Weighing weight

Effect of below-knee prosthetic inertial properties on gait

Ruud Willem Selles

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Weighing weight

Effect of below-knee prosthetic inertial properties on gait

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1. General introduction

1.1. Introduction

Leather straps for suspension, an iron socket with leather inside, an iron knee, a shank and foot with iron cover, an iron hinge between foot and shank and a steel spiral spring to flex the footplate during stance,[74] one of the first 'modern' prostheses, an above-knee prosthesis designed by Ambroise Paré around 1560, was high-tech for its time but considered very heavy: it weighted about 7 kg, depending on the size of the subject (see Figure 1.1). Developments over the centuries have drastically changed this. A typical modern prosthesis may have, for example, a carbon fiber socket, an aircraft aluminum knee and shank, a carbon fiber foot and foam cover. With these materials, an above-knee prosthesis does not need to weight more than 2.5 to 3 kg. However, despite this strong reduction in mass, many professionals working in the field of prosthetics still consider most modern prostheses too heavy and aim for further mass reduction.

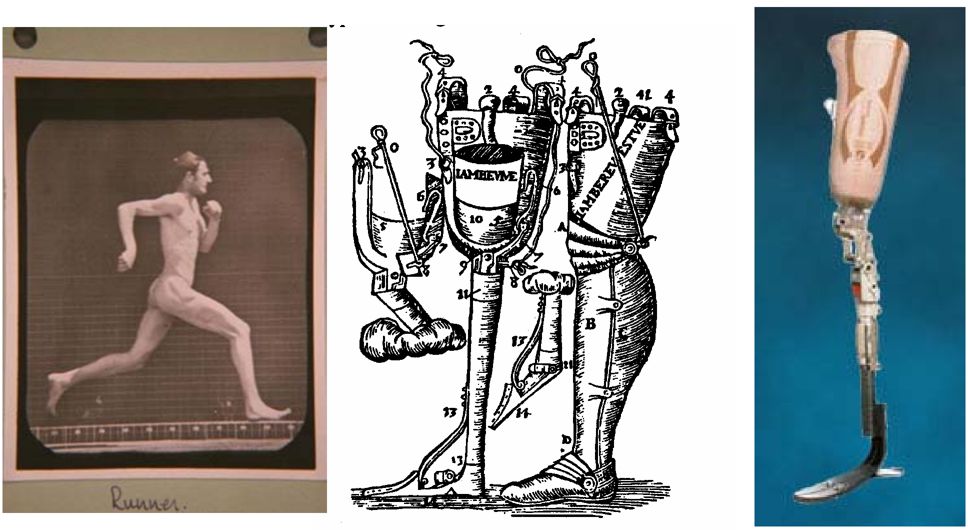

Figure 1.1. A non-amputated leg (picture by Muybridge, 1885, left frame), the above-knee prosthesis developed by Ambroise Paré (± 1560, middle frame) and a modern above-knee prosthesis by Ossur (2002, right frame). The average weight of a non-amputated leg is about 13kg, while the prostheses by Paré and Ossur weigh about 7kg and 3kg, respectively (excluding the mass of the residual limb).

A focus on lightweight design may have several disadvantages. One disadvantage is that lightweight materials can be expensive and may not be as strong, durable or comfortable as needed. In addition, new components may be rejected only because they are considered too heavy. While this was the case in former times in which heavy materials such as steel, wood and leather were used, the issue is still important when using, for example, heavier materials such as silicon. While many people believe that silicon liners provide a more comfortable socket fit, their weight of about 1 kg is assumed to be disadvantageous. The same holds for modern prosthetic knees including, for example, multiple-bar constructions and pneumatic, hydraulic or electronic swing phase controllers.

While these components may be beneficial for the gait of the amputation subjects, they may not be prescribed because of their additional weight.

Although, as far as we know, lightweight design has never been advocated in the present literature, there may be two main arguments for reducing prosthetic mass. The first is a mechanical argument: since a prosthesis is accelerated and decelerated during gait and since it takes less force or torque to accelerate a lighter object, decreasing the weight of an artificial limb will decrease the muscular cost of moving the leg. The second argument is based on the patient's perception of the weight: although almost all of the present prosthetic legs are lighter than the contralateral leg (e.g., [41]), most amputation subjects still report that they find their prosthesis too heavy. When asked, they normally believe that their prosthesis is heavier than their contralateral leg. Thus, while the prosthesis may not be heavy compared to the contralateral leg, it is perceived as such.

Despite the current practice of lightweight design, there are some anecdotes from professionals suggesting that this is still in debate. For example, there is the ‘headwind problem’, saying that subjects do not always like a lightweight prosthesis because it will involuntary flex their prosthetic knee. The ‘winter coat assumption’ (H.J. Stam, personal communication) concerns an analogy to a heavy, old-fashioned winter coat, the weight of which normally doesn’t bother anyone as long as the coat has the right size and ‘design’. The analogy to prosthetic design lies in the fact that a subject complaining about the prosthetic mass may suffer from a bad socket fitting rather than from too much weight. Finally, several professionals have reported to me that subjects sometimes, unexpectedly, consider a heavy prosthesis comfortable, while a lightweight prosthesis may still be perceived as heavy.

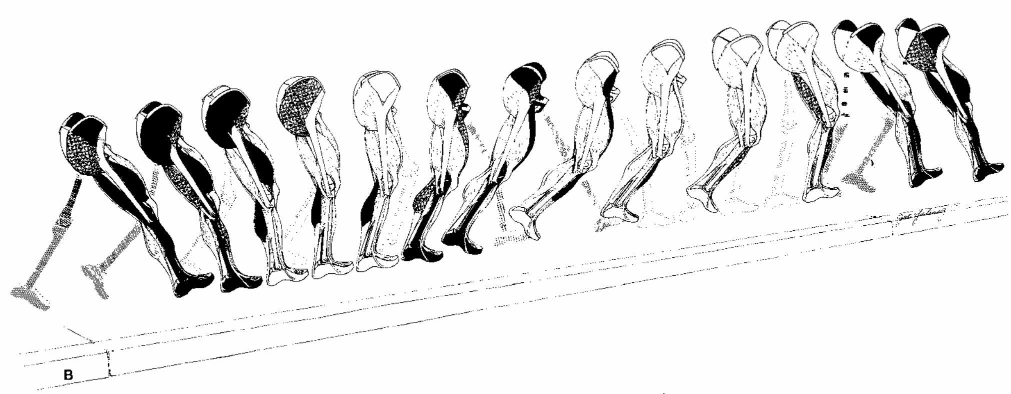

Figure 1.1. Muscle activity during a complete gait cycle. The darker the muscle, the more active the muscle is. Reprinted with permission from Rose and Gamble.[57]

In addition to these anecdotes, there are theoretical arguments to suggest that a lightweight design may not necessarily be desirable. In general, these arguments are based on the idea that during the swing phase the leg should not be understood as actively accelerated and decelerated, but as swinging largely without muscle input. This can be seen in Figure 1.2 in which muscle activity during the gait cycle is visualized. During the swing phase, most muscles are inactive: the leg swings forward like a pendulum. In the literature, this behavior has been referred to as passive (not influenced by muscle activity), pendular (only under the influence of inertial and gravitational forces, such as a pendulum) or ballistic (not regulated after initiation of the movement, such as in a bullet flight). The relation to prosthetic weight is that in a pendular movement, the kinematics of the swing phase are influenced by the inertial characteristics (mass, center of mass location and moment of inertia) of the leg in the same way as the natural frequency of a pendulum is influenced by its inertial properties. Based on the idea of the leg as a pendulum, several authors have suggested that asymmetry in inertial properties may lead to asymmetrical gait, thereby increasing the energetic cost of gait (e.g., [41, 43]). This inertial asymmetry between the prosthetic and contralateral leg in a modern prosthesis may be relatively large, since it can be calculated that the biological counterpart of an above-knee prosthesis weighs about 9 kg in an 80-kg male.[86] In this light, even the prosthesis of Ambroise Paré can be considered lightweight.

The aim of this thesis is to determine the optimal inertial properties of the prosthetic leg on the swing phase of transtibial amputation (TTA) subjects. We will focus only on TTA subjects, disregarding the transfemoral (TFA) and through-knee amputation (TKA) subjects. We expect the influence of mass perturbation to be different in these groups (for discussion, see Chapter 8) and focussed on TTA subjects because most studies reported these subjects to be the largest group of lower limb amputation subjects who, on average, most frequently use a prosthesis.[55, 56, 69] We will focus only on gait, assuming that walking is the most important dynamic activity performed with the prosthesis (e.g., [34, 35]). Within the gait cycle, we will mainly study the swing phase. This assumes that the inertial properties mainly influence the swing phase, during which the leg is moved forward and does not influence the stance phase, during which the prosthetic leg remains fixed on the ground. This assumption is in line with most of the present theoretical models on prosthetic mass which also focus on the swing phase (e.g., [20, 39, 43, 67]; see also Chapter 2)

1.2. Outline of this thesis

First, in **Chapter 2,** the present theoretical models on the relation between prosthetic inertial loading and amputee gait were reviewed. The aim was to summarize the present models and to derive specific predictions from these models on how mass perturbation will affect the gait of amputation subjects. The predictions were tested by systematically reviewing the experimental studies. **Chapter 3** tested the main assumption in all present theoretical approaches to prosthetic mass, that is, that the leg should be understood as ballistically swinging. To that end, four ballistic swing phase models were evaluated in terms of their ability to predict the kinematics of the swing phase in a group of healthy subjects. The same approach was extended in **Chapter 4** to determine the predictive validity of one of the swing phase models, the double pendulum model, in a group of TTA amputees. In this chapter, the inertial properties of stump and prosthesis were compared with the shank and foot of matched controls. We evaluated whether differences in these properties lead to differences in the kinematics of the double pendulum model and in the experimentally measured data. Adaptation strategies of TTA subjects to mass perturbation are evaluated in **Chapter 5**. It is hypothesized that there are two extreme adaptation strategies to mass perturbations possible: (1) a kinetical invariance strategy in which kinematics (joint angles) change while kinetics (joint torques) remain the same, or (2) a kinematical invariance strategy in which kinetics change while kinematics remain the same. To determine which strategy best describes the adaptations used, the effect of five mass conditions on the kinematics and kinetics of the swing phase were evaluated and compared to simulations predicting the effect of each strategy. Based on the adaptation strategy found in Chapter 5, in **Chapter 6**, the effect of a systematic set of mass perturbations was simulated for ten TTA subjects. For each subject, we determined which mass perturbations were beneficial in terms of muscular cost of the swing phase. In addition, we studied how mass perturbation influenced the estimated forces and torques between socket and stump during the swing phase. The aim of **Chapter 7** was to further analyze the effect of the different mass perturbations simulated in Chapter 6 and to explain the differences found between subjects. In addition, we determined whether body dimensions influence the effect of mass perturbation and walking speed. Finally, in **Chapter 8,** we discuss the main limitations of this study, the implications for studies on ballistic walking and the clinical implications of our findings.

2. Effects of prosthetic mass and mass distribution on kinematics and energetics of prosthetic gait: a systematic review

2.1. Abstract

Objective: To introduce the theoretical models used in literature that describe the relation between prosthetic inertial loading and amputee gait and to derive specific predictions from these models; to systematically review experimental studies on the relation between prosthetic inertial loading and energetics and kinematics of lower-limb prosthetic gait; and to compare the review outcomes with predictions derived from theoretical models. **Data Sources:** Studies selected from Medline and from examining references in the selected Medline publications. **Study Selection:** Theoretical models were selected that are used in the present literature to predict the effects of prosthetic mass and mass distribution on kinematics and energetics of prosthetic gait. Experimental studies were selected that investigate the effects of prosthetic mass or center of mass location on the economy, self-selected walking speed, stride length or stride frequency of lower limb amputee subjects. **Data extraction:** The design and methodological quality was assessed using a checklist of nine criteria. Data on economy, self-selected walking speed, stride frequency and stride length data extracted from the studies selected. **Data synthesis:** The predictions of the theoretical models suggest that inertial loading of the present lightweight prosthesis need not be decreased and may sometimes need to be increased to improve the gait of amputee subjects. The methodological quality of most experimental studies was limited. Review of the experimental studies suggests that the inertial loading of the present lightweight prosthesis need not further be reduced. The discrepancy between theoretical models and experimental findings may be related to both the poor methodological quality of the experiments as well as to the limited predictive value of the existing models.

2.2. Introduction

One objective in the development of lower limb prosthetics is to reduce prosthetic weight and moment of inertia. In the literature, however, there is no consensus on the optimal prosthetic inertial loading. For example, Van de Veen and associates[78] evaluated the influence of prosthetic mass and mass distribution on maximal stump load and concluded that lightweight shoe and prosthetic feet should be used. On the other hand, Donn and coworkers[16] investigated the effect of footwear mass on the left-right symmetry of various swing phase parameters of transtibial amputee (TTA) subjects and found that lightweight shoes do not necessarily provide the most symmetrical gait. In addition, 6 out of 10 TTA subjects preferred a shoe mass heavier than they normally wear.

In the section that follows, we discuss the main theoretical models in the literature that relate prosthetic inertial characteristics to prosthetic gait. These models can be divided into three types, that is, pendulum models, multi-segment models and energy exchange models. All three models have in common the fact that they do not model the leg as a load that has to be accelerated and decelerated, but as a system in which potential and kinetic energy are being exchanged.

2.2.1. Pendulum models

Holt and colleages[25, 26] studied leg swing by modeling the whole leg as a one-segment harmonic oscillator. Because the energy necessary to maintain swinging is minimal at the resonant frequency of the oscillator, they hypothesized that muscle force requirements should be minimal during walking at resonant frequency of the legs. In their 1990 study,[26] resonant frequency of the legs was determined by using the formula

$$\mathrm{f} = \frac{1}{\sqrt{\frac{I}{cmgd}}} \tag{2.1}$$

with f (sec^{-1}) the frequency of the swing, I ($\mathrm{kg{\cdot}m^2}$) the moment of inertia, c a constant, m (kg) the mass, g (m/sec2) the gravitational constant and d (m) the distance between the center of mass and the rotation point. Holt[26] found a strong correlation between the step frequency and resonant frequency in subjects walking at preferred rate with different masses added to their legs when c was equal to 2.

In another study in which stride frequency was systematically manipulated, Holt[25] observed that the resonant frequency of the legs coincided with the minimum in the U-shaped curve of oxygen consumption per meter. The idea of the leg swinging at resonant frequency is supported by electromyography (EMG) studies reporting very little electromyographic activity during walking at self-selected walking speed (SSWS) in the swing leg muscles of healthy subjects.

The formula used by Holt was derived from the formula for the frequency of an ideal pendulum,

$$f = \frac{1}{\sqrt{\frac{L}{g}}} \tag{2.2}$$

in which L (m) is the distance between the rotation point and the pendulum's center of mass and g ($\mathrm{m/s^2}$) the gravitational constant. In this formula, it can be seen that the natural frequency of a pendulum is only determined by the center of mass location. Changing the mass without changing the center of mass location does not change the pendulum's natural frequency.

Several authors have suggested that modeling the lower leg or the whole leg as a one-segment pendulum has important implications for prosthetic design.[2, 10, 16, 20, [21, 23, 24, 67] According to Donn,[16] Tashman and coworkers[67] and Godfrey and colleagues,[21] a prosthesis acts as a pendulum rotating around the knee. They predict therefore that the resonant frequency of a prosthesis decreases when inertial load increases, lowering the natural cadence and walking speed of persons with amputation. According to Holt,[25, 26] the pendulum model predicts that oxygen consumption is minimal when the stride frequency is the same as the natural frequency of the leg. For a through-knee (TK) prosthesis, Tashman[67] calculated that a change in center of mass location to 13cm more proximal resulted in a reduced pendulum swing period of .20 seconds, or 15%.

2.2.2. Multi-segment models

Others researchers[43, 45, 68, 83, 90] have introduced more complex models, in which the swinging leg is not considered a single pendulum but is modeled as a set of interacting linked rigid segments. In general, these biomechanical models can be divided into models in which (1) applied forces and torques are computed from limb kinematics (i.e., inverse dynamics), or (2) kinematics are computed from initial positions and angles in combination with known forces and torques (i.e., forward dynamics). One finding from the multi-segment modeling studies is that the swing phase during walking at comfortable walking speed is, to a large extent, ballistic. For example, Mochon and McMahon[45] have shown that a mathematical model of a body represented by a two-segment swing leg and a one-segment stance leg can predict swing time, ground-reaction forces and angles at comfortable walking speed without using muscle force as model input. Important inputs in the latter model are the segment velocities at toe-off. The same findings were reported by Mena and colleagues[43] who modeled the swing leg as a three-link rigid body representing the thigh, shank and foot. They found an almost normal swing phase pattern when positions and velocities at toe-off, hip trajectory and ankle torque were used as model inputs.

Several authors have used multi-segment models to evaluate the effects of inertial loading in lower limb amputees. Mena[43] studied the effects of systematically varying segment masses and moments of inertia in steps of 10% from the values of healthy subjects, keeping toe-off conditions and hip trajectory constant, and found that trajectories of all segments diverged from the normal pattern as a result. These changes were more pronounced when inertia was decreased instead of increased. For example, a reduction of the body segment inertial properties with 20% from the values of healthy subjects led to a deviation in thigh and shank angles of about 10-15%. A 30% increase in segment inertial properties led to angular deviations of only about 5%. According to Mena[43], this indicates that lightweight prostheses are not desirable. This conclusion was confirmed by Tsai and Mansour,[68] describing a model of the swing phase of transfemoral amputee (TFA) subjects in which the leg is represented by a thigh (stump and socket) and shank (prosthetic shank and foot). They report that in their model, inertial loading had a strong influence on the swing phase of prosthetic gait and that lightweight prostheses resulted in a greater deviation from normal kinematics than more heavyweight designs. In addition, adding mass proximally to the (simulated) prosthesis resulted in more normal kinematics than distally added mass. Beck and Czerniecki[2] developed a two-segment model of the swing phase of a TFA prosthesis in which they manipulated inertial loading and found that a heavy prosthetic shank with the center of mass relatively proximal minimized the hip work needed to produce normal kinematics.

2.2.3. Segment energy models

Another approach towards studying the influence of inertial loading on prosthetic gait focuses on the energy exchange between different body segments. Inman[27] suggested that the energy needed to accelerate the body forward is to a large extent supplied by the swinging leg during its deceleration phase. During push-off, the body is lifted and potential energy increases. At the end of the swing phase, kinetic energy of the swing leg is transferred across the hip, creating an acceleration of the upper trunk. Dillingham and colleagues[14] provided more evidence for Inman's proposition, showing a major forward impulse to the head, arms and trunk by the leg at the end of the swing phase. Several authors have formulated implications of the latter finding for prosthetic design. According to both Inman[27] and Dillingham,[14] in a lightweight prosthesis, less kinetic energy during swing phase can be fed back into the body to provide forward head, arm and trunk velocity. If the prosthesis is too heavy, however, the initiation of the swing requires too much energy. Therefore, the optimum should be a compromise between light and heavy weight designs. This was confirmed in an experimental study by Gitter and coworkers[20] who studied the mechanical work during the swing phase of above-knee amputee gait and reported that an increased energy transfer across the hip balanced the negative effects of increasing the prosthetic inertia. They concluded that adding mass up to 1.34 kg to the center of mass of prostheses does not effect the economy of prosthetic gait.

Modeling the influence of inertial characteristics on energy transfer using a two-segment body, Beck and Czerniecki[2] found a maximal energy transfer into the trunk in a relatively heavy prosthesis with a proximal location of the center of mass at all three walking velocities investigated (1.0, 1.2 and 1.4m/ses).

2.2.4. Model predictions

From these theoretical models, predictions can be derived on how prosthetic inertial loading of lower limb prostheses affects the SSWS, stride frequency, stride length, and economy (oxygen consumption per kilogram body mass per meter) of amputees. The application of pendulum dynamics predicts that when the center of mass location changes to more distal, either stride frequency decreases or, when the subject maintains the same stride frequency, economy increases. Applying this model, no predictions on stride length and SSWS can be made. Simulation studies of multi-segment models have predicted an increased SSWS and gait economy when the inertial loading of lightweight prostheses is increased. Studies on the energy exchange between body segments during gait have suggested that within a certain range of loading conditions, positive and negative effects of inertial loading are balanced. However, the energy exchange studies predict that both strongly decreased as well as strongly increased inertial loading would decrease SSWS and economy. The boundary conditions of the range in which both effects balance still have to be determined.

In this article we will systematically review the present literature to determine which of these models are most strongly supported by empirical evidence. We will focus on two techniques of changing the inertial loading, that is, changing prosthetic mass and changing center of mass location. The effects of inertial loading will be evaluated in terms of SSWS, stride length, stride frequency, and economy.

Table 2.1. Criteria used to assess design and methodological quality of the experimental studies. Abbreviations: TT, transtibial; TK, through-knee; TF, transfemoral; SSWS, self-selected walking speed; SF, stride frequency; SL, stride length.

	Criteria	Scoring
1.	Number of subjects measured	Number
2.	Level of amputation of the subjects	TT/TK/TF
3.	Research design used	Experimental/retrospective/ single case
4.	Independent variable investigated	Mass/center of mass
5.	Outcome measures used	SSWS/SF/SL/economy
6.	Did all subjects use the same prosthetic components	Yes/no/not applicable
7.	Were the different trials were randomized	Yes/no/not applicable
8.	Was a statistical analysis performed	Yes/no/not applicable
9.	The time the subjects were given to adapt to changes in the inertial loading of their prosthesis.	Time

2.3. Methods

We performed a Medline search using the Pubmed database between 1966 and November 1998. “Prosthetic design”, “amputees”, “gait” and “artificial limbs” were used as key words. In addition, references in the selected Medline publications were examined.

Studies included in the present study: (1) were written in English; (2) studied the gait of TTA, TK amputee (TKA) or TFA subjects; (3) investigated the influence of changing the prosthetic mass or center of mass location; (4) used economy, SSWS, stride frequency or stride length as outcome variables, or when these variables can be calculated on the basis of the data presented; and (5) were published in a book or journal. Abstracts were not included. It should be noted that changing prosthetic mass can result in a changed center of mass location. Studies investigating the influence of center of mass location were only recognized as such when the prosthetic mass was constant.

The design and methodological quality of the studies was assessed with a checklist of nine criteria (Table 2.1). In this checklist, experimental design is defined as a design in which the researcher manipulates the inertial loading of the prosthesis of the same groups of subjects, whereas in a retrospective design, the researchers compare subjects with different prosthetic inertial loading.

Table 2.2. Assessment of methodological aspects of the experimental studies. Abbreviations: TF, transfemoral; TT, transtibial; TK, through-knee; E, economy; SSWS, self-selected walking speed; SL, stride length; SF, stride frequency.

Reference	**N**	**Level**	**Design**	**Independent variable**	**Outcome measures**	**Components**	**Randomized**	**Statistical analysis**	**Adjustment time**
Czerniecki[10]	8	TF	E	Mass	E, SSWS	Yes	Yes	Yes	One week
Gailey[18]	39	TT	R	Mass	E, SSWS	No	NA	Yes	-
Gailey[17]	10	TT	E	Mass	E	No	Yes	Yes	-
Gitter[20]	8	TF	E	Mass	SL, SF	Yes	Yes	Yes	One week
Godfrey[21]	6	TF	E	Mass	SL	No	Yes	No	-
Hale[23]	10	TF	E	Mass	SSWS, SL,SF	Yes	Yes	Yes	"allowed to practice"
Hillery[24]	1	TT	S C	Mass	SSWS, SL, SF	NA	NA	NA	15 min.
Lehmann[37]	15	TT	E	Mass, CM	E, SSWS	Yes	Yes	Yes	"sufficient"
Skinner and Mote[65]	4	TF	E	Mass, CM	E, SSWS	Yes	No	Yes	-
Tashman[67]	1	TK	S C	CM	SSWS, SL, SF	NA	NA	NA	"a while"

The results of the review are presented in four sections, focussing on the effects of prosthetic mass on the economy and gait pattern (SSWS, stride frequency, and stride length), and the effects of the prosthetic center of mass location on the economy cost and gait pattern.

*Table 2.3. Effects of changes in prosthetic mass on the economy of prosthetic gait. Abbreviations: NS, non-significant; S, significant. * walking speed of 1.26 ms^{-1}.*

Reference (number of subjects)	**Mass condition**	**Economy (l/kg/m)**	**Statistical outcome**
Czerniecki[10] (8)	0 kg added	.187 (.020)	NS
	.68 kg added	.190 (.021)	
	1.34 kg added	.189 (.021)	
Gailey[18] (39)	Light prosthesis (<2.27 kg; average 2.0 kg)	.185	NS
	Heavy prosthesis (>2.27 kg; average 2.7 kg)	.181	
Gailey[17] (10)	0 kg added	.168*	NS
	.454 kg added	.172*	
	.907 kg added	.177*	
Lehmann[37] (15)	Lightweight prosthesis (2.02 kg)	.177 (.35)	NS
	Intermediate weight prosthesis (3.00 kg)	.178 (.29)	
	Heavyweight prosthesis (3.50 kg)	.178 (.32)	
Skinner and Mote[65] (4)	0 kg added		S (p<.05)
	1.70 kg added		
	2.84 kg added		
	3.97 kg added		

2.4. Results

Ten studies met the inclusion criteria. Two of these studies used data from the same experiment.[10, 20] Therefore, redundant information in the second report was excluded in the analysis. Table 2.2 shows the assessment of the methodology items in the different studies. All studies were with adult subjects, except for Tashman[67] who measured the gait of a 13-year-old subject. The paper by Skinner and Mote[65] was a progress report, describing only method and preliminary results. Because of the limited number of studies, we did not make a further distinction between different subject groups (TTA, TKA, and TFA).

2.4.1. *Effects of prosthetic mass on economy*

Czerniecki and colleagues[10] studied the effects of systematically varying prosthetic mass in eight TFA subjects at four different walking speeds. Loads of .68 and 1.34 kg was attached to the center of mass of the prosthetic shank. No significant differences in economy were found between the mass conditions in any of the four different walking speeds used (Table 2.3).

Gailey and colleagues[18] studied the effects of heavier or more lightweight prostheses in a retrospective design in 39 TTA subjects. After controlling for stump length, age, baseline metabolic cost, and walking speed, no statistically significant differences between the groups were found. In another study, Gailey and colleagues[17] investigated the effects of weights of .454 and .907 kg added to shank of the prosthesis of six TTA subjects walking on a treadmill at 1.26 m/sec. No significant differences in economy were found between the different weight conditions.

Lehmann and associates[37] studied the effects of changes in prosthetic mass in 15 TTA subjects. Subjects were instructed to walk at SSWS as well as at 2.0 m/sec using prostheses weighting 2, 3 and 3.5 kg. No significant differences were found between the different mass conditions at both walking speeds. In a preliminary report, Skinner and Mote[65] describe a study in which oxygen consumption is measured at SSWS in four TFA subjects with weights up to 3.97 kg added to their prosthesis. A significant effect of load on economy was reported. It was found that energy use was minimal when 1.70 kg was added to the prosthesis.

2.4.2. *Effects of prosthetic mass on gait pattern*

Swing phase kinematics of ten TFA subjects were studied by Hale and coworkers[23] during three different mass conditions (Table 2.4). They found no significant differences in SSWS, stride frequency, and stride length between walking with light (1.75 kg), intermediate (3.15 kg), or heavy (4.13 kg) prostheses. It was reported that 4 of the 6 subjects preferred the intermediate weight prosthesis while 2 subjects preferred the lightest mass condition.

Studying the effects of adding masses of .68 and 1.34 kg to the center of mass of the prostheses of 10 TFA subjects, Gitter and associates[20] found no statistically significant differences in stride length or stride frequency between the different mass conditions. Gailey and colleagues[18] studied SSWS in 39 TFA subjects wearing prosthesis with different weights. They found that the SSWS of amputees with a heavier prosthesis (2.7 kg on average) was 8% higher than the SSWS of subjects with a lightweight prosthesis (2.0 kg on average); however, this difference was not statistically significant.

Hillery and colleagues[24] assessed gait kinematics of a male TFA subject while masses up to 1.460 kg were added to the distal foot portion of the prosthesis. They found small increases in SSWS and stride length when mass was increased. Stride frequency showed a small decrease with increasing mass. No statistical analysis was performed.

Lehmann[37] studied the effects of changing the prosthetic mass on SSWS in 15 TTA subjects using prostheses weighting 2, 3 and 3.5 kg. No significant differences in SSWS were found between the mass conditions. Godfrey and coworkers[21] studied the effects of attaching weights of .113 and .226 kg to the prosthetic foot of six TFA subjects. No differences in SSWS and stride length were found between the different mass conditions. Skinner and Mote[65] studied the addition of up to 3.97 kg of mass to the prostheses of four TFA subjects and reported no significant influence on SSWS.

2.4.3. Effects of prosthetic center of mass location on metabolic cost

Lehmann[37] changed the center of mass location of TFA prostheses from 47% to 60% distal to the knee without changing prosthetic mass. Two different prosthetic masses were used at both locations while subjects walked at SSWS as well as at 2.0m/sec. Significant decreases in economy were found at both walking speeds and mass conditions in which the center of mass was changed to a more distal position (Table 2.5).

Skinner and Mote[65] studied changes in center of mass location on economy and reported significant effects. They found that of the four mass locations investigated, the two intermediate mass locations (17 and 25 cm distal to the knee) were the most energy efficient.

2.4.4. Effects of prosthetic mass on gait pattern

Tashman[67] built an experimental TKA prosthesis in which the distal mass could be reduced. This enabled changes in distance of the center of mass location of the prosthesis (1.9 kg.) from 31 to 19 cm distal of the knee. Stride parameters of a male 13-year-old TKA were measured. No differences in SSWS, stride length or stride frequency were reported between the different conditions (Table 2.6). When the subject was instructed to walk ‘fast’, walking speed increased by 7% when the mass was placed more distally and a faster cadence and a longer stride length were reported.

Lehmann[37] changed the prosthetic center of mass location of 15 TTA subjects from 47% to 60% distal to the knee without changing prosthetic mass, and found that the SSWS did not significantly change when the center of mass location was altered.

Table 2.4. Effects of changes in prosthetic mass on gait patterns. Abbreviations: SSWS, self-selected walking speed; SL, stride length, SF, stride frequency; Stat, statistical outcome; NS, non-significant.

Reference (number of subjects)	**Mass condition**	**SSWS (m/s)**	**SL (m)**	**SF (1/s)**	**Stat**
Hale[23] (10)	Lightweight prosthesis (1.75kg)	.98 (.23)	1.40 (.21)	.71	NS
	Intermediate weight prosthesis (3.15 kg)	.99 (.22)	1.40 (.20)	.71	
	Heavyweight prosthesis (4.13 kg)	.98 (.23)	1.41 (.20)	.72	
Gitter[20] (8)	0 kg added	1.21 (.09)	1.36	.89	NS
	.68 kg added	1.22 (.06)	1.37	.89	
	1.34 kg added	1.23 (.08)	1.35	.91	
Gailey[18] (39)	Light prosthesis (<2.27 kg; average 2.0 kg)	1.11			NS
	Heavy prosthesis (>2.27 kg; average 2.7 kg)	1.20			
Hillery[24] (1)	0 kg added	1.47 (.03)	1.76 (.020)	.87 (.01)	
	.530 kg added	1.57 (.04)	1.81 (.031)	.87 (.01)	
	1.460 kg added	1.61 (.03)	1.91 (.032)	.84 (.02)	
Lehmann[37] (15)	Lightweight prosthesis (2kg)	1.46			NS
	Intermediate weight prosthesis (kg)	1.46			
	Heavyweight prosthesis (3.50 kg)	1.46			
Godfrey[21] (6)	.113 kg added				NS
	.226 kg added				
Skinner and Mote[65] (4)	0 kg added				NS
	1.70 kg added				
	2.84 kg added				
	3.97 kg added				

Table 2.5. Effects of changes in the center of mass location on the economy of prosthetic gait. Abbreviations: S, significant.

Reference (number of subjects)	Prosthetic center of mass location	Economy (l/kg/m)	Statistical outcome
Lehmann[37] (15)	Intermediate weight, proximal	.178	S(p<.05)
	Intermediate weight, distal	.186	
	Heavy weight, proximal	.178	
	Heavy weight, distal	.192	
Skinner and Mote[65] (4)	Load located 17 cm distal of the knee		S (p<.05)
	Load located 25 cm distal of the knee		
	Load located 33 cm distal of the knee		
	Load located 7 cm proximal of the knee		

2.5. Discussion

The introduction of this study describes three different theoretical models on how prosthetic inertial loading influences prosthetic gait; from these, specific predictions were derived. The pendulum model predicts that when the center of mass location changes to more distal, either the stride frequency decreases or, when the subjects maintain the same stride frequency, the economy (oxygen cost) of walking increases. The simulation studies of multi-segment models predict an increased SSWS and gait economy when the inertial loading of lightweight prostheses is increased. The studies on the energy exchange between body segments during gait have suggest that within a certain range of loading conditions, SSWS and economy are not influenced by inertial loading. The results of the systematic review will be discussed and the different model predictions evaluated.

The methods of most of the experimental studies were relatively poor. Of the 10 studies that meet the inclusion criteria (Table 2.2), two studies had a single case design and one a retrospective design. Of the 7 studies using an experimental design, 5 used uniform prosthetic components in all subjects, 6 performed statistical analysis and 6 were randomized trials. Only 4 out of the 7 experimental studies[10, 20, 23, 37] scored positive on all these items. The number of subjects was limited (ten or less) in most of the studies with an experimental design. We also assessed the time subjects were allowed to adjust to a changed inertial loading, although we are not aware of reports providing evidence for a minimal period that is required for subjects to adjust. Czerniecki[10] and Gitter[20] included a 1-week period for subjects to adjust. Hillery[24] describes a 15-minute adjustment period. All other authors mention more vague terms such as “sufficient”, “a while”, or do not specify a period at all. In addition, it should be noted that prosthetic gait may be influenced by the dynamical behavior of the prosthetic components such as knee and feet. It is not clear from the literature whether the latter aspect may have interacted with changes in inertial characteristics. Although possibly useful, we did not distinguish between TTA, TKA and TFA amputees because of the limited number of studies.

Five studies focussed on the effects of changes in inertial loading on the economy of prosthetic gait. Three of these studies reported no differences between conditions.[10, 17, 18] Lehmann[37] found a decreased economy when the center of mass location was more distal on the shank, but did not find an effect of adding mass at the center of mass location. On the other hand, Skinner and Mote[65] found an increased economy when mass was added to the prosthesis as well as when the center of mass was located more distally. These two studies mainly differed in the subjects used (TTA in the study by Lehmann, TFA in the study of Skinner and Mote), the number of subjects (15 in Lehmann, 4 in Skinner and Mote) and the variation in mass (1.5 kg in Lehmann, 3.97 kg in Skinner and Mote). In addition, the study of Skinner and Mote was only described in a preliminary report in which no data were presented and the research design was only briefly described.

Table 2.6. Effects of changes in the prosthetic center of mass location on gait patterns. Abbreviations: SSWS, self-selected walking speed; SL, stride length, SF, stride frequency; NS, non-significant.

Reference (number of subjects)	Prosthetic center of mass location	SSWS (m/s)	SL (m)	SF (1/s)	Statistical outcome
Tashman[67] (1)	Proximal center of mass location	.95	1.26	.76	NS
	Distal center of mass location	.95	1.26	.75	
Lehmann[37] (15)	Intermediate weight, proximal	1.46			NS
	Intermediate weight, distal	1.47			
	Heavy weight, proximal	1.46			
	Heavy weight, distal	1.46			

Eight studies investigated the effect of changes in inertial loading on SSWS,[10, 18, 20, 23, 24, 37, 65, 67] five on stride length[20, 21, 23, 24, 67] and four on stride frequency.[21, 23, 24, 67] No significant effects were reported. Hillery[24] used a single case design and found an increased SSWS and stride length when mass was added to the prosthesis, but did not perform a statistical analysis. In addition, a non-significant trend towards an increased SSWS with increased prosthetic mass was found by Gailey.[18]

2.5.1. Evaluating model predictions

The pendulum model predicts that when the center of mass location changes to more distal, either the stride frequency decreases or, when the subjects maintain the same stride frequency, the economy of walking increases. Predictions on SSWS or stride length can not be directly derived from the pendulum model.

The predictions of the pendulum model for prosthetic gait were not supported by the outcome of the present study. The effect of changes in the center of mass location on stride frequency was only investigated in a single case study by Tashman.[67] Although the authors calculated that the change in center of mass location resulted in a changed resonant frequency of .20 seconds (15%), no differences in stride frequency were found at comfortable walking speed. The comfortable walking speed also did not change in this experiment. The effect of changes in the center of mass location on walking economy was investigated by Lehmann,[37], who reported a statistically significant decreased economy when weight was located more distally. Skinner and Mote[65], on the other hand, reported a significant increase in economy in the two intermediate mass locations.

Simulation studies of multi-segment models have predicted an increased SSWS and gait economy when the inertial loading of lightweight prostheses is increased[2, 43, 68]. No significant effects were reported to support an increased SSWS, although, in the studies of Gailey[18] and Hillery,[24] a trend towards an increased SSWS with increased inertial loading was reported. Of the five studies investigating the effects of prosthetic inertial loading on economy, three found no differences[10, 17, 18] whereas Lehmann[37] reported an increase, and Skinner and Mote[65] a decrease in economy as the result of an increased inertial loading.

Studies on the energy exchange between body segments during gait have suggested that within a certain range of loading conditions, SSWS and economy are not influenced by inertial loading. These predictions seem to be supported by the findings of the present review. The energy exchange approach also predicts that both strongly decreased as well as strongly increased inertial loading would decrease the SSWS and EC. However, because the boundary conditions have not been specified, it is not possible to test this prediction.

2.5.2. Implications

The present review focuses on SSWS, stride length, stride frequency and economy as outcome measures to determine the influence of prosthetic inertial loading on persons with amputation. These parameters were chosen because they are accepted as important variables and are known to reveal differences between amputee subjects and controls. Moreover, these parameters are most often used in the literature to determine the influence of inertial loading, although other variables, such as gait symmetry, joint angles during walking or patient satisfaction, can also be of interest.

Prostheses have always been designed to reduce inertial loading. Despite this long tradition, we were not able to find a theoretical framework in the present literature supporting this goal. This review found, from both a theoretical and an experimental point of view, that reducing inertial loading of the present lightweight prostheses is not beneficial to the gait of amputee subjects. None of the three theoretical frameworks predict positive effects of reducing inertial loading. In addition, the review of the experimental data did not show a clear benefit of reducing inertial loading.

All three theoretical models include the idea of the leg as a pendulum in which an exchange between potential and kinetic energy occurs. The first two models focus on kinematics, the third one on energetics. Therefore, information in the first two models can be complementary to the third. The pendulum model can be used to make prediction on stride frequency and economy, whereas in the energy exchange model, predictions on SSWS and economy can be made. Only the multi-segment models are able to predict all four variables evaluated in the present review. In addition, some of the multi-segment models can also predict the influence of inertial loading on other variables such as joint-angle characteristics and gait symmetry.

The present review has shown that the pendulum and multi-segment models were not able to predict the outcomes of the experimental research. This may be related to the methodological quality of some of the experimental research as well as to the predictive values of the presented models. The poor methodological quality of most of the experimental research may have resulted in a limited statistical power for detecting actual differences between prosthetic configurations (type II error). Also, in some studies, changes in inertial loading may have been too small to cause detectable differences. In addition, the models discussed are relatively simple, including only some basic aspects of human gait. Some of the models refer to the ‘six determinants of gait’, as formulated by Saunders and assiciates,[58] and include a number of these determinants. Further research should indicate the need for including additional determinants in the models. With respect to model inputs, Tsai,[68] Mena[43] and Beck and Czerniecki[2] all use kinematics obtained from the gait of healthy subjects as reference data. Because of, for example, the lack of active plantar-flexion in amputee subjects may result in different toe-off conditions between amputee subjects and control subjects. A detailed analysis of the gait patterns of amputee persons appears to be necessary to improve the predictive values of these models.

2.6. Conclusion

Different theoretical models were described that predict the influence of inertial loading on prosthetic gait. All models agree on modeling a swinging leg or prosthesis as a system in which potential and kinetic energy are efficiently exchanged. Several predictions were derived from these models. Although our review of the literature did not provide uniform results, it suggests that further reducing inertial loading of the present lightweight prosthesis should not be the main goal of prosthetic design. The empirical findings suggest that, within the range of masses studied, kinematics and energetics of prosthetic gait do not change. This finding, however, may be related to poor methodological quality of the gait studies, decreasing the statistical power for detecting actual differences. Some of the theoretical models, as well as empirical results, suggest that increasing the inertial loading might increase economy of prosthetic gait and change kinematics towards a more normal pattern. Therefore, future research should focus on testing detailed predictions derived from simulation studies to obtain the most optimal inertial loading of prostheses.

3. Comparing predictive validity of four ballistic swing phase models of human walking

3.1. Abstract

It is unclear to what extent ballistic walking models can be used to qualitatively predict the swing phase at comfortable walking speed. Different study findings regarding the accuracy of the predictions of the swing phase kinematics may have been caused by differences in (1) kinematic input, (2) model characteristics (e.g. the number of segments), and (3) evaluation criteria. In the present study, the predictive validity of four ballistic swing phase models was evaluated and compared, that is, (1) the ballistic walking model as originally introduced by Mochon and McMahon, (2) an extended version of this model in which heel-off of the stance leg is added, (3) a double pendulum model, consisting of a two-segment swing leg with a prescribed hip trajectory, and 4) a shank pendulum model consisting of a shank and rigidly attached foot with a prescribed knee trajectory. The predictive validity was evaluated by comparing the outcome of the model simulations with experimentally derived swing phase kinematics of six healthy subjects. In all models, statistically significant differences were found between model output and experimental data. All models underestimated swing time and step length. In addition, statistically significant differences were found between the output of the different models. The present study shows that although qualitative similarities exist between the ballistic models and normal gait at comfortable walking speed, these models cannot adequately predict swing phase kinematics.

3.2. Introduction

In 1980, Mochon and McMahon[44, 45] reported two ballistic models of the swing phase of gait, assuming that lower extremity movements are influenced only by gravity. In models of ballistic walking, muscles act only to establish an initial position and velocity of the legs at the beginning of the swing phase, and then remain inactive throughout the rest of the swing phase.

Ballistic walking models have been proposed to provide a basic understanding of the coordination of walking[8, 44, 49] as well as to establish the determinants of pathological gait in, for example, persons with lower leg amputations.[43, 59] However, to what extent ballistic walking models can be used to quantitatively predict the swing phase of human walking is still undefined. For example, Mochon and McMahon[44, 45] as well as Mena et al.[43] describe qualitatively similar patterns between their ballistic models and kinematic data in subjects walking at comfortable walking speed. However, the simulations reported by Piazza and Delp[50] suggest that without incorporating muscle activity in a ballistic walking model, the swing phase kinematics of normal walking can not be accurately predicted.

The various ballistic walking models differ mainly in the amount of segments included. Both models developed by Mochon and McMahon consist of a double-segment swing leg connected to a one-segment stance leg. In the first model, a point mass, representing the Head, Arms and Trunk (HAT), was connected to the hip while the stance leg rotated around a fixed point on the ground, resembling an inverted pendulum. The second model includes heel-off and knee flexion of the stance leg to better predict ground reaction forces and swing phase kinematics. An alternative to these models is to model the swing leg as a double pendulum without including a stance leg.[33, 43, 83] In the latter model, translation of the hip is prescribed and forward dynamics are applied to calculate how the swing phase would evolve. Donn et al.[16] modeled the movement of only the shank and foot of amputee subjects as a ballistically swinging pendulum to predict the effects of changes in footwear mass on prosthetic gait on the basis of the natural frequency of the pendulum.

It remains unclear to what extent ballistic walking models can accurately predict kinematics of the swing phase of human walking. Most of the reported studies have focused primarily on model development, paying little attention to its predictive validity with regard to the actual movement patterns (kinematics) during swing phase.[8, 33, 40, 44, 45, 48, 49, 83] For example, both Piazza and Delp[50] and Mochon and McMahon[44, 45] use data derived from different literature sources as 1) model input and 2) a reference for their model output. In the study of Mena et al.,[43] the model input is based on single subject data. In addition, it is unclear whether differences in model characteristics, such as the number of segments included, will lead to different simulation outcomes.

The aim of the present study was to evaluate and compare the predictive validity of four ballistic walking models reported in the literature, i.e. (1) the ballistic walking model as originally formulated by Mochon and McMahon (MM model), (2) an extended version of the ballistic walking model in which heel-off of the stance leg is added (MMH model), (3) a double pendulum model, consisting of a two-segment swing leg with a prescribed hip trajectory (DP model), and (4) a shank pendulum model, consisting of only a shank with a prescribed knee trajectory and a rigidly attached foot (SP model). It was hypothesized that, (1) adding heel-off to the MM model (the MMH model) will increase the predictive validity of the model, (2) the DP model will more accurately predict the recorded data than the MM model, since its hip trajectory is prescribed, and (3) the SP model will better predict the recorded kinematics than the DP model, since its knee trajectory is prescribed. The predictive validity of models is evaluated as the extent to which subject-specific ballistic models can accurately describe the kinematics of the swing phase of gait.

3.3. Methods

Six subjects (three males, three females, age range 21 to 54 yrs) participated in the study. No subjects had any cardiopulmonary, neurological or orthopedic disorders that may influence their walking. Body length, body mass and lengths of thigh, shank and foot were measured in all subjects. These data were used to calculate mass, center of mass location and moment of inertia of the different segments.[86]

The assessment was carried out in a gymnasium on a 15-meter straight course. The subjects passed a light gate before entering the measurement field of an opto-electronic system, left that field and passed a second light gate. Distance between the light gates was 6m, whereas the distance covered by the camera was about 3m. All subjects walked at comfortable walking speed and all completed the course eight times. Before the start of the experiment, subjects had three practice trials.

The kinematics of the lower extremities were recorded with a one-camera MacReflex infrared system (Qualisys, Sweden, accuracy ± 2 mm), using reflective markers. Sample frequency of the camera was 50 Hz. Markers were placed according to Winter[86] on the greater trochantor, the lateral femoral condyle, the lateral malleolus and the fifth metatarsal head. On the opposite leg, markers were placed on the medial malleolus and the first metatarsal head. A sensor connected to a portable data recorder (Vitaport 2™) recorded flashes from the light gates.

An interpolation algorithm (MacReflex software) was used to fill marker dropouts. For each subject, six complete gait cycles were selected based on foot contact. Marker trajectories of these six cycles were normalized to the shortest cycles. Then, the trajectories were averaged using Matlab algorithms. In the averaged data, toe-off was selected using the vertical velocity profile of the fifth metatarsal head marker. Walking speed was determined by dividing the distance between the light gates (6m) by the average time taken by the subject to cover the distance. To compare walking speed between subjects, dimensionless walking speed was calculated following Wagenaar and Beek[80] as

$$\widehat{v} = \frac{v}{\sqrt{gh}} \tag{3.1}$$

where v represents walking speed, g acceleration due to gravity, and h leg length defined as the distance between the greater trochanter and the lateral malleolus. For all subjects, step length was calculated as the distance covered by the ankle marker during the swing phase. Swing duration was defined as the time between toe-off and foot contact. The four swing phase models were implemented using automatic dynamic analysis of mechanical systems (ADAMS) software (Mechanical Dynamics Inc). In all models, segment length, mass, center of mass location and moment of inertia were specified based on the anthropometric data obtained from each subject and using regression data described by Winter[86]. The only force specified in all simulations was gravity.

The MM model (Figure 3.1A) consisted of a stance leg, a thigh, a shank, two feet and a HAT. Both feet are rigidly attached to the shank at a 90^{o} angle. The HAT was defined as a point mass rigidly attached to the stance leg at the hip. The hip and the knee joint, as well as the connection between heel and ground are modeled as frictionless revolute joints.

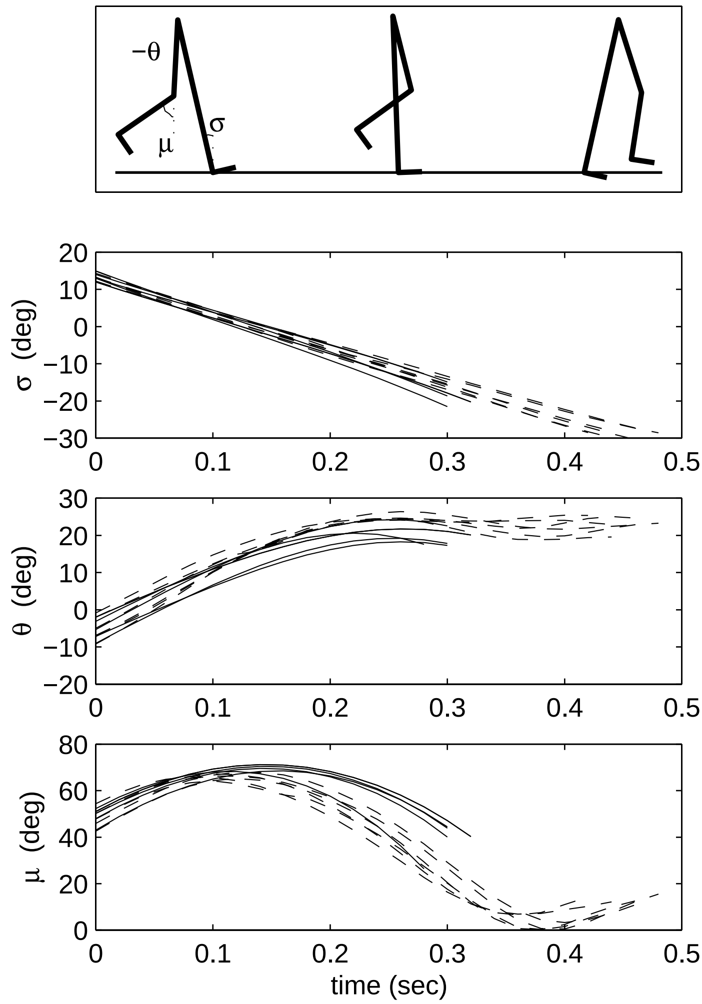

Figure 3.1. Simulation of the MM model. A, stick figure for the model during the simulation. B-D, the simulated (———) and measured (- - -) time series for the angles φ, θ and μ, respectively, for all individual subjects.

The model output was comprised of the angular displacement and velocity between (1) stance leg and vertical axis (σ), (2) thigh and vertical axis (θ), and (3) thigh and shank (μ). From the averaged kinematics, similar data were derived using medial malleolus, greater trochantor, lateral femoral condyle and lateral malleolus markers. The values of the recorded angular displacements and velocities at toe-off were used as model input.

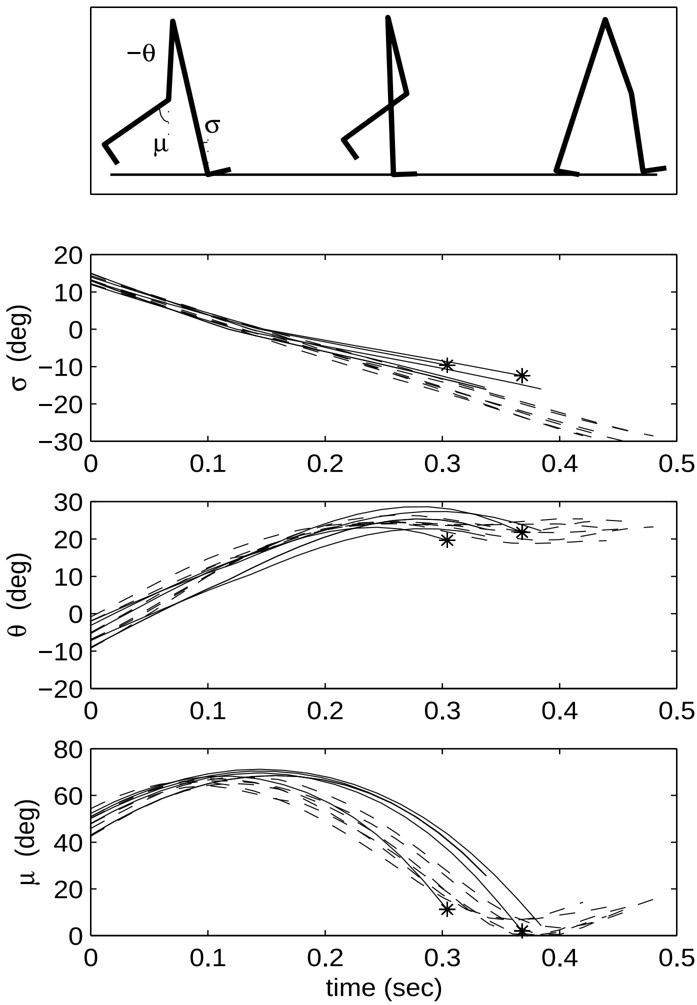

*Figure 3.2. Simulation of the MMH model. A, stick figure for the model during the simulation. B-D, the simulated (———) and measured (- - -) time series for the angles φ, θ and μ, respectively, for all individual subjects. * indicates the trials in which the simulation was stopped because full knee extension (μ=0) was reached.*

The MMH model (Figure 3.2A) was implemented similar to the MM model. In addition, heel-off of the stance leg was added by changing the joint connecting stance leg and ground from the heel location to the toe at the moment the foot is flat on the ground.

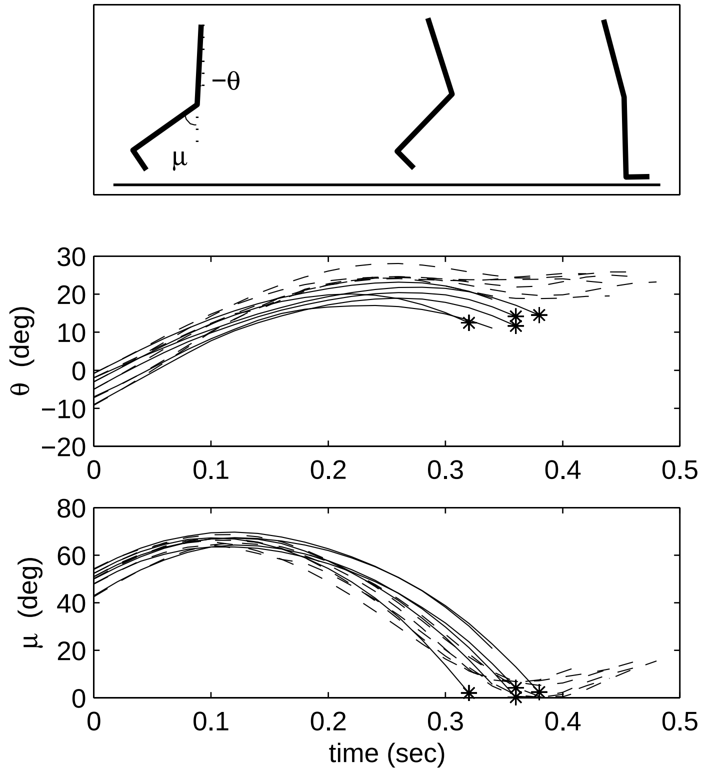

*Figure 3.3. Simulation of the DP model. A, stick figure for the model during the simulation. B and C, the simulated (———) and measured (- - -) time series for the angles ω and μ, respectively, for all individual subjects. * indicates the trials in which the simulation was stopped because full knee extension (μ=0) was reached.*

The DP model (Figure 3.3A) comprised of a thigh connected to the shank by a revolute joint and a foot rigidly attached to the shank. For each subject, the movement of the hip is prescribed, using the recordings from the greater trochantor marker. The model output consisted of the angular displacement and velocity between (1) thigh and vertical axis (θ), and (2) thigh and shank (μ). Similar data were derived from the recordings of the greater trochantor, the lateral femoral condyle and the lateral malleolus markers. Again, values of the angular displacements and velocities at toe-off were used as model input. The SP model consisted of a rigidly attached shank and foot (Figure 3.4A). In the model, the position of the knee is prescribed. The orientation of the model is defined by the angle υ between the shank and the vertical axis. For each subject, the movement of the knee is prescribed, using the recordings from the lateral femoral condyle marker. The angular displacement and velocity at toe-off were used as model input.

Forward-dynamic numerical simulations were performed for each subject separately using ADAMS. In ADAMS, the equations of motions are solved applying a range of different numerical algorithms (i.e., stiff solution methods using implicit, backward difference formulations to solve the algebraic equations, and non-stiff solution methods using explicit formulations to solve the differential equations). Iterative techniques are used to correct or improve upon the predicted solutions. The simulations were stopped when the heel of the swing leg hit the ground or when the knee reached full extension (μ=0). Output of the simulations was created in steps of 0.02 sec.

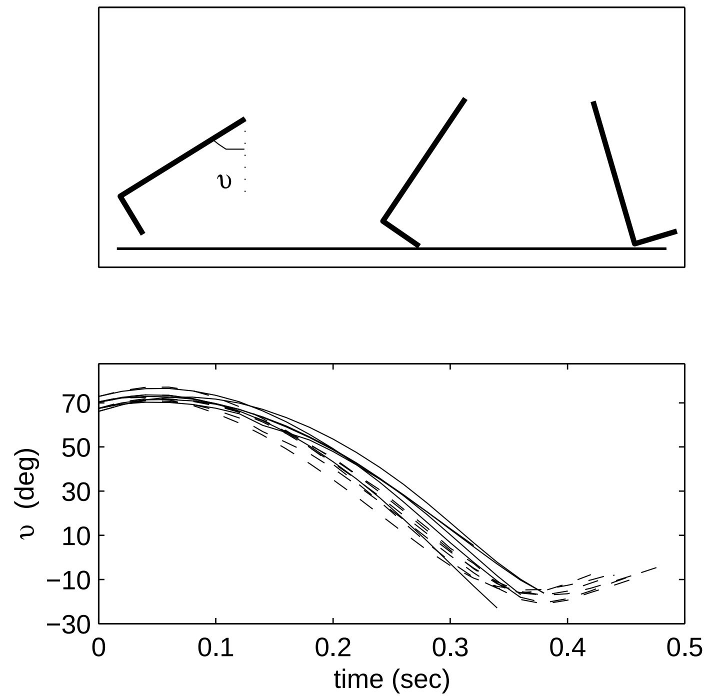

Figure 3.4. Simulation of the SP model. A, stick figure for the model during the simulation. B, the simulated (———) and measured (- - -) time series for the angle υ for all individual subjects.

The simulations generated the above-mentioned angles as a function of time. Since the models applied do not accurately predict swing time (see Results), kinematic data were not normalized for swing time. In addition to the angular time series, step length, swing time, maximum knee flexion and the time after toe-off at which maximum knee flexion occurred (hereafter designated as: time of maximum knee flexion) were calculated for each subject. To compare experimental and simulation time series, average root mean squares (RMS) were calculated from the first 0.3 sec of the swing phase as

$$RMS = \sqrt{\frac{\sum_{i=1}^{n}(x_i - y_i)^2}{n}} \tag{3.2}$$

where x represents the simulation time series, y the experimental time series and n the number of samples of each variable. The measure indicates the average absolute angular difference between both time series. RMS during the first 0.3 sec was chosen because this was the minimal swing phase duration found in all time series.

To statistically test the difference between the time series of experiments and simulations, a Wilcoxon Signed Rank test was used to compare all above-mentioned angles at t=0.3 sec after initiation of the swing phase and to compare maximum knee flexion angle, time of maximum knee flexion, step length and swing time. The same test was applied to compare the RMS of the different models. In all tests, a level of significance of 0.05 was used.

Table 3.1. Characteristics of the subjects, their comfortable walking speed (mean and standard deviation of the different trials) and the dimensionless walking speed.

Subject number	**Age (yrs)**	**Gender**	**Body height (m)**	**Comfortable walking speed (m/s)**	**Dimensionless walking speed**
1	54	Male	1.80	1.40 (0.03)	0.47
2	26	Female	1.61	1.10 (0.06)	0.41
3	27	Female	1.63	1.29 (0.03)	0.46
4	21	Male	1.88	1.32 (0.04)	0.44
5	28	Male	1.83	1.39 (0.03)	0.48
6	28	Female	1.69	1.22 (0.01)	0.45
Group mean	30.7		1.74	1.29	0.45

3.4. Results

Table 3.1 presents characteristics of the subjects including their absolute and dimensionless comfortable walking speed. The initial angular displacements and velocities at toe-off used as model input are given in Table 3.2.

Table 3.2. Angles and angular velocities at toe-off for each subject derived from the averaged experimental data and used as model input. In addition, group mean and standard deviation (SD) are presented.

Subject number	σ	θ	μ	υ	dσ/dt	dθ/dt	dμ/dt	dυ/dt
1	13	-1	54	55	-108	153	259	106
2	13	-3	52	55	-99	158	270	111
3	15	-7	50	57	-119	137	259	122
4	12	-2	51	53	-102	140	273	133
5	14	-5	48	53	-103	174	285	112
6	13	-9	43	52	-118	166	308	141
Group mean	13	-5	50	54	-108	154	276	121
SD	1.0	3.7	3.8	1.8	8.5	15.3	18.6	13.8

Figure 3.1A shows a stick figure of subject 1 for the MM simulation at the moments of toe-off, mid-stance and foot contact. Compared to the experimental data, the simulation time series revealed a significantly smaller σ, no significantly different θ and a significantly larger μ at t=0.3 sec (Figures 3.1B-D; Tables 3.3 and 3.4). In addition, a significantly larger maximum knee flexion angle and an increased time of maximum knee flexion were found (Table 3.3). Predicted swing time and step length in the MM model were significantly smaller than the experimental data (25% and 33%, respectively; Table 3.3).

Figure 3.2A shows a stick figure of subject 1 for the MMH simulation at the moments of toe-off, mid-stance and foot contact. Compared to the experimental data, the simulation time series revealed a significantly larger σ, no significantly different θ and a significantly larger μ at t=0.3 sec (Figures 3.2B-D; Tables 3.3 and 3.4). In addition, a significantly larger maximum knee flexion angle and an increased time of maximum knee flexion were found (Table 3.3). The MMH model predicted a significantly smaller swing time and step length than the experimental data (14% and 19%, respectively).

The RMS of the time series of σ and θ (Table 3.4) were statistically significantly smaller (both $p=0.03$) for the MM than the MMH model. No significant difference was found for μ ($p=0.75$). The MMH model predictions were significantly better than the MM model for swing time (p=0.03) and step length (p=0.03).

Table 3.3. Angles σ, θ, μ and υ at t=0.3 sec, maximum knee flexion and time after toe-off of maximum knee flexion, swing time and step length in the experiment and simulations. Data are the average group values, the standard deviation and the p-value of the Wilcoxon test comparing the simulation with the experimental data.

	Experiment Mean (SD)	**MM model** Mean (SD), p-value	**MMH Model** Mean(SD), p-value	**DP model** Mean (SD), p-value	**SP model** Mean (SD), p-value
$\sigma_{t=0.3\ sec}$	-16 (1.3)	-18 (2.6), 0.05	-9 (1.9), 0.03	-	-
$\theta_{t=0.3\ sec}$	24 (1.0)	22 (4.0), 0.25	25 (2.7), 0.46	18 (3.1), 0.03	-
$\mu_{t=0.3\ sec}$	21 (4.0)	50 (11.4), 0.03	59 (6.8), 0.03	37 (7.2), 0.17	-
$\upsilon_{t=0.3\ sec}$	3.7 (2.6)	-	-	-	12.2 (3.1), 0.03
Maximum knee flexion (^{o})	66 (1.7)	71 (2.0), 0.05	70 (1.7), 0.05	65 (2.4), 0.14	-
Time of maximum knee flexion (sec)	0.10 (0.02)	0.14 (0.01), 0.04	0.14 (0.02), 0.04	0.10 (0.03), 0.25	-
Swing time (sec)	0.44 (0.02)	0.33 (0.01), 0.03	0.37 (0.02), 0.03	0.35 (0.02), 0.03	0.39 (0.01), 0.03
Step length (m)	1.27 (0.10)	0.85 (0.09), 0.03	1.03 (0.12), 0.03	0.96 (0.08), 0.03	1.16 (0.09), 0.03

Figure 3.3A shows a stick figure of subject 1 for the DP model at toe-off, mid-stance and foot contact. Compared to the experimental data, the simulation time series revealed a significantly smaller θ at t=0.3 sec, whereas the predicted μ was not significantly different at t=0.3 sec (Figures 3.3B,C; Table 3.3 and 3.4). No significant differences were found in the maximum knee flexion angle and the time of maximum knee flexion between the simulation and the experimental data. Swing time and step length were significantly smaller in the DP model than in the experimental data (20% and 24%, respectively; Table 3.3).

Table 3.4. Root Mean Squares (RMS) of the difference between experimental and simulation time series, averaged over the subjects. The standard deviation (SD) indicates the variation between subjects. For abbreviations of models, see text.

Model		**MM**			**MMH**		**DP**		**SP**
Angle	σ	θ	μ	σ	θ	μ	θ	μ	v
RMS	1.30	2.85	15.68	3.22	4.08	15.8	3.56	5.97	8.94
SD	0.43	2.19	6.12	0.65	2.25	5.42	1.68	3.31	2.51

No significant difference in RMS was found between the θ time series of the DP model and the MMH model ($p=0.25$). However, the RMS of μ was significantly smaller in the DP model ($p=0.03$). The predicted swing time was significantly better in the MMH model than in the DP model ($p=0.03$), whereas no significant difference was found in step length ($p=0.115$).

Figure 3.4A shows a stick figure of subject 1 for the SP model at toe-off, mid-stance and foot contact. The simulation time series revealed a significantly larger υ at t=0.3 sec compared to the experimental data (Figures 3.3B,C; Table 3.3). The SP predictions for swing time and step length were significantly smaller than the experimental data (11% and 24%, respectively; Table 3.3).

The SP model predictions for swing time ($p=0.03$) and step length ($p=0.03$) were significantly better than those of the DP model.

3.5. Discussion

That human walking is to a large extent ballistic is currently an appealing topic. Work on 'passive dynamic walking' robots show clear similarities in gait kinematics between human walking and robots.[19, 42, 71] The gait studies by Holt et al.[25] indicate that the step frequency at comfortable walking speed can be predicted from the natural frequency of the leg when modeled as a compound pendulum (see also [81]). In addition, others have reported qualitative similarities between ballistic swing phase models and the kinematics at comfortable walking speed.[43-45] Despite these reports, few systematic comparisons have been made between the outcomes of the models and actual subject data. The present study confirms the qualitative similarities between the ballistic models and normal gait at comfortable walking speed. However, all four ballistic models showed statistically significant differences between the angular time series and the recorded data. In addition, all models underestimated swing time and step length. This indicates that none of the ballistic models is capable of predicting the exact kinematics, length and duration of the swing phase at comfortable walking speed and that additional passive or active torques need to be included to simulate human walking (see Figure 3.5).

This confirms the findings of Piazza et al.[50] that muscle activity needs to be incorporated in forward dynamic models to predict the swing phase kinematics of normal walking.

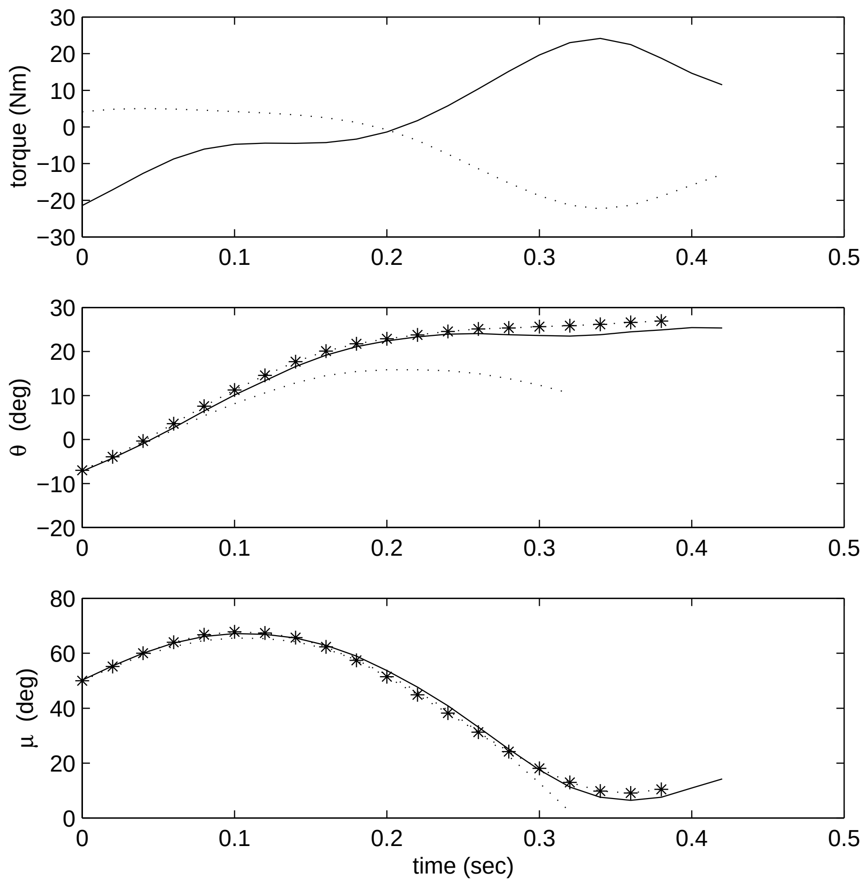

Figure 0.5. (A-C). Example of how torques around hip and knee affect the ballistic swing phase. An inverse dynamics approach was applied to the swing leg using the same segment model as in the DP model. A, the calculated hip- (—) and knee (···) torques. B and C, the experimental time series(—) , the DP model simulation (···) and the 'active simulation' (***) in which the torques around hip and knee were added. Adding the torques to the ballistic model reduced the RMS of θ from 4.9 to 1.3 and μ from 4.3 to 0.9. Adding the same knee torque to the SP model reduced the RMS of υ from 8.9 to 0.8.

Large differences between simulated and experimental data can be seen at the end of the swing phase, suggesting that significant torques are needed to model this part of the swing phase. (Figures 1-4). These differences are in agreement with findings on knee and hip torques during swing phase (Figure 3.5 and Winter, 1991). These torques may be caused by muscle activity at the end of the swing phase (e.g., [53, 87]). However, the work of Riener and Edrich[53] has shown that significant knee and hip torques can result from passive elastic joint and muscle properties before complete knee extension. Adding these torques to the present models will increase the predictive validity of the models in the last part of the swing phase and would delay the early full knee extension found in some subjects in the MMH and DP model. The present study does not allow to determine the relative contribution of passive and active torques to the last part of the swing phase.

The present study shows that the four ballistic models can lead to different model predictions. Although these differences were sometimes relatively small, they were the same for all subjects and, therefore, statistically significant. Three specific hypotheses concerning the outcomes of the different models were tested. First, it was hypothesized that adding heel-off to the MM model (in the MMH model) would increase its predictive validity. While swing time and step length were better predicted in the MMH model, the MM model better predicted two of the three angular time series (Table 3.3 and 3.4). The hypothesis that the DP model would better predict experimental data was confirmed by a better prediction of the knee angle. However, the swing time was better predicted in the MMH model. As expected, the SP model showed a higher predictive validity than the DP models in terms of swing time and step length.

Ballistic walking models have often been referred to as useful for understanding normal and pathological gait. For example, they have been used to investigate the influence of prosthetic mass on amputee gait.[43, 68] Differences in outcomes between the different ballistic models, as above-mentioned, suggest that applying different models to, for example, prosthetic gait may lead to different simulation outcomes. Future studies need to address this problem.

In the literature, a few subject-specific forward dynamic models of human walking have been reported. For example, Riley and Kerrigan[54] developed subject-specific swing phase models for patients with stiff-legged gait and were able to predict the individual differences in the kinematic data. In the present study, small differences were found between subjects in terms of experimental data and model predictions (Table 3.3 and 3.4). However, in one subject, the simulated angles were strikingly more similar to the experimental data (see Figures 3.1-4). This difference may be related to the fact that this subject had the lowest dimensionless walking speed (Table 3.1) and suggests that walking may be most ballistic in subjects walking relatively slow. Future research should investigate the predictive validity of ballistic models at different walking speeds.

Ballistic walking implies a relation between segment inertial characteristics and gait kinematics and suggests that changing inertial characteristics will change the kinematics of human walking. This may have important implications for, amongst others, prosthetic and orthotic design. The findings of the present study indicate that walking at comfortable walking speed is not completely passive. Future research should focus on the relative contribution of inertial properties and muscle activation patterns to gait kinematics.

4. Leg inertial properties in transtibial amputees and control subjects and their influence on the swing phase during gait

4.1. Abstract

Objective: To compare prosthetic leg inertial properties and kinematics and kinetics of the swing phase of transtibial amputation subjects with matched able-bodied controls and to evaluate whether subject-specific double-pendulum models of the swing phase can explain gait differences between both groups. **Design:** The swing phase of transtibial amputation subjects and controls was simulated using a subject-specific double pendulum model based on the individual kinematic data and leg inertial properties. Simulation outcomes were compared to gait analysis data. **Setting:** A gait laboratory. **Subjects:** Ten transtibial amputation subjects and ten matched healthy controls. **Main Outcome Measure(s):** Inertial properties of the lower leg; kinematics and kinetics of the swing phase; kinematics of double-pendulum model simulations. **Results:** In all transtibial amputation subjects, inertial properties were reduced. There were no differences between groups in kinematics, while hip and knee joint torques and powers were reduced in the amputation subjects. Deviations between the double pendulum model and experimental data were larger in the amputation subjects than in the control subjects. **Conclusions:** Current lightweight prostheses are less optimal in terms of their pendular behavior. However, lightweight design leads to smaller joint torques needed to influence the pendular trajectory. Therefore, optimal inertial properties in terms of swing phase kinematics and kinetics will be a compromise between pendular properties and ‘efficient control’.

4.2. Introduction

To date, experimental and modeling studies have not revealed optimal values for the inertial properties (mass, mass distribution and moment of inertia) of lower-limb prostheses (for review, see[59]). Although most designs have aimed at making prostheses as light as possible, many authors (e.g., [2, 16, 20, 21, 67]) have suggested that this may not be beneficial for prosthetic gait because the swing phase is largely pendular or 'ballistic', that is, uninfluenced by joint torques.

A typical example of a pendular system is a one-segment pendulum swinging around a pivot point. In such a pendulum, the inertial properties determine its natural frequency, and thus the kinematics. Therefore, when modeling the legs as a simple pendulum,[25, 26] the movement is influenced by inertial characteristics, but it is not directly clear that the lowest prosthetic mass leads to the most symmetrical or energy efficient gait pattern.

Mena et al.[43] studied a double pendulum model of the lower extremities consisting of an upper and lower leg with prescribed kinematics of the hip joint. They showed that reduced inertial properties of the leg, as in a lightweight prosthesis, lead to abnormal simulation kinematics. Although other modeling studies have suggested the same disadvantage of decreased inertial properties, experimental findings do not uniformly support the idea of either light or heavier prostheses.[59]

Recent studies have questioned the assumption that the swing phase of walking is uninfluenced by muscle activity. For example, in a study on the predictive validity of various ballistic walking models in healthy subjects, Selles et al. [60] reported that although qualitative similarities between the models and normal gait exist, the models can not adequately predict the swing phase kinematics at self-selected walking speed without incorporating joint torques. Whittlesey et al.[84] and Piazza et al.[50] came to the same conclusion studying models of the swing phase of gait.

In the presence of nonzero joint torques, the relationship between inertial properties and kinematics is not as straightforward as in a pendular motion. Whereas in a pendulum changes in inertial properties directly lead to different kinematics, in a movement influenced by joint torques, changes in inertia may be accompanied by adaptations in joint torques and therefore by changes in the 'cost of control' of the swing phase. More specifically, whereas in prosthetic gait reduced inertial properties may be disadvantageously in terms of pendular properties, it may have the advantage of smaller joint torques necessary to impose control.

Although several studies have suggested that pendulum models may be useful for understanding the effect of mass perturbation in lower-limb amputation subjects,[16, 20, 21, 67] the ability of pendulum walking models to predict the exact kinematics of the swing phase of individual transtibial amputation (TTA) subjects has not been studied. The study of Mena et al.[43] suggests that a lighter prosthesis leads to deviations from a normal swing phase. However, the influence of joint torques during the swing phase suggests that studying the swing phase of prosthetic gait from the perspective of a pendulum system alone is not sufficient.

In the present study, lower-limb inertial properties and gait of ten TTA subjects were measured and compared with matched control subjects. We investigated whether (1) the inertial properties of stump and prosthesis in TTA subjects differ from those of shank and foot in matched controls, (2) experimentally derived kinematics and kinetics of the prosthetic leg in the TTA subjects differ from those of the controls, and, if so, whether (3) this can be explained by different kinematics of subject-specific double-pendulum models of the swing phase.

4.3. Methods

4.3.1. Subjects

Ten TTA subjects and ten healthy control subjects participated in the study (Table 4.1). A control subject was matched to each TTA subject for age, height, gender and body mass. All control subjects were free of cardiopulmonary, neurological or orthopedic problems that may influence their walking. TTA subjects were included if they could walk unassistedly for at least five minutes and had no skin problems of the stump. Although not selected as such, all subjects were traumatic amputees. The hospitals' Medical Ethical Commission approved the study and all subjects signed an informed consent.

4.3.2. Measurements

In the TTA subjects, all anthropometric and kinematical measurements were performed on the prosthetic leg; in the control subject, the left leg was used. Body height, mass and the lengths of thigh, shank and foot were measured and used to calculate mass, center of mass location and moment of inertia of all segments[86] except for the TTA subjects' stump and prosthesis. The anthropometric properties of the stump were determined using a geometric model.[31, 89] In addition, the combined prosthetic socket, shank and foot were weighed, balanced on a straight edge to determine the center of mass location, and swung in a pendulum with a small amplitude to determine moment of inertia.[24] Combined stump and prosthesis inertial properties were calculated and normalized to compare TTA subjects and controls. Lower leg mass was normalized to body mass, which was corrected in the TTA subjects for leg amputation. Center of mass and radius of gyration around the knee were normalized for each subject based on the measured segment length.

The assessment was carried out in a gait analysis laboratory on a 15-meter straight track. All subjects walked at comfortable walking speed, wearing their preferred walking shoes and all completed the course eight times. Before start of the experiment, subjects had at least three practice trials.

The kinematics of the lower extremities were recorded with a three-camera ProReflex infrared system (Qualisys, Gothenburg, Sweden) using reflective markers. Sample frequency of the camera was 50 Hz. The camera system covered about three meters in the middle of the track. On the prosthetic leg, markers were placed on the greater trochantor and on the equivalent of the lateral femoral condyle and lateral malleolus locations;[88] in the healthy subjects, the same locations on the left leg were used.

4.3.3. Data analysis

An interpolation algorithm (ProReflex software) was used to fill marker dropouts. A maximum gap of 8 samples was found, occurring in the greater trochantor marker because of arm swing. Marker trajectories were filtered with a second order low pass Butterworth filter with a cut-off frequency of 9 Hz using Matlab algorithms. For each subject, from six recordings, a complete gait cycle was selected based on consecutive foot strikes using the ankle kinematics.[51] Marker trajectories of these six cycles were normalized to the shortest cycle and averaged. The averaged marker trajectories were used to determine walking speed based on the greater trochantor trajectory and stride length and duration using the lateral malleolus marker. In the averaged data, toe-off was selected using the fifth metatarsal head kinematics.[51] Net joint torques around knee and hip during the swing phase were calculated using a linked segment model in which foot and shank together were modeled as a single segment in the controls,[4, 84] and stump and prosthesis as a single segment in the amputation subjects. Net joint power during the swing phase across hip and knee was calculated from the joint torques and angular velocities.

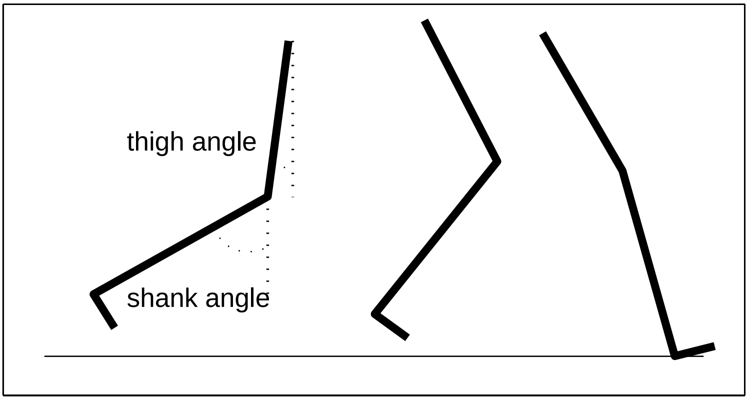

Figure 4.1. Stick figure illustrating the swing phase of subject 1, indicating the definition of the thigh and shank angle.

Two measures were used to quantify the amount of torque and power in hip and knee joint during swing phase for each subject. The 'torque effort' in hip and knee ($TE_{hip+knee}$) was defined as

$$TE_{hip+knee} = \int_{to}^{hs} (|T_{hip}|)dt + \int_{to}^{hs} (|T_{knee}|)dt \qquad (4.1)$$

where T_{hip} and T_{knee} are the net joint torques in hip and knee and to and hs refer to the time of toe-off and foot strike. The mechanical energy efficiency in hip and knee ($MEE_{hip+knee}$) was defined following Zatsiorsky and Gregor,[91]

$$MEE_{hip+knee} = \int_{to}^{hs} (|P_{hip}|)dt + \int_{to}^{hs} (|P_{knee}|)dt \qquad (4.2)$$

where P_{hip} and P_{knee} are the net joint powers. The measures use absolute net torque and power assuming that energy is not recoverable.[7, 32]

4.3.4. Modeling

The Newtonian equations of motion for the forward dynamical (ballistic) double pendulum (DP) model (Figure 4.1) were derived and implemented in Matlab. The model consisted of a thigh connected to the shank by a revolute joint and a foot rigidly attached to the shank. The hip trajectory is prescribed for each subject as a function of time using the recordings from the greater trochantor marker. The orientation of the model is described by two angles, that is, 1) the angle of the thigh with the vertical axis, and 2) the angle of the shank with the vertical axis. The thigh and shank angle and angular velocity at toe-off were derived from the experimental data for each subject and defined the state of the model at toe-off. Forward-dynamic numerical simulations of the equations of motion were performed separately for each subject, using a variable stepsize variable order Adams-Bashford-Moulton integration algorithm. No additional constraints were included in the simulation and each simulation had the same duration as the swing phase of the corresponding subject. Output of the simulations was created in steps of 0.02 s.

*Table 4.1. Subject characteristics and inertial properties. * indicates statistically significant differences ($p<0.05$) between subjects and controls.*

	TTA subjects Mean (SD)	**Controls Mean (SD)**	**p-value**
Age (yrs)	38 (10.4)	35 (12.4)	0.31
Height (m)	1.81 (0.1)	1.81 (0.1)	0.59
Gender (M/F)	8/2	8/2	
Mass (kg)	85 (17)	81 (14)	0.64
Leg & foot mass/body mass	0.047 (0.009)	0.061	0.002*
Leg & foot center of mass from knee/leg length	0.464 (0.071)	0.606	0.001*
Leg & foot radius of gyration around knee/leg length	0.658 (0.066)	0.735	0.005*

4.3.5. Data comparison

For visual presentation as well as to statistically test for differences in kinematic data between TTA subjects and controls, swing time was normalized to 100% using linear interpolation. Groups were compared using a combination of variables derived from these data, that is, values at toe-off, mid swing values (50% of the swing phase), foot strike (100% of the swing phase) and minimum and maximum amplitude.

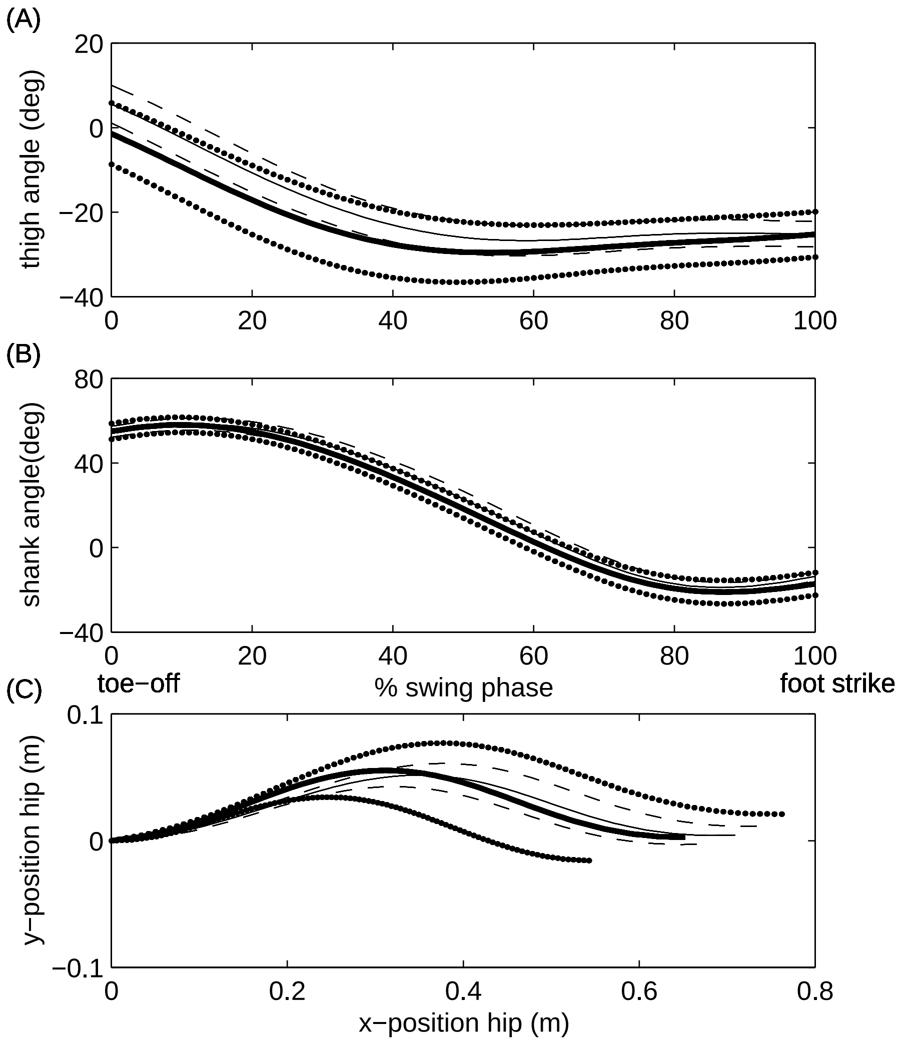

Figure 4.2. Experimental data of thigh angle (A), shank angle (B) and hip trajectory (C) during the swing phase in controls (Mean —, ± 1 SD, - - -) and TTA subjects (Mean, ▬▬, ± 1 SD, ●●●).

Correspondence between simulation and experimental time series were quantified for each subject using root mean squares (RMS):

$$RMS = \sqrt{\frac{\sum_{i=1}^{n}(x_i - y_i)^2}{n}} \quad (4.3)$$

where x and y represent the time series and n the number of values (101) of the normalized swing phase time series.

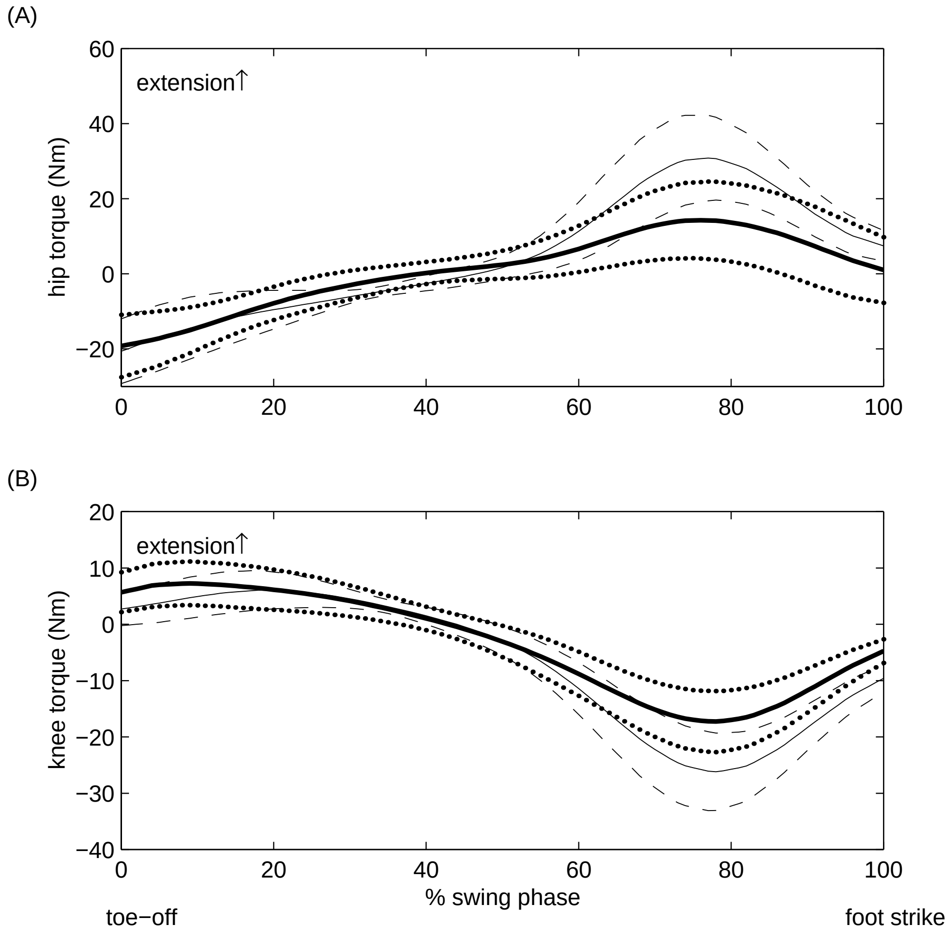

Figure 4.3. Hip (A) and knee (B) torque during the swing phase in controls (Mean —, ± 1 SD, - - -) and TTA subjects (Mean, ▬▬, ± 1 SD, ●●●).

4.3.6. Statistical analysis

Wilcoxon Signed Rank tests were used to compare above-mentioned variables between TTA subjects and controls. Anthropometric data of the TTA subjects were compared to the known values of Winter[86] with a one sample T-test. A p-value smaller than 0.05 was considered significant.

In order to assess the accuracy of the simulations, hip and knee torques obtained from the inverse dynamical analysis were applied to the model to see whether the experimental time series were regained. The absolute difference at the end of the swing phase between simulation and experimental data, averaged over thigh and shank in all subjects, was .0062 degrees. From this we concluded that accumulation of numerical integration error was not a problem in this application.

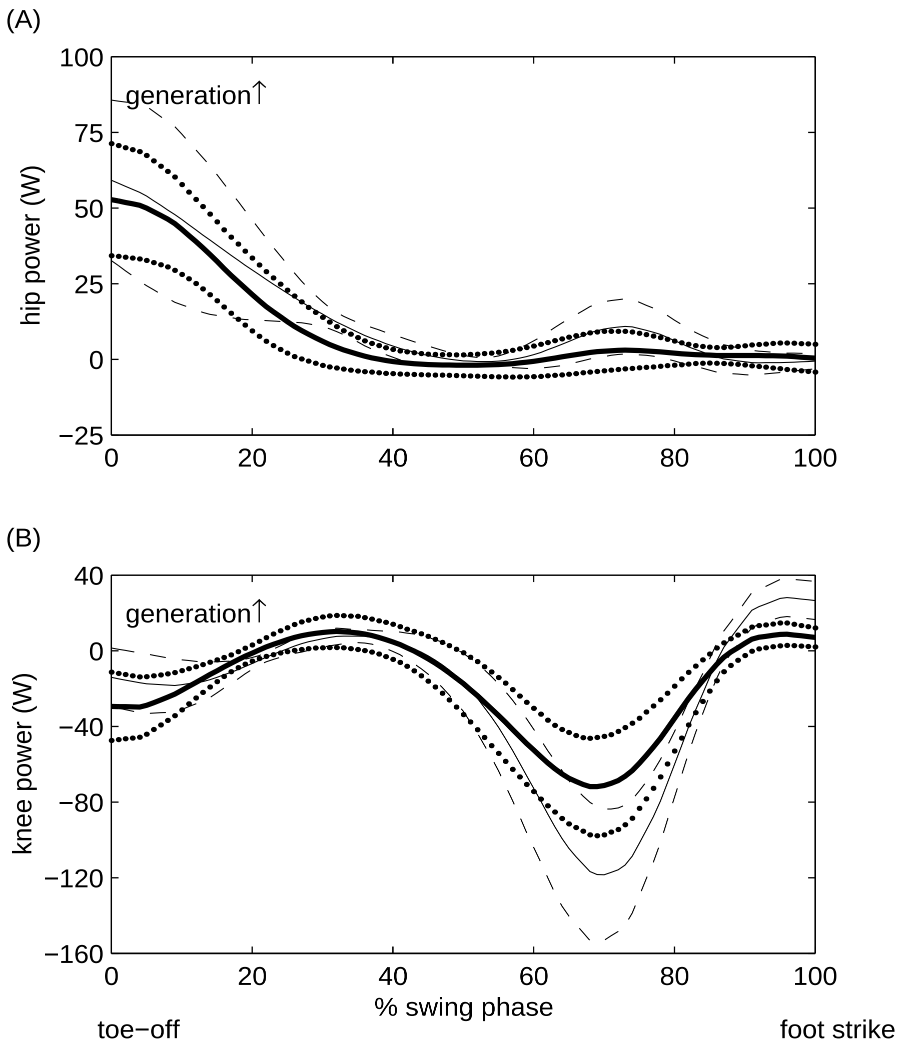

Figure 4.4. Hip (A) and knee (B) power during the swing phase in controls (Mean —, ± 1 SD, - - -) and TTA subjects (Mean, ▬▬▬, *± 1 SD, ●●●). Positive values indicate energy generation.*

4.4. Results

4.4.1. Inertial properties

The characteristics of the study population are given in Table 4.1. All TTA subjects used a silicon liner suspension system and an energy storing foot except for subject 7 using a SACH foot. The normalized lower leg mass (Table 4.1) and radius of gyration around the knee were reduced in the TTA subjects compared to controls by 23 and 10%, respectively. The center of mass location was significantly more proximal (23%) in the subjects than in the controls.

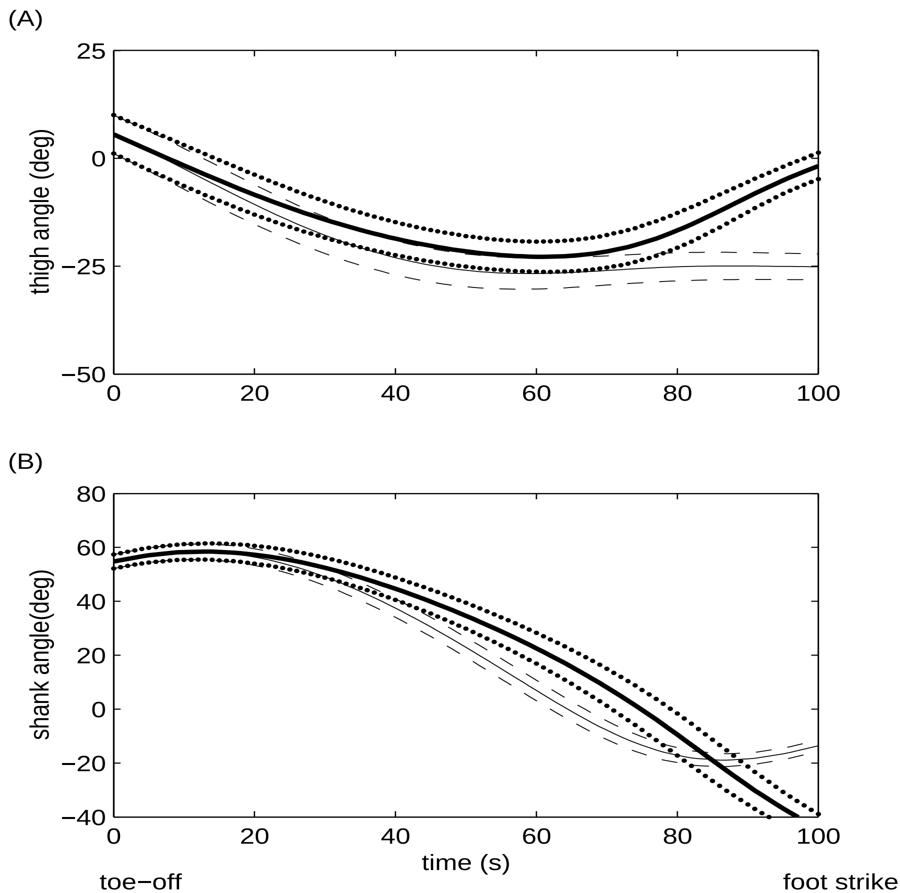

Figure 4.5. Experimental (Mean —, ± 1 SD, - - -) and (ballistic) DP model simulation (Mean, ▬▬▬, ± 1 SD, ●●●) data of thigh (A) and shank (B) angle during the swing phase in the healthy control subjects.

*Table 4.2. Comfortable walking speed, stride duration, stride length and thigh, shank and hip kinematics during swing phase in TTA subjects and controls. Hip kinematics are expressed as the displacement in vertical and horizontal direction after toe-off. * indicates statistically significant differences (p<0.05) between TTA subjects and controls.*

		TTA subjects Mean (SD)	**Controls Mean (SD)**	**p-value**
	Walking speed (m/s)	1.34 (0.24)	1.40 (0.16)	0.59
	Stride duration (s)	0.89 (0.07)	0.92 (0.04)	0.39
	Stride length (m)	1.50 (0.20)	1.61 (0.09)	0.04*
Thigh	Toe-off angle (deg)	-1.4 (7.3)	5.6 (4.5)	0.02*
	Toe-off angular velocity (deg/s)	-162.3 (24.6)	-164.4 (16.5)	0.79
	Max angle (deg)	-1.4 (7.3)	5.6 (4.5)	0.02*
	Min angle (deg)	-30.4 (6.4)	-27.2 (3.6)	0.06
	Angle at mid swing (deg)	-29.4 (7.4)	-26.0 (3.8)	0.09
	Angle at foot strike (deg)	-25.2 (5.3)	-25.1 (3.0)	0.72
Shank	Toe-off angle (deg)	54.8 (2.6)	55.0 (3.7)	0.58
	Toe-off angular velocity (deg/s)	134.6 (21.7)	131.4 (23.1)	0.65
	Max angle (deg)	58.2 (3.5)	58.5 (2.8)	0.45
	Min angle (deg)	-21.2 (5.4)	-19.0 (2.4)	0.29
	Angle at mid swing (deg)	18.3 (4.4)	22.9 (3.7)	0.05
	Angle at foot strike (deg)	-17.2 (5.4)	-13.6 (2.1)	0.11
Vertical hip displacement	Maximal (m)	0.057 (0.02)	0.052 (0.01)	0.50
	mid swing (m)	0.055 (0.02)	0.052 (0.01)	0.59
	foot strike (m)	0.003 (0.02)	0.004 (0.01)	0.80
Horizontal hip displacement	Maximal (m)	0.65 (0.11)	0.71 (0.04)	0.11
	mid swing (m)	0.32 (0.05)	0.35 (0.02)	0.05
	foot strike (m)	0.65 (0.11)	0.71 (0.04)	0.11

4.4.2. Kinematics

There was no significant difference between groups in walking speed and stride duration, whereas there was a significant reduction in stride length in the TTA subjects compared to the controls (Table 4.2). Figure 4.2 shows that the shank and thigh kinematics in subjects and controls were similar. This was confirmed by non-significant differences in most measures derived from the shank and thigh angle. Only the thigh angle at toe-off (Table 4.2) was significantly decreased in the TTA-subject group. The trochantor major displacement after toe-off, used in the DP model to prescribe the hip trajectory, was similar in TTA subjects and controls.

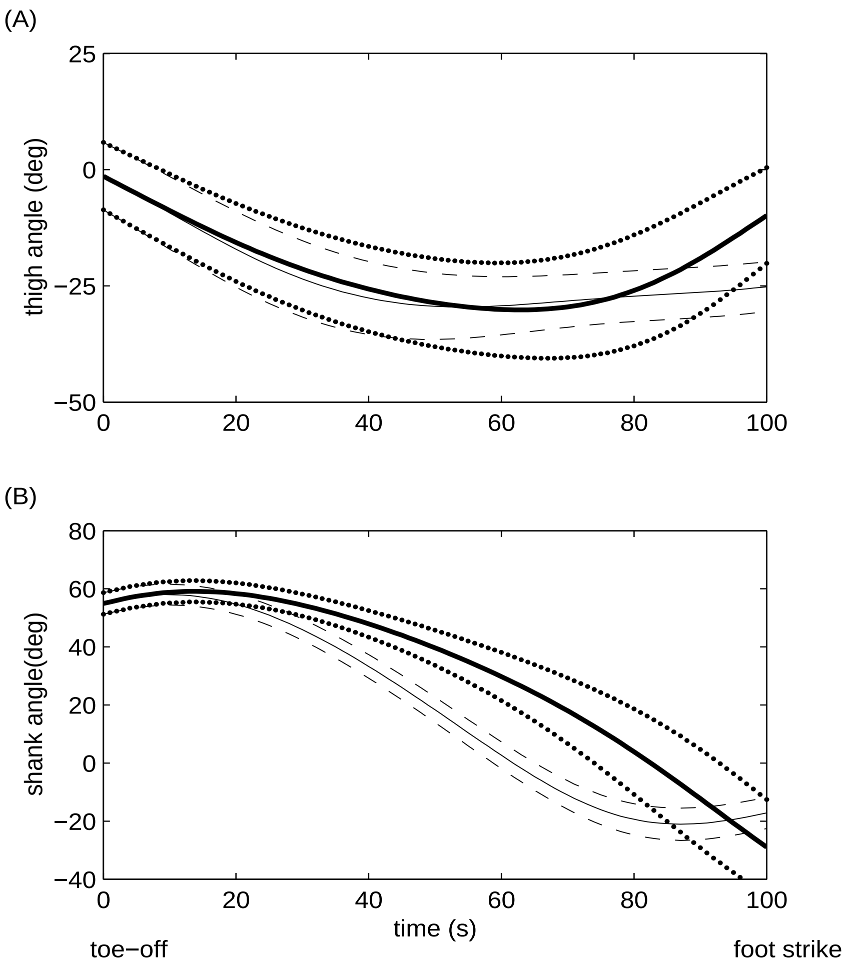

Figure 4.6. Experimental (Mean —, ± 1 SD, - - -) and (ballistic) DP model simulation (Mean, ▬▬, ± 1 SD, ●●●) data of thigh (A) and shank (B) angle during the swing phase in the TTA subjects.

4.4.3. Kinetics

Similar patterns in net hip and knee torques in TTA subjects and controls were found in the first part of the swing phase (Figure 4.3), while in the second part of the swing phase, maximum hip extension and knee flexion torques were smaller in the TTA-subject group. This was supported by non-significant differences between groups in maximum hip flexion and maximum knee extension torques found in the first part of the swing phase, and significant differences in maximum hip extension and maximum knee flexion torques found in the second part of the swing phase (Table 4.3). Maximum knee power generation and absorption were significantly reduced in the TTA subjects while the maximum hip power generation and absorption were not significantly different (Figure 4.4 and Table 4.3). Total amount of both torque and power in hip and knee joint, quantified by means of the $TE_{hip+knee}$ and $MEE_{hip+knee}$, was significantly lower in the TTA subjects than in controls (Table 4.3). Both parameters were reduced with 27% in the TTA group.

4.4.4. Double pendulum model simulations

DP model simulations and experimental data are compared in Figures 4.5 and 4.6 for controls and TTA subjects, respectively. A relatively good fit between experimental and simulation data was found in both groups in the thigh angle, except for the last part of the swing phase. In the shank angle, deviations were larger and occurred throughout the swing phase. Table 4.3 shows that the RMS of the difference between simulation and experimental data in the TTA-subjects group was significantly higher for the shank angle compared to controls. There was no significant difference in RMS for thigh angle between TTA subjects and controls.

4.4.5. Additional simulations

The similar kinematic data in both groups suggests that the higher RMS between DP model simulations and experimental data in the TTA subjects is the result of the reduced inertial properties. To check whether the increased RMS as well as the decreased $TE_{hip+knee}$ and $MEE_{hip+knee}$ can be explained by the reduced inertial properties in the TTA subjects or is the result of the small kinematic differences between groups reported in Table 4.2, additional simulations were performed.

In these simulations inertial properties of each TTA subject were replaced by those of their matched control. It was found that the RMS for the shank angle was reduced from 18.5 to 13.0, while $TE_{hip+knee}$ increased from 7.7 to 10.8 and $MEE_{hip+knee}$ from 16.0 to 21.6. None of the measures were significantly different from the values of the healthy controls ($p=0.94$, $p=0.87$ and $p=0.89$, respectively).

*Table 4.3. RMS between experimental and simulation angles of thigh and shank for TTA subjects and controls and the torque effort and mechanical energy efficiency during swing phase. * indicates statistically significant differences between subjects and controls.*

		TTA subjects Mean (SD)	**Controls Mean (SD)**	**p-value**
Hip	Max extension torque (Nm)	14.5 (10.3)	31.4 (12.0)	0.005*
	Max flexion torque (Nm)	-19.3 (8.3)	-20.7 (8.5)	0.88
	Max power generation (W)	43.9 (13.8)	56.7 (30.7)	0.14
	Max power absorption (W)	-3.7 (3.0)	-1.7 (1.9)	0.05
Knee	Max extension torque (Nm)	7.6 (3.9)	6.4 (3.5)	0.33
	Max flexion torque (Nm)	-17.4 (5.4)	-26.4 (6.9)	0.005*
	Max power generation (W)	13.8 (8.4)	28.8 (10.7)	0.02*
	Max power absorption (W)	-75.7 (24.9)	-122.0 (37.1)	0.005*
$TE_{hip+knee}$		7.7 (3.0)	10.6 (3.5)	0.005*
$MEE_{hip+knee}$		16.0 (4.8)	22.0 (7.5)	0.005*
Thigh	RMS	7.3 (3.4)	8.6 (2.3)	0.17
Shank	RMS	19.6 (6.9)	11.9 (2.3)	0.01*

4.5. Discussion

4.5.1. Primary outcomes

The present study investigated the relationship between lower leg inertial properties in TTA subjects and experimental and simulation data. In all TTA subjects, the mass of the combined stump and prosthesis was reduced compared to shank and foot in matched controls. In addition, the center of mass was more proximal and the radius of gyration around the knee was reduced. No differences between TTA subjects and controls were found in measured walking speed and most of the kinematic parameters evaluated, while joint torques during gait were reduced in the TTA subjects. Deviations between DP model simulations and experimental kinematics were larger in the TTA subjects than in the controls.

The larger deviations between DP model simulations and experimental kinematics in our TTA subjects corresponds with findings reported by Mena et al.[43] who showed that lightweight prostheses lead to double pendulum swing phase kinematics that deviate from normal swing phase kinematics. It may be expected that a larger discrepancy between the double pendulum kinematics and the actual swing phase kinematics would lead to larger joint torques. However, similar hip and knee torques were found during the first part of the swing phase, while in terminal swing, torques were closer to zero in the TTA subjects. This may be explained in line with Van Soest et al.,[76] who argued that changing inertial properties not only affects the pendular behavior of the system, but also the torque needed to "correct" the pendular trajectory. To test this, we performed additional simulations in which the inertial properties of the TTA subjects were substituted by those of the controls, which resulted in a closer match between simulation and experimental data but larger torques and powers during the swing phase. This supports the hypothesis that lightweight prostheses lead to less optimal pendular properties, but to lower joint torques necessary to control the swing phase.

4.5.2. Methodological issues

The segment properties in this study were determined using a combination of techniques, i.e., regression data described by Winter,[86] a geometric model developed by Yeadon at al.[89] and a pendulum technique described by Hillery et al.[24] Kingma et al.[31] compared the first two techniques and found that they can lead to significantly different outcomes. For example, the regression model estimated the combined lower leg and foot mass 7% higher than the geometric model. Therefore, it is useful to compare our data with others. Lehmann et al.[37] reported light and intermediate weight prostheses of 42 to 70% of a normal leg mass, which is even more lightweight than those in the present study (77%). Lehmann et al. reported a relative proximal center of mass location of 0.47, which is very similar to the 0.464 in the present study. Czerniecki et al.[11] found a combined shank and foot mass, normalized to body mass, of 0.051 for five TTA subjects (0.047 in this study). From these similarities with other reports, we conclude that the differences in inertial properties between TTA subjects and controls found in the present study are not the result of differences in measurement techniques. For the anthropometric comparison, it may have been better to compare the prosthetic leg with the contra-lateral leg. However, because of possible compensatory mechanisms in the kinematics and kinetics of the latter leg, we chose to compare the TTA subjects to matched controls.

The similar walking speed in TTA subjects and controls is in contrast with reports that walking speed is reduced in TTA subjects (e.g., [64]). However, it should noted that the TTA subjects in the present study were a selected part of the TTA population, i.e., they are all traumatic, relatively young, and physically fit. In addition, our finding is not unique. For example, Lehmann et al.[37] reported walking speeds of 1.46 m/s in a group of 15 TTA subjects.

The amount of joint torque and power during the swing phase was quantified using two measures based on the time integral of the absolute non-normalized net joint torques and powers. The optimal way of determining the amount of work performed by the subject would be to perform calculations based on individual muscle energetics. However, this is presently considered difficult or impossible in normal subjects[22, 91] and may be even more difficult in amputees, in which a standard musculoskeletal model may not apply. Zatsiorksy and Gregor[22, 91] extensively discuss the determination of energy cost from net joint torques and powers and proposed the $MEE_{hip+knee}$ used in this study. Although in our opinion the $MEE_{hip+knee}$ is better described as a measure of cost rather than efficiency, we believe that, in combination with the $TE_{hip+knee}$, it is gives a reasonable indication of the cost of control of the swing phase.

4.5.3. Implications

The optimal prosthetic mass and inertial properties have been discussed for a long time. Inman proposed in 1967[27] that the optimal prosthetic mass should be a compromise between advantages and disadvantages of prosthetic mass. As an advantage of a heavy prosthesis, he describes increased kinetic energy, part of which can be fed back into the body at the end of the swing phase. As a disadvantage, he mentioned increased energy needed to initiate the swing phase. The comparison between TTA subjects and controls in the present study suggest a disadvantage of reduced inertial properties in the TTA subjects in terms of a less optimal pendular swing. However, an advantage of reduced inertial properties was found in terms of less joint torques necessary to correct upon this trajectory.

In a recent review, Selles et al.[59] found that none of the eight studies investigating the effects of changing the prosthetic inertial properties reported significant changes on comfortable walking speed, while only two of the five studies relating the inertial properties to economy of gait found significant differences. This finding was not in line with the modeling studies proposed in the literature. The present study, however, may explain that while lightweight prostheses may be less optimal in terms of their pendular behavior, they may be superior in terms of the joint torques needed to control the swing phase. In other words, adding mass to a prosthesis may improve the pendular behavior, but will require larger corrective joint torques; thus, the net effect of adding mass on a prosthesis may be small or even zero, depending on the location.

It is unlikely that the effect of adding mass will always be zeroed out by the above-mentioned mechanism. Modeling the lower leg as a single pendulum suggests that adding mass proximally will increase the natural frequency of the leg, whereas placing it distally may decrease the natural frequency, depending on the initial configuration.[76] In terms of control, adding mass close to the knee leads only to a relatively small increase in the torque needed to obtain the same acceleration, whereas the same mass added more distally will require a much higher torque to retain the same acceleration. An important prosthetic design implication of this may be that changing the mass of a liner around the stump may have a very different effect on amputee gait than changing ankle or shoe mass by the same amount. Thus, further study of the balance between pendular properties and ‘efficient control’ is needed to determine the optimal prosthetic inertial properties.

The present study focuses on the effect of inertial properties on kinematics and kinetics of the swing phase. However, it should be noted that other variables may also play a role. For example, while a lightweight prosthesis may need little energy to control, a prosthesis swinging more as a pendulum may need a less complex neural input and may thus impose a lower cognitive load on the subject. Another relevant aspect may be that a strong asymmetry between prosthetic and contra-lateral leg mass may be subjectively perceived as unnatural. In addition, prosthetic mass may influence the forces on the stump-socket interface. It is not obvious that all these variables will be optimal with the same prosthetic inertial configuration at all possible walking speeds. However, clarifying the relationship between these variables and the inertial properties of the prosthesis will make the choices more transparent.

4.6. Conclusion

In terms of mass, moment of inertia and center of mass location, the present TTA prostheses can be considered lightweight compared to a normal leg. The present study indicates that this leads to a disadvantage in terms of pendular properties at comfortable walking speed. However, this has the advantage of needing smaller joint torques to correct the unforced trajectory. Simulations indicate that changing the inertial properties to those of a normal leg leads to a more normal pendular swing phase, but increases the effort necessary to perform the swing phase. Therefore, the optimal inertial properties in terms of kinematics and kinetics of gait will be a compromise between pendular properties and ‘efficient control’.

5. Adaptations to mass perturbations in transtibial amputation subjects: kinetic or kinematic invariance?

5.1. Abstract

While most theoretical studies on the relation between prosthetic inertial properties and amputee gait predict adaptations in swing phase kinematics as a result of mass perturbation, these predictions are not supported by empirical evidence. In the present study, it was hypothesized that two extreme adaptation strategies to mass perturbation are possible: (1) a kinetic invariance strategy in which kinetics (joint torques) remain the same while kinematics (joint angles) change, or (2) a kinematic invariance strategy in which kinematics remain the same while kinetics change. We investigated which of these strategies best describes the adaptation to mass perturbation in a group of ten persons with a transtibial amputation. A gait analysis was performed during 5 different mass conditions and condition effects in kinetics and kinematics were evaluated. In addition, simulations were performed to predict the effect of each strategy. We found that mass perturbation induced more significant changes in the torques than in the angles. In addition, simulations assuming kinematic invariance were significantly related to the measured data, in contrast to the simulations assuming kinetic invariance. It is concluded that adaptation to mass perturbation in transtibial amputation subjects is better characterized as a kinematic invariance strategy than as a kinetic invariance strategy. This suggests that manipulating prosthetic inertial properties is not a tool to influence kinematics and that inertial properties should be evaluated in terms of the energetic cost of the swing phase.

5.2. Introduction

While establishing the effect of mass perturbation on amputee gait is important for determining the optimal inertial properties (mass, center of mass location and moment of inertia) of lower limb prostheses, the results of experimental data on mass perturbation in these subjects are not straightforward. In a review, Selles et al.[59] reported that none of the eight experimental studies investigating the effects of mass perturbation in amputation subject reported significant changes in comfortable walking speed, while only two of the five studies found significant differences in gait economy.

The reports on mass perturbation in amputation subjects are not in line with the dominant theoretical approaches which model the swing phase as a ballistic movement, that is, uninfluenced by muscle activity (e.g., [2, 16, 43]), and predict a direct influence of inertial properties on the swing phase kinematics. This discrepancy may be explained recent findings that ballistic models cannot predict the exact kinematics of the swing phase in healthy subjects without incorporating joint torques.[50, 60, 84] In addition, Selles et al.[62] confirmed these findings for a group of transtibial amputation (TTA) subjects and found that reduced inertial properties compared to control subjects led to different ballistic walking kinematics, whereas experimentally measured kinematics were highly similar in both groups.

While ballistic swing phase models predict an effect of mass perturbation only in the kinematics, in a movement influenced or controlled by muscle activity, adaptations in joint torques are also possible. Therefore, we theorized that mass perturbation may affect the swing phase of TTA subjects in two ways: at one extreme, kinetics (joint torques) remain the same while kinematics (joint angles) change: a kinetic invariance strategy. At the other extreme, kinematics remain the same while kinetics change: a kinematic invariance strategy. These extremes are in fact the opposite ends of a continuum; a response in which both kinetics and kinematics are affected is also possible.

In the present study, we investigated whether the effects of mass perturbation in TTA subjects can predominantly be described as a kinetic or a kinematic invariance strategy. The analysis is based on measured data on the one hand and, on the other, simulations predicting the effect of each strategy (i.e. kinetic invariance versus kinematic invariance).

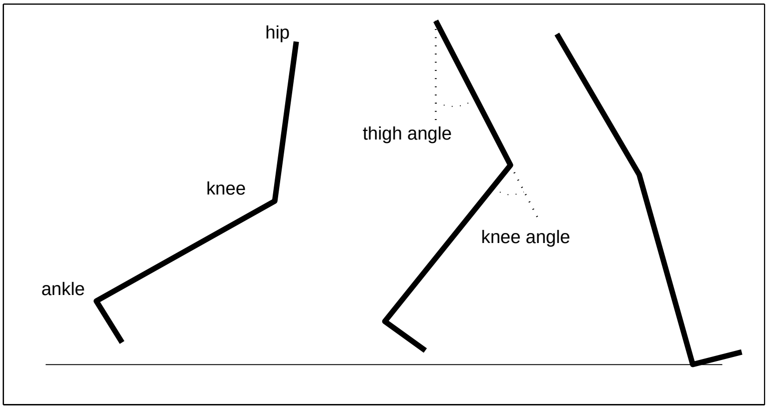

Figure 5.1. Stick figure of the swing phase of subject 1 during beginning, mid and end of the swing phase, indicating the definitions of the thigh and knee angle.

5.3. Methods

5.3.1. Subjects

Ten unilateral TTA subjects participated in the study (see Table 5.1). The subjects were included if they could walk without assistance for a prolonged period, had no cardiopulmonary, neurological or orthopedic disorders other than their amputation, had no skin problems of the residual limb, and were at least one year after discharge from the rehabilitation program. The hospital's Medical Ethical Commission approved the study and all subjects signed an informed consent.

5.3.2. Measurements

The assessment was carried out in a gait analysis lab on a 15-meter straight track. Five mass conditions were chosen, i.e., (1) no additional mass, (2) 1kg added to the prosthesis on a location similar to the lateral malleolus position of the normal leg, (3) 1kg just distal of the knee, (4) 1kg at the center of mass of the combined residual limb and prosthesis, and (5) 2kg at the center of mass of the combined residual limb and prosthesis. The 2kg condition was always performed last to avoid that, due to fatigue, not all 1kg conditions could be completed; the 1-kg conditions were randomly assigned. The locations were chosen to cover the whole shank. The mass was added using lead strips of 1kg that were bent around the leg. The inner side of the lead strips was overlay with foam to protect the prosthesis. When necessary, the strips were held in place with an elastic band. Before measurement in each condition, subjects walked around the lab to get used to the added mass; this adaptation period lasted until subjects reported that they were used to the condition, with a minimum of 5 minutes. Between conditions, subjects sat down until they indicated they were sufficiently rested. For each mass condition, subjects were instructed to walk at the speed that felt most comfortable. All subjects wore their preferred walking shoes and completed the track eight times in each condition.

The kinematics of the prosthetic leg were recorded unilaterally with a 3D 3-camera ProReflex infrared system (Qualisys, Sweden) using reflective markers and a sample frequency of 50 Hz. The camera system covered about three meters in the middle of the track. Markers were placed on the greater trochantor, the lateral femoral condyle and on the equivalent of the lateral femoral condyle and lateral malleolus locations.[88] Body height, body mass and the lengths of thigh, shank and foot were measured and used to calculate mass, center of mass and moment of inertia of all segments.[86] The anthropometric properties of the residual limb were determined using a geometric model.[31, 62, 89] In addition, the combined prosthetic socket, shank and foot were weighted, balanced on a straight edge to determine center of mass location and swung in a pendulum with a small amplitude to determine the moment of inertia.[24, 62] Combined residual limb+prosthesis properties were then calculated. In the added mass conditions, anthropometric properties were calculated by modeling the additional mass as a point mass.

5.3.3. Data analysis

A spline-fitting interpolation algorithm (Qview software, Qualisys, Sweden) was used to fill marker dropouts. A maximum gap of 6 samples was found, occurring in the greater trochantor marker because of arm swing. Marker trajectories were filtered in Matlab (The MathWorks, Inc., USA) with a second order low pass zero-phase-lag Butterworth filter with a cut-off frequency of 9 Hz. For each subject in each condition, six complete gait cycles were selected by marking foot contact in the ankle kinematics [51]. After visual inspection of the individual trajectories, marker trajectories of the six cycles were normalized to the shortest cycle and averaged. In the averaged data, toe-off was selected using the fifth metatarsal head kinematics.[51] Net joint torques around hip and knee during the swing phase were calculated using a linked segment model, in which foot and shank together were modeled as a single segment Kinematic and kinetic variables were calculated in the sagital plane only.

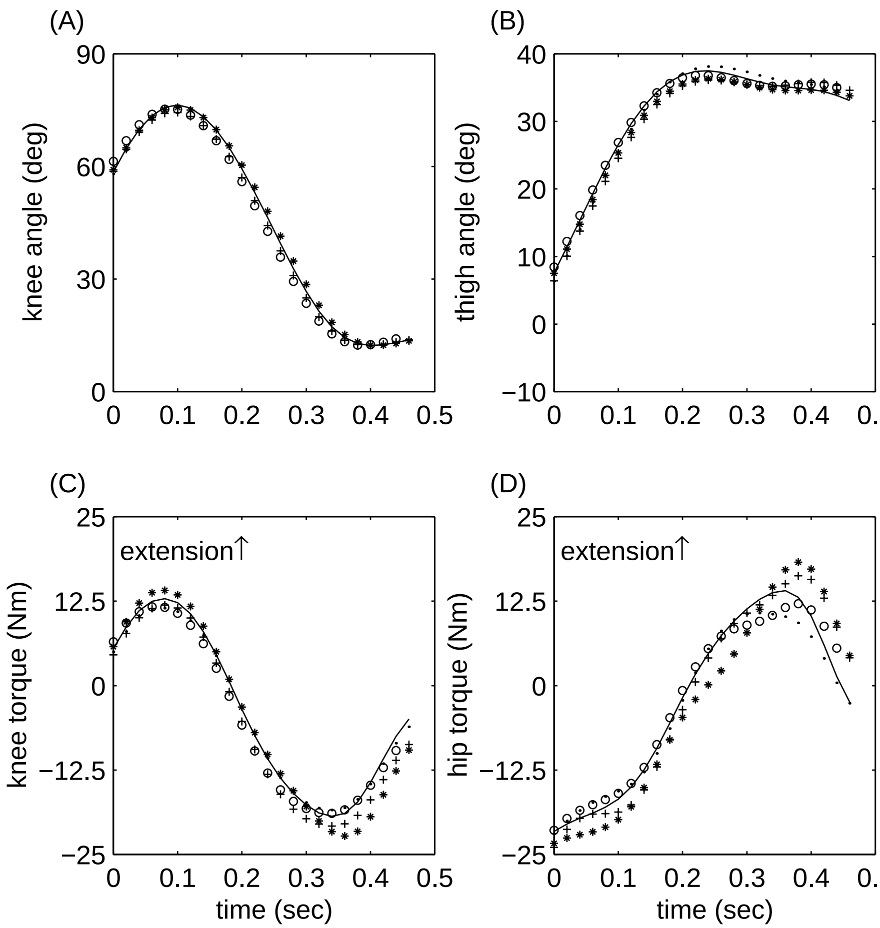

*Figure 5.2. Typical example of the knee (A) and thigh (B) angle and the knee (C) and hip (D) torque during the swing phase for subject 1 during the five conditions: zero-added-mass (—), 1kg added to the ankle (*), 1kg added to the knee (o), 1kg added to the center of mass of the lower leg (•) and 2kg added to the center of mass (+).*

5.3.4. Simulations

Calculations were performed to predict the effects of the two extreme adaptation strategies to mass perturbation, i.e., 1) the kinetic invariance strategy or 2) the kinematic invariance strategy.

Table 5.1: Characteristics, reason for amputation, prosthetic mass as well as the walking speed, stride length and stride frequency of the transtibial amputation subjects. Abbreviations: V, Vascular; T, Traumatic.

TTA	**Age (yr)**	**Height (m)**	**Weight (kg)**	**Cause of amp.**	**Prosthetic mass (kg)**	**Walking speed (m/s)**	**Stride length (m)**	**Stride freq (1/s)**
1	38	1.78	83	T	3.0	1.28	1.43	.89
2	38	1.88	113	T	3.4	1.41	1.62	.87
3	48	1.76	77	T	2.0	1.31	1.44	.91
4	42	1.81	72	V	2.5	1.41	1.62	.87
5	51	1.78	93	T	2.4	1.17	1.33	.88
6	71	1.57	74	V	2.3	1.36	1.36	1.00
7	35	1.86	71	T	2.5	1.39	1.61	.86
8	44	1.68	74	V	2.4	.74	.97	.76
9	25	1.85	69	T	2.7	.88	1.10	.79
10	50	1.84	105	V	3.0	1.29	1.36	.94
Mean (SD)	44 (12)	1.78 (.1)	83 (15)		2.6 (.4)	1.22 (.23)	1.39 (.22)	.88 (.07)

In the simulations of kinetic invariance, forward dynamical simulations of the swing phase were performed for all subjects and all mass conditions based on the joint torques calculated in the condition without added mass. The added mass was modeled as a point mass. The model (Fig. 5.1) consisted of a thigh connected to the shank by a revolute joint and a foot rigidly attached to the shank. The hip trajectory was prescribed for each subject as a function of time using the recordings from the greater trochanter marker in the zero-added-mass condition and cubic spline interpolation. The thigh and shank angle and angular velocity at toe-off were derived from the zero-added-mass condition and defined the initial state of the model. The Newtonian equations of motion for the model were derived and implemented in Matlab using MUSK software.[3, 75] Forward-dynamic numerical simulations of the equations of motion were performed separately for each subject, using a variable stepsize variable order Adams-Bashford-Moulton integration algorithm. Each simulation had the same duration as the swing phase of the corresponding subject and output was created in steps of .02s. An estimation of the numerical integration error of a similar model has been reported earlier and was well below one degree at the end of the swing phase.[62]

In the simulation of kinematic invariance, inverse dynamics calculations were carried out for all subjects and all mass conditions to calculate joint torques based on the kinematic data from the zero-added-mass condition.

Table 5.2. Mean difference with the zero-added-mass condition for the kinematic variables in each condition as a percentage of the range observed in the time series. "Time of maximum angle" and "time of minimum angle' refer to the time after toe-off that the event occurs. The p-values indicate the result of the MANOVA testing for a general condition effect. ¶ indicates a significant condition effect at a .01 significance level. Abbreviations: dis, distal mass location; mid, mid-shank location; prox, proximal mass location.

	Knee					**Thigh**				
	1kg dis	1kg mid	1kg prx	2kg mid	p-value	1kg dis	1kg mid	1kg prx	2kg mid	p-value
Initial angle (%)	3	1	0	2	.203	3	1	0	-1	.540
Initial angular velocity (%)	-2	-2	0	-2	.102	0	-1	0	0	.913
Final angle (%)	-1	-1	-1	-1	.162	-2	1	0	4	.008¶
Final angular velocity (%)	3	1	-2	1	.144	7	3	0	4	.001¶
Max angle (%)	-1	-2	-1	-3	.016	-5	-3	-1	-4	.001¶
Time of max angle(%)	0	-2	0	-1	.241	-2	0	-1	1	.707
Min angle (%)	-2	-1	-1	-1	.079	3	1	0	-1	.540
Time of min angle (%)	0	-1	1	-2	.099	0	0	0	0	.354
Max angular velocity (%)	-2	-2	0	-2	.102	-6	-5	-2	-4	.001¶
Min angle velocity (%)	0	0	1	0	.151	1	2	0	4	.288
Angle at 50% swing time (%)	2	-4	6	-1	.175	-5	-2	-1	0	.033

5.3.5. Data comparison

To determine condition effects in the measured kinematics and kinetics of the swing phase, specific events were derived from the time series (see Tables 5.2 and 5.3) to cover all relevant aspects of the signals. To indicate the magnitude of the effects, for each event in each condition, the difference with the zero-added-mass condition was calculated, normalized by the range observed in the time series.

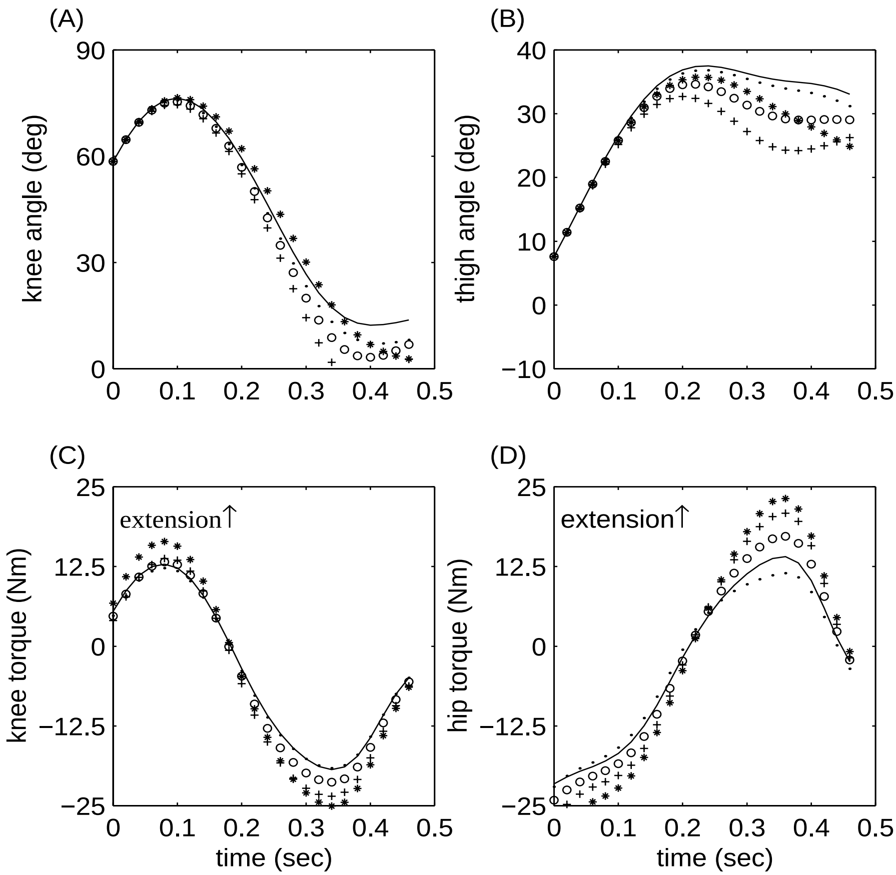

*Figure 5.3. A, B: Predicted knee and thigh angle during the swing phase assuming a kinetic invariance strategy in the five mass conditions. C, D: Predicted knee and hip torque during the swing phase assuming kinematic invariance in the same five mass conditions. Data are based on the same transtibial amputee as Figure 5.2 to indicate that the effects of the simulations assuming kinematic invariance (C, D) are more similar to the measured data than simulations assuming kinetic invariance (A, B): zero-added-mass (—), 1kg added to the ankle (*), 1kg added to the knee (o), 1kg added to the center of mass of the lower leg (•) and 2kg added to the center of mass (+).*

5.3.6. Statistical analysis

Systematic condition effects in the measured data were tested for using a Multiple Analysis of Variance (MANOVA) in SAS (SAS Institute Inc., Cary, NC, USA). A "missing at random" procedure was used to deal with the 2-kg condition missing in three subjects. To compare simulations with measured data, the relation between these changes in simulation and measured data with the zero-mass condition was determined using linear regression in SPSS (SPSS Inc., Chicago, Illinois, USA). The use of a kinematic invariance strategy would result in a significant relation between the changes in joint torques with the zero-added-mass condition in simulations and in measured data with a slope (a) of one and an intercept (b) of zero for the regression line Y=aX+b. Similarly, the use of a kinetic invariance strategy would results in a significant relation between the changes in joint angles with a slope of one and an intercept of zero. Because of the large number of statistical tests, a p-value smaller than .01 was considered significant.

Table 5.3. Mean difference with the zero-added-mass condition for the kinetic variables in each condition as a percentage of the range observed in the time series varies. "Time of maximum torque", "time of minimum torque', and 'time of first zero-crossing' refer to the time after toe-off that the event occurs. The p-values indicate the result of the MANOVA test for a general condition effect. ¶ indicates a significant condition effect at a .01 significance level. Abbreviations: dis, distal mass location; mid, mid-shank location; prox, proximal mass location.

	Knee					**Hip**				
	1kg dis	1kg mid	1kg prx	2kg mid	p-value	1kg dis	1kg mid	1kg prx	2kg mid	p-value
Initial torque (%)	2	-1	-2	-2	.003¶	1	1	-1	3	.002¶
Initial value torque derivative (%)	3	0	0	2	.027	0	-4	-5	-10	.187
Final torque (%)	-13	-7	-5	-14	.003¶	-26	-11	-1	-18	.001¶
Final value torque derivative (%)	7	2	-4	1	.004¶	2	0	-9	-2	.100
Max torque (%)	2	-2	-3	-3	.002¶	1	0	0	3	.002¶
Time of max torque (%)	-1	-2	0	0	.483	-18	-19	-1	0	.149
Min torque (%)	-11	-4	1	-7	.001¶	-7	-2	3	-8	.002¶
Time of min torque (%)	2	-1	-1	-1	.001¶	8	0	-3	1	.001¶
Time of first zero-crossing (%)	-2	-3	-1	-4	.092	5	3	-2	3	.006¶
Torque at 50% swing time (%)	-2	-3	1	-5	.466	8	4	2	7	.001¶

5.4. Results

The subjects included in the study were relatively divers in terms of age, weight, height and walking speed (Table 5.1). All subjects used a regular Kondylen Bettung Munster (KBM) or patellar-tendon-bearing (PTB) prosthesis. All subjects finished the 1kg-conditions. Three of the TTA subjects did not perform the last (2kg) condition because of general fatigue due to the walking.

Significant condition effects were found in only 4 of the 22 measured angular (kinematic) variables (18%), compared with 13 out of the 20 torque (kinetic) variables (65%; see Figure 5.2 for a typical example and Table 5.2 and 5.3). No significant effects of mass condition were found in the measured walking speed ($p=.27$), stride length ($p=.22$) and stride time ($p=.36$). Generally, the influence of mass perturbation on the measured kinetic and kinematic variables (Tables 5.2, 5.3, and Figure 5.2) were relatively small considering the large increase in prosthetic mass. The condition effects of the kinetic invariance simulations were generally larger than found in the measured data (Figure 5.3A, 5.3B, and Table 5.4; compare Figure 5.2A, 5.2B, and Table 5.2) while the effects of kinematic invariance simulations were more similar to the measured data (Figure 5.3C, 5.3D, and Table 5.4; compare Figure 5.2C, 5.2D, and Table 5.3). Generally, the kinematic invariance strategy predicted effects on the amplitude of the joint torques, while the timing remains unchanged.

Table 5.4. Mean difference with the zero-added-mass condition for the kinematic variables in each condition assuming kinetic invariance as a percentage of the range observed in the time series varies. Abbreviations: dis, distal mass location; mid, mid-shank location; prox, proximal mass location.

	Knee				**Thigh**			
	1kg dis	1kg mid	1kg prox	2kg mid	1kg dis	1kg mid	1kg prox	2kg mid
Initial angle (%)	0	0	0	0	0	0	0	0
Initial angle vel (%)	3	3	3	3	-2	-3	-3	-2
Final angle (%)	-15	-13	-9	-16	-29	-15	-6	-22
Final angular vel (%)	-10	9	3	20	-2	15	1	34
Max angle (%)	0	-2	-1	-3	-10	-12	-4	-21
Time of max angle (%)	0	-1	0	-2	-6	-8	-5	-5
Min ang (%)	-13	-15	-9	-23	0	0	0	0
Time of min angle (%)	9	0	1	-3	0	0	0	0
Max angle vel (%)	3	3	3	3	-3	-3	-1	-4
Min angle vel (%)	-2	-8	-3	-14	-12	-5	-2	-12
Angle at 50% swing time (%)	9	-13	-7	-34	-20	-19	-6	-35

We selected minimum and half-time knee angle, maximum and final thigh angle, and maximum and minimum knee and hip torques for further analysis because they were strongly influenced by the mass perturbation in the simulations (Figure 5.3 and Table 5.2-5).

Direct comparison of measured data and simulations for the selected variables confirmed that the simulations of the kinematic invariance strategy better describes the measured data than the simulations of the kinetic invariance strategy (Figure 5.4 and Table 5.6). No significant relations were found between measured changes in angles compared to the zero-mass condition and predicted changes assuming kinetic invariance. In contrast, significant relations were found between measured and predicted torques assuming kinematic invariance. The estimated intercepts were close to zero both for all variables. However, while for the angular variables the slopes did not differ from zero, the estimated slopes for the torque variables where all larger than zero although a slope of one was outside the confidence interval.

Table 5.5. Predicted difference with the zero-added-mass condition for the kinetic variables in each condition assuming kinematic invariance as a percentage of the range observed in the time series varies. Abbreviations: dis, distal mass location; mid, mid-shank location; prox, proximal mass location.

	Knee				**Hip**			
	1kg dis	1kg mid	1kg prox	2kg mid	1kg dis	1kg mid	1kg prox	2kg mid
Initial torque (%)	3	-3	-2	-5	12	6	3	11
Initial value torque derivative (%)	9	2	-1	2	2	-4	-6	-10
Final torque (%)	-5	-2	0	-4	-5	-1	4	-4
Final value torque derivative (%)	12	4	0	5	16	7	-3	11
Max torque (%)	7	-1	-2	-2	11	6	5	10
Time of max torque (%)	0	1	1	3	-19	-10	0	0
Min torque (%)	-20	-8	-1	-12	-23	-9	3	-16
Time of min torque (%)	0	0	0	0	1	0	-1	1
Time of first zero-crossing (%)	0	-2	0	-4	3	0	-2	-1
Torque Angle at 50% swing time (%)	-5	-3	-1	-7	6	1	-3	1

5.5. Discussion

The present study investigated whether the adaptation in TTA subjects to mass perturbation of the lower leg is better described by a kinetic or a kinematic invariance strategy by evaluating systematic mass conditions effects in measured joint angles and torques during the swing phase as well as by comparing the measured data to simulations of each strategy.

The measured data revealed more systematic changes after mass perturbation in the joint torques than in the angles, indicating a kinematic invariance strategy. Additionally, no significant relations were found between changes after mass perturbation in kinetic invariance simulations and measured data, while relations between kinematic invariance simulations and measured data were significant, although not directly one-to-one. Taken together, the results indicate that the adaptation to mass perturbation is better characterized as a kinematic invariant than as a kinetic invariant strategy.

Table 5.6: Linear regression between simulated changes assuming kinetic or kinematic invariance and measured changes in selected angular or torque variables. All changes are relative to the zero-added-mass condition. ¶ indicates a significant condition effect at a .01 significance level

	R	**p-value**	**Slope (95% CI)**	**Intercept (95% CI)**
Max thigh angle (ang)	.284	.053	.13 (-.01:.26)	-.42 (-.86:.01)
Thigh angle at half-time (%)	.087	.562	.04 (-.08:.15)	.36 (-.86:.17)
Min knee angle (ang)	.197	.185	.04 (-.02:.09)	.40 (-1.05:.26)
Knee angle at half-time (%)	.122	.414	.07 (-.10:.23)	.21 (-.55:.97)
Max hip torque (%)	.451	.001¶	.29 (.12:.46)	-.22 (-.85:.40)
Min hip torque (%)	.421	.003¶	.23 (.08:.37)	.04 (-.91:.99)
Max knee torque (%)	.711	.001¶	.54 (.38:.70)	-.29 (-.52:-.07)
Min knee torque (%)	.725	.001¶	.47 (.34:.60)	.01 (-.45:.47)

Despite the fact that we carefully designed the study to avoid an influence of fatigue, for the 2-kg condition, its effects can not be ruled out. Subjects were included only if they could walk for a prolonged period and therefore were relatively young and physically able. Three of the ten subjects preferred not perform the last 2kg condition because of fatigue. None of the subjects rejected this condition because of the amount of mass. Since we expected mass location to be more important then mass magnitude, we decided to start with the randomized 1-kg and zero-added-mass conditions and end with the 2kg condition.

During the experiments, subjects reported getting used to the different conditions within a few steps. The majority of the subjects were not able to name a single condition which they preferred, although the 2kg condition was often considered too heavy, while the 1kg added to the lateral malleolus felt most different from the no-added-mass condition. The finding that TTA subjects systematically maintained their kinematic pattern suggests that they had adjusted to the new mass conditions. However, this does not rule out the possibility of long-term kinematic adjustments to mass perturbation, optimizing, for example, efficiency or walking speed.

In this study, we focussed on TTA subjects, disregarding the transfemoral amputation (TFA) and through-knee amputation (TKA) subjects. Distinguishing these groups was considered necessary because we expect the influence of prosthetic mass to be different. Since TFA and TKA subjects do not have direct muscular control of the knee joint, using a kinematical invariance strategy similar to TTA subjects may not be possible for these subjects.

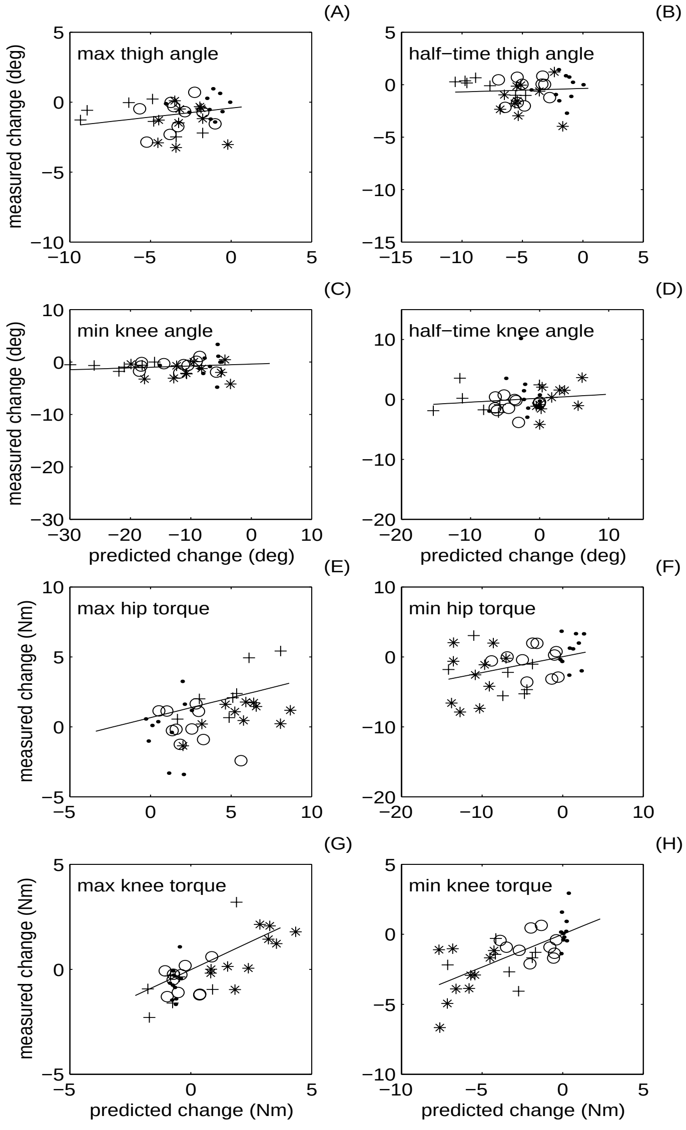

*Figure 5.4. A-D: Relations between change in simulated (x-axis) and measured (y-axis) angular variable compared to the zero-added-mass condition, assuming kinetic invariance. E-H: Correlations between change in simulated and measured torque variables for the simulations compared to the zero-added-mass condition, assuming kinematic invariance. Symbols: *, 1kg added to the ankle, o, 1kg added to the knee, •, 1kg added to the center of mass of the lower leg and +, 2kg added to the center of mass.*

Though significant, the changes in torques after mass perturbation were relatively small. Whereas the 2kg condition increased the prosthetic mass by 59-100%, the effect on the joint torques was on the order of maximally 10-20% (see Figure 5.2 and Table 5.3). These small effects compared to the inter- and intra-subject variations may partly explain the relatively low percentage of explained variance of the linear regression between simulations and measured data. Although the explained variance may increase with larger mass perturbations, adding more mass to a prosthesis (for example, 5 kg) may lead to unnatural situations largely increasing, for example, the forces between socket and residual limb. However, the linear regression data may also indicate that the kinematics are not be completely invariant in all parts of the swing phase.

Several studies have shown that a significant part of the joint torques during the swing phase can result from passive elastic joint and muscle properties at the extremes of the joints range of motion (e.g., [38, 53, 79]). The present study does not allow to determine the relative contribution of passive and active torques. However, the torques found throughout the swing phase, and the systematic torque adaptations after mass perturbation suggest that torques did result from passive elastic joint and muscle properties alone.

The kinematic invariance strategy suggests that mass perturbation affects the energetic cost of gait rather than walking speed or kinematics, which may be in line with most of the experimental literature. In a recent review, Selles et al.[59] reported that none of the eight experimental studies investigating the effect of mass perturbation in amputation subject reported significant changes in comfortable walking speed, while two of the five studies found significant differences in gait economy. Recently, Mattes et al.[41] matched the prosthetic inertial properties of TTA subjects to their contralateral leg by adding, on average, 1.7 kg distally on the lower leg. As a result, step length was not influenced by mass addition, while swing time significantly increased by about 4%, the latter finding in contrast with the kinematical invariance strategy. However, in line the kinematical invariance strategy, the largest changes were found in the metabolic cost, which increased by 7%.

Findings similar to the kinematic invariance strategy have been reported in studies on mass perturbations in non-amputation subjects during walking and running. For example, Donker et al.[15] found that adding 1.7 kg to the right ankle during walking at different walking speeds did not influence stride frequency. Skinner and Barrack[63] added 1.82kg to the ankles of ten able-bodied subjects and reported a significant increase in oxygen cost per meter, while no significant changes were found in walking speed, cadence, stride length and double-limb support. Cavanagh and Kram[5] concluded that, during running, even small loads on the feet dramatically increase the metabolic cost but had little effect of the stride frequency and stride length.

The nature of the kinematic invariance strategy was not investigated in this study. While the present data suggest that minimization of energy expenditure is not a driving mechanism during the swing phase, it may still apply for the complete gait cycle. A model of the complete gait cycle is necessary to further investigate this strategy. The kinematical invariance strategy could also have a more behavioral nature. Subjects may aim at maintaining their kinematic pattern because they learned to walk this way during rehabilitation. In addition, walking kinematically similar to non-amputees, or matching the prosthetic leg kinematics to the contralateral leg may be an implicit or explicit ‘goal’ for amputation subjects.

The present findings may have important implications for models predicting the optimal prosthetic inertial loading, since they suggest that the relation between prosthetic inertial properties and amputee gait can not be understood using ballistic walking models only. Therefore, alternative models need to be developed to understand and predict the effects of mass perturbations. Because mass perturbations mainly affected kinetics, the present data suggest that prosthetic inertial properties should be evaluated in terms of the "cost" for the subjects to generate their preferred kinematic pattern.

6. Predicting the optimal prosthetic inertial properties in transtibial amputation subjects

6.1. Abstract

Present models used to predict the influence of prosthetic mass on the gait of amputation subjects are based on the assumption that the swing phase is uninfluenced by muscle activity. However, predictions of these models are not supported by empirical evidence. In this study, we developed a new model based on the assumption that subjects adapt to mass perturbations by changing joint torques while the kinematics remain invariant. In the model, the effect of mass perturbations of the lower leg is evaluated in terms of net joint hip and knee torques as well as in the net joint reaction forces and torque between socket and stump during the swing phase of walking. The magnitude of the mass perturbation was varied from minus 2.5 kg (removing mass) to plus 2.5 kg (adding mass), while the location of the mass perturbation was varied from directly below the knee to the heel. It was found that both size and direction of the effect of mass perturbation on muscular cost depended on the location of the perturbation. In 9 of the 10 transtibial amputees, cost decreased after distally removing mass as well as after proximally adding mass to the lower leg. In contrast, net joint forces and torques between stump and socket always decreased when mass was removed and increased when mass was added. It was concluded that, in particular, the mass of distally located components (e.g., foot, ankle, shoes) is related to the estimated muscular cost and the forces and torque in the stump-socket interface. A comparison with the experimental literature on mass perturbation in below-knee amputees suggests that the present simulation data better describe the experimental data than the predictions derived from ballistic walking models.

6.2. Introduction

While it is current practice to minimize prosthetic mass, from a theoretical point of view, it has been claimed that this may not lead to an optimal gait pattern (e.g., [2, 43, 67, 68]). The rationale behind this claim is that prosthetic inertial properties mainly affect the swing phase of gait and that this swing phase is ballistic, that is, only under the influence of gravitational and inertial forces. In a ballistic swing phase, kinematics are influenced by mass perturbation in the same way that the natural frequency of a simple pendulum is influenced by its inertial properties. For this reason, a different mass and mass distribution, as in a lightweight prosthesis, can lead to a different swing phase. It has been proposed, therefore, that the optimal inertial prosthetic properties should be determined by optimizing the pendulum characteristics of a prosthesis (e.g., [16, 21, 43, 67]).

However, the prediction of ballistic models that swing phase kinematics (joint angles) change when the prosthetic inertial properties are changed is not confirmed by most of the experimental literature. In a recent review, Selles et al.[59] reported that none of the eight studies investigating the effect of prosthetic mass perturbation reported significant effects on walking speed, stride length and stride frequency. In addition, the ballistic swing phase assumption itself has been challenged by recent findings that the swing phase cannot be understood as completely ballistic and that muscle activity needs to be included to accurately model the swing phase. This was reported both in healthy subjects[50, 60, 84] as well as in transtibial amputation (TTA) subjects.[61]

Recently, Selles et al.[61] hypothesized that there are two extreme adaptations to mass perturbations possible: (1) a kinetical invariance strategy in which kinematics (joint angles) change while kinetics (joint torques) remain the same, or (2) a kinematical invariance strategy in which kinetics change while kinematics remain the same. Studying ten TTA subjects, Selles et al.[61] found that the response to mass perturbation was best characterized as a kinematical invariance strategy. This has implications for modeling the effect of prosthetic inertial loading because ballistic swing phase models can not explain this finding. In addition, the use of a kinematical invariance strategy suggests that inertial properties of lower limb prosthesis should not be optimized in terms of gait kinematics and walking speed, since the subject (consciously or unconsciously) keep these parameters constant. Instead, it suggests that inertial properties should be optimized in terms of a cost function related to the muscle activity required to obtain 'normal' gait kinematics.

While optimal prosthetic mass has mostly been studied from the perspective of gait kinematics, kinetics and energetics, other variables may also be important for evaluating the effect of prosthetic mass perturbation, such as, for example, forces in the stump-socket interface, complexity of the neural control, perception of asymmetry between both leg and proprioceptive feedback from the prosthetic leg. These variables will not necessarily be optimal in the same prosthetic configuration. If the effects of prosthetic inertial properties on these variables can be quantified, then the optimal prosthetic design can, at least in principle, be expressed as a function of the relative weighting of the individual cost terms.

The aim of the present study is to develop a new method for predicting the effects of prosthetic mass, center of mass location and moment of inertia on two variables important for determining optimal prosthetic properties, that is, swing phase kinetics and forces in the stump-socket interface. The method will not be based on pendulum characteristics but evaluates the effects of a wide range of different inertial properties based on the assumption that TTA subjects adapt to mass perturbations by keeping the swing phase kinematics the same.

6.3. Methods

The experimental data used in the present study have been described more extensively elsewhere[61] and will be summarized here.

6.3.1. Subjects

Ten TTA subjects participated in the study. The subjects were included if they could walk unassistedly for a longer period, had no cardiopulmonary, neurological or orthopedic disorders other than their amputation and had no skin problems of the residual limb. All subjects were measured at least one year after discharge from the rehabilitation program.

6.3.2. Measurements

The TTA subjects walked in a gait analysis lab at comfortable walking speed wearing their preferred walking shoes. Kinematics of the lower extremities were recorded with a 50 Hz, three-camera ProReflex infrared system (Qualisys, Sweden). Reflective markers were placed on the greater trochantor, the lateral femoral condyle and on the prosthesis on the equivalent of the lateral malleolus location.[88] Average marker trajectories were calculated for each subject based on 6 complete gait cycles.

For each subject, height, weight and the lengths of thigh and shank were measured and used to calculate mass, center of mass and moment of inertia of the thigh. [86] The anthropometric properties of the residual limb were calculated based on a geometric model.[31, 89] Weighing, balancing on a straight edge and a pendulum test determined the prosthetic anthropometric properties.[24]

6.3.3. Simulations

Simulations were performed to evaluate the effect of mass perturbations of the lower leg on the swing phase of TTA subjects. A linked segment model of the prosthetic leg was used with the prosthesis (foot and shank), residual limb and thigh as separate segments (Figure 6.1). Prosthesis and stump were modeled as separate segments in order to allow calculation of the net joint reaction forces and torque in their interface. Relative motion between stump and prosthesis was assumed to be negligible.

The simulations are based on inverse dynamics calculations in which net joint torques and reaction forces are calculated from the measured kinematics and the segment inertial properties. Based on the assumption that kinematics do not change when mass is perturbed, the effect of mass perturbations is evaluated for each subject in each mass condition based on the kinematic data measured without additional mass.

Mass perturbation was parameterized in terms of adding and removing point masses of varying magnitude, at varying locations. The amount of mass was varied in 11 steps of 0.5 kg from minus 2.5 kg (removing mass) to plus 2.5 kg (adding mass). The location was varied in 12 steps of 10% of the knee-ankle length from directly below the knee to 110%, the latter position estimating the location of the heel. The whole lower leg was studied because changes in sockets and liners can affect the proximal part of the lower leg. Combining all locations and masses led to 132 mass conditions. In the analysis, only conditions were considered in which the perturbed segments had a positive mass and positive moment of inertia relative to the segment center of mass.

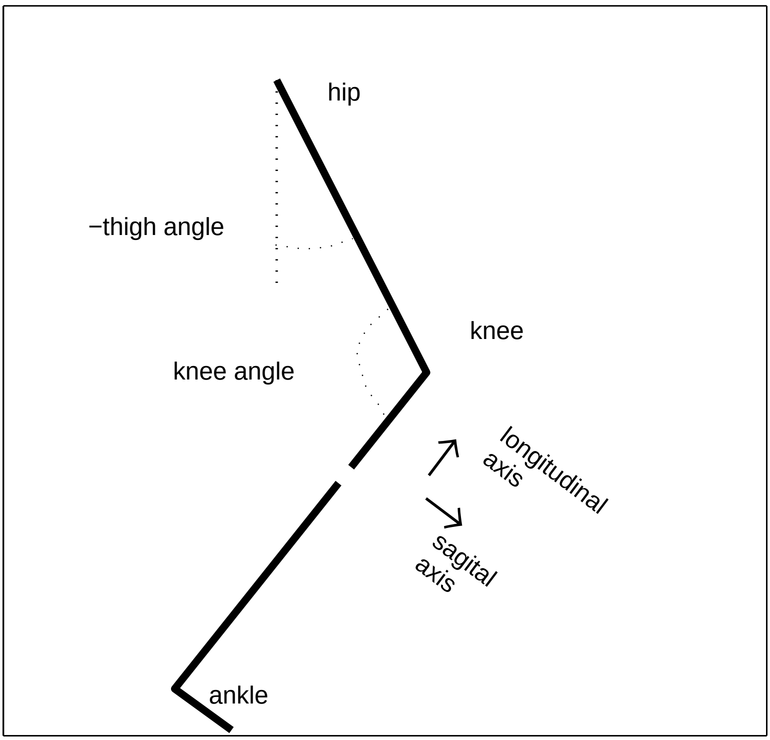

Figure 6.1. Segment model of the prosthetic leg, indicating the definition of the segments, angles and axes.

The outcome of each inverse dynamics simulation were the net joint reaction forces and torque in the stump-socket interface as well as the net joint torques in hip and knee. To improve inter-subject comparability, all torques were normalized to body weight and leg length[30] and the joint reaction forces to body weight. An estimate of the total muscular cost during the swing phase in the hip and knee was made using the 'torque effort' (TE) in the hip and knee joint,

$$TE_{hip+knee} = \int_{to}^{hs} (|T_{hip}|)dt + \int_{to}^{hs} (|T_{knee}|)dt \qquad (6.1)$$

where T_{hip} and T_{knee} are the normalized net joint torques in knee and hip and to and hs refer to the time of toe-off and heel strike. This definition of "torque effort" assumes the cost of muscle contraction to depend on muscle force only.[91]

Table 6.1. Characteristics of the transtibial amputation (TTA) subjects, walking speed, prosthetic mass and the lower leg (residual limb and prosthesis) inertial properties, normalized to body mass and lower leg length.[86] Reference values for the inertial properties of control subjects[86] are indicated.

TTA subject	Age (yr)	Height (m)	Weight (kg)	Walking speed (m/s)	Mass prosthesis (kg)	Lower leg mass/body mass	Lower leg center of mass from knee/leg length	Proximal lower leg radius of gyration/leg length
1	38	1.78	83	1.28	3.0	0.052	0.538	0.725
2	38	1.88	113	1.41	3.4	0.042	0.633	0.765
3	48	1.76	77	1.31	2.0	0.036	0.461	0.651
4	42	1.81	72	1.41	2.5	0.049	0.545	0.750
5	51	1.78	93	1.17	2.4	0.039	0.614	0.779
6	71	1.57	74	1.36	2.3	0.050	0.630	0.788
7	35	1.86	71	1.39	2.5	0.047	0.612	0.783
8	44	1.68	74	0.74	2.4	0.044	0.510	0.700
9	25	1.85	69	0.88	2.7	0.052	0.531	0.726
10	50	1.84	105	1.29	3.0	0.052	0.489	0.647
Mean (SD)	44.2 (12)	1.781 (0.1)	83.05 (15)	1.22 (0.23)	2.62 (0.41)	0.046 (0.01)	0.56 (0.06)	0.73 (0.05)
controls						0.061	0.606	0.735

Since the stump-socket interface was modeled as a rigid joint, the torque in this joint is needed to maintain prosthesis and residual limb parallel. However, in reality, this torque is established by a set of forces acting between socket and stump. Therefore, the torque should be interpreted as the net result of many forces additional to the joint reaction force between socket and stump. An estimate of the total torque between residual limb and socket was calculated as

$$TE_{stump-socket} = \int_{to}^{hs} (|T_{stump}|)dt \tag{6.2}$$

in which $TE_{stump\text{-}socket}$ was the torque (normalized to body weight and leg length) in the interface between the socket and stump.

6.4. Results

The characteristics of the subjects are given in Table 6.1, as well as the initial mass of their prosthesis and the anthropometric properties of the combined lower leg (residual limb and prosthesis).

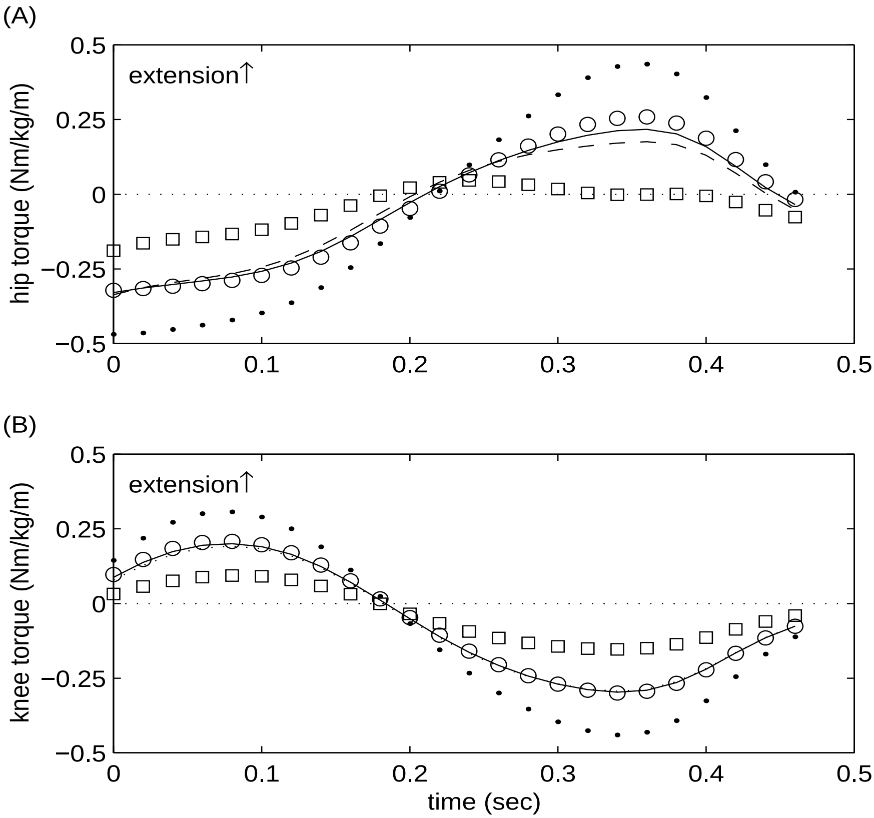

Figure 6.2. Net joint torque in hip (A) and knee (B) during the swing phase for subject 1 in five simulated conditions. —, unperturbed condition; --, adding 1 kg just below the knee; o, removing 1 kg from just below the knee; •, adding 1 kg to the heel; □, removing 1 kg from the heel. Torques are normalized to body weight and leg length.

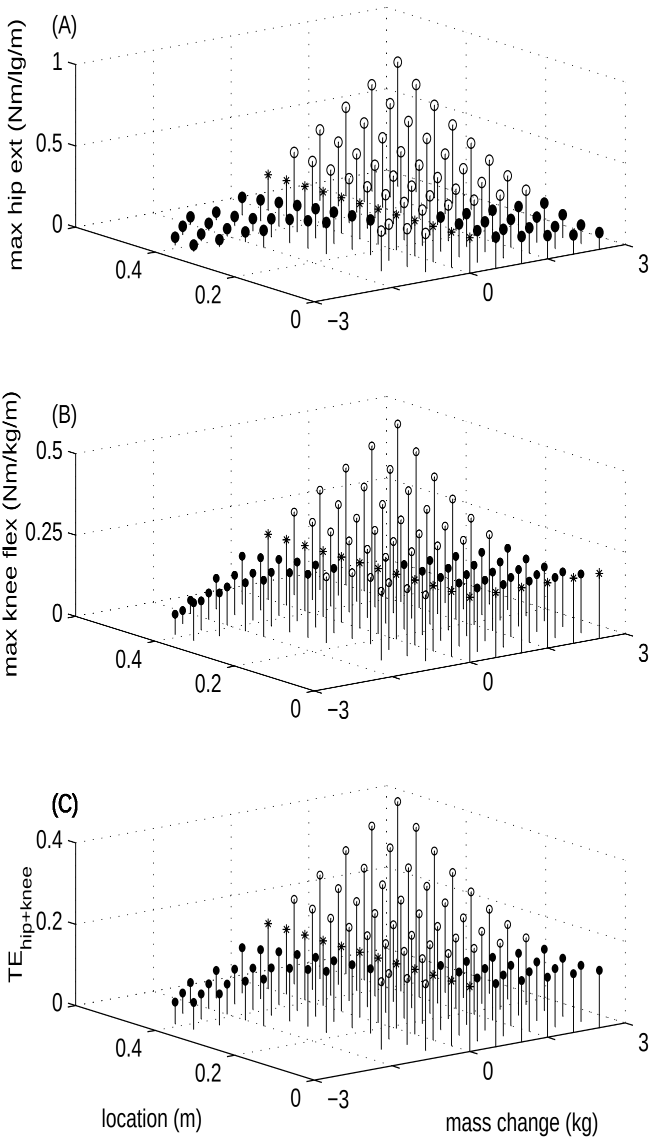

*Figure 6.3. Effects for subject 1 of adding and removing point masses ranging from -2.5 to +2.5 kg from locations between knee and heel on two events derived from the joint torques (maximum hip extension torque (A) and maximum knee flexion torque (B)), as well as $TE_{hip+knee}$ (C). *, zero-mass conditions; o, increase-; •, decrease in the torques and $TE_{hip+knee}$ compared to the zero-mass condition.*

6.4.1. Muscular cost of the swing phase

Figure 6.2 shows a typical example of the normalized hip and knee torques during the unperturbed condition, as well as during four simulated conditions. Removing mass at the heel decreased the maximum flexion and extension torque in hip and knee compared to the zero-mass condition, whereas adding mass at the heel increased these variables. In contrast, removing mass at the knee increased the flexion and extension torques while adding mass decreased the torques needed to maintain the same movement pattern. The magnitude of the effect of perturbation around the knee was much smaller than at the heel. The same effects were found when evaluating all mass perturbations for the same subject (Figure 6.3): $TE_{hip+knee}$ decreased when mass was removed from the heel or added to the knee; in contrast, it increased when mass was removed from the knee or added to the heel. Again, it should be noted that the effects of mass perturbations in the distal part are much larger than in the proximal part of the lower leg.

Table 6.2. Maximum changes in the muscular cost of the swing phase that can be obtained through mass perturbations of maximally 2.5 kg.

TTA subject	**$TE_{hip+knee}$**			
	Maximum decrease (%)		**Maximum increase (%)**	
	Adding mass	**Removing mass**	**Adding mass**	**Removing mass**
1	-13	-69	152	5
2	-16	-62	161	7
3	-9	-55	319	2
4	0	-51	187	0
5	-12	-54	201	9
6	-12	-70	166	4
7	-20	-70	209	5
8	-2	-41	226	2
9	-3	-51	200	2
10	-5	-49	158	2
Mean	-9.2	-57	198	4

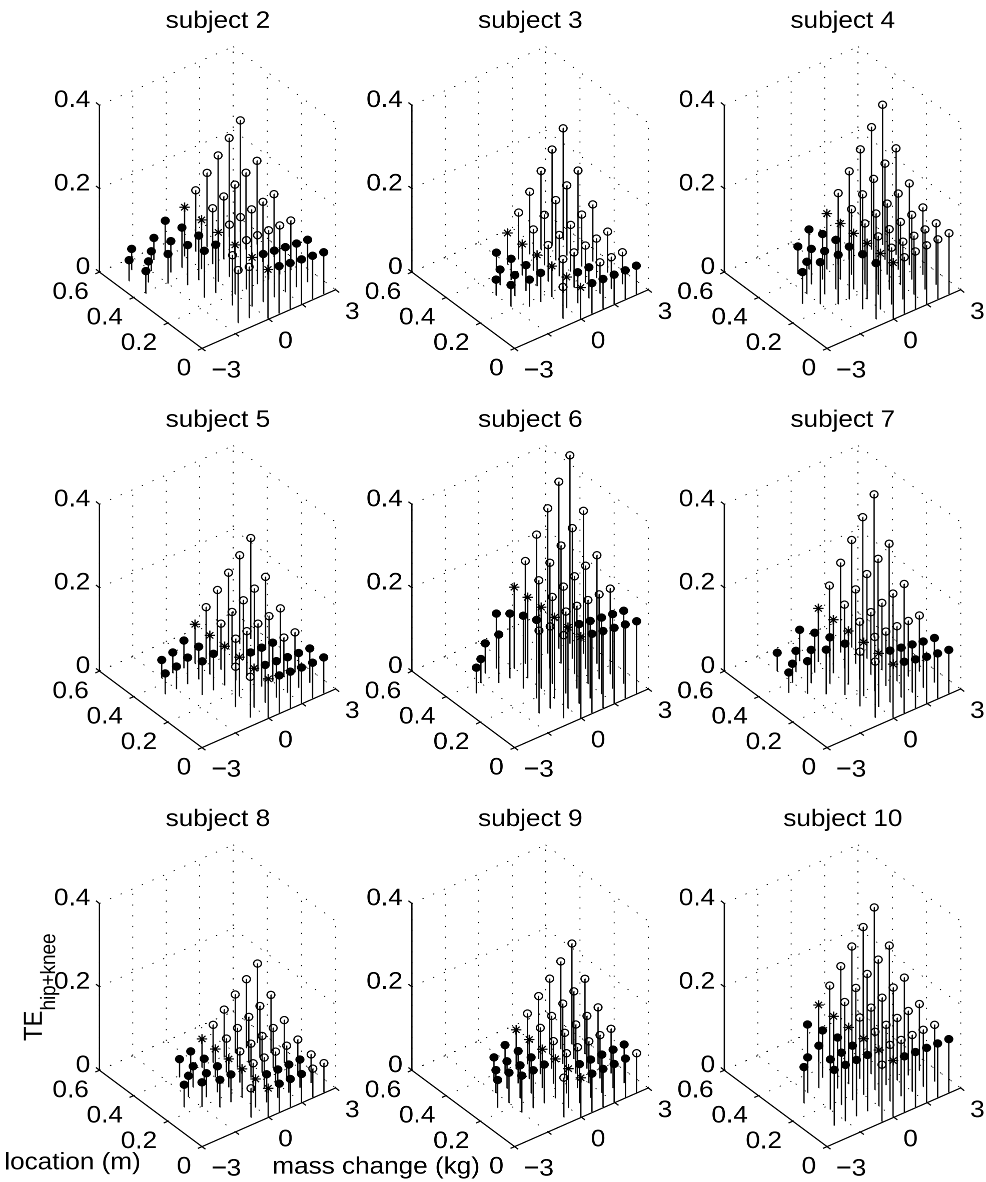

*Figure 6.4. Graphs similar to Figure 6.3C for the remaining nine subjects, i.e., the effect of adding and removing point masses ranging from -2.5 to +2.5 kg from locations between knee and heel on $TE_{hip+knee}$. For visualization, only half of the simulated locations are shown. *, zero-mass conditions; o, increase-; • decrease in $TE_{hip+knee}$ compared to the zero-mass conditions.*

Comparing the effect of mass perturbation for all individual subjects (Figure 6.3C for subjects 1 and Figure 6.4 for the other nine subjects), differences were found in the number of mass perturbations possible without obtaining unrealistic physical properties of the total segments, related to the initial anthropometric values of residual limb and prosthesis. In terms of the effect of the mass perturbation on $TE_{hip+knee}$, most subjects showed the same four regions in the graphs as in Figure 6.3C. Only one subject (TTA subject 4) showed a pattern in which mass addition always increased and mass removal always decreased $TE_{hip+knee}$. In all subjects, $TE_{hip+knee}$ could be more strongly decreased by removing mass than by adding mass (Table 6.2).

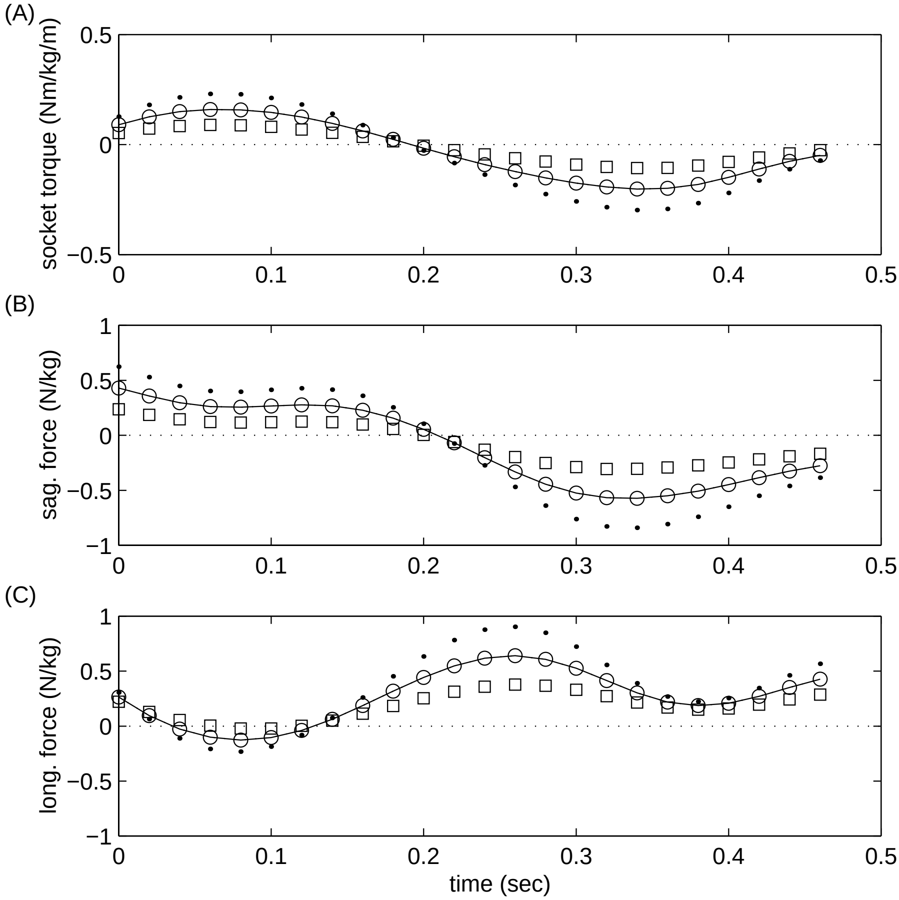

Figure 6.5. Typical examples of the net joint torque (A), and the sagittal- (B) and longitudinal (C) joint reaction force between stump and socket during the swing phase for subject 1 in five simulated conditions: —, unperturbed condition; --, adding 1 kg on the residual limb just below the knee; o, removing 1 kg from the residual limb just below the knee; •, adding 1 kg to the heel; □, removing 1 kg from the heel.

6.4.2. Forces between stump and socket

Figure 6.5 shows a typical example of the net joint torque at the stump-socket interface as well as the resultant sagittal and longitudinal reaction forces. As expected, adding weight to the knee did not affect the forces and torque, since only the properties of the stump are perturbed. On the prosthesis, adding mass increased the amplitude of all time series while removing mass decreased the amplitudes (Figure 6.6). Averaged over all subjects (Table 6.3), the maximum decrease that could be obtained by removing mass ranged from 61% to 86% for the three variables. The increase that could be obtained by adding mass ranged from 127% to 153%.

Table 6.3. Maximum changes possible in the peak longitudinal (F_{long}) and sagittal (F_{sag}) forces between stump and socket as well as the TE_{stump} during the swing phase that can be obtained through mass perturbations of maximally 2.5 kg.

TTA subject	**Maximum decrease (%)**						**Maximum increase (%)**					
	Adding mass			**Removing mass**			**Adding mass**			**Removing mass**		
	F_{long}	F_{sag}	TE_{stump}	F_{long}	F_{sag}	TE_{stump}	F_{long}	F_{sag}	TE_{stump}	F_{long}	F_{sag}	TE_{stump}
1	0	0	0	-88	-92	-70	103	117	116	0	0	0
2	0	0	0	-81	-91	-94	91	102	118	0	0	0
3	0	0	0	-75	-85	-55	161	245	255	0	0	0
4	0	0	0	-80	-81	-51	150	157	129	0	0	0
5	0	0	0	-84	-88	-52	124	140	122	0	0	0
6	0	0	0	-88	-89	-55	130	142	128	0	0	0
7	0	0	0	-86	-91	-87	124	145	146	0	0	0
8	0	0	0	-85	-90	-68	135	186	175	0	0	0
9	0	0	0	-77	-79	-55	131	160	141	0	0	0
10	0	0	0	-68	-69	-26	118	136	132	0	0	0
Mean	0	0	0	-81	-86	-61	127	153	146	0	0	0

6.5. Discussion

The aim of the present study was to develop a new method to predict the optimal prosthetic inertial properties for TTA subjects which is not based on the assumption of a ballistic swing phase. Assuming invariant kinematics during the swing phase, the effects of a systematic set of mass perturbations were evaluated in terms of the net joint torques in hip and knee as well as the net joint reaction forces and torque between socket and residual limb needed to obtain the measured swing phase kinematics.

6.5.1. Muscular cost of the swing phase

In contrast to the forces between socket and residual limb, in the muscular cost of the swing phase, not only the size but also the direction of the effect of mass perturbation depends on the location of perturbation. In 9 out of 10 subjects, muscular cost decreased when removing mass distally as well as when adding mass proximally. On the other hand, muscular cost increased after adding mass distally and removing mass proximally. The magnitude of the effects was much larger when mass was perturbed distally compared to proximally (Figure 6.4 and Table 6.2). In one subject (Figure 6.4), adding mass always increased the muscular cost of the swing phase.

The validity of the predictions in this study is related to the assumptions made. One of the main assumptions was that TTA subjects adapt to mass perturbations by maintaining the same kinematic pattern: the kinematical invariance strategy.[61] The data on which this assumption is based have been discussed more extensively elsewhere.[61] In addition, because the predictions are based on inverse dynamical simulation, the limitations of inverse dynamical calculations (e.g., [70, 86]) also apply to our data. Another limitation relates to the measure of muscular cost. Ideally, the cost of the swing phase would be evaluated in terms of the total cost of all muscle action involved. However, this is presently considered difficult or impossible in normal subjects and may be even more difficult in amputation subjects, in which a standard musculoskeletal model may not apply. In the present study, muscular cost was estimated using $TE_{hip+knee}$. A comparison with three alternative measures for quantifying mechanical cost of the swing phase (Figure 6.7) revealed that the main conclusion was not affected by the chosen measure.

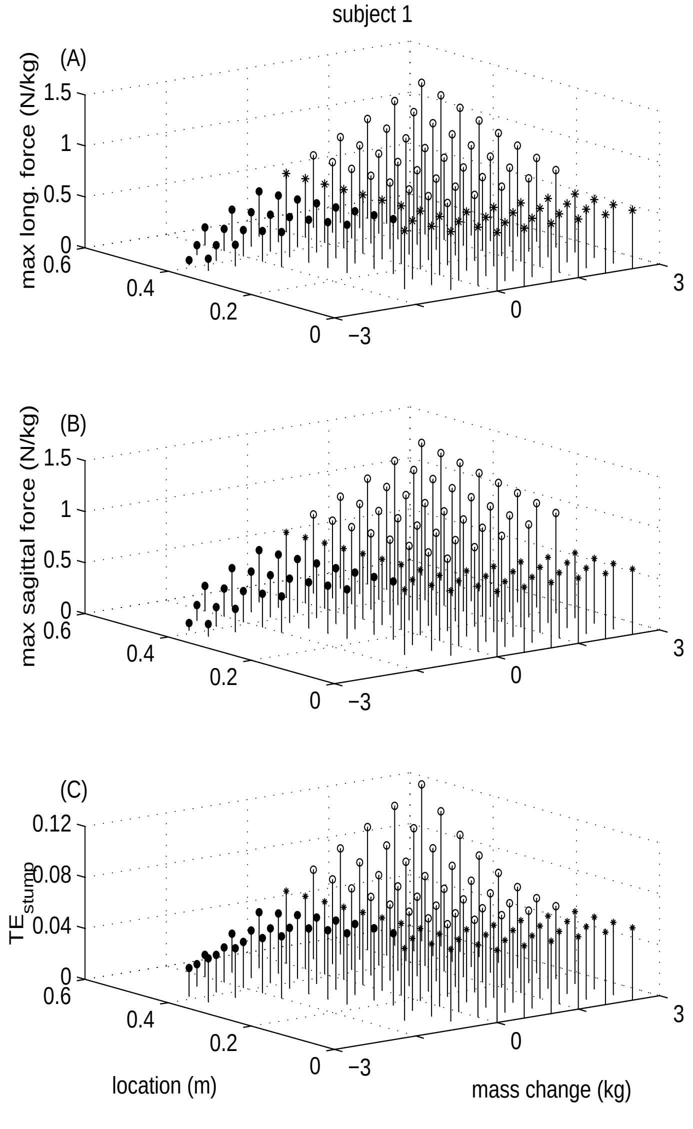

*Figure 6.6. Effects of adding and removing point masses ranging from -2.5 to +2.5 kg from locations between knee and heel on two events derived from the joint reaction force time series (maximum longitudinal (A) and sagittal (B) joint reaction forces) and $TE_{stump\text{-}socket}$ (C). The forces are normalized to body weight, the torques to body weight and leg length. *, zero-mass conditions; o, increase-; •, decrease in the net joint reaction forces and $TE_{stump\text{-}socket}$ compared to the zero-mass condition.*

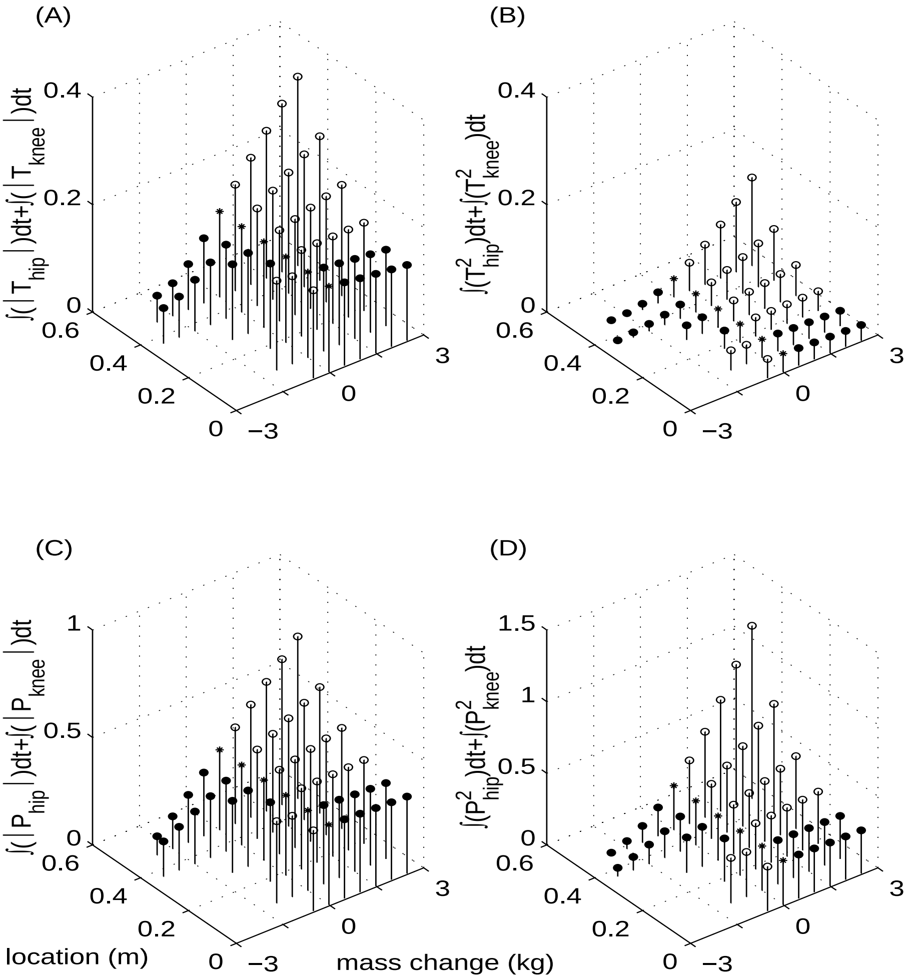

Figure 6.7. $TE_{hip+knee}$ (A) and three alternative estimates (B-D) of the total muscular cost of the swing phase for subject 1. The integrals indicate the time integral from toe-off to heel strike. For visualization, only half of the simulated locations are shown. Abbreviations: T_{knee}, knee torque; T_{hip}, hip torque; P_{knee}, knee power; P_{hip}, hip power.

In the present study, we did not experimentally test the simulation predictions. However, there is a large body of literature on mass perturbations of lower limb prostheses with which to compare our data. In a literature review on prosthetic mass perturbations, Selles et al.[59] reported significant effects of mass perturbations in two of the five studies on the economy of gait, one reporting a positive effect of adding mass, the other a negative effect. These conflicting results may be explained by finding that the effect of mass perturbation depends on the location of the perturbation, as well as by differences in effects found between subjects. Recently, Mattes et al.[41] investigated the effect of matching the prosthetic inertial properties of TTA subjects to the inertial properties of their contralateral leg. On average, 1.7 kg was added relatively distal to the lower leg. As a result, step length was not influenced by mass addition, while swing time significantly increased by about 4%, the latter finding in contrast with the kinematical invariance strategy. In addition, metabolic cost per second significantly increased by about 7%. Simulating lower leg inertial symmetry in our data confirmed this increased metabolic cost: $TE_{hip+knee}$ increased in all subjects by, on average, 25% (SD 18). It should be noted that the percentages can not be compared directly because the 25% in our study concern the increase in muscular cost of only the swing phase of the prosthetic leg.

6.5.2. Forces between stump and socket

The effects of the mass perturbations on the stump-socket interface were relatively straightforward, that is, forces and torque always decreased when mass was removed and increased when mass was added. As in the muscular cost, the magnitude of the effects was larger when mass was perturbed more distally.

The analysis of the effects of mass perturbations on the stump-socket interface should be considered a first estimate, indicating only the resultant forces, which are the combined effect of all shear and pressure forces acting on many locations. Currently, little is known about the forces between socket and stump and how they relate to skin problems such as blisters and sores. Most of the experimental studies, as well as the finite element models, on pressure and shear inside a socket have focused only the stance phase. An estimate of the forces during the swing phase can be found in a study by Appoldt et al.,[1] showing that pressures measured at several locations at the stump during gait in above-knee amputees were low compared to the stance phase. In contrast, a more recent study by Williams et al.,[85] measuring both pressure and shear indicates that the forces during swing phase are significant, although lower than during the stance phase.

In the present study we did not make assumptions on how the resultant forces and torques are distributed over the residual limb, since this may strongly depend on the socket type used. For example, a liner locked into the socket with a pin may distribute the resultant forces and torques very differently over the residual limb than a PTB prosthesis in which the locking during the swing phase is mainly obtained by pressing the socket over the condyles of the knee. The aim of the present study was to show that, in contrast to the effects on muscular cost, the resultant forces always increased when adding mass and decreased when removing mass. Future studies should investigate how these forces are distributed and whether the magnitude of the effects of mass perturbation is relevant for prosthetic design.

6.5.3. Implications for prosthetic design

The aim of the method developed in this study was to clarify the relation between prosthetic inertial properties on the one hand and the muscular cost of the swing phase and the forces in the stump-socket interface on the other hand.

In terms of the forces in the stump-socket interface, this study suggests that more heavyweight prosthetic designs will increase the total resultant forces and torques. Whether the magnitude of the effect has a clinical relevance remains to be answered. In terms of muscular cost, the shaded areas in Figures 3C and 4 indicate the mass perturbations that decrease the total muscular cost of the swing phase. Despite the individual differences, a general pattern can be seen. In most subjects, only very small decreases in muscular cost can be obtained by adding mass, while adding mass at the wrong locations, that is, distally, largely increases the muscular cost. This is illustrated in Figure 6.8, in which the same data as in Figure 6.3C are translated into specific manipulations that may be relevant for prosthetic design or prescription. In general, it can be seen that component mass may not be important for proximally located components (e.g., socket, liners, locks), while distally located components (e.g., foot, ankle, shoes) strongly affect muscular cost.

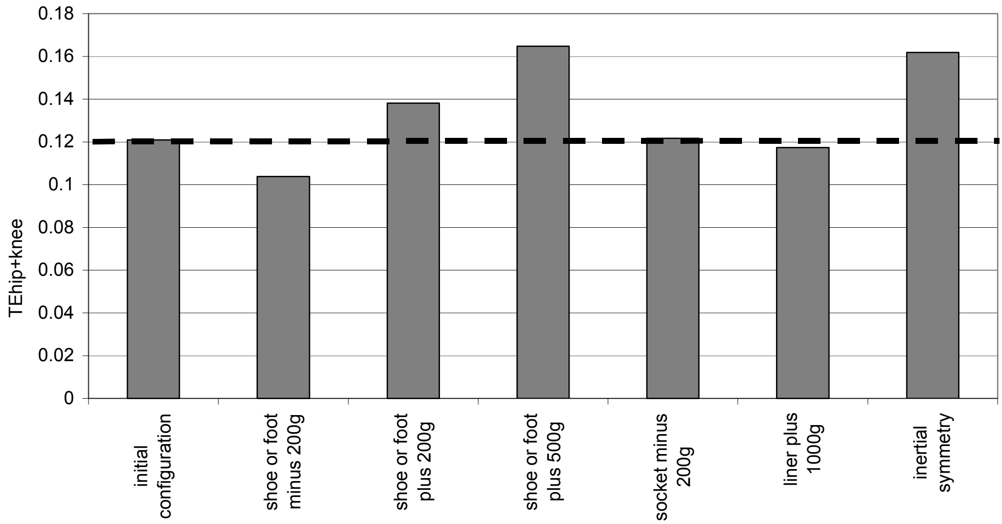

Figure 6.8. Transformation of the data of Figure 6.3C to indicate the effects of typical alterations of prosthetic components on the muscular cost ($TE_{hip+knee}$) of the swing phase. Alterations in foot and shoe mass were modeled as point mass perturbations at the heel, while the effect of socket and liner alterations were modeled as point mass perturbations halfway the residual limb. The inertial symmetry was modeled based on the contralateral lower leg mass, center of mass location and moment of inertia.

7. Mass perturbation of below-knee prostheses: analysis of the equations of motion and effects of anthropometric properties and walking speed

7.1. Abstract

The aim of this study was to (1) clarify why in most TTA subjects proximal mass addition decreases the muscular cost of the swing phase while distal mass addition increases the muscular cost, (2) clarify why in some TTA subjects, proximal mass addition increases the muscular cost, and (3) determine whether the effect of mass perturbation is influenced by individual differences in body dimensions and walking speed. From a group of ten TTA subjects, anthropometric and kinematic data were measured at self-selected and fast walking speed. The equations of motion of the prosthetic leg were derived and the values of the equation of motion components (EMCs) during the swing phase were calculated. Finally, the effect of mass perturbations was compared at self-selected and fast walking speed. It was found that for proximal mass addition, positive and negative effects on the joint torques were practically balanced. This resulted in a small increase in muscular cost in some subjects, and a small decrease in cost in some others. The change in hip and knee torque time series did not depend on initial anthropometric properties, but was only determined by the swing phase kinematics and the size and location of the mass perturbation. For distal mass perturbation, this led to a linear increase in amplitude of the torques. For proximal mass perturbation, small mass addition first decreased the amplitude of the torques. However, for larger mass perturbations, the torques changed from flexion to extension and vice versa and the amplitude increased again. The large mass (about 15 kg in the subject studied) needed to obtain the minimum amplitude and, therefore, the minimum muscular cost, makes this configuration clinically irrelevant. Finally, in this study, we found qualitatively similar effects of mass perturbation at self-selected and fast walking speed, indicating that, for the velocities studied, the same prosthetic inertial properties are beneficial.

7.2. Introduction

In a study (Chapter 6) predicting the effect of lower leg mass perturbation in transtibial amputation (TTA) subjects, we found in all subjects that distal mass addition increased and distal mass removal decreased the muscular cost of the swing phase (see also Figure 7.2A). However, for proximal mass addition, in these subjects, we found the opposite: proximal mass addition decreased the muscular cost, while proximal mass removal increased the muscular cost. While the effect of proximal mass addition in nine of the ten subjects was opposite to the effect of distal mass addition, in one subject, both proximal and distal mass addition increased the muscular cost of the swing phase.

The analysis in Chapter 6 did not explain why proximal mass addition decreases the muscular cost of the swing phase in most subjects. Although several recent studies have indicated that the swing phase can not be understood as completely ballistic, the pendulum-like behavior of the leg may account for this pattern. Van Soest et al.[77] studied the effect of mass perturbation on the undamped natural frequency of a single segment such as a lower leg and showed that proximal mass addition increases the natural frequency of the leg, while distal mass addition decreases this natural frequency. As a result, changing leg inertia can change the natural frequency of the leg, which will change the amount of control needed to obtain the desired movement pattern.

The analysis in Chapter 6 also did not explain the nature of the individual differences in the effect of mass perturbation. Since the joint torque during the swing phase is determined by its kinematics as well as by the segment inertial properties, individual differences in kinematical variables (e.g., walking speed, joint angles) as well as in anthropometric variables (e.g., body length, body mass, stump length) may account for this. Establishing the nature of the individual differences is important for clinical practice. If anthropometric differences between subjects influence the effect of mass perturbation and therefore the optimal prosthetic mass, it may be possible to determine for each subject the optimal prosthetic inertial properties based on the subject's body dimensions in combination with the unperturbed prosthetic inertial properties. If differences in walking speed and gait kinematics influence the effect of mass perturbation, a gait analysis may be necessary to determine the optimal prosthetic inertial properties for individual subjects.

The aim of the present study is to clarify the above-mentioned issues. First, we will analyze why in most TTA subjects proximal mass addition decreases the muscular cost of the swing phase while distal mass addition increases the muscular cost. Then, we will apply a similar analysis to the data of a subject in which both proximal and distal mass addition increases the muscular cost. Finally, we will study whether the effect of mass perturbation is influenced by individual differences in body dimensions and swing phase kinematics.

7.3. Methods

Data measurement and part of the data analysis have been described earlier (see Chapter 5) and will be summarized here.

Ten TTA subjects participated in the study. The subjects were included if they could walk without assistance for a prolonged period, had no cardiopulmonary, neurological or orthopedic disorders other than their amputation, and had no skin problems of the residual limb. All subjects were measured at least one year after discharge from the rehabilitation program. The hospital’s Medical Ethical Commission approved the study and all subjects signed an informed consent.

7.3.1. Measurements

The subjects walked in a gait analysis lab wearing their preferred walking shoes in two different conditions. In the first condition, subjects were asked to walk at their self-selected walking speed, in the second, they were asked to walk faster without running. How much faster the subjects needed to walk was not specified. Lower extremities kinematics were recorded with a 50 Hz, three camera ProReflex infrared system (Qualisys, Sweden) with reflective markers on the greater trochantor, the lateral femoral condyle and on the prosthetic equivalent of the lateral malleolus location. Average marker trajectories were calculated for each subject based on 6 complete gait cycles and the swing phase was selected based on the kinematic data. For each subject, height, weight and the lengths of thigh and shank were measured and used to calculate mass, center of mass and moment of inertia of the thigh.[86] The anthropometric properties of the residual limb were calculated based on a geometric model.[31, 89] Weighing, balancing on a straight edge and a pendulum test determined the prosthetic anthropometric properties.[24]

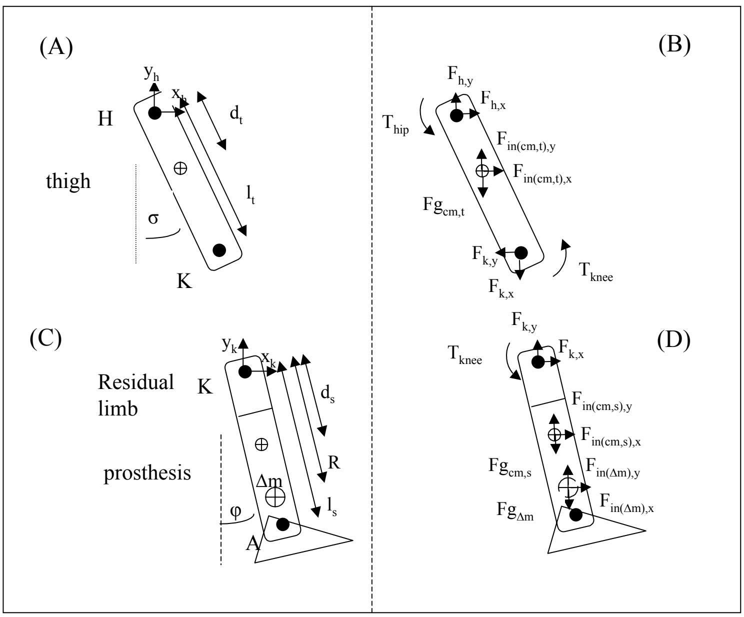

Figure 7.1. Parameters used to describe thigh (A) and shank (combined residual limb plus prosthesis, C) segments and the mass perturbation. The forces and torques acting on both segments are shown in Figures B and D, respectively. For the thigh, the frame of reference moves with the hip, for the shank, the frame of reference moves with the knee. For symbols: see Appendix A, page 117.

7.3.2. Simulations

Simulations were performed to evaluate the effect of mass perturbations of the lower leg on the swing phase of the subjects. The simulations are similar to those described in Chapter 6. A linked segment model of the prosthetic leg was used with the prosthesis (foot and shank), residual limb and thigh as separate segments (Figure 7.1). We assumed the relative motion between stump and prosthesis to be negligible. The simulations are based on inverse dynamics calculations in which net joint torques are calculated from the measured kinematics and the segment inertial properties. Based on the assumption that kinematics do not change when mass is perturbed (see Chapter 5), the effect of mass perturbations is evaluated for each subject in each mass condition based on the kinematic data measured without mass perturbation.

Mass perturbation was simulated by adding and removing point masses in steps of 0.5 kg from minus 2.5 kg (removing mass) to plus 2.5 kg (adding mass) from locations varying in steps of 10% from directly on the knee to the heel. In the analysis, only conditions were considered in which the perturbed segments (residual limb or prosthesis) had a positive mass and positive moment of inertia relative to the segment center of mass.

From the simulations, the muscular cost in hip (TE_{hip}) and knee (TE_{knee}) joint separately and the total muscular cost ($TE_{hip+knee}$) in the swing leg were estimated using:

$$TE_{hip} = \int_{to}^{hs} (|T_{hip}|)dt \tag{7.1.1}$$

$$TE_{knee} = \int_{to}^{hs} (|T_{knee}|)dt \tag{7.1.2}$$

$$TE_{hip+knee} = \int_{to}^{hs} (|T_{hip}|)dt + \int_{to}^{hs} (|T_{knee}|)dt \tag{7.1.3}$$

where T_{hip} and T_{knee} are the normalized net joint torques in knee and hip and *to* and *hs* refer to the time of toe-off and heel strike.

7.3.3. Equations of motion

The equations of motions for thigh and shank were derived in which, for the shank, the origin of the frame of reference translates with the knee joint, while for the thigh, the origin moves with the hip. The parameters are defined in Figures 7.1A,C; the forces and net torques acting on both segments are shown in Figures 7.1B,D.

For the perturbed system, the rotational equation of motion for the shank (residual limb plus prosthesis) relative to knee is (taking g positive) is

$$M_{Fg,s} + M_{Fin,s,x} + M_{Fin,s,y} + T_{knee} = I_s \cdot \ddot{\varphi} \tag{7.2}$$

with

$$M_{Fg,s} = -(m_s \cdot d_s + \Delta m \cdot R) \cdot g \cdot \sin(\varphi)$$

$$M_{Fin,s,x} = -(m_s \cdot d_s + \Delta m \cdot R) \cdot \ddot{x}_k \cdot \cos(\varphi)$$

$$M_{Fin,s,y} = -(m_s \cdot d_s + \Delta m \cdot R) \cdot \ddot{y}_k \cdot \sin(\varphi$$

$$I_s = I_{s,k} + \Delta m \cdot R^2$$

in which $I_{s,k}$ is the moment of inertia of the shank around the knee (for all symbols, see Appendix A, page 117). The equation of motion for the unperturbed system is obtained by setting Δm to zero.

The equation of motion of the thigh is indirectly influenced by mass perturbation of the lower leg by means of changes in $T_{knee,t}$, $M_{fk,x}$ and $M_{Fk,y}$ in the rotational equation of motion for the thigh relative to hip:

$$M_{Fg,t} + M_{Fin,t,x} + M_{Fin,t,y} + M_{Fk,x} + M_{Fk,y} + T_{knee,t} + T_{hip} = I_{t,h} \cdot \ddot{\sigma} \tag{7.3}$$

with

$$M_{Fg,t} = -m_t \cdot d_t \cdot g \cdot \sin(\sigma)$$

$$M_{Fin,t,x} = -m_t \cdot d_t \cdot \ddot{x}_h \cdot \cos(\sigma)$$

$$M_{Fin,t,y} = -m_t \cdot d_t \cdot \ddot{y}_h \cdot \sin(\sigma)$$

$$M_{Fk,x} = -F_{k,x} \cdot l_t \cdot \cos(\sigma)$$

$$M_{Fk,y} = -F_{k,y} \cdot l_t \cdot \sin(\sigma)$$

$$T_{knee,t} = -T_{knee}$$

and

$$F_{k,x} = m_t \cdot \ddot{x}_{cm,s} + \Delta m \cdot \ddot{x}_{dm}$$

$$F_{k,y} = m_t \cdot \ddot{y}_{cm,s} + \Delta m \cdot \ddot{x}_{dm} - (m_t + \Delta m) \cdot g$$

To analyze the opposite effects of proximal and distal mass addition as well as the individual differences in the effect of proximal mass addition, we compared the values of the equation of motion components (EMCs) during the unloaded condition with two loaded conditions. In the first loaded condition, we simulated proximal mass addition of 1.5 kg at 20% of the knee-ankle length (Δm=1.5, R=0.2l_s), in the second condition, we simulated distal loading of 1.5 kg at the ankle (Δm=1.5, R=l_s).

To evaluate how the two loaded conditions influence the knee and hip torque, equation 7.2 and 7.3 were rewritten to

$$T_{knee} = I_s \cdot \ddot{\varphi} + (-M_{Fg,s}) + (-M_{Fin,s,x}) + (-M_{Fin,s,y}) \tag{7.4.1}$$

and

$$T_{hip} = (-T_{knee,t}) + (-M_{Fk,x}) + (-M_{Fk,y}) \tag{7.4.2}$$
$$(+I_{t,h} \cdot \ddot{\sigma} - M_{Fg,t} - M_{Fin,t,x} - M_{Fin,t,y})$$

7.3.4. Statistical analysis

The larger part of this study will be a qualitative description based on two subjects selected as typical examples from the total group of ten TTA subjects. In the last part of this study, comparing the effect of mass perturbation at self-selected and fast walking speed, for each ten subjects on each walking speed, two curves were fitted from the simulations of the mass perturbations (see Figure 7.8). The first curve describes all mass perturbation with R=0.8*l_s, the second curves all perturbations with Δm=1.5. The following second order polynomials were fitted to the curves:

For $R = 0.8 * l_s$:

$$TE_{hip+knee} = a + b \cdot \Delta m + c \cdot \Delta m^2 \tag{7.5.1}$$

For $dm = 1.5kg$:

$$TE_{hip+knee} = d + e \cdot R + f \cdot R^2 \tag{7.5.2}$$

A Wilcoxon sign rank test was used to compare the components (*a-f*) at self-selected and fast walking speed. A level of significance of 0.05 was used.

7.4. Results

Table 7.1 presents the characteristics of the subjects. All subjects used an ICEROSS Comfort liner with a pin lock and either an ICEX pressure cast socket or a regular cast. In addition, all subjects used an energy storing foot except for subjects 4 and 5 who used a Multiflex foot. From these subjects, subject 1 was selected as a typical example in which proximal mass addition decreased the muscular cost, while subject 2 was selected as a typical example in which both proximal and distal mass addition increased the muscular cost of the swing phase (see Figure 7.2).

Table 7.1. Characteristics, prosthetic mass as well as the self-selected and increased walking speed of the transtibial amputation subjects. Abbreviations: M, male; F, Female; SSWS, self-selected walking speed.

Subject	Age (yr)	Gender (male/ female)	Weight (kg)	Height (m)	Prosthetic mass (kg)	SSWS (m/s)	Fast walking speed (m/s)
1	52	M	77	1.72	3.0	0.97	1.12
2	33	F	100	1.72	2.7	1.38	1.79
3	51	M	73	1.86	2.9	1.18	1.40
4	78	M	75	1.78	2.5	0.93	1.15
5	68	M	86	1.82	2.3	0.77	0.94
6	61	F	85	1.63	2.0	1.16	1.36
7	71	M	72	1.57	2.9	1.34	1.54
8	61	M	94	1.78	2.4	1.21	1.43
9	53	M	79	1.78	2.3	1.02	1.27
10	72	F	66	1.56	1.6	1.03	1.02
Mean (SD)	61 (13)		82 (10.6)	1.74 (0.10)	2.6 (0.4)	1.11 (0.19)	1.33 (0.26)

7.4.1. Proximal and distal mass addition: opposite effects on muscular cost

The left side of Figure 7.2 shows the muscular cost of subject 1 in each simulated mass condition. Separate analysis of hip and knee (Figures 7.2C&E) showed a similar pattern in the individual joints as in the total muscular cost (Figure 7.2A), that is, proximal mass addition decreased and distal mass addition increased the total muscular cost of the swing phase. The EMC values of equation 7.2 in the unloaded condition (Δm=0) for subject 1 are visualized in the left part of Figure 7.3. The four components of the shank that add up to $I \cdot \ddot{\varphi}$ are generally in the same direction (Figure 7.3A), indicating that the joint torque (T_k) accelerates the shank in a direction similar to the gravitational (M_{Fg}) and inertial forces ($M_{Fin,s,x}$ and $M_{Fin,s,y}$). For the thigh (equation 7.3; Figure 7.3C), some of the EMCs are in opposite directions and $I \cdot \ddot{\sigma}$ is, in contrast to the shank, relatively small.

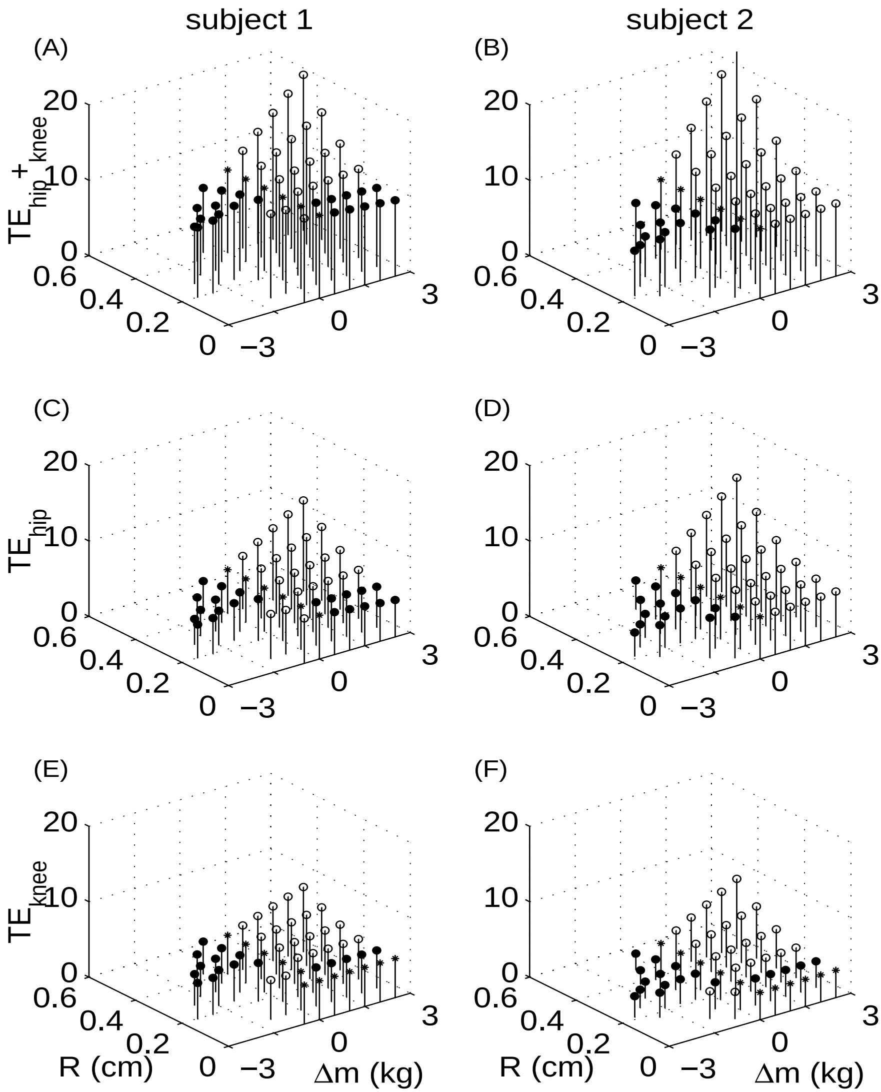

*Figure 7.2. Effects of adding and removing point masses (Δm) ranging from -2.5 to +2.5 kg from location R on $TE_{hip+knee}$, TE_{hip} and TE_{knee} for subject 1 (left side) and subject 2 (right side). For visualization, only half of the simulated locations are shown. *, Δm=0 conditions; o, increase-; •, decrease in TE compared to the zero-mass condition. Mass reductions that led to unrealistic inertial properties of residual limb or prosthesis, such as a negative mass, were excluded.*

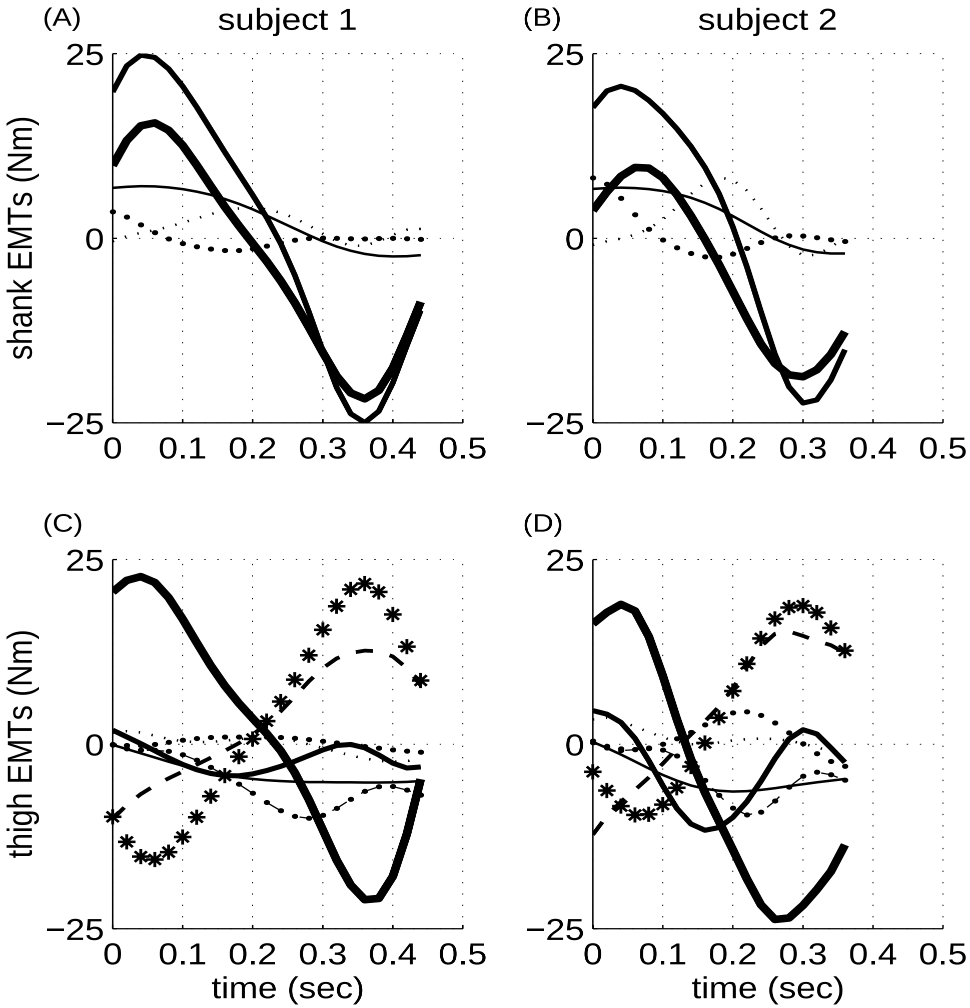

*Figure 7.3 A: EMC values of the shank (equation 7.2) for subjects 1 (left side) and 2 (right side). —, $M_{Fg,s}$; ···, $M_{Fin,s,x}$; ●●●, $M_{Fin,s,y}$; ♉ ,T_{knee} and ▬, $I \cdot \ddot{\varphi}$. B: EMC values of the thigh (equation 7.3) for the same subjects. —, $M_{Fg,t}$; ···, $M_{Fin,t,x}$;●●●, $M_{Fin,t,y}$; — —, M_{Fkx},; —•-, $M_{Fk,y}$,;***, $T_{knee,t}$; ♉T_{hip}; ▬▬, $I \cdot \ddot{\sigma}$.*

For the thigh, the EMC values from equation 7.4.2 are shown on the right side of Figure 7.4. The last four components are not shown because they are not influenced by mass perturbation of the lower leg. Again, the effect of mass perturbation on T_{hip} can be understood as the combined effect in all three EMCs.

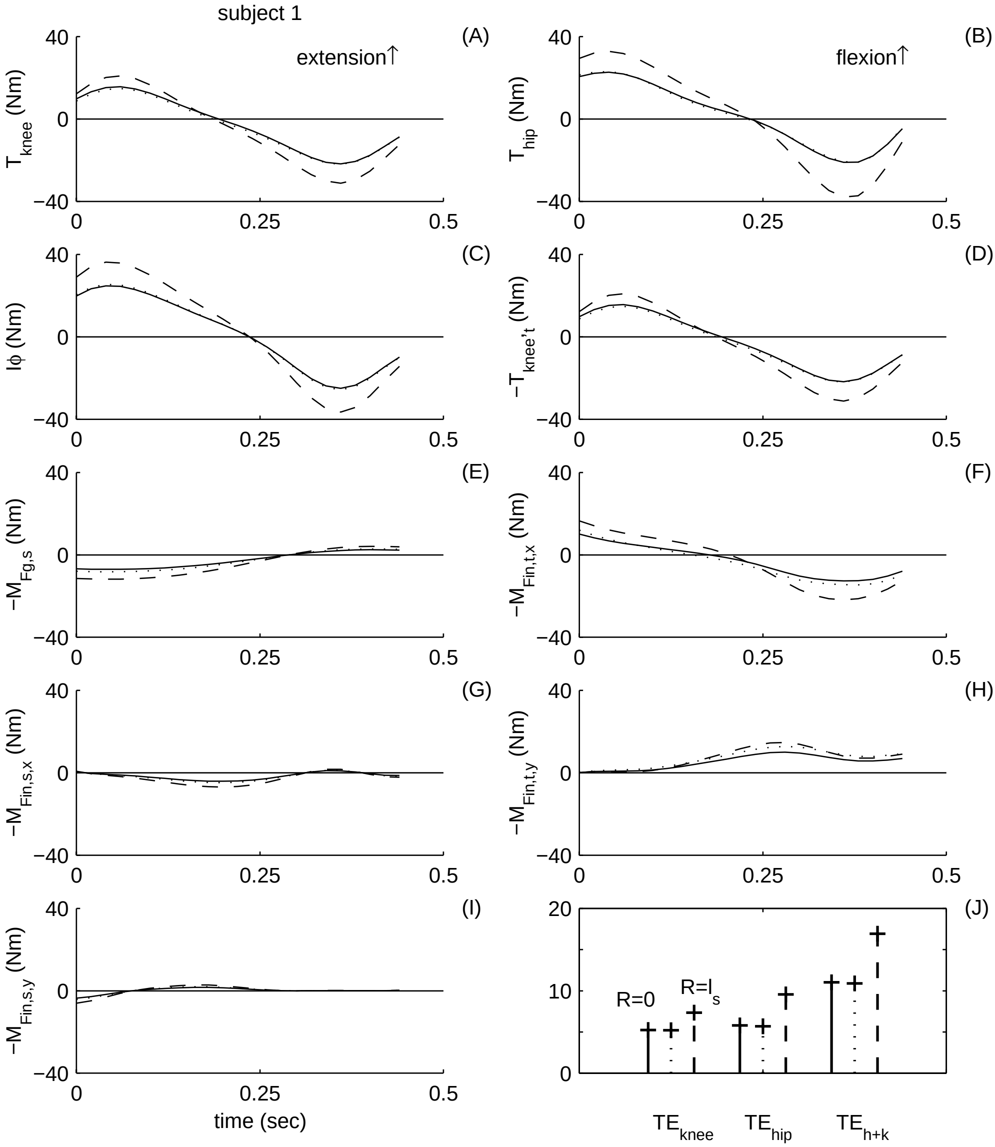

Figure 7.4. EMC values of shank (left side) and thigh (right side) during three mass conditions for subject 1: —, unloaded; ..., proximal loading (Δm=1.5, R=0.2l_s); ---, distal loading (Δm=1.5, R=l_s). Both for shank and thigh, the components are visualized in such a way that the changes add up to the change in T_{knee} and T_{hip} (see equations 7.4.1&7.4.2). The last plot shows TE_{knee}, TE_{hip} and $TE_{hip+knee}$ for the three conditions.

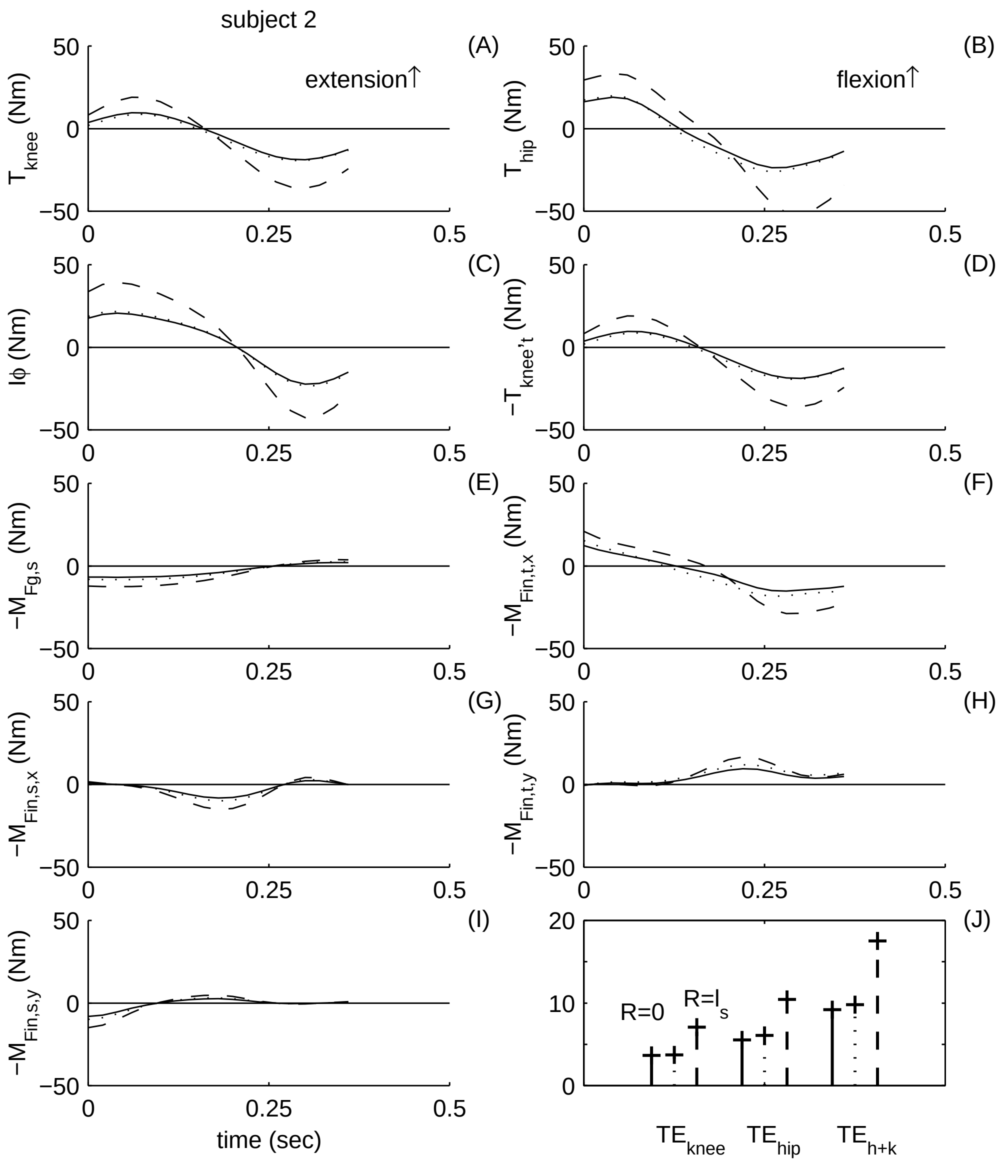

Figure 7.5. EMC values of shank (left side) and thigh (right side) during three mass conditions for subject 2: —, unloaded; ..., proximal loading (Δm=1.5, R=0.2l_s); ---, distal loading (Δm=1.5, R=l_s). Both for shank and thigh, the EMCs are visualized in such a way that the changes add up the to change in T_{knee} and T_{hip} (see equations 7.4.1&7.4.2). The last plot shows TE_{knee}, TE_{hip} and $TE_{hip+knee}$ for the three conditions.

7.4.2. Proximal and distal mass addition: similar effects on muscular cost

Subject 2 was selected as a typical example in which both proximal and distal loading increase the total muscular cost (see Figure 7.2). In this subject, in the knee torque, proximal mass addition decreased the muscular cost, as in subject 1. In the total muscular cost, however, this effect is dominated by the increased muscular cost at the hip. The EMC values of equation 7.2 and 7.3 in the unloaded condition (Δm=0) for this subject are visualized in the right part of Figure 7.3. Qualitatively, the patterns are very similar to subject 1.

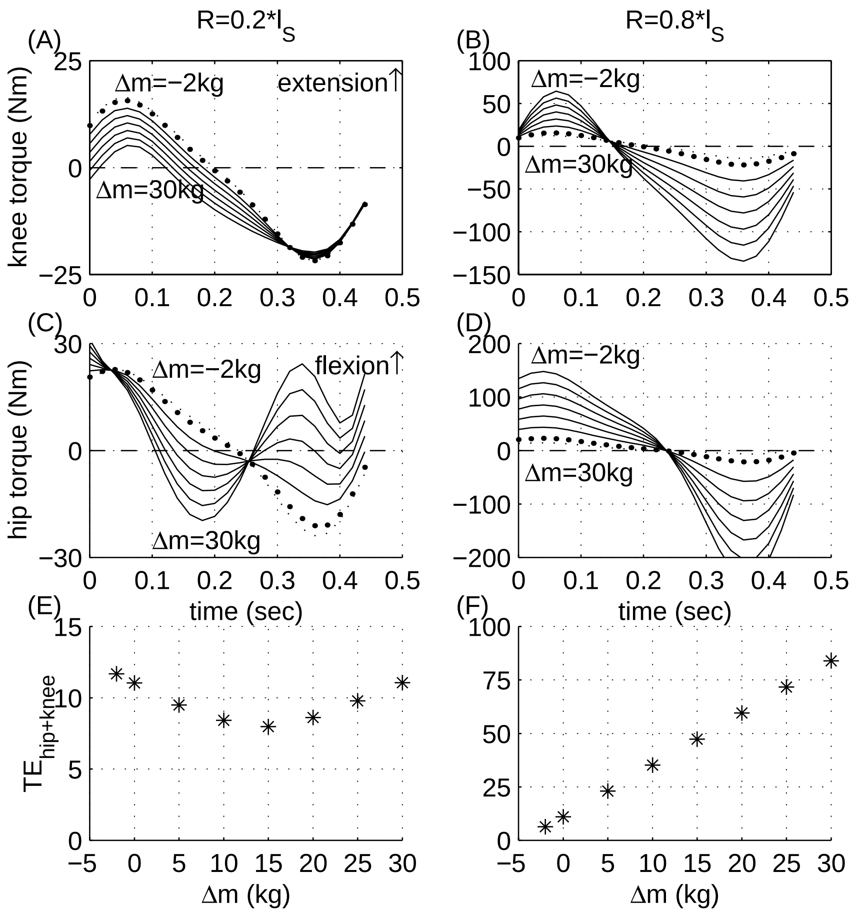

*Figure 7.6. Knee (A-B) and hip (C-D) torque in the unloaded condition (●●●) and simulations of mass removal of 2 kg (--) and 6 mass additions (—; Δm=5, 10, 15, 20, 25 and 30 kg) at a proximal (R=.8*l_s; left side) and distal (R=.8*l_s; right side) location. Mass reduction larger than 2 kg was not simulated because it would lead to a negative prosthetic mass. E-F: $TE_{hip+knee}$ during the simulated mass conditions.*

Figure 7.5 shows the EMC values of thigh and shank (equations 7.4.1 and 7.4.2) during the three conditions for subject 2. Again, qualitatively, the values are very similar to subject 1 (Figure 7.4). Comparing both subjects, it can be seen that the difference in the effect of proximal loading on the total muscular cost between both subjects is the result of small differences in balance between the different EMCs.

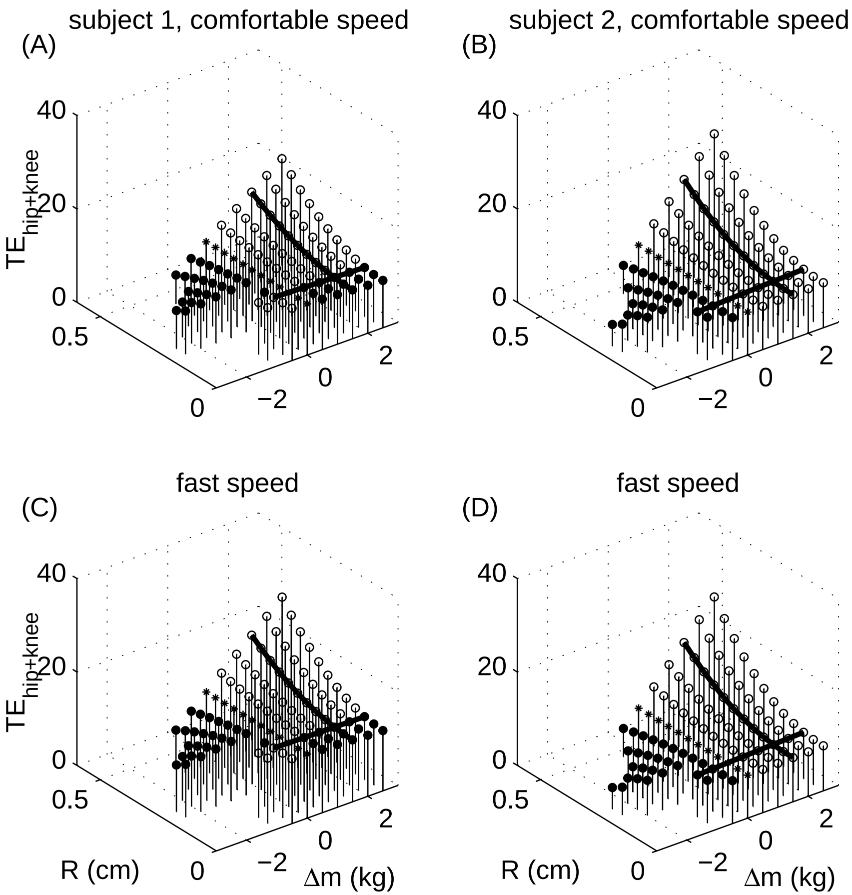

*Figure 7.7. Effects of adding and removing point masses (Δm) ranging from -2.5 to +2.5 kg from location R on $TE_{hip+knee}$ for subject 1 and 2 at self-selected walking speed (upper two) and increased walking speed (lower two). * indicate the Δm=0 conditions, o an increase and • a decrease in TE compared to the zero-mass conditions. The lines indicate all mass perturbations at $R=0.2*l_s$ and all mass perturbations with Δm=1.5.*

Equations 7.2 and 7.3 indicate how the initial anthropometric properties (defined by m_s, d_s, I_s, m_t, d_t, l_t and I_t) and the properties of the mass perturbation (Δm and R), influence the EMCs and therefore the torques in hip and knee. It can be seen that the change in T_{knee} and T_{hip} after mass perturbation depends on Δm and

R, in combination with the kinematical variables $(\varphi, \ddot{\varphi}, \sigma, \ddot{\sigma}, \ddot{x}_k, \ddot{y}_k, \ddot{x}_{dm}, \ddot{y}_{dm}, \ddot{x}_h, \ddot{y}_h)$, but is not influenced by the initial anthropometric properties of thigh and shank. The same can be seen from Figure 7.6, that the hip and knee torque time series change linearly after large proximal and distal mass perturbations.

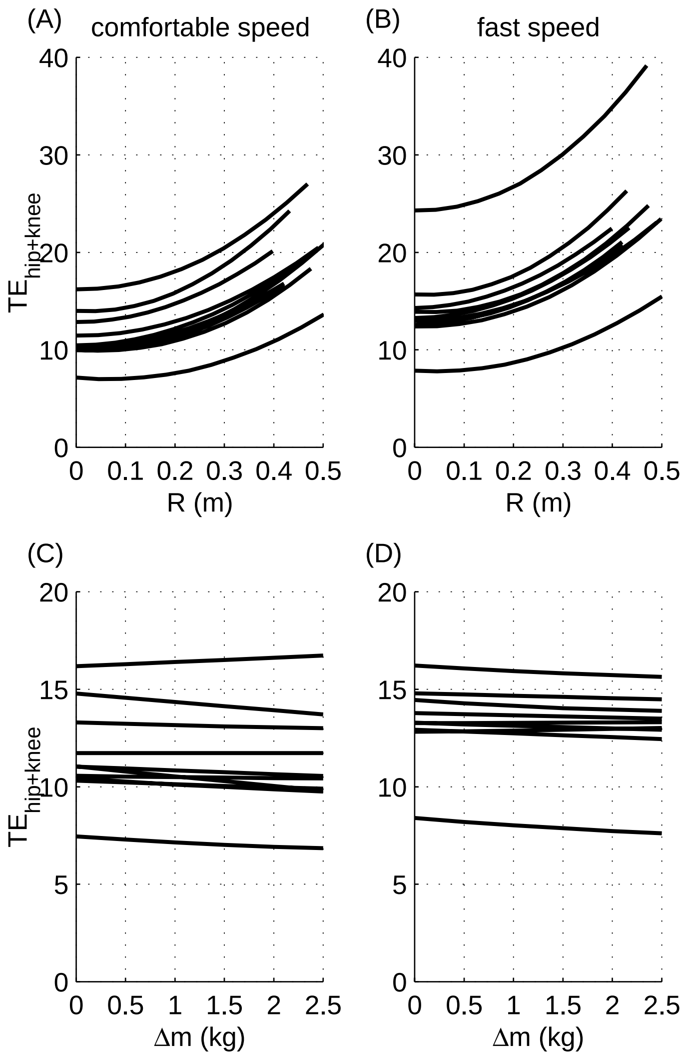

*Figure 7.8. Same lines as in Figure 7.7 for all subjects at self-selected (left side) and fast walking speed (right side). The upper two graphs show $TE_{hip+knee}$ during all mass perturbations with Δm=1.5, the lower two graphs show $TE_{hip+knee}$ during all mass perturbations at R=0.2*l_s.*

When analyzing total muscular cost (Figures 7.6E&F) for the distal mass perturbation ($R=0.8*l_s$), this leads to a linear increase in the muscular cost ($TE_{hip+knee}$) of the swing phase. For the proximal mass perturbation, however, the effect is not linear; a minimum in $TE_{hip+knee}$ was found at about $\Delta m=15$ kg.

The linear effect of distal mass perturbation indicates that the effect of mass perturbation at this location is independent of initial inertial properties and prior mass perturbation: each new perturbation of Δm at R changes $TE_{hip+knee}$ with the same amount in the same direction. For the proximal loading, however, initial inertial properties and prior mass perturbations do influence the effect of mass perturbation, as can be seen from the non-linear trend in $TE_{hip+knee}$.

7.4.3. Effect of walking speed

Since the effect of mass perturbation depends on the kinematics of the swing phase, walking speed may influence the effect of mass perturbation. Figure 7.7 shows the effect of mass perturbation for subject 1 and 2 at self-selected walking speed and fast walking speed. Qualitatively, the effect of mass perturbation is similar at both walking speeds, although $TE_{hip+knee}$ is generally increased at the higher walking speeds. The curves plotted from all mass perturbations with $R=0.8*l_s$ and with $\Delta m=1.5$ for all subjects are shown in Figure 7.8 and the average values of the coefficients of the polynomials fitted on the curves are shown in Table 7.2. It was found that *a* and *d* were increased at the higher walking speed, indicating that $TE_{hip+knee}$ was generally increased. From the other coefficients, only c was significantly increased at the higher walking speed, indicating a steeper incline in muscular cost with more distal location of the same mass at higher walking speeds. The other components were not significantly different between both walking speeds.

*Table 7.2. Average values for the components from the polynomial (equation 7.5.1) describing the relation between TE and R at dm=1.5 from the polynomial (equation 7.5.2) describing the relation between TE and dm at $R=0.2*l_s$. Group mean and standard deviation are given at self-selected and fast walking speed. P-values indicate the statistical test comparing self-selected and fast walking speed. * indicates a statistically significant difference between the coefficients at self-selected and fast walking speed.*

	a	b	c	d	e	f
Mean self-selected walking speed (SD)	10.9 (2.6)	-1.0 (2.4)	44.6 (8.4)	11.4 (2.5)	-.3 (.2)	.03 (.03)
Mean fast walking speed (SD)	13.8 (4.3)	-0.9 (2.4)	51.9 (11.4)	14.2 (4.0)	-.25 (.25)	.03 (.03)
p-value	.005*	.799	.005*	.005*	.114	.385

7.5. Discussion

The aim of the present study was to analyze why in most TTA subjects proximal mass addition decreases the muscular cost of the swing phase while distal mass addition increases the muscular cost. In addition, we studied why in some subjects, both proximal and distal loading increase the muscular cost. Finally, we investigated how the initial inertial properties influence the effect of mass perturbation and whether walking speed influences the effect of mass perturbation.

To explain why proximal loading decreases and distal loading increases the muscular cost of the swing phase in most subjects, we compared the values of the equation of motion components (EMCs) of thigh and shank during proximal and distal loading with the unloaded condition. It was found that both proximal and distal loading generally changed the EMCs in the same direction compared to the unloaded condition. The opposite effect of proximal and distal loading on the hip and knee torque resulted from a different balance between the different components: in the proximal loading condition, the components lowering the torques dominated the other components, while in the distal loading condition, the components increased the joint torques. The dominance of $I \cdot \ddot{\varphi}$ in the distal loading condition can be understood from equation 7.2, showing that Δm is multiplied by R squared only in $I \cdot \ddot{\varphi}$.

The EMC values for subject 1 were compared with subject 2 to establish why in the latter subject, both proximal and distal loading decreased the muscular cost of the swing phase. We found that the mass addition led to qualitatively similar changes in all EMCs as in subject 1 and that the difference in hip and knee torque between both subjects resulted from small differences in the summed effect of all changes. Overall, the present data indicate that in proximal mass perturbations, negative and positive effects are practically balanced. As a result, proximal mass addition leads to a small increase in muscular cost in some subjects, but to a small decrease in some others.

Calculating the EMCs during gait to determine their relative contribution is in line with the work of, amongst other, Whittlesey and colleages[83, 84] and Putnam.[52] The size of the EMCs is comparable to those reported by Whittlesey et al.,[84] who also found that the passive (gravitational) components during the swing phase are relatively small compared to the active torques and concluded that the swing phase is not a passive movement. The same conclusion can be drawn from the present data, that is, knee and hip torque importantly contribute to the total swing phase torques.

The large contribution of hip and knee torque may seem to contradict studies of, amongst others, Mochon and McMahon[44, 45] and Mena et al.[43] who showed that the swing phase can be simulated relatively accurately without including joint torques. This may partly be explained by wrong estimates of kinematic data used in the above-mentioned study as model input, as suggested by Piazza and Delp.[50] However, it should also be noted that for the shank, the direction of the knee torque is in the same direction as the other torques, indicating that the shank would move in the same direction without a knee torque. In addition, the large amplitude and out-of-phase relation between T_{hip} and $T_{knee,t}$ suggests that most of the muscle activity around the hip is used to balance the knee torque. Thus, absence of the knee torque as in a ballistic movement would largely decrease the hip torque needed to obtain the same movement (see Figure 7.3). Therefore, the presented data are in line with studies indicating that the kinematics of the swing phase can be approximated using a double pendulum model. However, it also indicates that from this similarity, it can not be concluded that the swing phase is pendular.

In the present study, we found that the change in hip and knee torque time series after mass perturbation is independent of the subjects' initial mass, mass distribution and moment of inertia as well as from prior mass perturbations. For the total muscular cost of the swing phase, for distal loading, the same was found. The implication for prosthetic design is that distally removing mass will always be beneficial in terms of muscular cost, while distally adding mass will always increase the muscular cost. For proximal loading, however, this is not the case. Proximal mass addition of, for example, 5 kg to the unperturbed systems decreases the muscular cost. However, proximally adding 5 kg when already 20 kg has been added proximally increases the muscular cost. Visual inspection of the hip and knee torque time series in Figures 7.6A,C explains this pattern: each mass perturbation changes the torques in the same direction. First, this decreases the total cost. However, when more mass is added, the torque will change from flexion to extension or vice versa. From there on, the total muscular cost will start to increase again. This indicates that for proximal mass perturbation, initial inertial properties and prior mass perturbations do influence the effect of mass perturbation. For subject 1 shown in Figure 7.6, this leads to an 'optimal' inertial configuration when 15 kg is added at $.2*l_s$. It should be noted, however, that the effects are very small and that this mass perturbation may not be clinically feasible because of negative effects of such large mass addition on a wide range of other variables (e.g., stump-socket interface forces, the energy during non-cyclic activities or during walking up a slope).

Since the effect of mass perturbation on the hip and knee torque during the swing phase depends on the swing phase kinematics, we also investigated the effect of walking speed and found that speed did not systematically influence the effect of mass perturbation. At higher walking speeds, an overall increase in muscular cost was found. However, qualitatively, the effects of mass perturbation at the higher walking speeds are very similar, that is, similar mass perturbations have a similar effect on the muscular cost. This suggests that, for the velocities studied, the same prosthetic inertial properties are beneficial.

7.6. Conclusion

The present study indicates that the effect of proximal mass addition is almost zero because of a balance between positive and negative effects. As a result, in some subjects muscular cost is decreased, while in others muscular cost is increased. In addition, we found that for proximal loading, the effect of mass perturbation is dependent on initial inertial properties and that, theoretically, an ‘optimal’ configuration can be found. However, the large mass needed to obtain this optimal configuration makes it clinically irrelevant. For distal mass perturbations, the effect is independent of initial inertial properties. For these perturbations, any mass reduction will have the same beneficial effect of muscular cost. Finally, in this study, we found qualitatively similar effects of mass perturbation at self-selected and fast walking speed, indicating that, for the velocities studied, the same prosthetic inertial properties are beneficial.

7.7. Appendix A: symbols

H, K, A	hip, knee, ankle
l_t, l_s	lengths of thigh and shank
d_t, d_s	distance from center of mass of thigh and shank to proximal joint
m_t, m_s	mass of thigh and shank
σ, φ	angle of thigh and shank with vertical
Δm	amount of mass added to the segment
R	distance between mass perturbation and the proximal joint (knee)
F	reaction force
T_{hip}, T_{knee}	torque in hip and knee
to, hs	toe-off and heel strike
G	gravitational acceleration
$I_{t,h}$, $I_{s,k}$	moment of inertia of thigh and shank around hip and knee, respectively
$x_{cm,t}$, $y_{cm,t}$, $x_{cm,s}$, $y_{cm,s}$	x and y positions of the center of mass of thigh and shank
TE_{hip}, TE_{knee}, $TE_{hip+knee}$	torque effort in hip, knee and combined hip and knee

8. Discussion and concluding remarks

The aim of this thesis was to determine the optimal prosthetic inertial properties for the swing phase of transtibial amputation (TTA) subjects. In the present chapter, we will discuss some of the limitations of the methodology used in these studies as well as theoretical implications of our work. We will finish with the main implications of this thesis for the clinical practice of designing and prescribing prosthetic components. Several of the issues have already been discussed in previous chapters. The aim here is to bring these issues together, to discuss them more extensively and to introduce some new issues.

8.1. Limitations of this study

8.1.1. Above-knee and through-knee amputation subjects

In this study, we focussed on transtibial amputation (TTA) subjects, disregarding the transfemoral amputation (TFA) and through-knee amputation (TKA) subjects. We chose to evaluate TTA subjects mainly because they are the largest group of lower limb amputation subjects and, on average, most frequently use a prosthesis.[55, 56, 69] Distinguishing TTA, TKA and TFA subjects was necessary because we expected the influence of prosthetic mass on gait in TFA and TKA to be different between groups. In contrast to TTA subjects, TFA and TKA subjects do not have direct muscular control of the knee joint. As a result, they can only indirectly control knee flexion and extension by means of muscular activity around the hip and by accelerating their hip with the rest of their body.

Because most prosthetic knees are only stable during stance if the knee is fully extended, controlling knee flexion and extension during swing phase may be especially important in TFA and TKA subjects. This has led to many developments in prosthetic knees, such as hydraulic knee dampers and, in recent years, knee dampers which electronically adapt to walking speed, such as in the C-leg by Otto Bock and the Intelligent Prosthesis (IP) by Blatchford. In terms of the models used in this study, the swing leg of TFA and TKA subjects using such a prosthesis can be characterized as a double pendulum with a damped knee joint and a moving hip joint that can be controlled by muscular activity.

The simulations in this study suggest that mass perturbation can have similar effects on the swing phase as knee damping, that is, it can change the flexion and extension velocity of the knee. However, whereas in a mechanical damper energy is dissipated, mass perturbation can influence the kinetic and potential energy during the swing phase without energy loss. Several studies have indicated that energy exchange between kinetic and potential energy during gait is a very energy efficient mechanism in gait (e.g., [11, 14, 20]). Therefore, changing prosthetic mass to influence the flexion and extension velocity of swing phase in TFA and TKA subjects may warrant further study.

We believe that a study design similar to the one presented in this thesis may be used to study the influence of prosthetic mass on the gait of TFA and TKA subjects. In practice, the experiments of Chapter 5 would need to be repeated to determine the adaptation strategy of these subjects to mass perturbation. Because of the lack of direct knee joint control, using a kinematical invariance strategy may be impossible for TKA and TKA subjects. When this strategy is determined, depending on the outcomes, simulations can be performed to predict optimal inertial properties.

8.1.2. Gait and other activities

In this thesis, we focussed only on gait, disregarding the other activities that subjects perform with their prosthesis. One reason for this is that walking is considered the most important dynamic activity performed with a prosthesis. In two recent studies, Legro and coworkers[34, 36] investigated the recreational activities lower-limb amputation subjects reported as most important as well as the issues of importance related directly to their prosthesis. For nine of the ten most reported recreational activities, walking is prerequisite. When asked directly for important issues related to their prosthesis, amputation subjects score the fit of the prosthesis highest (average: 98.1 out of 100), while weight of the prosthesis scored 85.7. The energy required for walking, which we found to be influenced by prosthetic weight, scored an average 88.8.

Another reason for focussing on gait is that gait may be the only activity in which mass reduction is not necessarily beneficial. In gait, as shown in this thesis, specific mass additions can decrease the muscular cost of the swing phase as a result of the pendular characteristics of the leg. However, while gait is a cyclic, highly optimized movement, in more discrete movements such as stepping over an object or turning during standing, these mechanisms may not apply, while the negative effects of mass addition when walking slopes or stairs seem obvious. Therefore, for the clinical interpretation of the findings of this study, it should be kept in mind that we studied only gait. If weight reduction is beneficial for most or all other dynamic activities, then small beneficial effects of mass addition for gait efficiency may not be worthwhile from a clinical point of view.

8.1.3. Swing phase vs. complete gait cycle

Within the gait cycle, this study focuses on the swing phase, assuming no influence of prosthetic weight on the stance phase. It should be noted, however, that prosthetic mass might have some (indirect) influence on the stance phase. One of the reasons why the leg can swing with relatively little muscle activity is because of the initial angular velocities of the body at toe-off (see Chapters 3 and 4). Since these angular velocities need to be created or preserved during the stance phase, changing prosthetic mass may influence the amount of energy needed to do this.

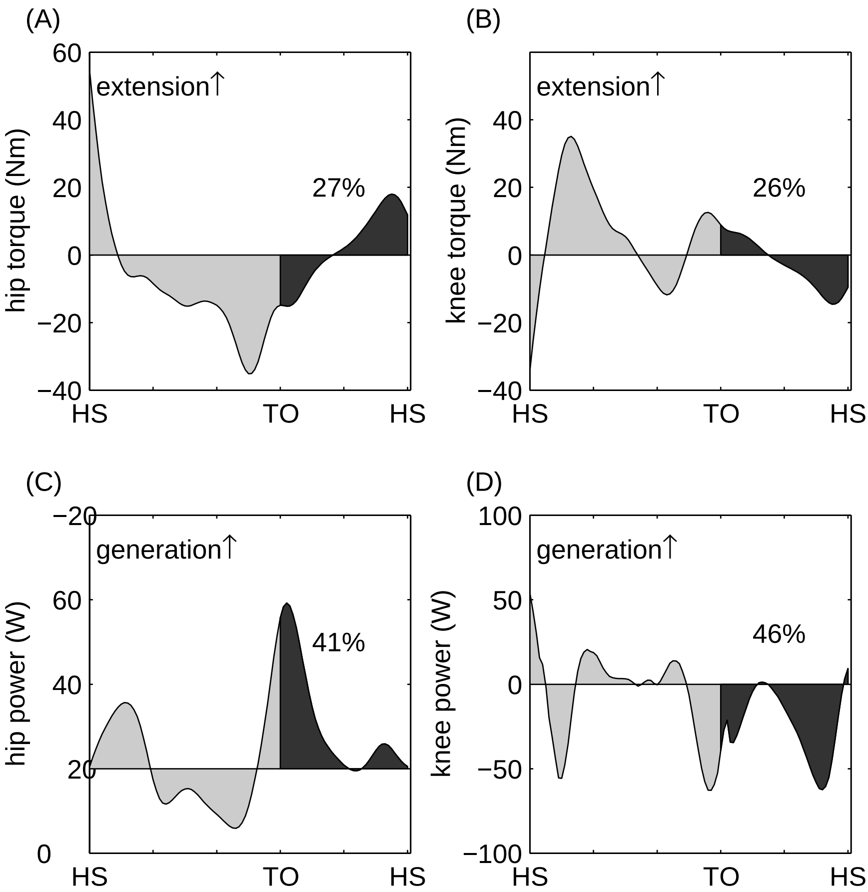

Figure 8.1. Torque and power in hip and knee joint during the complete gait cycle in a healthy adult, derived from Winter.[86] The lighter areas indicate the energetic cost during the stance phase, the darker areas the energetic cost of the swing phase. The percentages shown in the graph indicate the size of the dark area (swing phase) as a percentage of the total area (stance plus swing phase). Abbreviations: HS, heel strike, TO, toe-off.

The limitation to the swing phase should be kept in mind when evaluating the effect of prosthetic mass on the complete gait cycle. Figure 8.1 shows literature data of hip and knee kinetics during the complete gait cycle of a healthy male adult,[86] showing that the energetic cost of the swing phase is a significant part of the total kinetics. Since our study determines only the muscular cost of the prosthetic leg during swing phase, an increase in this cost of, for example, 25% of the swing phase energetic cost increases the total muscular or metabolic cost during gait cycle with a significantly smaller percentage.

8.1.4. Kinematical invariance and energetic cost minimization

In Chapter 5, we found that TTA subjects keep their swing phase kinematics invariant after mass perturbation, indicating that they do not use an energy minimization strategy during the swing phase. This may have important implications for biomechanical modeling in which minimization of some kind of energy or muscle cost function is often used as an optimization criterion (e.g., [6, 12]). The present data indicate that applying such a model to the swing phase during walking may not provide valid solutions.

While energy minimization may not apply for the swing phase, it may still apply for the complete gait cycle. During the stance phase, the knee is slightly flexed, which decreases the vertical movement of the center of gravity during gait. (e.g., [9, 58]) Therefore, both too much and too little knee flexion will increase energetic cost of the stance phase. If the kinematics of the swing phase would change as a result of mass perturbation, the optimal knee flexion during stance phase may be disrupted. Therefore, the kinematical invariance strategy may be a strategy in which additional muscular input during the swing phase is used in such a way that the energetic cost of the complete gait cycle is minimized. A model of the complete gait cycle is necessary to further investigate this strategy.

Findings in line with the kinematical invariance strategy can be found in literature on running. In 1990, Cavanagh and Kram[5] reviewed the literature and experimentally studied the relation between anthropometric measures (body height, leg length, body mass, leg mass, and moment of inertia) and found that all these variables were poorly correlated to stride length during running. In addition, they found that masses added to the feet during running had little effects on both stride length and stride frequency, while the metabolic cost dramatically increased with added mass. This indicates the use of a kinematical invariance strategy in running, measured over the complete gait cycle. Cavanagh and Kram[5] were not able to give a clear explanation for these findings.

The kinematical invariance strategy may have a more behavioral than biomechanical nature. Subjects may aim at maintaining their kinematic pattern because they learned to walk this way during rehabilitation. In addition, walking kinematically similar to non-amputees may be an implicit or explicit ‘goal’ for amputation subjects. Many amputation subjects reported, proudly, that people hardly see from their gait that they use a prosthesis. This interpretation of the kinematical invariance strategy may be in line with the findings of Mulder and coworkers (e.g., [13, 46]) who showed in a series of experiments that various patient groups maintained a symmetrical or normal movement pattern (such as gait) when large gait deviations were expected. However, when a cognitive load is imposed, in the same patients, the symmetrical or normal movement pattern broke down, indicating that this pattern was cognitively imposed.

Further study of adaptation to mass perturbation, as well as of adaptation to differently designed prosthetic feet, shoes or knees, may be important for prosthetic design. Present design approaches in prosthetics are often based on assuming how subjects will adapt to the perturbations. In the case of mass perturbation, it was always believed this would influence gait kinematics, which led to different predictions of the optimal prosthetic mass than assuming kinematical invariance. Studying adaptation strategies will indicate why subjects walk the way they do, which may not only provide us with better tools to develop prosthetic components, but may also lead to better gait training and rehabilitation tools.

8.1.5. Other variables relevant for evaluating prosthetic mass

In the present thesis, we assumed the kinetics of the swing phase to be an important variable of interest for prosthetic mass. In addition to kinetics, we estimated the influence of prosthetic mass on the forces in the stump-socket interface during the swing phase. The latter variables were strongly influenced by prosthetic mass. Although recent developments in measuring stump-socket forces may provide us with more information on this, the present literature suggests that the stump-socket interface forces during the swing phase are small compared to the stance phase, indicating that they may not be a very important parameter for evaluating prosthetic mass.

There may be additional variables important for evaluating prosthetic mass. One such variable could be the complexity of the neural control. While in a completely ballistic swing phase no neural control over the swing leg muscle is needed, in this study we found significant hip and knee torques. These torques have a ‘cost’ not only in terms of energetics, but also in terms of neural input. Therefore, mass perturbation may influence the complexity of the control needed. At present, it is unclear whether this neural complexity is an important issue.

Prosthetic inertial properties may also influence the perception of asymmetry between both legs. Professionals working in the field of prosthetics have reported that a relatively heavy prosthesis is not always perceived as such, while a lightweight prosthesis may feel heavy. In our experiments, we did not systematically score the subject's perception of the prosthesis. However, during the experiments we always asked the subjects how they ‘liked’ the mass perturbation and found that both non-amputation subjects as well as TTA subjects found it very difficult to verbalize their experience of the added mass and to decide which mass condition they preferred.

Related to this is the proprioceptive information obtained from the prosthesis. In upper limb prosthesis, much research is performed on providing amputation subjects with proprioceptive information about the position and orientation of their prosthesis. In fact, O’Riain and Gibbons[47] suggested that the absence of position information of their prosthesis is an important reason for the low acceptance of powered upper-extremity prosthesis. In lower limb prosthesis, less is known about proprioceptive information. Isakov et al.[28] have shown that TTA subjects are significantly less stable when standing with both eyes closed or open and suggested that these differences are related to the proprioceptive deficit as a result of partial limb loss. A very lightweight prosthesis may provide the subject with little information about its position in space, especially in above-knee amputation subjects is which proprioceptive information about the knee angle and its derivatives is lacking. However, as in neural complexity, this variable is very difficult to measure and, therefore, it remains unclear whether changes in prosthetic mass importantly influence the proprioception.

8.2. Theoretical implications

8.2.1. Is human walking ballistic?

While the work of Wilhelm and Edward Weber,[82] dating from 1836, is often considered the first modern scientific study of gait, it can also be considered the first study of passive or ballistic walking. In their study, Weber and Weber measured the swing time of a leg during normal walking and during free swing while subjects stood on a platform. In addition, they measured the natural frequency of cadaver legs. Based on the similarity between the measured swing times, Weber and Weber assumed the forward swing during walking to be a pendular movement in which musculature plays no role. According to Weber and Weber, during the swing phase, '*...the leg hangs from the trunk and, together with it, is carried forwards by the opposite leg. During this period, the former leg shares the movement with the trunk and, additionally, carries out another movement: rotating about its upper extremity, moved by its own weight, like a pendulum, the leg swings from behind forwards*' (page 18).

Forty-seven years later, in 1883, Duchenne challenged this hypothesis of the Weber brothers. He observed that patients with paralyzed leg muscles needed to lift the swing leg to obtain sufficient foot clearance (see [29]) and argued that if gravity alone was responsible for the movement, no compensation would have been necessary. In 1953, in a review on gait analysis, Steindler[66] assumed the ballistic walking hypothesis contested and disproved.

Nowadays, the idea of ballistic walking is popular again, with much work performed either from a more robotic or a more biological point of view. In robotics, researchers aim at the development of passive or quasi-passive bipedal robots that are as energy efficient as humans (e.g., [42, 72]). In biology and human movement sciences, ballistic walking models are used to model and understand animal or human walking (e.g., [32, 44]). The importance of the ballistic walking models is often expressed in terms of the assumed reciprocal nature between robotics and biology: robotic gait may be improved by applying principles of human walking, while the understanding of human walking may increase when comparing it to the gait of ballistic walking robots.

The ballistic walking models of Mochon and McMahon published in 1980 ([44, 45], see also Chapter 3) have been important in the revival of ballistic walking.

Although Mochon and McMahon did not test whether their ballistic model could exactly predict the kinematics of human gait, they reported close agreement with published data. In Chapter 3, we investigated whether four different ballistic walking models were able to predict the kinematics of the swing phase in healthy subjects and found that the four models were not able to exactly predict swing phase kinematics. In addition, in Chapter 7, we studied in detail the size of the different torques that contribute to the swing phase dynamics and found that both hip and knee torque significantly contribute to the equations of motions of thigh and shank. These results are in line with several other recent studies in which the ballistic swing phase assumption has been put to the test,[50, 84] indicating that ballistic walking models cannot exactly predict kinematics of human walking without incorporating joint torques.

8.2.2. Can ballistic walking models predict the effect of inertial properties on gait?

Although the swing phase is not completely ballistic, ballistic walking models may still be useful to provide insight into important features of human walking or predict essential features of walking. One such feature that is often mentioned in ballistic walking literature is predicting the effect of changes in segment inertial properties on gait characteristics (e.g., [19, 73]).

In Chapter 4, we found that inertial properties of the lower leg of TTA subjects were reduced compared to the shank and foot of matched controls. As a result, the ballistic model simulation predicted different kinematics between TTA subjects and controls. However, these predictions were not supported by the experimental data reported in the same Chapter. In Chapter 5, we studied the adaptation of TTA subjects to mass perturbations and found that swing phase kinematics did not change as a result of mass perturbation; instead, subjects adapted to the perturbations by changing their muscular input in such a way that the kinematics remained unchanged. Both of these findings indicate that swing phase kinematics and adaptations to mass perturbation can not be predicted using ballistic walking models.

Although we have mainly stressed findings that can not be explained by ballistic walking models, this study still indicates that ballistic or pendular characteristics are influential and, therefore, that studying the behavior of ballistic walking models may provide important insights into human walking. For example, the data in Chapters 3 and 6 show that a relatively large part of the swing phase kinematics are obtained ‘for free’, that is, without the need of additional muscle activity. In addition, the findings in Chapters 5 and 6 show decreased joint torques in most subjects after proximal mass addition, which can only be explained based on the pendular characteristics of the leg.

In sum, the present study indicates that the swing phase during walking at self-selected walking speed in normal and TTA subjects is strongly influenced by gravitational and inertial forces and, for this reason, reducing prosthetic mass is not necessarily beneficial for prosthetic gait. However, muscle input plays a significant role and, more importantly, is systematically adapted in such a way as to obtain the desired swing phase kinematics. Therefore, modeling prosthetic gait using pendulum equations or more complex ballistic models is insufficient to understand and predict the relation between the inertial properties of the leg and the kinematics and kinetics during gait.

8.3. Clinical implications

The most direct implications of this thesis for prosthetic design can be derived from the Figures 6.3, 6.4 and 6.6, showing the effects of changing prosthetic mass of ten TTA subjects on the muscular cost of the swing phase as well as on the forces in the stump-socket interface.

The effects on the forces and torque in the stump-socket interface (Figures 6.5 and 6.6) are straightforward, that is, forces increase when mass is added and decrease when mass is removed. In addition, the effects are largest after distal perturbation. As noted before, however, this analysis only provides a first estimate of the stump-socket forces. In addition, because the present literature suggests that forces in the stump-socket interface during the swing phase are small compared to the stance phase, drawing conclusions for clinical practice from these results seems far stretched. Overall, the present data show no reason to change the current practice to make prostheses as lightweight as possible from the point of view of stump-socket interface forces.

In terms of the muscular cost of the swing phase, a distinction should be made between distal and proximal mass perturbations. For distal mass perturbation, mass reduction always decreases the muscular cost while addition always increases the muscular cost. For proximal loading, the effect is less straightforward (see Chapters 6 and 7), because the direction of the effect is different in different subjects. However, the analysis in Chapter 7 showed that the individual differences in the direction of the effect are the results of small differences in joint torques and that the effect of proximal mass perturbation can best be interpreted as an almost complete balance between positive and negative effects.

For clinical practice, these results indicate that for components located in the upper 15 to 20 cm of the lower leg, such as liners and sockets, mass is not of primary importance. In this region, prosthetic mass can be increased without increasing the muscular cost of the swing phase (see also Figures 6.3 and 6.4). For more distally located components, such as prosthetic feet and shoes, the muscular cost increases with increasing mass. The size of the effects on the swing phase muscular cost can be estimated from the data in Chapters 6 and 7.

8.3.1. Influence on clinical practice

At the start of this project, there was a relatively large body of literature claiming that prostheses need not be lightweight. Statements can be found such as *'lightweight footwear does not necessarily provide the most symmetrical gait …or the most acceptable gait as preferred by the amputee*'[16] and '*…prosthesists can design limbs using heavier components without significantly increasing the amount of energy necessary to ambulate*'.[17] Godfrey et al.[21] concluded in 1977 that the mass of prosthetic feet can be increased '*without changing gait patterns*', and in a modeling study, Mena et al.[43] concluded in 1981 that '*a "lightweight" prosthesis would be less desirable than a "heavy" prosthesis, while a prosthesis that had the same inertial properties as the removed leg may be most desirable*'.

The above-mentioned literature did not strongly influence the aim in prosthetic design to reduce prosthetic mass. This may be explained by the limited quality of most experimental studies (see Chapter 2) as well as by the simplistic modeling approaches, which were not clearly supported by empirical evidence. Another important reason, however, may be that the conclusions in these studies differed too much from the clinical experience of professionals working in the field of prosthetics. For example, Hilmar Janusson, at the time head of Research and Development Department of Össur stated in April 2000 (personal communication) that there was a discrepancy between literature and subjects' perception: subjects like lightweight, while the literature suggests otherwise. This thesis suggests that distal components need to be lightweight, while the mass of proximal components does not importantly influence the swing phase energetics. Future developments will indicate whether this outcome will become more widely accepted and whether prostheses designed using this principle are positively appreciated by their users.

Summary

In this thesis, the influence of prosthetic inertial properties (mass, mass distribution and moment of inertia) on the gait of transtibial amputation (TTA) subjects is studied. **Chapter 1** introduces the present ideas on prosthetic mass. It describes that the general design effort has always been, and still is, to reduce prosthetic mass. However, as far as we know, lightweight design has never been advocated in the present literature. The Chapter introduces the opposite view, found in a relatively large body of literature, that lightweight design might not be beneficial for prosthetic gait. The aim of this thesis, therefore, is to determine the optimal inertial properties of the prosthetic leg.

Chapter 2 presents a systematic review of the literature. First, theoretical models are reviewed that describe the relation between prosthetic inertial loading and amputee gait. We found that the present theoretical models can be divided into (1) pendulum models, (2) multi-segment models, and (3) segment energy models. All three models have in common that they assume the swing phase or the complete gait cycle to be more or less ballistic; that is, uninfluenced by muscle activity. In addition, all three models predicted that inertial loading of the present lightweight prosthesis need not be decreased while some models predicted that gait may improve when mass is added. The model predictions were compared with experimental reports in which prosthetic mass was perturbed. It was found that the methodological quality of most studies was limited. The majority of the studies did not report a significant effect of adding mass on economy, self-selected walking speed, stride frequency and stride length, while some studies reported small beneficial effects of adding mass. It was concluded that there is a discrepancy between model predictions and experimental data, which may be related to both the poor methodological quality of the experiments as well as the limited predictive value of the models.

Because of the discrepancy between models and experimental data found in the review, in **Chapter 3** we studied the validity of the main assumption of all three above-mentioned models, that is, that the swing phase is ballistic. To that end, we quantified to what extent the swing phase kinematics at self-selected walking speed in six healthy subjects could be predicted using (1) the ballistic walking model as originally introduced by Mochon and McMahon, (2) an extended version of this model including heel-off of the stance leg, (3) a double pendulum model, consisting of a two-segment swing leg with a prescribed hip trajectory, and 4) a shank pendulum model consisting of a shank and foot with a prescribed knee trajectory. Statistically significant differences between model output and experimental data were found in all four models. All models underestimated swing time and step length. In addition, significant differences were found between the different models. Despite qualitative similarities between the ballistic models and normal gait at self-selected walking speed in healthy subjects, we concluded that these models can not exactly predict swing phase kinematics. Therefore, ballistic walking models may not be directly applicable to predict the effect of mass perturbation, as proposed in the literature.

Although the swing phase in healthy subjects is not completely passive, adaptation to mass perturbation may be in the same direction as predicted by a ballistic walking model. Therefore, in **Chapter 4**, we measured the inertial properties of the combined residual limb and prosthesis of ten TTA subjects and compared them to the lower leg inertial properties of ten matched controls. In addition, we measured swing phase kinematics and kinetics in the TTA subjects and controls, and investigated whether differences in the swing phase between both groups can be predicted using subject-specific double-pendulum models. We found that in all TTA subjects, inertial properties of the lower leg were reduced compared to the matched controls. However, kinematics were similar in both groups, while the joint torques and powers were reduced in the TTA subjects. The double pendulum model simulations revealed larger deviations between the model and experimental data in the TTA subjects. It was concluded that the prosthetic leg of TTA subjects is lightweight (reduced mass and moment of inertia, more proximal center of mass) compared to a matched non-amputated leg and that the lightweight design is less optimal in terms of the pendular behavior of the leg. However, lightweight design leads to smaller joint torques needed to influence the pendular trajectory. Therefore, the optimal inertial properties in terms of kinematics and kinetics of gait may be a compromise between pendular properties and 'efficient control'.

If the swing phase would be ballistic, adaptation to mass perturbation could directly be predicted using forward dynamical models or pendulum equations. However, we found that ballistic walking models can not exactly predict swing phase kinematics in controls and TTA subjects and that muscular activity is significant. Therefore, we defined two extreme adaptation strategies to mass perturbation: (1) a kinetical invariance strategy in which kinematics (joint angles) change while kinetics (joint torques) remain the same, or (2) a kinematical invariance strategy in which kinetics change while kinematics remain the same. In **Chapter 5**, we investigated whether TTA subjects predominantly use one of these strategies. A gait analysis was performed during five different mass conditions and condition effects in kinetics and kinematics were evaluated. In addition, the effect of each strategy was predicted using forward- and inverse dynamical models. We found more significant changes in the net joint torques than in the joint angles. In addition, in contrast to the simulations assuming kinetical invariance, simulations assuming kinematical invariance were significantly correlated with the experimental data. We concluded that adaptation to mass perturbation in TTA subjects is better characterized as a kinematical invariance strategy than as a kinetical invariance strategy.

The kinematical invariance strategy (adapting joint torques after mass perturbation in such a way to maintain a kinematical pattern) suggests that ballistic walking models can not adequately describe the effect of mass perturbation and that an alternative model to determine optimal prosthetic inertial properties is needed. Therefore, in **Chapter 6**, we developed a new model in which the effect of mass perturbation on the joint reaction forces and torques is calculated using known kinematical data measured in the condition without additional mass and the inertial properties after mass perturbation. Outcomes of the simulation were the net torques in hip and knee as well as in the net joint reaction forces and torque between socket and stump. We found that both size and direction of the effect of mass perturbation on the muscular cost depends on the location of the perturbation. In 9 of the 10 TTA subjects, cost decreased after distally removing mass as well as after proximally adding mass to the lower leg. In contrast, net joint forces and torques between stump and socket always decreased when mass was removed and increased when mass was added. A comparison with the experimental reports on mass perturbation in below-knee amputees suggests that the present simulation data better describe the experimental data than the predictions derived from ballistic walking models.

The aim of **Chapter 7** was to (1) clarify why in most TTA subjects proximal mass addition decreases the muscular cost of the swing phase while distal mass addition increases the muscular cost, (2) why in some TTA subjects, proximal mass addition increases the muscular cost, and (3) determine whether the effect of mass perturbation is influenced by individual differences in body dimensions and walking speed. From a group of ten TTA subjects, anthropometric and kinematic data were measured at self-selected and fast walking speed. The equations of motion of the prosthetic leg were derived and the values of the equation of motion components during the swing phase were calculated. Finally, the effect of mass perturbations was compared at self-selected and fast walking speed. It was found that for proximal mass addition, positive and negative effects on the joint torques were practically balanced. This resulted in a small increase in muscular cost in some subjects, and a small decrease in cost in some others. The change in hip and knee torque time series did not depend on initial anthropometric properties, but was only determined by the swing phase kinematics and the size and location of the mass perturbation. For distal mass perturbation, this led to a linear increase in amplitude of the torques. For proximal mass perturbation, small mass addition first decreased the amplitude of the torques. However, for larger mass perturbations, the torques changed from flexion to extension or vice versa and the amplitude increased again. The large mass needed to obtain the minimum amplitude (about 15 kg in the studied subject), and therefore the minimum muscular cost, makes this configuration clinically irrelevant. Finally, in this study, we found qualitatively similar effects of mass perturbation at self-selected and fast walking speed, indicating that, for the velocities studied, the same prosthetic inertial properties are beneficial.

In **Chapter 8,** some methodological issues related to our study are discussed. We motivate our choice of focussing on transtibial amputation subjects and provide some suggestions on the importance of prosthetic mass for transfemoral and through-knee amputation subjects. In addition, we explain our focus on gait, and, within the gait cycle, on the swing phase. We also discuss why subjects may use a kinematical invariance strategy and suggest that this strategy may have implications for biomechanical modeling because energy or muscle force minimization may not be an optimization criterion during the swing phase, as often is assumed. We also discuss additional variables that may be important for prosthetic mass, such as the complexity of the neural control, the perception of inertial asymmetry and proprioceptic information obtained from the prosthesis. Then, in the same Chapter, the implication of this thesis for ballistic walking models is discussed, concluding that the swing phase at self-selected walking speed is strongly influenced by gravitational and inertial forces, but that muscle input plays a significant role and is systematically adapted after mass perturbation. Therefore, modeling prosthetic gait using ballistic models is insufficient to understand and predict the influence of leg inertial properties on the kinematics and kinetics during gait. Discussing some clinical implications concludes chapter 8. We emphasize that we only provide a first estimate of the forces and torque in the stump-socket interface, which indicates that there is no reason from this point of view to change the current practice of lightweight design. The effect of proximal mass perturbation on the muscular cost of the swing phase can best be interpreted as an almost perfect balance between positive and negative effects, while for distal mass addition, the negative effects are stronger than the positive effects. For clinical practice, these results indicate that for components located in the upper 15 to 20 cm of the lower leg, such as liners and sockets, mass does not significantly affect the swing phase energetics, while for more distally located components, such as prosthetic foot and shoes, the muscular cost increases with increasing mass.

Samenvatting

In dit onderzoek bestudeerden we de invloed van de traagheidseigenschappen (gewicht, gewichtsverdeling en traagheidsmoment) van onderbeenprothesen op het gangbeeld. **Hoofdstuk 1** introduceert de huidige ideeën over de invloed van prothesegewicht. Het beschrijft dat gewichtsvermindering altijd een belangrijk criterium is geweest bij het ontwikkelingen van prothesen, en dat nog steeds is. Echter, in de literatuur is het nut van gewichtsvermindering nooit onderbouwd. Het hoofstuk introduceert het alternatieve gezichtspunt, regelmatig in de literatuur beschreven, dat lichtgewicht design niet optimaal is voor het gangbeeld. Het doel van dit proefschrift is dan ook om te bepalen wat de optimale traagheidseigenschappen van onderbeenprothesen zijn.

Hoofdstuk 2 presenteert een systematische review van de bestaande literatuur. Allereerst zijn de theoretische modellen over de relatie tussen de traagheidseigenschappen van prothesen en het gangbeeld beschreven, onderverdeeld in (1) pendulum modellen, (2) multi-segment modellen, en (3) segment-energie modellen. De drie modellen hebben gemeen dat ze de zwaaifase of de complete gangcyclus als min of meer ballistisch (niet beïnvloed door spieractiviteit, of ‘passief’) opvatten, en alle drie voorspellen dat het looppatroon kan verbeteren als prothesemassa wordt vergroot. De voorspellingen van de modellen werden vergelijken met experimentele studies waarin prothesegewicht wordt verstoord. De methodologische kwaliteit van de experimentele studies bleek beperkt. In meerderheid rapporteerden ze geen significante invloed van gewicht op energetische kosten, comfortabele loopsnelheid, schredenfrequentie en schredenlengte tijdens het lopen, terwijl een aantal studies kleine positieve effecten van extra massa vond. We concludeerden dat er een discrepantie is tussen de modellen en de experimentele studies die zowel verklaard zou kunnen worden door de matige methodologische kwaliteit van de experimentele studies als door de beperkte voorspellende waarde van de modellen.

Om de in het literatuuronderzoek gevonden discrepantie tussen modellen en experimentele data te verklaren hebben we in **hoofdstuk 3** de validiteit van de belangrijkste assumptie van bovengenoemde modellen onderzocht, namelijk dat de zwaaifase ballistisch is. Van zes gezonde proefpersonen kwantificeerden we in hoeverre de zwaaifase kinematica tijdens het lopen op comfortabele loopsnelheid bij voorspeld kan worden op basis van (1) het ballistische model geïntroduceerd door Mochon en McMahon, (2) een uitgebreide versie van dit model waarin hiel-lift van het standbeen is toegevoegd, (3) een dubbel-pendulum model bestaand uit een twee-segment zwaaibeen met voorgeschreven heuptraject, en (4) een pendulum model bestaand uit een onderbeen en voet met voorgeschreven knietraject. Statisch significante verschillen tussen simulaties en experimentele data werden gevonden in alle vier de modellen. Alle modellen bleken de zwaaiduur en staplengte te onderschatten. We concludeerden dan ook dat, ondanks een aantal kwalitatieve overeenkomsten, de ballistische modellen niet exact de zwaaifase kinematica bij gezonde personen kunnen voorspellen. Dit suggereert dat deze modellen niet direct toepasbaar zijn voor het voorspellen van het effect van massaverandering, zoals in de literatuur wordt beweerd.

Ondanks dat de zwaaifase bij gezonde proefpersonen niet geheel ballistisch is zouden aanpassingen na massaverstoring in dezelfde richting kunnen zijn als ballistische modellen voorspellen. In **hoofdstuk 4** hebben wij daarom de traagheidseigenschappen van de gecombineerde stomp plus onderbeenprothese gemeten en vergeleken met de traagheidseigenschappen van het onderbeen van tien vergelijkbare controle proefpersonen. Daarnaast hebben we de kinematica en kinetica in beide groepen gemeten en onderzocht of verschillen in de zwaaifase voorspeld konden worden met proefpersoon-specifieke dubbel-pendulum modellen. We vonden dat in de traagheidseigenschappen van het prothesebeen waren afgenomen vergeleken met de controlegroep. Echter, de kinematica van de zwaaifase was gelijk in beide groepen, terwijl de gewrichtsmomenten en vermogens waren afgenomen in de prothesegroep. De simulaties van het ballistische dubbel-pendulum model liet grotere verschillen met de experimentele data zien in de prothesegroep. We concludeerden dat de stomp plus prothese 'lichter' zijn (vermindert gewicht en traagheidsmoment, meer proximaal zwaartepunt) dan een niet-geamputeerd been en dat dit minder optimaal is in termen van het ballistische gedrag van het been. Echter, bij een lichtgewicht prothese zijn kleinere gewrichtsmomenten nodig om het pendulumtraject bij te sturen. Daarom zullen de optimale traagheidseigenschappen in termen van kinematica en kinetica een compromis moeten zijn tussen ballistische eigenschappen en ‘efficiënte controle’.

Als de zwaaifase compleet ballistisch was geweest, dan zou de adaptatie na gewichtverstoring direct voorspeld kunnen worden door middel van ballistische voorwaarts dynamische modellen of pendulum vergelijkingen. Echter, hoofdstukken 3 en 4 lieten zien dat ballistische modellen niet in staat zijn de kinematica te voorspellen en dat spieractiviteit een belangrijke rol speelt tijdens de zwaaifase. Daarom definieerden wij twee mogelijke adaptatiestrategieën na massaverstoring: (1) een kinetische invariantie strategie waarin de kinetica (gewrichtsmomenten) gelijk blijft terwijl de kinematica (gewrichtshoeken) verandert, of (2) een kinematische invariantie strategie waarin de kinematica gelijk blijft terwijl de kinetica verandert. In **hoofdstuk 5** onderzochten we of de aanpassingen na massaverstoring van een onderbeenprothese beschreven kan worden door een van beide strategieën. Een gangbeeldanalyse werd uitgevoerd met vijf massacondities en effecten op de kinetica en kinematica werden geëvalueerd. Daarnaast werd het effect van de beide strategieën voorspeld door middel van inverse- en voorwaarts dynamische modellen. We vonden meer significante verschillen in de netto gewrichtsmomenten vergeleken met de gewrichtshoeken. Daarnaast waren, in tegenstelling tot simulaties gebaseerd op kinetische invariantie, de simulatiedata gebaseerd op de kinematische invariantie strategie significant gerelateerd aan de experimentele data. We concludeerden dan ook dat aanpassingen na massaverstoring beter beschreven worden door een kinematische invariantie strategie dan met een kinetische invariantie strategie.

De kinematische invariantie strategie (aanpassing van gewrichtsmoment aan massaverstoring waardoor de kinematica hetzelfde blijft) impliceert dat ballistische gangbeeld modellen het effect van massaverstoring niet kunnen beschrijven. Daarom hebben we in **hoofdstuk 6** een nieuw model ontwikkeld dat het effect van massaverstoring op de reactiekrachten en de gewrichtsmomenten tussen bovenbeen, stomp en prothese voorspelt op basis van gemeten kinematica in de conditie zonder extra massa en de traagheidseigenschappen na massaverstoring. We vonden dat zowel grootte als richting van het effect van massaverstoring op de geschatte energetische kosten afhankelijk zijn van de locatie van verstoring. In negen van de tien proefpersonen namen de energetische kosten af na het distaal verwijderen van gewicht en na het proximaal toevoegen van gewicht. De netto gewrichtskrachten en het moment tussen stomp en koker namen altijd af na massaverwijdering en toe na massatoevoeging. Een vergelijking tussen de uitkomsten van het model en de experimentele literatuur suggereert dat het ontwikkelde model de experimentele data beter beschrijft dan ballistische modellen.

Het doel van **hoofdstuk 7** was om (1) duidelijk te maken waarom bij de meeste proefpersonen de energetische kosten van de zwaaifase afnamen na proximale massatoevoeging aan een onderbeenprothese, terwijl na distal massatoevoeging deze kosten toenemen, (2) waarom in sommige proefpersonen na proximale massatoevoeging de energetische kosten ook toenemen, en (3) of het effect van massaverstoring beïnvloed wordt door individuele verschillen in lichaamsmaten en loopsnelheid. Lichaamsmaten werden bepaald in een groep van tien proefpersonen met een onderbeenamputatie en de kinematica werd gemeten tijdens lopen op comfortabele en hoge loopsnelheid. De bewegingsvergelijkingen van het prothesebeen werden opgesteld en de waarden van de verschillende componenten van de bewegingsvergelijkingen tijdens de zwaaifase werden uitgerekend. Tot slot werden de effecten van massaverstoring vergeleken op comfortabele en hoge loopsnelheid. We vonden dat na proximale massatoevoeging de positieve en negatieve effecten op de energetische kosten praktisch gelijk waren. Het resultaat was een kleine afname bij sommige proefpersonen en een kleine toename in anderen. De verandering in de heup- en kniemomenten na massaverstoring bleek onafhankelijk van de initiële traagheidseigenschappen van het been en werd slechts bepaald door de kinematica van de zwaaifase en de grootte en richting van de massaverstoring. Na distale massatoevoeging leidde dit tot een lineaire toename van de gewrichtsmomenten en dus in de energetische kosten. Kleine proximale massatoename verkleinde de amplitude van de momenten. Echter, bij grote proximale massatoename veranderen momenten van richting, waardoor de amplitude weer toenam. De grote hoeveelheid massa die toegevoegd moest worden om de minimale energetische kosten te bewerkstelligen (ongeveer 15 kg in de onderzochte proefpersoon) maakt deze configuratie klinisch niet relevant. Tot slot vonden we in dit hoofdstuk kwalitatief vergelijkbare effecten van massaverstoring op comfortabele en hoge loopsnelheid, wat aangeeft dat dezelfde traagheidseigenschappen voor prothesen optimaal zijn voor de onderzochte loopsnelheden.

Hoofdstuk 8 bespreekt een aantal methodologische aspecten van dit onderzoek, waaronder (1) onze keuze voor onderbeenprothesen en het mogelijke belang van traagheidseigenschappen van bovenbeen- of disarticulatieprothesen; (2) de focus op het gangbeeld en, binnen de gangbeeldcyclus, op de zwaaifase; (3) de kinematische invariantie strategie en de mogelijke implicaties van deze strategie voor biomechanische modellen omdat minimalisatie van energie of spierkracht geen optimalisatiecriterium lijkt te zijn, zoals vaak wordt aangenomen; (5) andere variabelen die mogelijk van belang zijn voor prothesegewicht, zoals de complexiteit van de spieraansturing, de perceptie van asymmetrie in traagheidseigenschappen en de proprioceptieve informatie over de prothese; (5) de implicaties van dit proefschrift voor ballistische modellen en waarbij we concluderen we dat de zwaaifase op comfortabele loopsnelheid weliswaar sterk beïnvloed wordt door de zwaartekracht and traagheidskrachten, maar dat spieractiviteit toch een belangrijke rol speelt en systematisch aangepast wordt na massaverstoring. Dit toont aan dat ballistische modellen niet voldoen om de zwaaifase te verklaren of om het effect van de traagheidseigenschappen op de kinematica en kinetica te voorspellen.

We eindigen hoofdstuk 8 met de klinische implicaties van dit proefschrift. We benadrukken dat, ondanks dat we slechts een eerste inschatting hebben gemaakt van de krachten en het moment in de stomp-koker interface, er geen reden bestaat om de huidige praktijk van lichtgewicht design te veranderen. Voor wat betreft de energetische kosten kan het effect van proximale gewichtsverstoring het best geïnterpreteerd worden als een balans tussen positieve en negatieve effecten, terwijl bij distale massa toevoeging de negatieve effecten overheersen. Voor de klinische praktijk suggereert dit dat gewicht geen belangrijk aandachtspunt is voor componenten in de bovenste 15 tot 20 centimeter van het onderbeen, zoals kokers en liners, terwijl voor meer distale componenten zoals prothese voeten en schoenen, de energetische kosten toenemen als de componenten zwaarder zijn.

List of Publications

RW Selles, S Korteland, AJ Van Soest, JBJ Bussmann, & HJ Stam. Lower Leg Inertial Properties in Transtibial Amputees and Control Subjects and their Influence on the Swing Phase during Gait. Arch Phys Med Rehabil (in press).

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Curriculum Vitae

Ik zag voor het eerst het licht op 23 oktober 1972, negen hoog te Amersfoort. Na een aantal omzwerving onder de bezielende begeleiding van mijn ouders rondde ik achtereenvolgens de HAVO en het VWO af op het Christelijk Lyceum te Gouda. In de zomer van 1992 begon ik aan de studie Bewegingswetenschappen, Vrije Universiteit Amsterdam. Voor deze studie deed ik onderzoek bij de Dienst Fysiotherapie van het VU ziekenhuis en bij het 'Centre de Recherche' van het 'Institut de Readaptation de Montreal' in Canada. Na mijn afstuderen in de zomer van 1997 werkte ik kortstondig als junior-onderzoeker bij Fysiotherapie van het VU ziekenhuis en als junior-kelner voor BBB Uitzendbureau.

Mijn promotieonderzoek startte januari 1998 aan het Instituut Revalidatie van de Erasmus Universiteit Rotterdam op een project getiteld 'het karakteriseren en optimaliseren van het looppatroon van beenprothesegebruikers met behulp van principes uit de niet-lineaire dynamica'. Anders dan de titel doet vermoeden komt dit proefschrift hier rechtstreeks uit voort. In 2001 is mijn aanstelling als AiO voor zeven maanden onderbroken om op dezelfde afdeling onderzoek te doen naar de 'ICEX koker' van Ossur, een nieuwe koker voor onderbeenprothesen. Daarnaast werkte ik deze periode aan het internationale onderzoeksproject 'Monitoring Amputee Progress with Sensor Socket' (MAPS), waarin een aantal meetinstrumenten in een liner worden ingebouwd om prothese en stomp continu te monitoren.

Naast mijn werk als AiO werkte ik de laatste jaren als docent 'Methodologie en Onderzoeksmethoden' voor de opleiding 'Adviseurs voor Arbeid en Gezondheid' van de Transfergroep Rotterdam, als docent van de cursus 'Evidence Based Practice' voor fysiotherapeuten en ontwikkelde ik de website van de 'European Society of Physical Medicine and Rehabilitation'.

Vanaf oktober 2002 zal ik als postdoctoral fellow werkzaam zijn aan het Rehabilitation Institute of Chicago en het Department of Physical Medicine and Rehabilitation van Northwestern University in Chicago.

Dankwoord

Het laatst geschreven maar het meest gelezen: het dankwoord. En voor een dankwoord geldt het cliché dat de meeste clichés waar zijn: vele mensen hebben bijgedragen aan dit proefschrift. Zonder hen had dit proefschrift er duidelijk anders, of niet, uitgezien.

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Knoek van Soest wil ik bedanken voor zijn bijdrage aan, grofweg, de tweede fase van dit onderzoek. Je betrokkenheid leverde ons op meerdere punten nieuwe inzichten en kennis, vooral op het gebied van de biomechanica en simulatie studies. Deze inzichten, in combinatie met je creatieve ideeën, waren zeer waardevol en leerzaam.

Henk Stam was, wat meer op de achtergrond, door zijn combinatie van wetenschappelijk inzicht en klinische ervaring ook zeer belangrijk bij het tot stand komen van dit proefschrift. De open en inspirerende sfeer, de vele mogelijkheden die jij en Hans mij boden, en het vertrouwen dat je in mijn werk had waren voor mij zeer belangrijk.

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Michiel de Bakker, Henk van der Rijst, die het resultaat niet meer heeft mogen zien, en Peter Janssens van het Zuiderziekenhuis wil ik bedanken voor hun hulp bij het zoeken naar geschikte proefpersonen voor de verschillende experimenten. Daarnaast kon ik bij Michiel als orthopedisch instrumentmaker altijd terecht voor mijn vragen op het gebied van beenprothesen.

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Als er één groep was zonder wie dit proefschrift er natuurlijk nooit was gekomen dan zijn het de proefpersonen. Ondanks dat voor zowel de niet-geamputeerde 'controls' als de proefpersonen met een beenprothese er geen directe voordelen aan deelname verbonden waren, waren jullie allemaal graag bereid deel te nemen en je van alles te laten welgevallen. Dank daarvoor!

Het leukste van onderzoek is wat mij betreft het puzzelen. Als mensen zich afvragen waarom je in hemelsnaam 4 jaar nodig hebt om 150 bladzijden vol te schrijven, dan is het antwoord, in ieder geval in mijn geval: in de vele dingen die je uitprobeert maar die tot niets blijken te leiden. Tijdens de loop van dit promotieonderzoek bleken veel van de vooraf gestelde hypothesen te moeten worden verworpen, inclusief de daarbij in het hoofd al geschreven artikelen. Het opstellen van nieuwe hypothesen en het doen van de analyses om deze hypothesen te toetsen (waarvan de meeste weer worden verworpen) kost veel tijd, is af en toe frustrerend, maar toch vooral erg leuk.

Zeer veel mensen hebben ergens in dit proces, bewust of onbewust, een rol gespeeld. Door het meedenken over vraagstukken, het aandragen van suggesties of gewoon door het hebben van een goed of minder-goed gesprek. Om te beginnen wil ik de onderzoekers van de afdeling bedanken voor hun gezelligheid, ideeën en kennis, en speciaal mijn lange-termijn kamergenoten Fabiënne en Nanne. Daarnaast waren vele anderen van de afdeling Revalidatie, van verschillende disciplines, op verschillende momenten zeer behulpzaam, waarvoor mijn dank. De mede-onderzoekers van LOPEN, de 'jonge honden' uit heel Nederland op het gebied van het menselijk lopen, en speciaal mijn medeorganisatoren Stella Donker en Mirjam Pijnappels, wil ik bedanken voor de leuke en leerzame ideeën en discussies. En tot slot, maar misschien toch wel het meest belangrijk, alle familieleden en vrienden (een paar heb ik al genoemd) voor het delen van successen en frustraties. Ondanks dat er veel meer zijn, wil ik een aantal speciaal bedanken, namelijk: Pier, Krijna, Anke, Martine, Ann, Walter, Sicco, Iva, Liesbeth, Carlo, Geert, Wouter en Jaap.

En tot slot, natuurlijk, Marjolijn: Samen gingen we vanuit Amsterdam naar het door jouw meer dan door mij geliefde Rotterdam (het is met mij nog helemaal goed gekomen) en hadden we het hier erg leuk. Nu storten we ons een nieuw avontuur: Chicago. En omdat we weer samen gaan wordt het zeker weer leuk!

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