisch perspectief. Gedeelten van de dataset die gebruikt zijn voor haar proefschrift, zijn ook gebruikt voor het huidige proefschrift.

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Expand

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Research Department of Rehabilitation Medicine
Center for Rehabilitation UMCG

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Extremities, Pain and Disability

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EXPAND focuses on two spearheads: research on the conditions and amputations of the extremities with emphasis on body functions and structures, activities and participations, and chronic pain and work participation. EXPAND contributes to Healthy Aging, the focus of the UMCG.

Wetenschappelijk onderzoek afdeling Revalidatiegeneeskunde
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PhD thesis

Predictions on how not to walk symmetrically with a mechanical prosthetic limb

Model and measurement studies on stages of prosthetic gait

Helco G. van Keeken, 2013