Identification of Design Requirements for a High-Performance, Low-Cost, Passive Prosthetic Knee Through User Analysis and Dynamic Simulation

by

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Submitted to the Department of Mechanical Engineering in partial fulfillment of the requirements for the degree of

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Abstract

In January 2012, a partnership was initiated between the Massachusetts Institute of Technology and Bhagwan Mahaveer Viklang Sahayata Samiti (BMVSS, a.k.a., Jaipur Foot) to design a high-performance, low-cost, passive prosthetic knee for transfemoral amputees in India. The knee was primarily intended to improve the walking gait of amputees relative to existing low-cost devices.

This thesis aimed to identify detailed design requirements for the prosthetic knee through user analysis and dynamic simulation. User analysis identified the needs and constraints of numerous stakeholders in the prosthesis development process. Members of the Indian biomechanics, prosthetics, and rehabilitation communities were interviewed to identify general requirements for the design, manufacturing, evaluation, and fitting of a prosthetic knee, and a structured survey of Indian amputees was conducted to quantify the demographics, functional capabilities, and functional needs of future end users.

Dynamic simulation identified methods to enable transfemoral amputees to walk with reduced energy expenditure and normative gait kinematics. 2-dimensional inverse dynamics simulations were used to calculate the effects of inertial alterations of a prosthetic leg on the energy expenditure required to walk with normative kinematics. In addition, simulations were performed to compute the effects of inertial alterations on the knee moment required to walk with normative kinematics. Mechanical power analysis, sensitivity analysis, and optimization were used to formulate a passive mechanical model that could accurately reproduce the specified knee moment. The effects of walking cadence on critical results were also examined.

Through the identification of user-centered and biomechanical requirements, the thesis provides a blueprint for the mechanism design comprising the next phase of the project.

Thesis Supervisor: Amos G. Winter, V
Title: Assistant Professor of Mechanical Engineering

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Contents

Chapter 1

Introduction

> "It requires some sophisticated thinking to arrive at a simple solution[...] What we want is more, and not less, science in the developing world." - P.K. Sethi, co-inventor of the Jaipur Foot[1]

In 2011, Bhagwan Mahaveer Viklang Sahayata Samiti (BMVSS, a.k.a., "Jaipur Foot") and the Massachusetts Institute of Technology (MIT) initiated a collaboration to develop a prosthetic knee for amputees in India. BMVSS, a non-governmental organization (NGO) based in Jaipur, India, is a major developer and distributor of prosthetic, orthotic, and assistive devices throughout India and the developing world. Using outside funding sources, they distribute all their products free of charge to amputees. By the time the collaboration began, they had already developed and distributed several types of prosthetic knees, but they desired a new prosthetic knee that allowed amputees to walk with improved gait.

In January 2012, an initial project meeting was held with BMVSS, and the following design requirements were given:

1. Allows normal gait on flat ground
2. Provides stability on uneven terrain
3. Costs less than $100 to manufacture

In January 2012, August 2012, and January 2013, additional meetings at BMVSS and a number of other organizations were conducted in order to expand the design requirements. These meetings were held with amputees, technicians, engineers, physicians, professors, and administrators at prosthesis fitment centers, rehabilitation hospitals, and academic institutions across India. Specifically, the organizations were Manav Seva Sannidhi, a prosthesis fitment organization hosting a fitment camp in Valsad, Gujarat; the rehabilitation department of the Sawai Man Singh Hospital in Jaipur; MUKTI, a prosthesis fitment organization in Chennai; the mechanical engineering department of the Indian Institute of Technology, Delhi; the departments of bioengineering and physical medicine at the Christian Medical College in Vellore; Otto Bock Healthcare in Mumbai; and Dow Chemical International Pvt Ltd in Mumbai.

Based on these meetings, the following design requirements were added:

1. Provides stability when standing
2. Resists buckling during stumbles
3. Allows squatting, kneeling, and cross-legged sitting
4. Aesthetically pleasing to Indian amputees
5. Complies with international standards (ISO 10328)[2] for strength testing
6. Can be mass-manufactured at high quality
7. Can be manufactured using locally sourced materials
8. Easy for technicians (non-prosthetists) to fit and align
9. Can be fit to amputees with long residual limbs
10. Lasts 3-5 years without maintenance or replacement

The goal of the present thesis was to use the tools of biomechanics, mechanical design, and user-centered design to translate some the above requirements, particularly those concerned with functionality, into a more detailed set of design requirements that can drive the construction of an alpha prototype. In this chapter, a brief review of gait and prosthetics is conducted, and an outline of the thesis is presented.

1.1 Biomechanics of human gait

The periodic motion of walking is referred to as the "gait cycle." Qualitatively, the gait cycle is often divided into phases based on whether one or more legs are in contact with the ground. "Stance" is when the foot of a specified leg is in contact with the ground, and "swing" is when the foot of the leg is off the ground. Stance and swing of one leg alternate with those of the other. "Single limb support" occurs when a single leg is on the ground, and "double limb support" occurs when both legs are on the ground. These terms are illustrated in Figure 1-1.

The gait cycle can also be divided into phases based on the forward progression of the body. The Rancho Los Amigos Gait Analysis Committee[3] proposed a taxonomy that is commonly used in the literature. The phases are summarized as follows:

1. **Initial contact**: foot contacts the ground
2. **Loading response**: weight is transferred to the leg
3. **Mid-stance**: body progresses over the leg

4. **Terminal stance**: body progresses ahead of the leg
5. **Pre-swing**: leg pushes off the ground and opposite foot contacts the ground
6. **Initial swing**: leg lifts off the ground
7. **Mid-swing**: leg moves ahead of the body
8. **Terminal swing**: leg lowers to the ground

Excellent illustrations and a detailed discussion of each phase are presented in Perry and Burnfield[4].

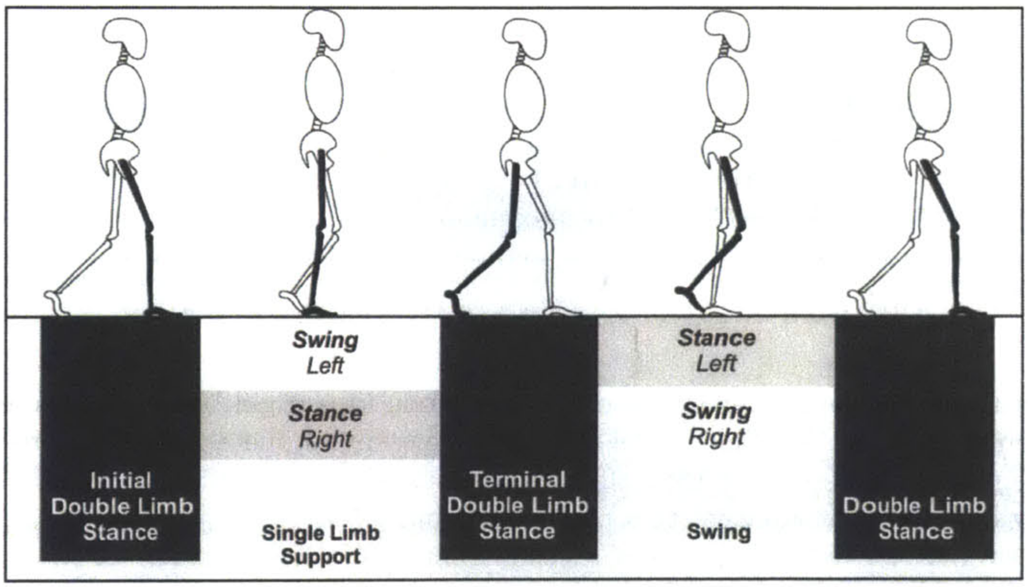

Figure 1-1: Phases of the gait cycle based on contact of the legs with the ground. From Perry and Burnfield[4].

Quantitatively, gait is often analyzed by investigating kinematics (motion), kinetics (forces and moments), and energetics (power and energy). Kinematic data is typically collected by placing reflective markers on the body and tracking them with a camera in a gait laboratory. By observing the motion of the markers, quantities like joint angles can be accurately estimated. Sample joint angle data for the knee are presented in Figure 1-2.

Kinetic quantities are computed by first measuring the external forces acting on the body. For normal walking, the external forces are simply the net force of the ground (i.e., ground reaction force, or GRF), which is measured using a force plate, and the gravitational force, which can be measured or estimated. A physical model of the body is then constructed based on the quantities that one wants to compute. For example, the leg is often represented using a 2-dimensional link-segment model[5] in

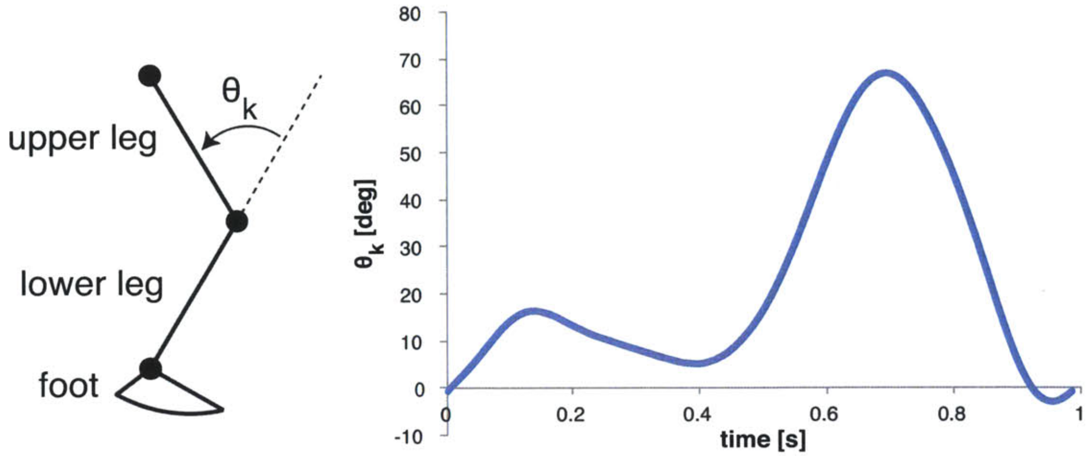

Figure 1-2: On left: convention used for knee joint angle (θ_k). Angular displacement in the positive direction is described as "flexion," whereas angular displacement in the negative direction is described as "extension." On right: knee joint angle v. time over one gait cycle. Peaks correspond to maximum flexion during loading response and maximum flexion during swing. Data adapted from Winter[5] for a 55.6 kg woman walking at a fast cadence.

which the upper leg, lower leg, and foot are modeled as rigid bodies connected via pin joints (Figure 1-3). This model allows one to easily estimate the "net" forces and moments that represent the total effect of the muscles, tendons, ligaments, and bones acting on a segment adjacent to a particular joint.

Once kinematics and external forces are measured and a model is created, joint forces and moments can be computed using a process called inverse dynamics. Conceptually, this process can be thought of as simply drawing a free-body diagram for each segment in the model (Figure 1-3), and then using the Newton-Euler equations to calculate unknown variables. The process of calculating inverse dynamics for 2- and 3-dimensional models is described in detail in Robertson et al[6]. One kinetic quantity often computed in biomechanics is a joint moment, which is the moment acting on a segment of the body adjacent to a particular joint. Sample moment data for the knee are presented in Figure 1-4.

Finally, energetic quantities are computed by combining kinematic data with the results of kinetic calculations. For instance, joint power is computed by multiplying the moment acting about a particular joint by its angular velocity over time. Sample joint power data for the knee is presented in Figure 1-5. The integral of joint power can then be taken to compute joint work, which is directly related to the energy of an adjacent segment through the work-energy theorem.

It is important to note that, although joint work is directly related to mechanical

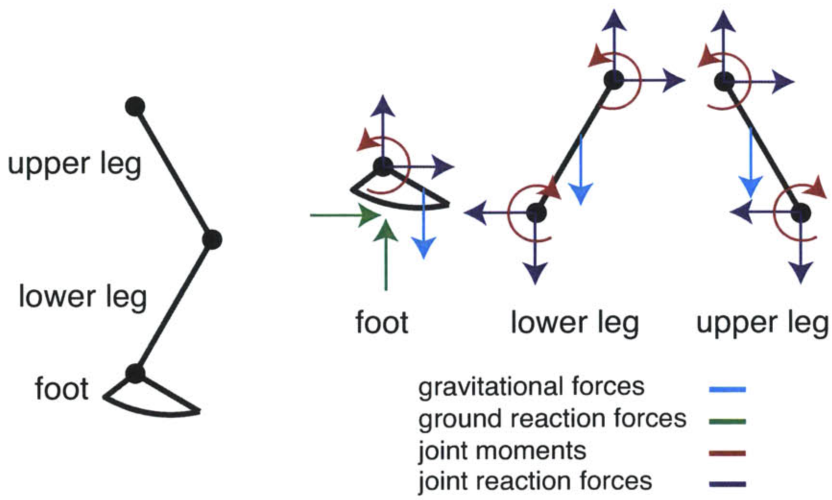

Figure 1-3: On left: sample 2-dimensional link-segment model for one leg. On right: free-body diagrams for all segments.

energy, it is not necessarily related to metabolic energy expenditure (i.e., the chemical energy that the body expends). A simple example of this fact is a person holding a weight at a fixed location. Although the weight is stationary and the person is performing no mechanical work, the muscles rapidly fatigue. The issue of correlating mechanical work with metabolic energy is addressed later in the thesis.

1.2 Existing prosthetic knee joints

An "above-knee prosthesis" is a prosthetic leg that has been designed for individuals amputated above the knee. Typically, an above-knee prosthesis consists of 5 major parts: the suspension, the socket, the knee, the shank, and the foot (Figure 1-6). The present thesis focuses primarily on the design of the knee.

In an able-bodied human, the knee allows a large range of motion, and the muscles of the leg (e.g., quadriceps and hamstrings) control the flexion of the knee as required for a given activity. During walking, normal knee kinematics are critical, as deviations from normal kinematics have been found to increase metabolic energy expenditure[8]. Unfortunately, above-knee amputees typically have reduced muscle function due to muscle loss and atrophy, making flexion of a prosthetic knee difficult to control. The ideal prosthetic knee not only allows a large range of motion, but also replaces lost muscle function by providing appropriate resistance and/or propulsion to allow normal kinematics during walking and other activities[9].

Many different types of prosthetic knees have been designed thus far. In the developing world, a few major categories of low-cost knee exist: manual locking knees, single-axis free-swinging knees, single-axis braking knees, and four-bar knees.

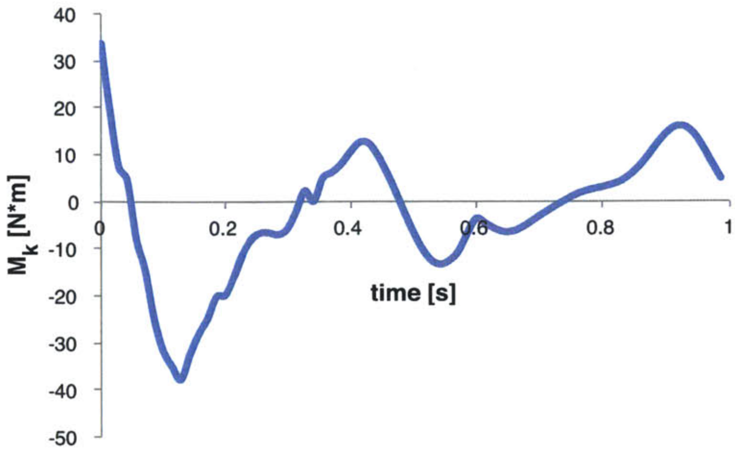

Figure 1-4: Knee joint moment (M_k) v. time over one gait cycle. Positive moments are described as "flexion moments," as they act to flex the knee, and negative moments are described as "extension moments," as they act to extend it. Data adapted from Winter[5] for a 55.6 kg woman walking at a fast cadence.

Two types of manual-locking knees used in the developing world are depicted in Figure 1-7. The BMVSS model is an exoskeletal knee joint, meaning that it is located on the exterior of the prosthesis. It is intended to be in a locked and extended position during walking, but it can be unlocked and flexed during sitting. BMVSS typically prescribes these knees to older patients who may not be able to control a joint that flexes while walking. These types of knees have several known disadvantages. First, they do not allow flexion at the beginning of stance. On flat ground, able-bodied humans flex their knees up to 20° during loading response, providing shock absorption[4]. Second, they do not allow flexion at the end of stance. On flat ground, able-bodied humans flex their knees up to 40° during pre-swing (approximately 67% of peak flexion during swing), facilitating clearance of the leg from the ground during swing[4]. Finally, they do not allow flexion during swing, which often forces the amputee to circumduct the leg (i.e., swing it in a circular motion while bringing it forward) to clear the ground. Such a problem becomes even more severe while walking up inclines. The International Committee of the Red Cross (ICRC) model is an endoskeletal knee joint, meaning that it is located along the centerline of the prosthesis. It is used in either the locked or unlocked position during walking. When locked, the knee behaves like the BMVSS knee, and when unlocked, the joint behaves like a free-swinging knee, which is described below.

A single-axis free-swinging knee is depicted in Figure 1-8. These knees typically resist flexion only through friction within the joint. Additionally, the BMVSS version is an exoskeletal joint that frequently comes with a band at the front of the knee, which resists excess flexion. Like a locked knee, one disadvantage of a free-swinging knee is that it does not allow flexion at the beginning of stance. Another disadvan-

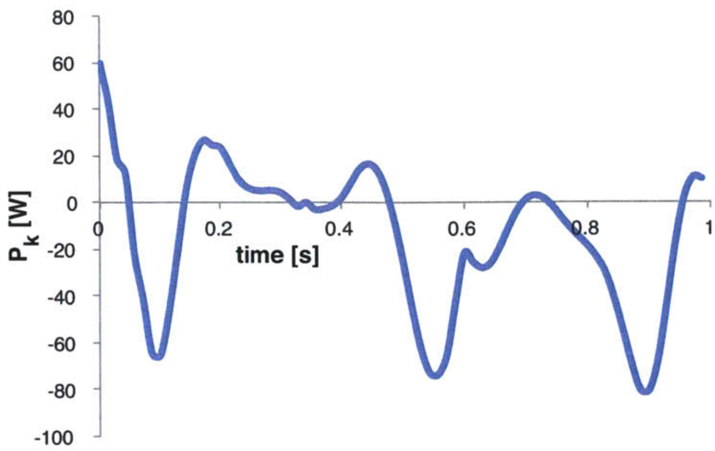

Figure 1-5: Knee joint power (P_k) v. time over one gait cycle. Positive power is generated power, whereas negative power is dissipated power. Data adapted from Winter[5] for a 55.6 kg woman walking at a fast cadence.

tage is that it can buckle during mid-stance when the GRF creates a flexion moment about the knee. Furthermore, a free-swinging knee does not allow the leg to swing with appropriate timing at multiple cadences[10, 11]. To mitigate buckling during stance, free-swinging knees are sometimes stabilized using a technique called "alignment stability," [12] in which the axis of the knee is placed behind the centerline of the prosthesis[11]. Alignment stability ensures that the GRF from early to late stance creates a large extension moment about the knee, preventing the knee from buckling. However, this technique may also delay the initiation of knee flexion during pre-swing to the beginning of swing phase, making it more difficult to clear the leg from the ground[13].

A braking knee, referred to sometimes as a "safety knee," locks upon weight-bearing, preventing flexion during most of stance. Like an alignment-stabilized free-swinging knee, this type of knee may also delay initiation of knee flexion during pre-swing[10]. In addition, it can make sitting-to-standing and standing-to-sitting transitions difficult, as body weight must be shifted to the intact leg at the beginning of the motion[11].

A four-bar knee is depicted in Figure 1-9. As the name suggests, a four-bar knee is constructed as a four-bar linkage, which characteristically have a center of rotation that changes with the angles of the links. Four-bar knees are typically designed such that the center of rotation is behind the knee during stance, behaving similarly to an alignment-stabilized free-swinging knee[13]. The BMVSS model (a.k.a., the Stanford-Jaipur Knee) was designed in conjunction with Stanford University in the late 2000s[14] and has recently been refined by D-Rev[15]. The LIMBS knee was designed by LeTourneau University in the mid-2000s and is one of the few low-cost

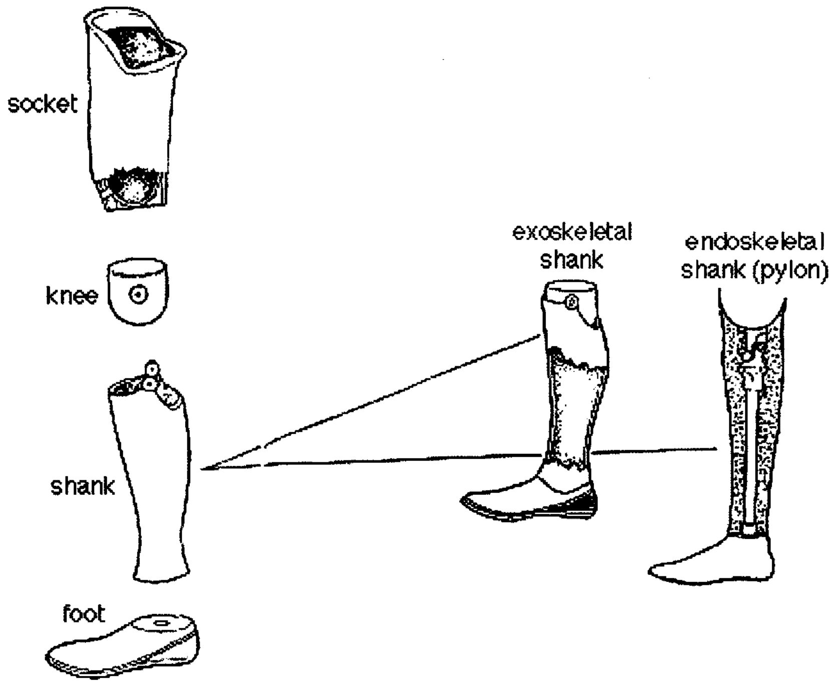

Figure 1-6: Drawing of a typical above-knee prosthesis. In an exoskeletal shank, load is borne upon a shell, whereas in an endoskeletal shank, load is borne primarily upon a pylon. As depicted, an endoskeletal shank may also have a cosmetic cover over the pylon. Adapted from [7].

knees for the developing world to meet ISO 10328 standards[16].

A comprehensive review of other developing-world knee technologies is provided by Andrysek[17]. One notable technology, developed by Andrysek himself, is the LCKnee[18, 19]. The knee locks at the end of swing and unlocks in late stance, mitigating the delayed initiation of flexion present in alignment-stabilized free-swinging knees, braking knees, and four-bar knees.

In the developed world, prosthetic knees can be divided into 2 major categories: "passive knees," which do not contain an energy source, and "active knees," which do. Like knees for the developing world, passive knees may be constructed in single-axis[20, 21] or four-bar[22] form, but they typically contain a resistive element that is not constant-friction, most commonly a hydraulic damper. Hydraulic knees have been designed that allow flexion at the beginning of stance phase[23]. In contrast to free-swinging knees, a hydraulic knee enables the leg to swing with appropriate timing at multiple cadences[10].

Active knees are typically powered by batteries. Like passive knees for the developed world, they also typically contain a hydraulic damper. Sensors are used to collect kinematic and kinetic data (e.g., knee angle, ankle moment, and knee reac-

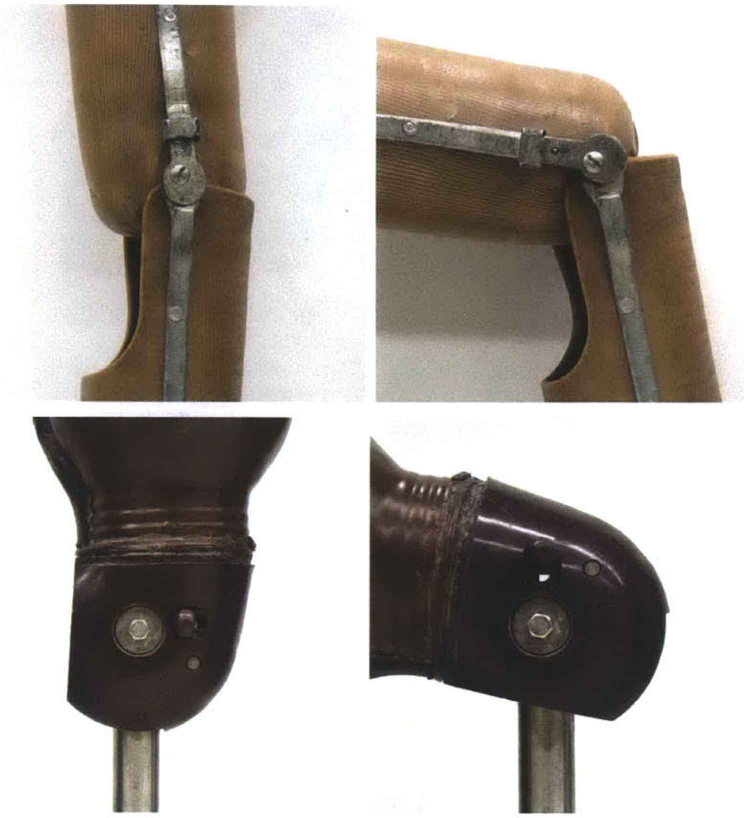

Figure 1-7: Two types of manual-locking knees. Top-left: BMVSS manual-locking knee, fully extended. Top-right: BMVSS manual-locking knee, flexed to nearly 90°. Bottom-left: ICRC manual-locking knee, fully extended. Bottom-right: ICRC manual-locking knee, flexed to nearly 90°.

Figure 1-8: BMVSS single-axis free-swinging knee. On left: fully extended position. On right: flexed to nearly 90°. A band to resist excess flexion is visible at the front of the knee.

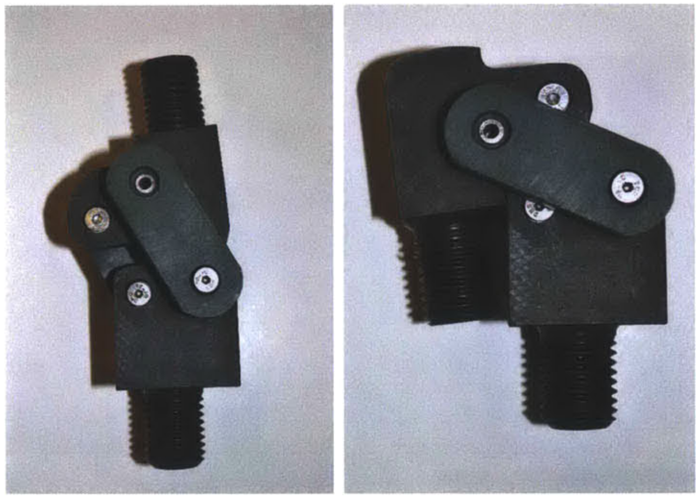

Figure 1-9: Stanford-Jaipur knee joint. On left: fully extended position. On right: fully flexed position.

tion force), and a microprocessor analyzes the data to identify whether the user is in stance or swing and what activity is being performed (e.g., walking fast, walking on uneven terrain, or climbing stairs). The microprocessor then commands motors to adjust valves in the damper, changing the resistance appropriately[9]. Variable damping has also been accomplished in prosthetic knees by using magnetorheological fluid, which changes viscosity based on the applied magnetic field[24, 25]. Recently, active knees have also been used not only to adjust mechanical elements within the knee, but also to actively propel the knee during power-intensive activities like stair climbing[26].

Active knees have been designed to allow flexion during loading response[23, 27], although experimental results have been mixed[27, 28]. Active knees have also been shown to allow flexion during pre-swing[23, 28] and reduce mechanical and metabolic energy expenditure of above-knee amputees[28, 29].

Table 1.2 presents costs for several types of knee joints used in the developing world or developed world. Based on available data, either manufacturing cost, retail price, or manufacturer's suggested retail price (MSRP) are reported for each knee, meaning that only approximate comparisons can be made. Regardless, it is clear that state-of-the-art prosthetic knees in the developed world cost several orders of magnitude more than prosthetic knees in the developing world. As an example, the Otto Bock C-Leg, the most widely used microprocessor knee in history[30], has an MSRP that is 2700 times as much as the manufacturing cost of the Stanford-Jaipur Knee.

Name	Organization	Type of knee	Cost	Type of cost	Source(s)
Manual-locking knee	BMVSS	Manual-locking	< $10	Manufacturing	[31]
LIMBS Knee	Limbs International	Four-bar	$15-20	Manufacturing	[32]
Stanford-Jaipur knee	BMVSS	Four-bar	$20	Manufacturing	[33]
LCKnee	Jan Andrysek	Single-axis, auto-locking	$50-100	Not reported	[34, 19]
Niagara Knee joint	Niagara Prosthetics & Orthotics	Single-axis, free-swinging	$147	Retail	[35]
802 Nylon Knee	Aulie Devices, Inc.	Single-axis, hydraulic	$2,060	Retail	[36]
C-Leg	Otto Bock	Single-axis, microprocessor, hydraulic	$54,510	MSRP	[37]
Genium	Otto Bock	Single-axis, microprocessor, hydraulic	$75,000	MSRP	[38]

Table 1.1: Costs of various prosthetic knees for the developing and developed world.

1.3 Outline of thesis

As described earlier, the present thesis aims to translate the general design requirements given by BMVSS and other Indian stakeholders into a detailed set of requirements to drive the design of a prototype. Quantitative and qualitative interviews of amputees were conducted to understand user requirements, and simulations and optimizations were performed in order to determine inertial properties and mechanical elements that would allow amputees to walk with normal knee kinematics and low energy expenditure. An outline of the thesis is as follows:

- **Chapter 2: User Factors** A structured survey of transfemoral (above-knee) amputees in India was conducted to understand the capabilities of current low-cost prostheses and identify the most critical areas for design improvement. Results from the survey are analyzed both quantitatively and qualitatively.

- **Chapter 3: Inertial Properties** Dynamic simulation and optimization techniques are used to determine inertial properties of an above-knee prosthesis that minimize energy expenditure. This section is presented in manuscript form and can be read independently.

- **Chapter 4: Component Selection** Dynamic simulation and optimization techniques are used to determine mechanical elements for a prosthetic knee that allow it to produce moments on the leg to replicate normal gait kinematics. This section is also presented in manuscript form, but it uses results from the previous chapter.

- **Chapter 5: Conclusion** The results of the thesis are summarized and connected, and suggestions for future work are outlined.

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Chapter 2

Identification and Analysis of Critical Areas for Design Improvement Based on Quantitative and Qualitative Feedback from Indian Amputees

2.1 Introduction

Significant literature has been published on design, manufacturing, dissemination, and evaluation considerations for prostheses in the developing world[1, 2, 3, 4, 5, 6]. Among the broader accomplishments of the literature is a clear definition of "appropriate technology"[7] as applied to the field of prosthetics: "a system providing proper fit and alignment based on sound biomechanical principles which suits the needs of the individual and can be sustained by the country at the most economical and affordable price"[2].

Identifying the "needs of the individual" is a challenge for designers of low-cost prostheses, particularly those in academic settings, as it often requires significant and frequent interaction with stakeholders who live hundreds to thousands of miles away from research laboratories. Although remarkable work has been conducted in designing appropriate prostheses for amputees in India[8], few studies have been published that provide specific feedback from Indian amputees on the strengths and weaknesses of existing devices or desired improvements. The most comprehensive study found was that of Narang et al[9] (of no relation to the author), who surveyed 500 lower limb amputees in Pune, India in 1984. His study recorded the demographics of amputees and evaluated their ability to perform a wide range of activities, such as dressing, bathing, sit-stand transitions, and walking on ramps.

The purpose of the present study was to extend the work of Narang et al to an alternate and recent population of transfemoral amputees. In addition, the goal was to use the results to formulate design requirements for a low-cost prosthetic knee with improved functionality relative to existing devices. A structured survey of 19 transfemoral amputees was conducted by the author in Jaipur, India in January 2013. The amputees were asked about their current ability to locomote in different conditions and the impact that additional abilities would have on their lives. Quantitative and qualitative results of the surveys are reported here, and the outcomes are used to identify additional functionality that an improved knee should possess.

2.2 Background

To provide context for the survey results and subsequent discussion, this section begins with a brief introduction to India and Indian amputees.

2.2.1 Background on India

India is a highly populous country with a diverse population, economic challenges, and varied terrain. Specifically, India has a population of 1.22 billion, making it the second-most populous country in the world. Approximately 30% of the population live in urban areas, whereas 70% live in rural areas. Fifty-two percent are male and 48% are female. An estimated 29.8% of people live below the poverty line, and 9.9% of people are unemployed. The two most practiced religions are Hinduism and Islam, with 80.5% and 13.4% of the population following each faith, respectively. The terrain varies from flat and upland plains in the North and South, to deserts in the West, to mountains in the Far North. The climate is temperate in the North and tropical monsoon in the South[10].

2.2.2 Demographics of Indian amputees

Few quantitative studies have been published on the demographics of amputees in India. The most comprehensive study found was the "Disabled Persons in India" report published by the National Sample Survey Organisation (NSSO) of India in 2003[11]. The NSSO surveyed a total of 7,991 cities, towns, and villages and 70,302 households in the latter half of 2002 to determine the demographics of physically and mentally disabled people across India.

According to the report, 1.0% of all people in India have locomotor disability. Among those with locomotor disability, 7.7% are amputees. Thus, 0.077% of all people in India are amputees, which corresponds to approximately 950,000 individuals when projected to the current population. The distribution of amputees by type of amputation is not available in the report. Narang and Jape[12] published an earlier study on 14,000 amputees treated in Pune, India, and found that 62% were lower-limb amputees, among which 43% were unilateral above-knee amputees, 1.4% were bilateral above-knee amputees, and 2.2% were bilateral amputees with one above-knee and one below-knee amputation. If these percentages are applied to the earlier estimation of the total number of amputees in India, then there are approximately 440,000 above-knee amputees in India today.

Among amputees, 25% live in urban areas and 75% live in rural areas, closely matching the geographic distribution of the general population. The regional distribution of people with locomotor disability is varied. Densities in continental India are highest in the North and Northwest (Punjab, Himachal Pradesh, Uttar Pradesh, and Haryana) and lowest in the Northeast (Mizoram, Nagaland, and Arunachal Pradesh). High densities are also present in other areas, such as Central India and the Far South[11].

76% of amputees are male and 24% are female, indicating a clear imbalance relative to the sex distribution of the overall population. Agewise, the number of people with locomotor disability decreases steadily from the 10-19 decade to the 50-59 decade, with the latter having 40% the number of affected people as the former[11].

The NSSO does not publish socioeconomic data on Indian amputees, and relevant data from other current sources is scarce. Mohan[13] published some statistics on the financial status of Indian amputees based on two earlier studies[14, 15] and concluded that the majority of amputees were experiencing poverty.

2.2.3 Activities of Daily Living (ADL) in India

'ADL' is a term in rehabilitation defined as "activities or tasks that people undertake routinely in their every day life." The term is divided into basic ADL (functional mobility and personal care) and instrumental ADL (domestic and community activities)[16].

Most ADL, such as bathing, dressing, and using the toilet, are generally no different in India than in other countries, but certain characteristics distinguish how Indian people perform the activities. Mulholland and Wyss[17] conducted one of the few studies to examine these characteristics in detail. The study noted that floor-sitting postures, specifically squatting, kneeling, and cross-legged sitting, were critical in South Asia to basic ADL like bathing and using the toilet, as well as instrumental ADL like praying and socializing. Some ADL are indeed different in India relative to the developed world, such as walking from place to place across uneven village terrain.

The subsequent survey evaluates the ability of Indian amputees to perform a set of fundamental activities observed to be relevant to basic ADL and instrumental ADL in India. For example, the survey evaluates the ability of amputees to carry heavy objects, which is critical for lifting materials in agricultural and industrial work.

2.3 Methods

A survey was administered by the author at the Bhagwan Mahaveer Viklang Sahayata Samiti (BMVSS, a.k.a., Jaipur Foot) limb fitment center in Jaipur, India during January 2013. The survey was approved by the Massachusetts Institute of Technology (MIT) Committee on the Use of Humans as Experimental Subjects (COUHES), and in accordance with COUHES requirements, a consent form was administered prior to each survey. Both the consent form and survey were presented orally, as many interviewees had limited reading and writing ability. The documents were presented with the assistance of a translator in English, Hindi, or Marwari, depending on the language the interviewee was most comfortable speaking. A total of 19 subjects were interviewed.

The survey focused on answering two major questions: 1) What activities are Indian transfemoral amputees able to perform easily with existing low-cost prostheses? and 2) Out of those activities they are unable to perform easily, would their lives be significantly improved if they were able to perform them? The responses to these questions allow a designer to identify critical activities for design improvement in current prostheses, defined here as activities that are both difficult for amputees to perform and important to their lives.

The questions were asked about each activity within a comprehensive list of 22 activities, since interviewees were often hesitant to name activities on their own. In addition, the questions were presented in a binary format (e.g., easy/not easy), as interviewees were generally uncomfortable with a 3- or 5-point scale. Figure 2-1 shows a simplified flow chart of the survey, and Appendix A contains the original consent form and survey. Descriptive statistics were performed on the collected data. In addition, to identify critical activities for design improvement, a metric was created here called the Potential Impact of Design Improvement (PIDI) score. The metric was defined simply as follows:

$$PIDI = D * I \tag{2.1}$$

where $PIDI$ is the PIDI score of the activity, D is the percentage of amputees who felt that the activity was difficult (i.e., not easy), and I is the percentage of those amputees who felt that the ability to perform the activity with a different prosthesis would significantly improve their lives. Comparing PIDI scores among different activities allows designers to identify which ones should be made easier to perform with new prostheses.

2.4 Results

2.4.1 Subject demographics

Table 2.4.1 presents characteristics of the subject population. The percentage of male subjects (100%) is 24% greater than the percentage of male amputees in India re-

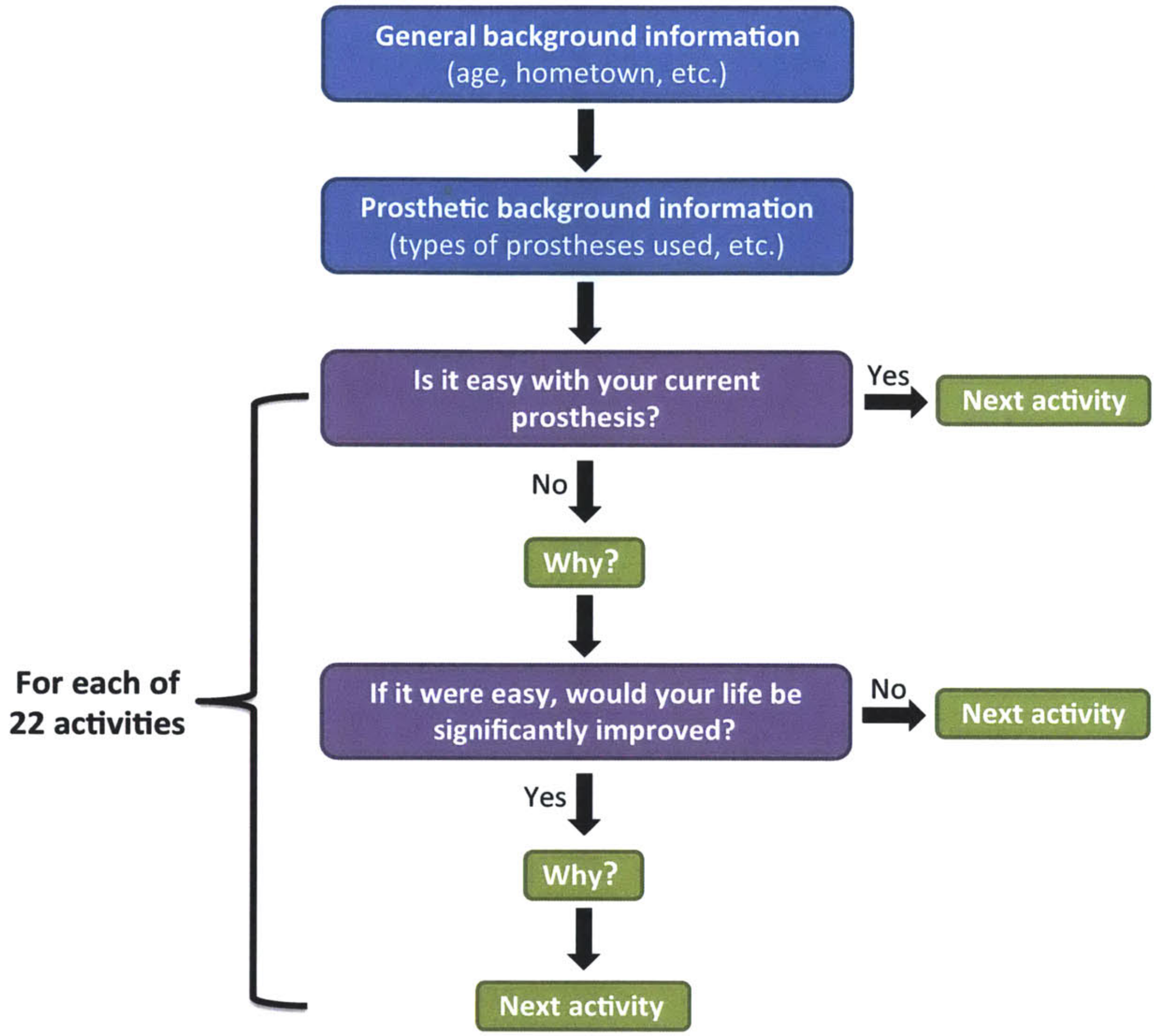

Figure 2-1: Survey flow chart

ported by the NSSO (76%), suggesting that a higher percentage of male amputees than female amputees are willing and able to obtain prostheses[11]. In addition, the percentage of males in the current study is 11.63% higher than that reported by Narang and Jape[12]. The precise reason for this discrepancy is unknown, but may be related to differences between Jaipur and Pune in societal norms that allow female amputees to obtain prostheses.

The average age of subjects (36) was 14 years less than the average age of amputees in the United States (50)[18]. The reason for this discrepancy can be illuminated by examining the cause of amputation. The present study found that 74% of amputations were caused by transportation accidents, supporting the trends reported by Narang and Jape[12]. On the other hand, only 37% of amputations in the United States are caused by trauma (which includes transportation accidents), and 34% are caused by vascular disease, which are typical of older patients[18]. Thus, the discrepancy is

Attribute	N	Value
Gender	19	100% male, 0% female
Age	19	36.3 $\pm$ 15.6 years
Home state	19	Uttar Pradesh (42%), Bihar (16%), Madhya Pradesh (16%), Chattisgarh(5%), Haryana (5%), Jammu (5%), Punjab (5%), Rajasthan (5%)
Hometown	17	Village (76%), City (18%), Town (6%)
Cause of amputation	19	Transportation accident (74%), Cancer (16%), Infection (5%), Violent crime (5%)
Occupation before amputation	19	Student (32%), Non-agricultural manual laborer (26%), Farmer (16%), Driver (11%), Manager (11%), Shop-worker (5%)
Current occupation	19	Unemployed (42%), Manager (21%), Non-agricultural manual laborer (11%), Student (11%), Security guard (11%), Farmer (5%)
Type of knee joint	19	Jaipur Foot locked exoskeletal (53%), Jaipur Foot single-axis (26%), Jaipur Foot four-bar (16%), ICRC locked (5%)
Walks barefoot outdoors	18	11% yes, 89% no
Uses assistive devices	19	64% yes, 36% no
Falls per month	18	Average: 0.65, Range: 0-4

Table 2.1: Summary of subject demographics

likely a result of increased incidence of traffic accidents in India, as well as shorter life expectancies.

Most subjects lived in the North and Northwest, aligning with previously mentioned statistics from NSSO data. However, this result is likely influenced by the fact that the BMVSS center itself is located in the Northwest. In addition, 76% of subjects in the present study were from villages, and previous computations based on NSSO data determined that 75% of amputees lived in rural areas. Since most Indian villages are in rural areas and rural areas primarily consist of villages, these numbers are in agreement. In addition, as 43% of Indian villages are not even linked with roads[19], it can be assumed that most Indian amputees walk frequently on uneven terrain. Conversations with many amputees at BMVSS supported this claim.

Whereas 100% of the non-student subjects were employed prior to amputation, only 53% of the non-student subjects were employed at the time of the interview. In addition, although questions about family and marital status were not explicitly asked, numerous subjects related stories about neglect or abandonment by spouses and children as a result of their disability. These results highlight the financial and social impact of amputation, depriving individuals of the ability and opportunity to be employed and sustain personal relationships.

95% of the subjects used prosthetic knees manufactured by BMVSS, and the results of this study may be influenced by functionality specific to prostheses made by this institution. Nevertheless, BMVSS has distributed approximately 400,000 artificial limbs in India[20], and throughout the author's travels in India in 2012 and 2013, the vast majority of low-cost above-knee prostheses were seen to be manufactured by BMVSS, derived from BMVSS technology, or similar in construction to BMVSS models. Thus, the results of this study may be generalizable to a large percentage of prosthesis users in India.

Numerical data is scarce, but Sethi et al[8] remarked that "the average rural Indian does not wear shoes." In context, the comment was applied to walking outdoors. The present study observed that 89% of subjects did not walk barefoot outdoors, including 87% of those living in villages. Sethi made the observation in 1978, and it may be possible that national economic growth and an increasing availability of footwear over the past 35 years has caused the percentage of amputees walking barefoot outdoors to significantly decrease.

64% of subjects said that they used assistive devices, such as canes, crutches, and wheelchairs, at least occasionally. The most commonly cited scenarios in which they used assistive devices were bathing and using the toilet, during which the limb was often removed and the devices were used for support. Subjects reported a low incidence of falls, averaging less than one per month. This statistic seemingly contradicts frequent comments by the subjects about stability problems with their prostheses. However, a lack of stability may lead to a high incidence of slips or trips rather than falls.

2.4.2 Difficulty of activities

Figure 2-2 shows the difficulty of each activity, measured as the percentage of subjects who felt that it was not easy to perform with their current prosthesis. The most difficult activity was walking on snow, as subjects reported that they frequently slipped. However, only two subjects had ever walked on snow before, and more data is needed to confirm the difficulty of the activity.

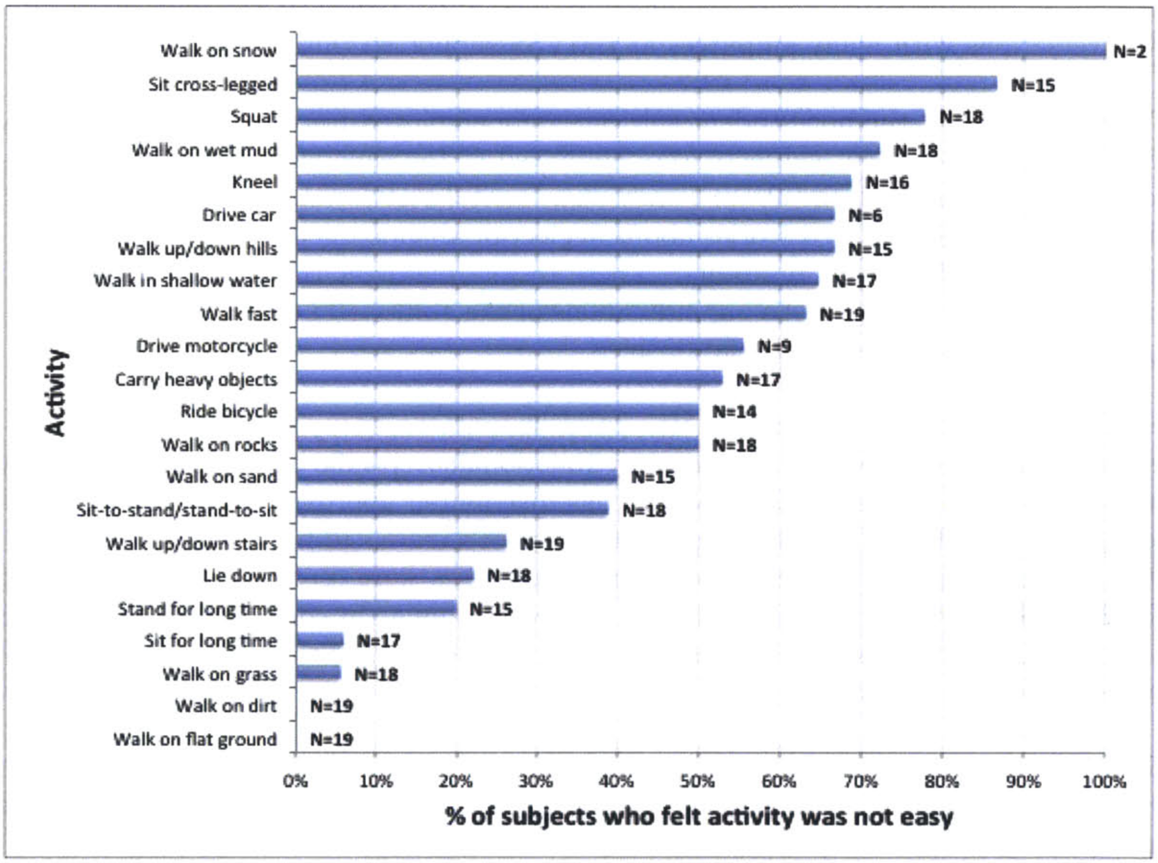

Figure 2-2: Difficulty of each activity, measured as the percentage of subjects who felt it was not easy. $N < 19$ for some activities primarily because 1) many subjects had never done certain activities before, such as walking on snow or driving a car, and 2) some surveys were interrupted one or more times.

The top five most difficult activities included the three floor-sitting postures (sitting cross-legged, squatting, and kneeling) discussed by Mulholland and Wyss[17]. On cross-legged sitting, subjects stated that the prosthesis "doesn't fold" in the varus (inward) direction, since the knee joint only allowed flexion in the sagittal (front-back) plain. On squatting, subjects stated that the prosthesis "doesn't flex," because the socket collided with the shank or, with the Stanford-Jaipur knee, one part of the knee collided with another (Figure 2-3). The collision obstructed the deep flexion required to position the center of gravity of the body over the feet, leading to instability. On kneeling, subjects stated again that collision in the prosthesis prevented the required flexion. Several subjects also expressed that they felt pain during one or more of the floor-sitting postures, typically from socket pressure at the upper thigh.

Figure 2-3: Illustration of squatting difficulties in a BMVSS above-knee prosthesis with the Stanford-Jaipur knee. Circled in red are collisions of 1) the socket with the shank and 2) the upper part of the knee with the lower part. Both collisions obstruct the deep flexion required for squatting.

At least 50% of subjects found that walking on certain types of uneven ground (wet mud, shallow water, and rocks) was difficult for various reasons. They stated that on wet mud, they got stuck or slipped; in shallow water, they slipped; and on rocks, they slipped or stumbled. On why they stumbled, a few subjects said that it was difficult to clear the leg from the ground during swing, which resulted in the leg catching the ground.

In addition, at least 50% of subjects also stated that walking up/down hills, walking fast, and carrying heavy objects were not easy to do with their current prostheses. Subjects frequently said that they felt unstable, feared falling, or felt pain and fatigue during these activities. Three subjects remarked that they had timing difficulties when walking fast, referring to the inability to lift and swing the leg fast enough at the beginning of swing and/or the inability to slow the leg down at the end of swing (Figure 2-4). From observations, these difficulties appeared to related to a few different factors:

- The weight of the prosthesis, which made it difficult to lift
- The tightness of the prosthesis at the upper thigh, which made it painful to lift
- The lack of sufficient flexion in the knee before toe-off, which then required a compensatory hip hike to clear the leg from the ground during swing
- The lack of flexion in the locked joint during swing, which then required circumduction for ground clearance
- The lack of sufficient resistance in the free-swinging knees during swing, which caused the leg to swing forward too quickly when walking fast

Figure 2-4: Time series showing how a subject stumbled while walking fast. The subject did not lift and swing the leg fast enough at the beginning of swing, causing the foot to drag on the ground and the subject to momentarily lose balance.

One subject also mentioned that he could not clear the leg from the ground while walking uphill. In addition, a few subjects wearing the Stanford-Jaipur knee displayed an abnormal kinematic pattern when walking down inclined surfaces, during which the knee would be forcefully extended and lowered at the end of swing to ensure that the leg did not buckle on weight bearing (Figure 2-5).

Figure 2-5: Photograph of the Stanford-Jaipur knee in a forcefully extended and lowered position. The subject walked down an incline and was initiating weight bearing on the prosthetic leg.

Finally, many subjects also said that driving a car, driving a motorcycle, and riding a bicycle were difficult activities. Relatively few subjects had operated cars or motorcycles, but several that had done so said that they struggled to manipulate the foot pedals with their prostheses. In addition, subjects stated that during cycling, it was difficult to flex the knee rapidly and fully due to excess resistance within the knee (likely from friction) and limited range of motion from collision in the prosthesis. In addition, they felt a general lack of balance.

Surprisingly, few subjects rated sit-stand transitions and walking up/down stairs as difficult activities. Based on observations at BMVSS, this result was likely due to the presence of supporting structures (e.g., armrests, rails, and walls) on most chairs and staircases, which allow subjects to grasp or lean against an object during the activities.

2.4.3 Importance of activities

Figure 2-6 shows the importance of each difficult activity, measured as the percentage of subjects who felt that the ability to perform the activity with a different prosthesis would significantly improve their lives. For every single activity rated as not easy, at least 70% of subjects felt that the ability to do the activity easily would significantly improve their lives. Furthermore, for 75% of the activities rated as not easy, 100% of subjects felt that the ability would lead to an improvement. The trend strongly suggests that the restoration of any major function that an able-bodied human can perform would be meaningful and significant to the lives of transfemoral amputees. At the same time, since designers of low-cost prostheses cannot yet restore all functions at once, additional surveys are needed in which the importance of various activities is evaluated with higher resolution.

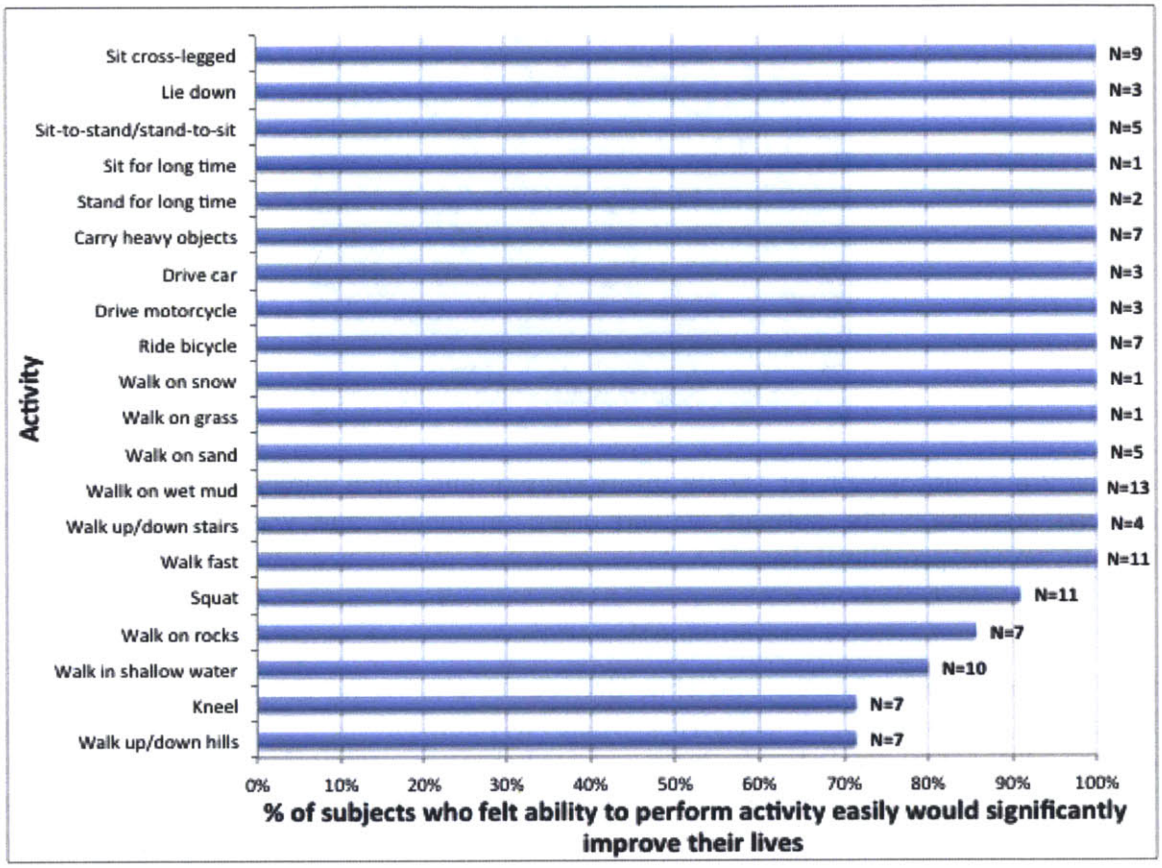

Figure 2-6: Importance of each activity, measured as the percentage of subjects who thought it was difficult and felt that the ability to do it easily would significantly improve their lives. Activities that no subject thought was difficult (i.e., walking on flat ground and walking on dirt) were not included.

When asked why the ability to do certain activities easily would significantly improve their lives, subjects typically provided reasons related to work and general well-being. About 72% specifically said that the ability to walk fast, walk on uneven terrain, carry heavy objects, and sit cross-legged easily would allow them to do more work or get a better job, and 50% said that the ability to do additional activities would give them confidence and allow them to "live a better life" and "feel that [one] is not impaired." Several subjects said that the ability to walk on uneven terrain easily would enable them to get to certain places (e.g., workplace, market, or school) around their hometowns and that operating vehicles would allow them to "go anywhere." As anticipated, many subjects also stated that the ability to squat easily would allow them to use Indian toilets or outdoor toilets without trouble, and that sitting cross-legged would allow them to eat properly or do work (e.g., tailoring) on the floor.

2.4.4 Potential impact of design improvement

Finally, Figure 2-7 shows a comparison of PIDI scores among the different activities. Activities for which $N < 10$ (walking on snow, driving a car, and driving a motorcycle) were excluded to ensure that rare activities were not overrepresented, and walking on grass and flat ground were excluded because no subjects rated them as difficult. "Tier 1" activities are hereby defined as those with a PIDI score in the top quartile (technically, top 24%), and "tier 2" activities are those in the second quartile. Tier 1 activities consisted of sitting cross-legged, walking on wet mud, squatting, and walking fast, and tier 2 activities consisted of carrying heavy objects, walking in shallow water, riding a bicycle, and kneeling.

The three floor-sitting postures, which were among the most difficult activities, were also among those with the highest PIDI score. However, sitting cross-legged and squatting were in tier 1, whereas kneeling was at the bottom of tier 2. The discrepancy is likely because none of the subjects interviewed followed the Muslim faith, in which kneeling is essential for prayer. Thus, a relatively small percentage of subjects felt that the ability to kneel would significantly improve their lives, resulting in a lower PIDI score.

Walking through mud and walking fast were both in tier 1. The difficulty of performing both activities was discussed earlier. Subjects stated that the ability to walk through mud would improve their lives because it would allow them to get directly to destinations rather than taking a detour, particularly after rainfall. As expected, subjects said that walking faster would allow them to get to places more quickly and save time.

Carrying heavy objects, which was not among the most difficult activities, was at the top of tier 2 because 100% of subjects who were not able to carry heavy objects easily felt that it would significantly improve their lives if they were able to do so. A few subjects explained the importance of the activity in detail, stating that it would allow them to lift objects for work or bags for school.

The remaining activities in the top tiers were walking in shallow water and riding a bicycle, which were both in tier 2. The difficulty of performing both activities was described earlier. Similar to the importance of walking on wet mud, subjects stated that the ability to walk in shallow water would significantly improve their lives because they would be able to walk outside after rainfall. In addition, subjects said that the ability to ride a bicycle would allow them to commute to and from the workplace

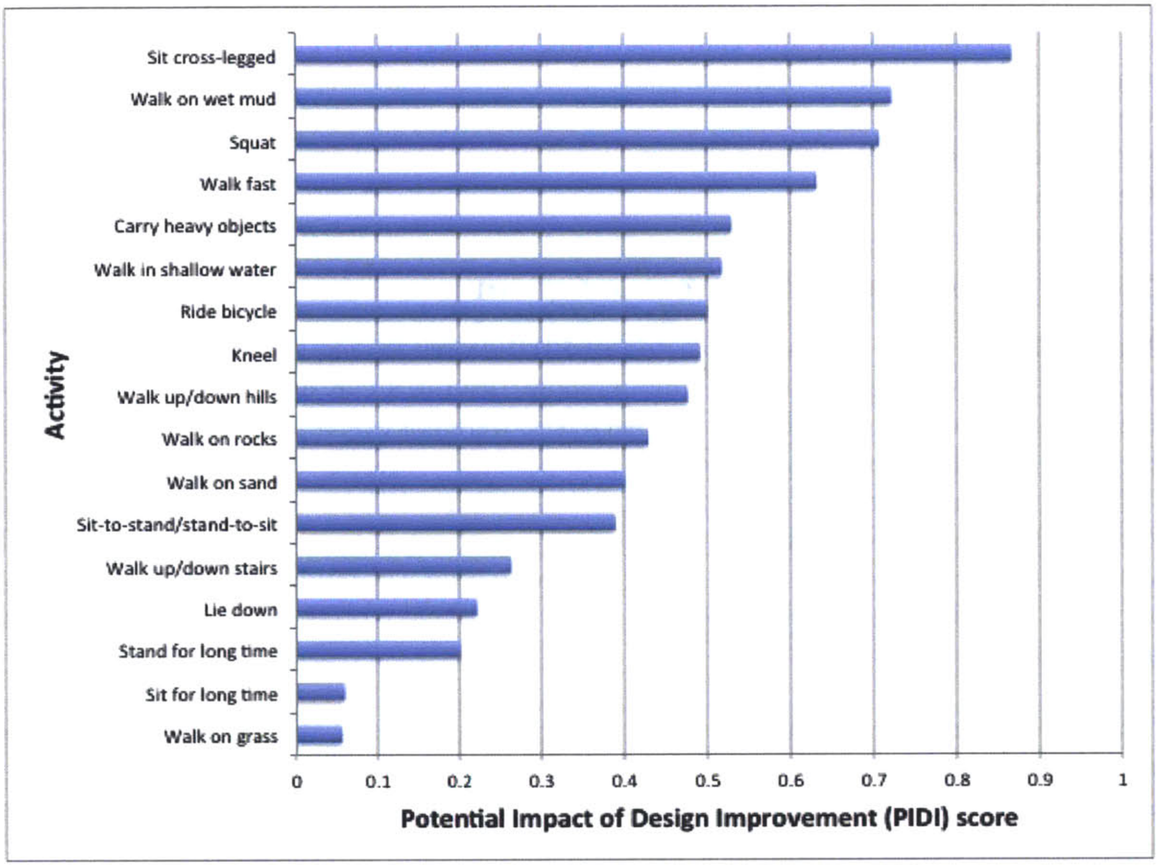

Figure 2-7: PIDI score for each activity, measured as the percentage of subjects who thought it was difficult multiplied by the percentage of those subjects who thought that the ability to do it easily would significantly improve their lives. To prevent overrepresenting activities that few subjects had ever performed, activities for which $N < 10$ (e.g., walking on snow and driving a car) were not included.

easily.

2.5 Discussion

The motivation of the present study was to survey a population of transfemoral amputees in India and use the results to formulate design requirements for a new low-cost prosthetic knee. The new knee aims to exceed the functionality of existing low-cost knees; as a result, the emphasis of the study was on identifying what additional functionality an improved knee should possess. In order to determine desired functionality, the difficulty and importance of a number of activities were evaluated, and the PIDI score was defined as the product of the two in order to identify the greatest areas for improvement.

Relative to existing solutions, the most critical activities for design improvement are

1. Cross-legged sitting
2. Walking on wet mud
3. Squatting
4. Walking fast

In addition, activities that are nearly as critical are

1. Carrying heavy objects
2. Walking in shallow water
3. Riding a bicycle
4. Kneeling

The kinematics and kinetics of cross-legged sitting, squatting, and kneeling have been examined in detail in the literature. Hemmerich et al[21] determined the maximum angular displacements for all the joints as subjects moved into and out of the three postures. For cross-legged sitting, the knee flexed an average of $150.0° \pm$ a standard deviation of $8.1°$. For squatting with the heels down (the only method of squatting observed by the present author in India), the knee flexed an average of $153.7 \pm 10.4°$. Finally, for kneeling with the feet dorsiflexed (instep facing the ground), the knee flexed an average of $154.9 \pm 8.6°$. Knee moments into, during, and out of deep squatting[22] and kneeling[23] have also been investigated and found to be significantly larger than the moments during normal walking. However, moments for cross-legged sitting were not found in the literature. An improved prosthesis must allow the appropriate ranges of motion in the knee joint, provide appropriate resistance while transitioning into and out of the given posture, and have a stable weight-bearing surface at the limits of the range of motion. Quantifying "appropriate resistance" can be performed using the cited kinematic and kinetic data in conjunction with the optimization techniques described in Chapter 4.

No studies could be found on the gait kinematics or kinetics of walking on mud. Still, one can use intuition to analyze the activity. Walking on mud is difficult for two major reasons: first, the foot sinks due to high pressure and low resistance at the foot-floor interface, and second, the mud collects and adheres to the foot or footwear. The former causes instability on weight-bearing, and both require greater work to be done by the muscles to lift the leg from the ground. While designing an above-knee prosthesis, the difficulty of walking in mud could be reduced by 1) decreasing the masses of the segments of the prosthesis, or 2) redesigning the foot or footwear to have reduced adherence to mud and reduced pressure on weight-bearing. The effects of decreasing mass are examined in Chapter 3, but design of the foot and footwear is left for later research.

The kinematics and kinetics of walking fast, walking in shallow water, and walking while carrying loads have been examined to various extents in the literature. Walking fast will be considered in detail in Chapter 3 and Chapter 4. Walking in shallow water has been examined for water at the level of the chest[24, 25], but studies for water at more commonly experienced levels (e.g., below the knee) were not found. Since subjects in the present study stated that walking in water was difficult primarily due to slipping, design improvements would likely require modification of the sole of the foot or footwear, which is not considered in this thesis. Walking while carrying loads has been examined extensively in the literature[26, 27, 28, 29, 30]. Generally speaking, the magnitudes of both flexion angles and moments increase with increasing load, and the moment-angle relationship changes, suggesting that the ideal resistance provided by a prosthetic knee may change with load. Again, quantifying the ideal resistance can be performed using optimization techniques described in Chapter 4.

The kinematics and kinetics of cycling have also been examined in the literature. However, the energetics of cycling are more informative to the present research, as it has been found that cycling requires significant energy generation over the gait cycle[31]. Thus, a low-cost, passive prosthesis may not allow an amputee to cycle with normative kinematics. However, microprocessor-controlled knees such as the Otto Bock C-Leg are sometimes programmed to provide minimal resistance during cycling[32] to grant the hip musculature full control of the prosthesis, a strategy which could be adopted easily by a passive prosthesis.

Another final area of design improvement is a general one. In reference to many activities, such as walking on uneven terrain and carrying heavy objects, subjects described instability that caused them to stumble and/or the knee to buckle. Increased resistance upon a sudden perturbation would allow them to successfully recover from a stumble using the "lowering strategy" of able-bodied humans during, during which the leg is quickly lowered to the ground after a perturbation in early swing. However, increased resistance may impede an alternate strategy called an "elevating strategy," during which the leg is lifted and brought forward after a perturbation in late swing[33, 34]. Which strategy transfemoral amputees typically use and should use is the subject of current research[35]. Regardless, the prosthetic knee should act with appropriate resistance to ensure that the subject can safely recover from a stumble.

Based on the preceding results and discussion, the additional design requirements that an improved low-cost knee should possess are as follows:

- Allows appropriate range of motion for cross-legged sitting (as defined earlier) and appropriate resistance when transitioning into and out of the sitting position
- Allows appropriate range of motion for squatting (as defined earlier), appropriate resistance when transitioning into and out of the squatting position, and weight-bearing stability at the limit of the range of motion
- Reduced mass of the knee and adjacent components to mitigate sinking of the leg while walking in mud
- Provides appropriate resistance during walking at fast speed, as determined in Chapter 4

- Enables (or does not impede) recovery from stumbling

Additional functional requirements that should also be considered include the following:

- Provides appropriate resistance during walking while carrying loads
- Provides appropriate (perhaps, negligible) resistance during cycling
- Allows appropriate range of motion for kneeling (as defined earlier), appropriate resistance when transitioning into and out of the kneeling position, and weight-bearing stability at the limit of the range of motion

As a final note, it should be mentioned that although the activities listed at the beginning of the discussion correspond to critical areas for design improvement, advancing the baseline performance of existing prosthetic knees should not be ignored. For instance, although all subjects in the present study stated that walking on flat ground was "easy," subjects frequently complained of timing difficulties, instability, pain, and fatigue. Since walking on flat ground is the primary use of an above-knee prosthesis for the vast majority of transfemoral amputees, this activity (as well as walking at various cadences) will be the focus of Chapter 3 and Chapter 4. The theoretical techniques developed in both sections can be easily applied to improve performance of the aforementioned activities as well.

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Chapter 3

The Effects of the Inertial Properties of Above-Knee Prostheses on the Energetics of Transfemoral Amputees

Abstract

Transfemoral amputees using above-knee prostheses expend significantly more metabolic energy than able-bodied humans during walking. Experimental and theoretical studies suggest that altering inertial properties of prostheses may reduce energy expenditure. The present study conducted a 2-dimensional inverse dynamics simulation over the entire gait cycle to determine the effects of the inertial properties of each segment of an above-knee prosthesis on the energy expenditure required to walk with normative kinematics. Ground reaction forces were estimated for the prosthetic leg, and energy expenditure was estimated through a measure of muscular effort at the hip. The simulation was conducted at multiple cadences. Masses of the segments, instead of moments of inertia (about centers of mass), were found to be the principal inertial determinants of energetic savings. Decreasing foot mass and lower leg mass reduced energy expenditure by up to 22%, and decreasing foot mass reduced energy expenditure by up to three times as much as decreasing lower leg mass. Minimizing masses of the segments resulted in minimal energy expenditure at all cadences. Major results are reported in the form of parametric illustrations that can be used by designers of prostheses.

3.1 Introduction

A major goal in the design of above-knee prostheses is to minimize the metabolic energy expenditure of transfemoral amputees, as they expend significantly more energy than able-bodied humans during walking. Specifically, unilateral transfemoral amputees consume 20%-119% more oxygen per unit distance than able-bodied controls[1, 2], whereas bilateral transfemoral amputees consume 52%-280% more oxygen per unit distance than able-bodied controls[3, 4]. The hypothesis that reducing the mass of a prosthesis can reduce energy expenditure has motivated the development of lightweight prostheses[5].

To investigate this hypothesis in detail, experimental and theoretical studies have examined the effect of various inertial properties (e.g., mass and moment of inertia) of a prosthesis on metabolic energy expenditure. Experimental studies have observed that adding mass to the center of mass (COM) of the prosthesis does not significantly affect energy expenditure in transfemoral amputees[6] and transtibial amputees[7, 8], whereas adding mass distal to the COM significantly increases energy expenditure in transtibial amputees[7, 9]. In addition, matching the mass and moment of inertia of a below-knee prosthesis to able-bodied values significantly increases energy expenditure[10].

Due to the difficulty of analytically calculating metabolic energy expenditure, most theoretical studies have estimated the effect of inertial properties on its value by computing various measures of muscular effort. Nevertheless, the results have been conflicting. Beck and Czerniecki[11] used a 2-segment forward dynamics simulation and found that a lower segment with high mass, high moment of inertia, and proximal COM resulted in minimum hip work and maximum energy transfer into the trunk. Selles et al (2004b)[12] used a 2-segment inverse dynamics simulation and determined that mass perturbations distal to the COM of the lower segment increased the total angular impulse of the hip and knee, whereas mass perturbations proximal to the COM decreased it. Srinivasan[13] used a 5-segment forward dynamic simulation and supported the results of Selles et al (2004b), but found that the patterns could not be fully extended to work at the hip and knee.

Although the previous experimental and theoretical work has significantly improved understanding of the relationship between inertial properties of a prosthesis and energy expenditure, it may be challenging to apply the results to the design of prostheses. Most studies examined mass perturbations that alter both mass and moment of inertia, making it difficult to predict how changing one property independently of the other would affect energy expenditure. In addition, several studies investigated these perturbations relative to the combined COM of multiple segments of a prosthesis, making it difficult to predict how changing the inertial properties of a particular segment independently of the other segments would affect energy expenditure (e.g., when selecting an appropriate footpiece for a prosthesis). Finally, among theoretical studies, only Srinivasan investigated muscular effort during both swing and stance, and only Beck and Czerniecki examined the effects of walking speed on optimal inertial properties.

The present study aims to theoretically determine how the inertial properties of an above-knee prosthesis affect the metabolic energy expenditure required for transfemoral amputees to walk with normative kinematics. Like previous theoretical studies, the effects on metabolic energy expenditure are estimated through a computation of muscular effort, specifically total hip work. In contrast to previous studies, the effects of varying mass and moment of inertia independently are calculated, and the effects of varying the inertial properties of each segment independently are determined. In addition, the results are computed over the entire gait cycle and at multiple cadences. The relationship between total hip work and metabolic energy expenditure is considered in the discussion. It was hypothesized that 1) decreasing mass would cause a greater reduction in energy expenditure than decreasing moment of inertia, 2) most of the energetic savings would occur during swing phase, 3) decreasing the masses of distal segments would cause a greater reduction in energy expenditure than decreasing the masses of proximal segments, and 4) minimizing the masses of all segments would cause the greatest reduction in energy expenditure at all cadences.

3.2 Methods

To determine total hip work for various inertial configurations, a model of a prosthetic leg was designed and inverse dynamics was performed. Due to the lack of complete normative data sets available in the literature for walking at multiple cadences, kinetic and kinematic data from multiple sources were carefully combined and unknown quantities were approximated. All calculations were conducted in MATLAB (R2012a, The MathWorks, Natick, MA).

3.2.1 Dimensions of model

A 2-dimensional, 4-segment link-segment model was designed to model the prosthetic leg of a transfemoral amputee wearing an above-knee prosthesis (Figure 3-1). The model consisted of a trunk segment, an "upper leg" segment (stump and socket), a "lower leg" segment (shank), and a foot segment. To model the foot segment, trial center of pressure (COP) data was acquired from Winter (2009). The COP data was transformed into the reference frame of the foot to compute a "foot roll-over shape"[14], and a circular arc was fitted to the data. Lengths of the segments were prescribed according to anthropometric ratios of able-bodied humans[15, 16] scaled to the average American body height[17].

3.2.2 Inertial properties of model

The inertial properties of the model were not constant, as they were altered to represent various inertial configurations of an above-knee prosthesis. The inertial properties of able-bodied humans were used as a reference, as they represented the extreme case in which the inertial properties of one or more segments of the prosthesis were the same as those of the corresponding able-bodied segments (e.g., upper leg mass of prosthesis equal to thigh mass of able-bodied human). The masses and moments of inertia (about COM) for able-bodied humans were based on anthropometric ratios of able-bodied humans[18, 19, 16] scaled to the average American body mass[17].

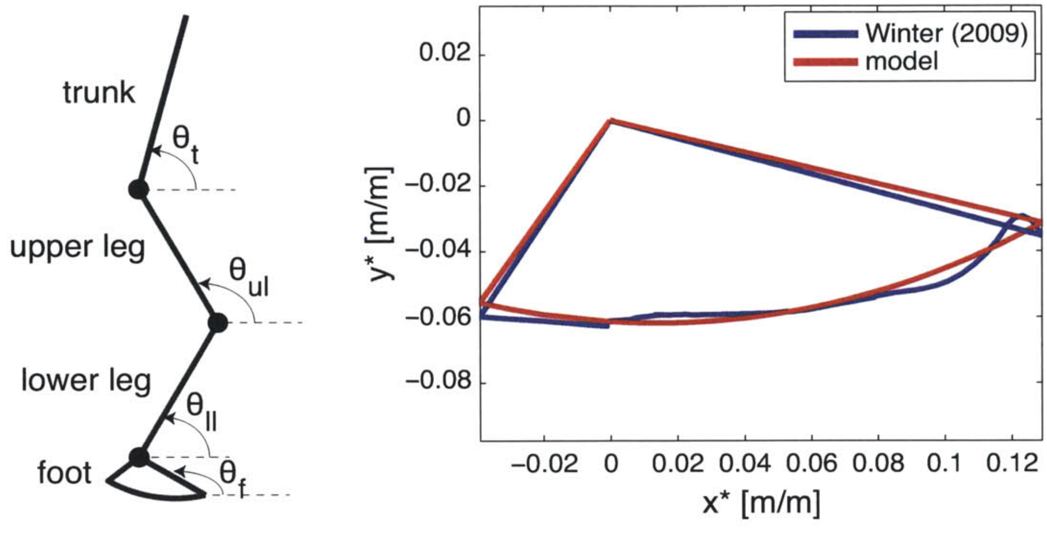

Figure 3-1: On left: A 2-dimensional model of the prosthetic leg. θ_t is the trunk angle, θ_{ul} is the upper leg angle, θ_{ll} is the lower leg angle, and θ_f is the foot angle, all with respect to the horizontal. On right: A roll-over model of the foot. The foot roll-over shape computed from Winter (2009) is compared to a circular approximation used in the model. The ankle joint is located at the origin of the graph. x* is x-coordinate normalized to body height, and y* is y-coordinate normalized to body height. $R^2 = 0.93$.

3.2.3 Kinematics of model

Since the present study aimed to determine the effect of inertial properties of a prosthesis on the energy expenditure required for an amputee to walk with normative kinematics, normative kinematics were obtained from the literature. Average normative lower-limb joint angles for the present model were acquired from Winter (1991)[20] for slow cadence (20 steps/min slower than natural cadence), natural cadence, and fast cadence (20 steps/min faster than natural cadence). However, since hip location, trunk angle, and COP were not reported in Winter (1991), these quantities were approximated at each cadence. Since the hip location only entered subsequent moment calculations through its derivatives, it was assumed to be stationary due to small forward and vertical velocities over the gait cycle[21]. Trunk angle was derived from sample data in Winter (2009)[16] for walking at a fast cadence. Thorstensson et al[22] found that the range of angular displacement of the trunk does not change with walking speed, but that the timing changes by -10% of the gait cycle from slow to fast walking. Thus, to estimate trunk angle at slow and natural cadence, the data from Winter (1991) were phase-shifted by approximately +10% and +5%, respectively. Finally, COP at each cadence was estimated from the foot roll-over shape calculated earlier. Hansen et al[14] determined that a circular arc fitted to a roll-over shape based on the knee, ankle, and foot was invariant to walking speed. Here it was assumed that the same pattern applies approximately to roll-over shape based on the foot alone. For all cadences, the COP at any time in stance was determined simply by identifying the lowest point of the foot segment at that time.

3.2.4 External forces on model

Only two external forces acted on the model: the gravitational force and the ground reaction force (GRF). Based on the mass properties of the model, the gravitational force on each segment can be easily calculated as $m_{seg}g$, where m_{seg} is the mass of the segment and g is the gravitational constant.

The GRF also depends on the mass properties of the model. From the dynamic equations for multi-rigid-body systems, the net external force on a system is equal to the sum of the mass-acceleration products of all the bodies. Applying this principle to a link-segment representation of the human body, we can write

$$\overrightarrow{GRF} = \sum_{i=1}^{N} m_i(\ddot{\vec{r}}_{COM} + g\hat{y})$$

where $\overrightarrow{GRF}$ is the ground reaction force, N is the number of segments representing the body, m_i is the mass of the i^{th} segment, $\vec{r}_{COM}$ is the position vector of the COM of the i^{th} segment relative to the origin of an inertial reference frame, g is the gravitational constant, and y is the unit vector in the positive vertical direction[16].

The difference between the GRF of an able-bodied human ($\overrightarrow{GRF}_{able}$) and the GRF of a unilateral above-knee amputee with a lightweight prosthetic leg ($\overrightarrow{GRF}_{amp}$) is simply the difference in the sum of the mass-acceleration products of their segments. Thus, we can write

$$\begin{aligned} \overrightarrow{GRF}_{amp} = & \overrightarrow{GRF}_{able} - \\ & \begin{pmatrix} (m_{ul} - m_{ul_p})(\ddot{\vec{r}}_{ul_{COM}} + g\hat{y}) & + \\ (m_{ll} - m_{ll_p})(\ddot{\vec{r}}_{ll_{COM}} + g\hat{y}) & + \\ (m_f - m_{f_p})(\ddot{\vec{r}}_{f_{COM}} + g\hat{y}) & \end{pmatrix} \end{aligned}$$

where m_{ul}, m_{ll}, and m_f are the masses of the upper leg, lower leg, and foot in an able-bodied human, m_{ul_p}, m_{ll_p}, m_{f_p} are the masses of the same segments in a prosthetic leg, $\vec{r}_{ul_{COM}}$, $\vec{r}_{ll_{COM}}$, and $\vec{r}_{f_{COM}}$ are the position vectors of the COMs of the upper leg, lower leg, and foot relative to the origin of an inertial reference frame, g is the gravitational constant, and y is the unit vector in the positive vertical direction. $\overrightarrow{GRF}_{able}$ for each cadence was acquired from Winter (1991)[20].

The GRF above is the total GRF acting on the body. To perform inverse dynamics on the model, it is necessary to calculate the GRF acting on the prosthetic leg alone. During single-support of the prosthetic leg, the total GRF is equal to the GRF on the prosthetic leg. However, during double-support, the total GRF is indeterminately distributed between the prosthetic leg and the unaffected leg. As an approximation, the GRF for the prosthetic leg was assumed to be 0% of the total GRF at the beginning of double-support and 100% of the total GRF at the end of double-support, with a linear interpolation in between. Assuming an average natural cadence of 105 steps/min[23], the length of double-support was estimated to be 15% of the gait cycle for slow cadence, 12.5% for natural cadence, and 10% for fast cadence based on trends calculated by Grieve and Gear[24] as reported in Inman et al[21].

3.2.5 Inverse dynamics

Using the previously described dimensions, inertial properties, kinematics, and external forces, a standard 2-dimensional inverse dynamics procedure[25] was applied to the model to compute joint moments. To evaluate the performance of the model, inertial properties were initially set to those of an able-bodied human, and resulting moments were compared to the average normative moments reported by Winter (1991)[20] for walking at natural cadence (Figure 3-2). In accordance with common practice, joint moments were normalized to body mass to reduce the effects of body height and weight on subsequent results[26]. Despite approximating the lengths of the segments, masses of the segments, hip location, trunk angle, COP location, and GRF, the model performed well. The moments predicted by the model have an R^2 value between 0.80 for the hip and 0.93 for the ankle relative to the moments reported by Winter (1991). To minimize the effects of any estimation error, all results in the present study are reported as percentage differences from a specified baseline, rather than as absolute magnitudes.

3.2.6 Testing of inertial properties

Masses and moments of inertia of the upper leg, lower leg, and foot were varied between 25% and 100% of their magnitudes in able-bodied humans, with 25 data points evenly distributed between the bounds for each segment. Hence, a total of 25^6 inertial configurations were tested.

For each configuration, GRF was adjusted to the mass properties of the leg, and inverse dynamics was performed to compute hip moment. Hip power (P_{hip}) was then calculated by multiplying the hip moment (M_{hip}) by the hip angular velocity (ω_{hip}). Total joint work, defined as the integral of the absolute value of joint power with respect to time, is an established measure of muscular effort[27]. In this study, total hip work was computed as

$$W_{hip}^{tot} = \int_{t_{HS}}^{t_{TO}} |P_{hip}| dt \tag{3.1}$$

where t_{HS} is the time at heel strike (0% gait cycle) and t_{TO} is the time at toe-off (100% gait cycle).

The calculations were performed at each of the three cadences examined. Opti-

	Altered variable(s)		
% of normal	m	I	m and I
25%	0.81	0.96	0.78
50%	0.87	0.97	0.84
75%	0.93	0.99	0.92
100%	1.00	1.00	1.00

Table 3.1: Effects of altering mass and/or moment of inertia on total hip work. Alterations in inertial parameters apply to all segments of the leg and are given with respect to able-bodied values (e.g., "25%" designates that the specified inertial parameters for the upper leg, lower leg, and foot are scaled to 25% of their corresponding able-bodied values). Total hip work is normalized to the total hip work of a prosthetic leg with able-bodied inertial properties.

mal inertial configurations were determined by identifying the configurations that minimized total hip work.

3.3 Results

3.3.1 Gross effects of inertial properties on energetics

Figure 3-3 shows the gross effects of altering inertial properties on hip power over the gait cycle. Decreasing both masses and moments of inertia of all segments of the leg relative to able-bodied values had a large effect on hip power, reducing peak hip power during stance by up to 26% and average hip power during swing by up to 74%. Decreasing masses and holding moments of inertia constant had a similar effect, reducing peak hip power during stance by up to 20% and average hip power during swing by up to 66%. However, decreasing moments of inertia and holding masses constant had a small effect on hip power, reducing peak hip power during stance by no more than 6% and average hip power during swing by no more than 8%. These results suggest that the masses of the segments are the principal inertial determinants of energetic savings in a prosthetic leg.

Table 3.3.1 further quantifies the previous results in terms of total hip work over the gait cycle. Decreasing both mass and moment of inertia of all leg segments reduced hip work by up to 22%. Only decreasing mass reduced hip work by nearly as much, 19%. On the other hand, only decreasing moment of inertia reduced hip work by no more than 4%. These results support the suggestion that the masses of the segments are the principal inertial determinant of energetic savings in a prosthetic leg. Subsequent analyses focus on the effects of alterations in mass, with the moment of inertia of each segment simply equal to its corresponding able-bodied value scaled by the mass of the segment.

3.3.2 Energetic savings during stance and swing

Figure 3-3 shows that the majority of energetic savings resulting from decreasing the masses of the leg segments relative to able-bodied values occurred during swing rather than stance. Moreover, although decreasing the masses appeared to significantly reduce the magnitude of power during late stance, this reduction was closely balanced by increased power in early- to mid-stance. In fact, when reducing the masses of all segments of the leg to 25%, 50%, or 75% of able-bodied values, approximately 98% of the total reduction in hip work occurred during swing and just 2% occurred during stance.

3.3.3 Effects of lower leg and foot mass on energetics

Since the stump of an above-knee amputee occupies a much larger volume than the prosthetic socket and typically has a higher mass density[16, 28], designers of prostheses have little control over mass reduction of the upper leg. Thus, it is practical to specify an upper leg mass and then investigate how alterations of lower leg and foot mass influence total hip work. Figure 3-4 shows how lower leg mass and foot mass affected hip work for four different upper leg masses. For all four upper leg masses, as lower leg mass and foot mass were decreased, hip work was reduced. When lower leg and foot mass were equal to 25% of their able-bodied values, hip work was reduced by 20-22%. Furthermore, the slope of the contour lines had a minimum magnitude of approximately 1/3, indicating that decreasing foot mass had up to a three times greater effect on reducing hip work than decreasing lower leg mass. Few notable differences exist between the results for the four upper leg masses, except that decreasing foot mass and lower leg mass was able to achieve up to a 2% greater reduction in hip work with higher upper leg masses.

3.3.4 Optimal inertial configuration at each cadence

The optimal lower leg and foot masses that minimized work total hip work were identified at the slow, natural, and fast cadences (defined earlier) for a prosthetic leg with upper leg mass equal to 50% of its corresponding able-bodied value. For all cadences, the optimal lower leg and foot masses were equal to the minimum values investigated (i.e., 25% of able-bodied values). At these masses, hip work was reduced by 13% for slow cadence, 19% for natural cadence, and 13% for fast cadence relative to the work for a prosthetic leg with able-bodied inertial properties.

3.4 Discussion

3.4.1 Analysis of major findings

The results supported all four hypotheses of the study. First, the results agreed with the hypothesis that decreasing mass would cause a greater reduction in energy expenditure than decreasing moment of inertia. Decreasing the masses of all the segments caused a large (up to 22%) reduction in total hip work, whereas decreasing the moments of inertia did not. Furthermore, the effect of decreasing mass on total hip work was nearly equivalent to the effect of decreasing both mass and moment of inertia. The results suggest that designers of prostheses can focus on decreasing the mass of each segment of the prosthesis rather than altering its mass distribution in order to reduce energy expenditure.

The results regarding total hip work align with experimental findings that adding mass to the shank of an above-knee or below-knee prosthesis increased muscular effort at the hip[29, 30, 11]. The results also support the theoretical findings of Beck and Czerniecki[11] that mass of the shank generally has a much larger effect on total hip work than moment of inertia. However, the assumption of the present study that an effect of mass on total hip work would indicate a corresponding effect on metabolic energy expenditure is contradicted by the experimental work of Lehmann et al[7], which found that the metabolic efficiency of transtibial amputees did not significantly change after mass perturbations at the COM of the prosthetic leg. However, Lehmann et al increased mass from 42% to 70%, and the mass was added to the lower leg. From Figure 3-4, our results predict just a 2-3% increase in total hip work over the same range.

The outcomes of the study also agreed with the second hypothesis, which stated that the majority of energetic savings would occur in swing rather than stance. Although decreasing mass reduced peak hip power in late stance by up to 20%, this reduction was balanced by an increase in the magnitude of power during early- to mid-stance. As a result, just a fraction (2%) of the reduction in total hip work occurred during stance. The small energy savings in stance relative to swing can be explained by observing that in normal walking, the changes in kinetic energy of the segments of the leg are relatively constant during stance, but highly variable during swing[31].

The present study also supported the hypothesis that reducing the masses of distal segments would cause a greater reduction in energy expenditure than reducing the masses of proximal segments. In particular, a decrease in foot mass was observed to cause up to a three times greater reduction in total hip work than an equivalent decrease in lower leg mass. This outcome agrees with the general physical principle that the work required to move a mass about an axis of rotation (in this case, the hip) increases as its distance from the axis of rotation increases. Furthermore, the outcome agrees with specific theoretical findings that increasing mass at or near the foot leads to an increase in muscular effort in amputees[11, 13] and experimental findings that doing so leads to an increase in metabolic energy expenditure in both amputees [9] and able-bodied humans[32, 33, 34]. The results suggest that designers of prostheses can focus on decreasing the mass of the foot in order to reduce energy expenditure. Furthermore, Figure 3-4 allows designers to quantify how much changing foot mass would affect energy expenditure for specific masses of the upper leg and lower leg.

Finally, the results of the study supported the fourth hypothesis, which stated that minimizing the masses of all the segments would cause the greatest decrease in energy expenditure at all cadences. When the masses of the upper leg, lower leg, and foot were all equal to 25% of able-bodied values, minimal total hip work was attained at all cadences, and the work reduction ranged from 13-20%. The results agree with the study of Beck and Czerniecki[11], which showed that minimal shank-foot mass corresponded to minimal hip work across a range of walking speeds.

3.4.2 Relationship with energy expenditure

The present study computed the effects of the masses of the leg segments on total hip work, which is a measure of muscular effort at the hip. The results can be used to decrease the muscular requirements of transfemoral amputees, who suffer between 40-60% atrophy of the hip muscles by volume, depending on the level of amputation[35].

An aim of the study was to apply the results to reduce not only muscular effort of transfemoral amputees, but also their metabolic energy expenditure. Since the hip musculature is the primary actuator of an above-knee prosthesis, it is intuitive that a correlation would exist between total hip work and metabolic energy expenditure; nevertheless, the existence of a correlation has been debated. Foerster et al[36] observed a lack of correlation between work and metabolic energy expenditure in transfemoral amputees, and Czernecki et al[6] found that adding mass to the COM of the shank of an above-knee prosthesis did not have a significant effect on energy expenditure. Gitter et al[30] explained the results of Czerniecki et al by proposing that a balance between positive joint work in late stance and negative hip force work in swing caused a negligible effect on energy expenditure. However, Foerster et al measured the sum of work at the hip, knee, and leg, which may not accurately represent energy expenditure in a transfemoral amputee due to the presence of mechanical components at the knee and ankle of a prosthesis. Gitter et al did not consider the effect of hip joint work during swing or energy transfer through the muscles, which may upset the observed balance. Furthermore, a link between leg mass and energy expenditure has been consistently observed in able-bodied humans[32, 33, 34]. More research is necessary to rigorously demonstrate a lack of correlation between total hip work and metabolic energy expenditure in transfemoral amputees.

3.4.3 Limitations of study

A major limitation of the study was the estimation of kinematic and kinetic parameters used as input for the inverse dynamics simulation. Specifically, hip location, trunk angle, and COP location were all estimated at multiple cadences due to the lack of complete data sets available in the literature, and GRF during double-support was estimated based on a linear transition of loading between the legs due to the indeterminate distribution of total GRF. Although the moments calculated by the model for able-bodied inertial properties were within range of average normative values, it is possible that inaccuracies in estimation could have weighted hip power and total hip work unrealistically at certain points in the gait cycle at one or more cadences. The accuracy of the model could be improved through extensive data collection and more accurate approximations of GRF during double support[37].

A second limitation of the study was an assumption implicit in the inverse dynamics calculations that the ankle moment of the prosthetic leg could reproduce normative kinematics of the foot. The energy produced at the ankle over the gait cycle was calculated to be positive for all inertial configurations of the prosthetic leg. Thus, the prosthetic leg could only be expected to reproduce normative kinematics if the ankle unit itself contained a power source. Powered ankle prostheses have been designed and tested in laboratory settings[38], but commercial units have only recently become available. Forward dynamic simulations may be required to model the kinematics and energetics of the leg when sufficient ankle power cannot be provided.

3.5 Conclusion

The goal of the present study was to investigate the effects of inertial alterations of an above-knee prosthesis on energy expenditure. The masses of the segments were found to be the principal inertial determinants of energetic savings, and foot mass was determined to have a significantly greater effect on energy expenditure than lower leg mass. Minimizing the masses of all segments of the prosthesis minimized energy expenditure at all cadences. In contrast to most previous work, the study conducted an inverse dynamics simulation throughout the entire gait cycle, examined the effects of mass and moment of inertia independently, examined inertial alterations of each segment independently, identified optimal inertial configurations at multiple cadences, and reported major results in the form of parametric illustrations that can be used by designers of prostheses. Future work should focus on further investigating the relationship between total hip work and metabolic energy expenditure of amputees wearing above-knee prostheses, as well as improving the accuracy of kinematic and kinetic approximations used in the model.

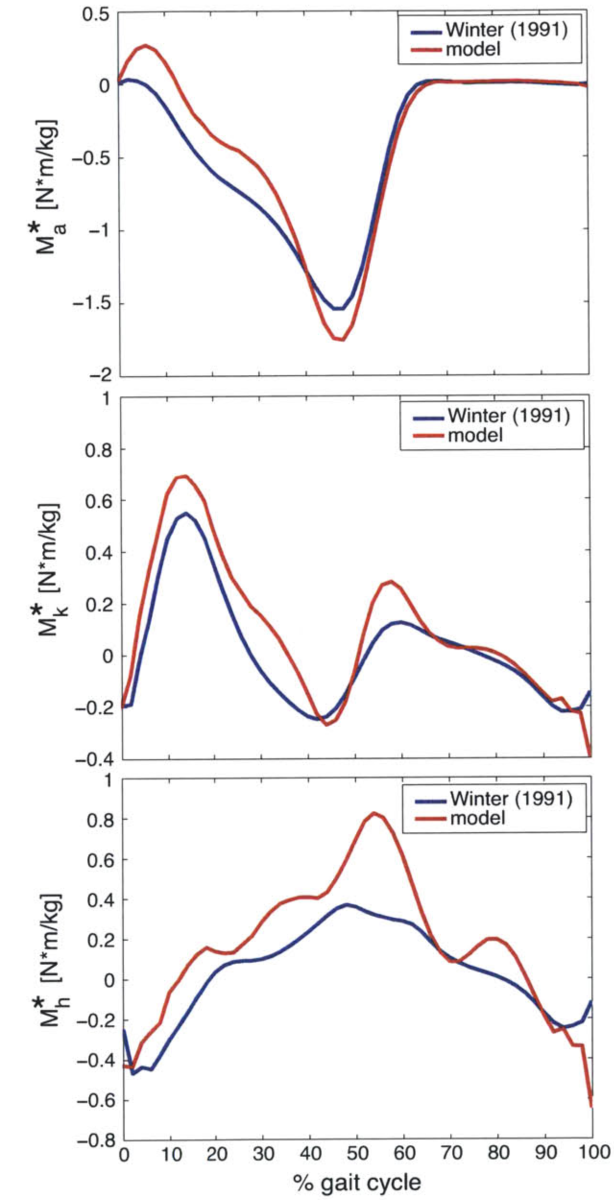

Figure 3-2: Comparison between moments calculated by model and moments reported by Winter (1991)[20] for walking at a natural cadence. M_a^* is ankle moment normalized to body mass, M_k^* is knee moment normalized to body mass, and M_h^* is hip moment normalized to body mass. From top to bottom, $R^2 = 0.93$, 0.91, and 0.80.

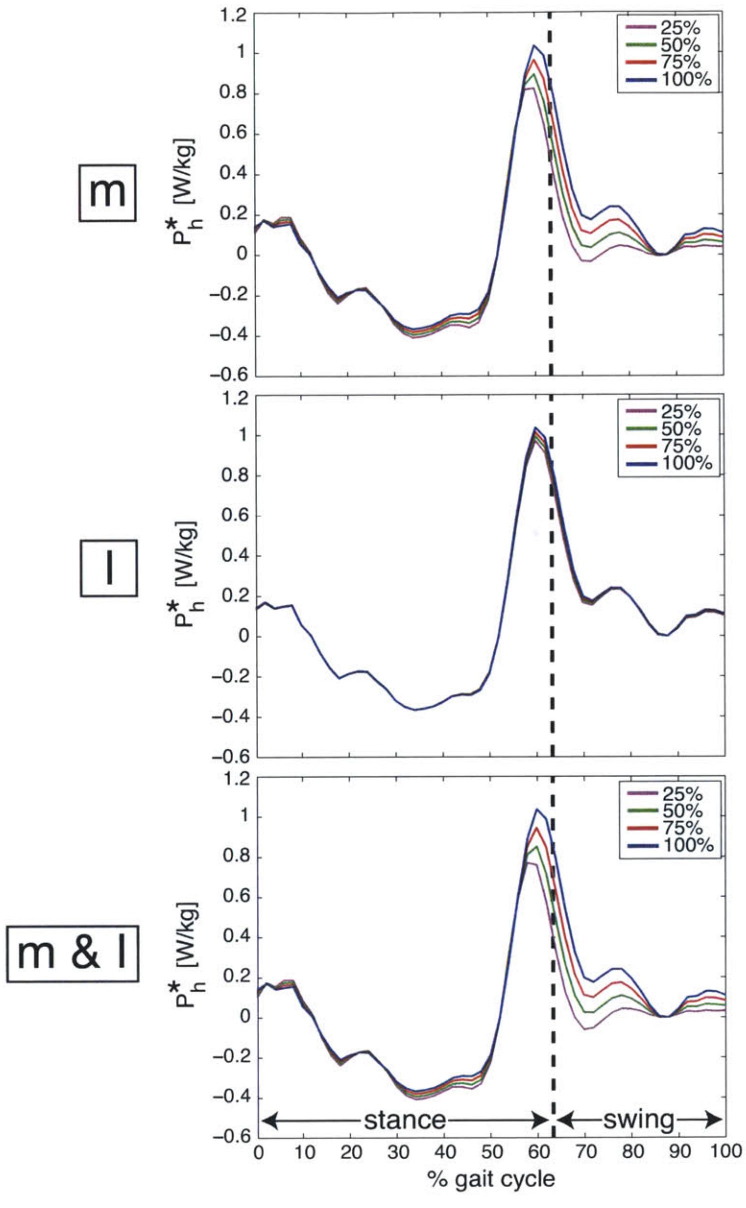

Figure 3-3: Gross effects of altering inertial properties on hip power. P_h^* is hip power normalized to body mass. The values of the inertial properties are given as percentages of able-bodied values. Top: masses of all leg segments were altered to the specified values and moments of inertia were held constant at 100%. Middle: moments of inertia of all leg segments were altered to the specified values and masses were held constant at 100%. Bottom: both masses and moments of inertia of all leg segments were altered to the specified values.

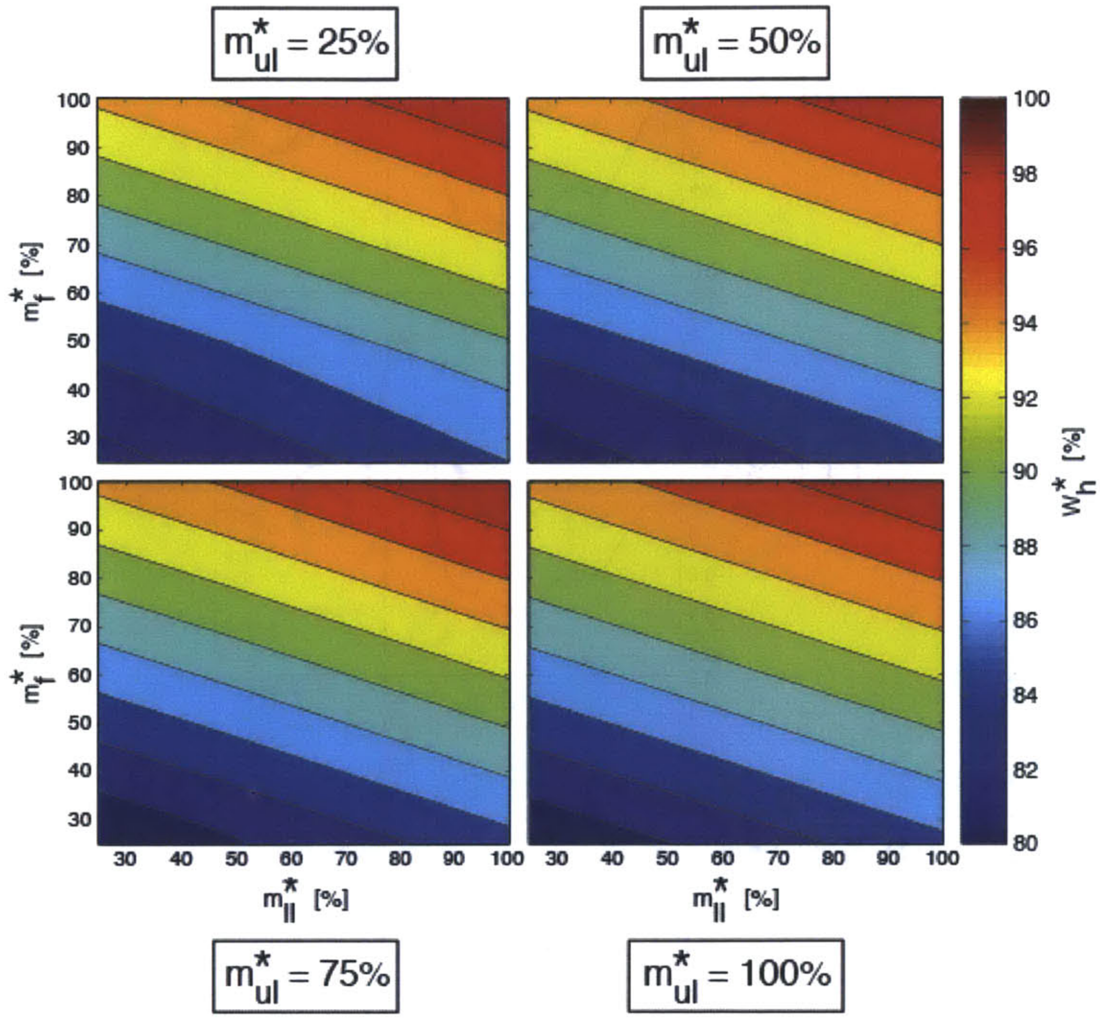

Figure 3-4: Effects of lower leg mass and foot mass on hip work for various values of upper leg mass. m^*_{ul} is upper leg mass normalized to able-bodied upper leg mass, m^*_{ll} is lower leg mass normalized to able-bodied lower leg mass, and m^*_f is foot mass normalized to able-bodied foot mass. W^*_h is total hip work normalized to the total hip work of a prosthetic leg with able-bodied inertial properties. Graphs are presented for m^*_{ul} values of 25%, 50%, 75%, and 100%.

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Chapter 4

The Effects of the Inertial Properties of Above-Knee Prostheses on Optimal Stiffness, Damping, and Engagement Parameters of Passive Prosthetic Knees

Abstract

A major goal of prosthetic knees is to allow amputees to walk with normative gait kinematics. The present study determined optimal stiffness, damping, and engagement parameters for a low-cost, passive prosthetic knee that accomplishes this goal. 2-dimensional inverse dynamics was used to compute knee moments required for an amputee with reduced inertial properties to walk with normative kinematics. Knee power was analyzed to divide gait into phases and select mechanical components for each phase. Sensitivity analysis was conducted to simplify the mathematical order of components. The coefficients of the components were optimized to reproduce moments required for normative kinematics, and the effects of prosthetic mass and walking cadence on optimal coefficients were identified. Reduction in prosthetic mass caused up to a 60% decrease in the moments required for normative kinematics. A simple mechanical model consisting of a first-order spring, two zero-order dampers, and three clutches accurately reproduced the required moments. Alterations in upper leg, lower leg, and foot mass had a large influence on optimal coefficients, increasing damping coefficients by up to 180%, and the effects were reported in the form of parametric illustrations that can be used by designers to select components. Walking cadence also influenced optimal coefficients, affecting values by up to 100%.

4.1 Introduction

A major goal in the design of prosthetic knees is to enable above-knee amputees to walk with the knee kinematics of able-bodied humans. When an able-bodied human walks, the muscles, tendons, and ligaments adjacent to the knee produce moments that flex and extend the knee in a metabolically efficient manner[1]. However, in an above-knee amputee, musculotendon function is impaired. Designers of prosthetic knees have attempted to provide the musculotendon function required for normal walking with the action of mechanical and/or electrical components[2, 3]. Passive prosthetic knees (i.e., knees without net energy sources), such as those for the developing world[4, 5], typically contain mechanical components alone and do not have electronic control systems to correct gait deviations. Thus, in designing passive knees, it is critical to carefully quantify the musculotendon function required for normal walking and select components to closely reproduce it.

Researchers have used multiple methods to quantify the musculotendon function required for normal walking. One established method is calculating the derivative of the moment-angle (DMA) graph of the able-bodied knee over the gait cycle to compute a measure of "joint stiffness"[6]. Frigo et al[7] selected three regions of the moment-angle graph over the gait cycle that were approximately linear and calculated the slope of a best-fit line in each region. Similarly, Shamaei and Dollar[8] and Shamaei et al[9] selected two regions of the moment-angle graph during the weight acceptance phase that were approximately linear and calculated the slopes of best-fit lines. Collectively, the studies demonstrated that the able-bodied knee has a relatively constant DMA during parts of the gait cycle and that the phenomenon is valid for multiple walking speeds and load carriage conditions. In the context of prosthesis design, the results suggest that a prosthetic knee could partially replicate the musculotendon function of an able-bodied knee during walking with torsional springs.

Another established method to quantify the musculotendon function required for normal walking is to design a mechanical model and optimize the coefficients of the components (referred to here as Mechanical Model Coefficients, or MMC) to allow the model to best reproduce the moment-time relationship of the able-bodied knee over the gait cycle. Sup et al[2] divided the gait cycle into four regions; modeled the knee as a first-order spring, third-order spring, and first-order damper in parallel; and optimized coefficients to allow the model to closely reproduce the moment-time relationship of the able-bodied knee during each of the regions. Similarly, Martinez-Villalpando and Herr[3] divided the gait into stance and swing, modeled the knee as two springs with clutches during stance and an actively driven damper during swing, and optimized coefficients to allow the model to closely reproduce the moment-time relationship of the able-bodied knee. In the context of design, each of the studies determined a configuration of components that allowed a prosthetic knee to accurately replicate the musculotendon function of an able-bodied knee during walking.

Although the results of the previous studies quantified musculotendon function required for normal walking and indicated components that could be used in a prosthetic knee, the studies have two major limitations with respect to prosthesis design. First, the studies quantified musculotendon function based on a leg with able-bodied inertial properties moving with normal kinematics. However, since the inertial properties of a prosthetic leg are typically lower[10], the knee moment required to produce normal kinematics is different. Second, the studies determining MMC did not report the sensitivity of the moment-time relationship of the model to the complexity of the components. It is possible that a simple configuration of zero- or first-order components can accurately reproduce the moment-time relationship for walking, potentially reducing the cost of the prosthesis.

The goal of the present study was to determine stiffness, damping, and engagement parameters for a low-cost, passive prosthetic knee by designing a mechanical model of the knee and determining MMC. In contrast to previous studies, MMC were optimized to allow the model to accurately reproduce the moment-time relationship of a *prosthetic* leg moving with normative kinematics. To help realize the goal of designing a low-cost knee, the sensitivity of the accuracy of the model to the complexity of components was investigated. In addition, the effects of inertial properties of the prosthetic leg and walking cadence on MMC were determined in detail. It was hypothesized that a simple mechanical model of the knee consisting of zero-order and first-order springs and dampers could accurately reproduce the moment-time relationship required for normative kinematics, and that MMC would be highly sensitive to changes in the mass of the prosthetic leg and walking cadence.

4.2 Methods

4.2.1 Gross effects of inertial properties on required knee moment

As described earlier, since the inertial properties of a prosthetic leg are typically lower than an able-bodied leg, the knee moment required to produce normative kinematics (M_{req}) is different. To determine M_{req} for various inertial configurations of the prosthetic leg, a model of the leg was designed and inverse dynamics was performed according to the methods of Narang and Winter[11] (i.e., Chapter 3). All calculations were performed in MATLAB (R2012a, The MathWorks, Natick, MA).

In summary, a 2-dimensional, 4-segment link-segment model[12] with a normative foot roll-over shape[13] was designed. The dimensions of the model were prescribed according to anthropometric ratios of able-bodied humans[14, 12] scaled to the average American body height[15]. The inertial properties of the model were prescribed according to the specific inertial configuration being examined. Normative kinematics for walking at a slow cadence (20 steps/min slower than natural cadence), natural cadence, and fast cadence (20 steps/min faster than natural cadence) were obtained and estimated from various sources in the literature[16, 12, 17, 18]. Normative ground reaction force data were obtained from Winter (1991)[16] and adjusted to the mass of the prosthetic leg. A standard 2-dimensional inverse dynamics procedure[19] was then performed, and M_{req} was calculated for various inertial configurations of the prosthetic leg. In accordance with common practice, joint moments were normalized to body mass to reduce the effects of body height and weight on later results[20].

Figure 4-1 shows the gross effects of inertial properties of the prosthetic leg on M_{req}. Decreasing the masses of all segments of the leg relative to able-bodied values had a large effect on M_{req}, increasing the peak magnitude during late stance by up to 43% and decreasing the peak magnitude during swing by up to 60%. Thus, M_{req} changes significantly with the inertial properties of the prosthetic leg, indicating that MMC should be optimized to reproduce M_{req} of the prosthetic leg rather than the knee moment of an able-bodied leg.

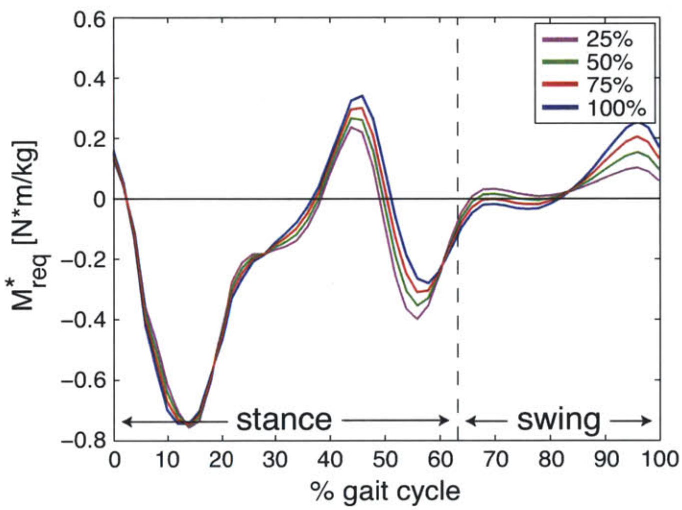

Figure 4-1: Gross effects of altering inertial properties on the knee moment required for a prosthetic leg to move with normative kinematics at a natural cadence. M^*_{req} is required knee moment normalized to body mass. The masses of all segments of the leg were scaled to the specified percentages of able-bodied values, and the moments of inertia (about the centers of mass of the segments) were scaled by the masses.

4.2.2 Validation and design of passive mechanical model

A mechanical model was designed to model the knee over the gait cycle. Because the aim of the present study was to determine stiffness, damping, and engagement parameters for a passive knee, a mechanical model consisting exclusively of passive elements was constructed. Prior to selecting components for the model, it was critical to determine whether a passive model could theoretically reproduce normative gait kinematics. To do so, knee power was computed as the product of knee moment with knee angular velocity, and net knee work was calculated as the integral of knee power over the gait cycle. For all inertial configurations of the prosthetic leg and walking cadences, net knee work was negative, indicating that kinetic energy was dissipated by the knee over the gait cycle. Thus, a passive mechanical model could theoretically reproduce normative gait kinematics, as no net energy source is required during gait.

Figure 4-2 shows a schematic of the general passive mechanical model used to model the knee. The components for the passive mechanical model were springs, to store and release energy; dampers, to dissipate energy; and clutches, to engage and disengage the springs and dampers.

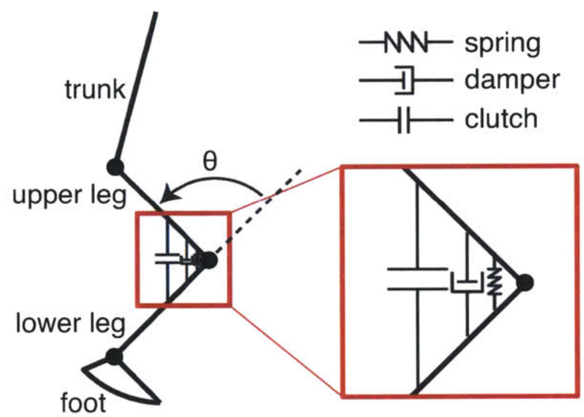

Figure 4-2: Schematic of general passive mechanical model used to model the knee. θ is the knee angle.

4.2.3 Identification of phases based on knee power

Inspired by the work of Gates[21] for the ankle, the power versus time graph for the knee was analyzed to determine regions of gait for which specific passive mechanical elements could be selected to model the knee. To simplify the model and subsequent optimization, only one element was selected for each region. For all inertial configurations of the prosthetic leg and walking cadences, the knee power versus time graph was observed to consist of three major phases–one in which the ratio of dissipated energy (E_{diss}) to generated energy (E_{gen}) was close to 1, and two in which energy was purely dissipated. Figure 4-3 shows the three phases for a prosthetic leg with a typical inertial configuration[10, 22, 23]. During phase 1, $\frac{E_{diss}}{E_{gen}} = 0.77$. During phases 2 and 3, $E_{gen} = 0$. Thus, a spring element with a clutch was selected to model the knee during phase 1, as a spring can mimic the dissipation and generation of energy through storage and release. A damper element with a clutch was selected to model the knee during both phase 2 and phase 3, as a damper purely dissipates energy.

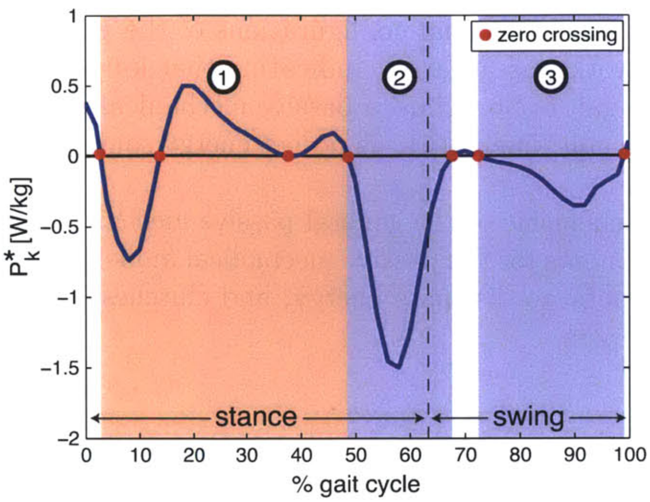

Figure 4-3: The three phases of gait. Power versus time graph is shown for a prosthetic leg with a typical inertial configuration (upper leg mass = 50% of able-bodied value, and lower leg and foot mass = 33% of able-bodied values) moving with normative kinematics at a natural cadence. P_k^* is power normalized by body mass. Phase 1 is an energy dissipation and generation phase ($\frac{E_{diss}}{E_{gen}} = 0.77$), and phases 2 and 3 are energy dissipation phases.

4.2.4 Mathematical representation of components

Before optimizing MMC to reproduce M_{req} over the gait cycle, the passive mechanical components in the model were first described mathematically. The moment produced by a general, second-order spring element with a clutch (M_{spr}) was written as

$$M_{spr} = \begin{cases} -sgn(\theta - \theta_{eq})k_0 - k_1(\theta - \theta_{eq}) - sgn(\theta - \theta_{eq})k_2(\theta - \theta_{eq})^2 & t(\theta_{dis}) \leq t(\theta) \leq t(\theta_{eng}) \\ 0 & t(\theta) < t(\theta_{eng}) \text{ OR } t(\theta) > t(\theta_{dis}) \end{cases}$$

where sgn was the signum function, k_0, k_1, and k_2 were the spring coefficients, θ_{eq} was the equilibrium angle of the spring, θ_{eng} and θ_{dis} were the engagement and disengagement angles of the clutch, and the function $t(\theta)$ described the time at which the knee angle was a given value. The spring coefficients were defined as non-negative, and the signum function was used to ensure that the zero-order and second-order terms produced a moment opposed to the angular displacement of the spring. Time (rather than angle) was used to describe engagement of a component because each point in time was associated with a unique angle, whereas each angle could be associated with more than one point in time. Physically, k_0 represented spring preload, k_1 represented linear spring stiffness, and k_2 represented nonlinear (quadratic) spring stiffness. To simplify later calculations, the arbitrary parameter θ_{eq} was assigned to be equal to θ_{eng}.

Analogously, the moment produced by a general, second-order damper element with a clutch (M_{dmp}) was written as

$$M_{dmp} = \begin{cases} -sgn(\dot{\theta})b_0 - b_1\dot{\theta} - sgn(\dot{\theta})b_2\dot{\theta}^2 & t(\theta_{dis}) \leq t(\theta) \leq t(\theta_{eng}) \\ 0 & t(\theta) < t(\theta_{eng}) \text{ OR } t(\theta) > t(\theta_{dis}) \end{cases}$$

where sgn was the signum function, b_0, b_1, and b_2 were the damper coefficients, and the remaining parameters were the same as before. The damper coefficients were defined as non-negative, and the signum function was used to ensure that the zero-order and second-order terms produced a moment opposed to the angular velocity of the knee. Physically, b_0 represented constant friction damping, k_1 represented linear viscous damping, and k_2 represented nonlinear (quadratic) damping.

4.2.5 Optimization of coefficients

As described earlier, a spring element with a clutch was selected to model M_{req} during phase 1, and a damper element with a clutch was selected to model M_{req} during both phase 2 and phase 3. MMC were optimized in each phase to minimize a cost function (C) defined as the least-squares error between the moment produced by the mechanical model (M_{mod}) and M_{req} over time. Mathematically, C was written as

$$\begin{aligned} C &= \sum_{i=1}^{N}(M_{req_i} - M_{mod_i})^2 \\ &= \sum_{i=1}^{N}(M_{req_i} - (M_{1_i} + M_{2_i} + M_{3_i}))^2 \end{aligned}$$

where N was the number of data points in M_{req} (determined by the kinematic and kinetic data used to compute it), M_1 was the moment produced by the spring in phase 1, M_2 was the moment produced by the damper in phase 2, and M_3 was the moment produced by the damper in phase 3. $i = 1$ corresponds to heel strike, and $i = N$ corresponds to toe-off.

Optimization of MMC was performed with the *ga* (genetic algorithm) tool in MATLAB. The *ga* tool was selected for its ability to identify global minima for non-smooth cost functions[24], such as those describing elements engaged and disengaged by clutches. Table 4.1 shows the lower and upper bounds prescribed for each optimized coefficient in each phase. The lower bounds for the stiffness and damping coefficients were all set to 0 to prevent the optimization from selecting negative coefficients. Although it cannot be proven that globally optimal values of the stiffness and damping coefficients did not exist outside the bounds, the optimal values determined by the genetic algorithm never approached any of the upper bounds, suggesting that the bounds indeed enclosed the global optima.

Finally, to ensure that the model was physically realistic, the following additional constraints were applied to the optimization:

1. For the clutch in each phase, $t(\theta_{dis}) > t(\theta_{eng})$ (The clutch was required to disengage *after* it engaged.)

		Coefficient				
Phase 1 (Spring)		$k_0[\frac{N*m}{kg}]$	$k_1[\frac{N*m}{kg*rad}]$	$k_2[\frac{N*m}{kg*rad^2}]$	$t(\theta_{eng})[s]$	$t(\theta_{dis})[s]$
	lower bound	0	0	0	t_{i1}	t_{f1}
	upper bound	5	30	200	t_{i1}	t_{f1}
Phase 2 (Damper)		$b_0[\frac{N*m}{kg}]$	$b_1[\frac{N*m*s}{kg*rad}]$	$b_2[\frac{N*m*s^2}{kg*rad}]$	$t(\theta_{eng})[s]$	$t(\theta_{dis})[s]$
	lower bound	0	0	0	t_{i2}	t_{f2}
	upper bound	3	0.7	0.2	t_{i2}	t_{f2}
Phase 3 (Damper)		$b_0[\frac{N*m}{kg}]$	$b_1[\frac{N*m*s}{kg*rad}]$	$b_2[\frac{N*m*s^2}{kg*rad}]$	$t(\theta_{eng})[s]$	$t(\theta_{dis})[s]$
	lower bound	0	0	0	t_{i3}	t_{f3}
	upper bound	0.7	0.1	0.02	t_{i3}	t_{f3}

Table 4.1: Lower and upper bounds for all the coefficients optimized in each phase. All coefficients are normalized by body mass. t_{i1} and t_{f1} are the initial time and final time in phase 1, t_{i2} and t_{f2} are the initial time and final time in phase 2, and t_{i3} and t_{f3} are the initial time and final time in phase 3.

2. For the clutch of the spring, $\theta_{eng} = \theta_{dis}$ (All energy stored in the spring was released by the end of the phase.)

4.2.6 Sensitivity of cost to spring and damper complexity

As described earlier, each spring and damper in the mechanical model was mathematically represented as the sum of a zero-order term, first-order term, and second-order term. However, the mathematical representations of the springs and dampers were general, and it is possible that not all terms for each component are necessary for the model to accurately reproduce M_{req} within each phase. The 'complexity' of a spring and damper is a relative term used here to describe the mathematical representation of the component, where representations with fewer terms and lower orders are 'simple,' and representations with more terms and higher orders are 'complex.' Complexity also generally coincides with difficulty of physical implementation (e.g., a zero-order damper, which is a friction damper, is easier to implement in a design than a first-order damper, which is a viscous damper). Since one of the goals of the present study was to determine a mechanical model that could reproduce M_{req} in a low-cost prosthetic knee, simple components were sought.

A sensitivity analysis was performed in which all possible mathematical representations of each spring and damper were evaluated. For each representation, the coefficients were optimized to allow the component to best reproduce M_{req} for a prosthetic leg with a typical inertial configuration (mass of upper leg = 50% of able-bodied value, and masses of lower leg and foot = 33% of able-bodied values) walking at a natural cadence, and C was computed. Values of C were then compared to determine the simplest possible representation of each spring and damper that allowed it to accurately reproduce M_{req} during its corresponding phase.

	Configuration						
	Z	F	S	ZF	FS	ZS	ZFS
Phase 1 (Spring)	0.21	0.08	0.09	0.08	0.07	0.07	0.07
Phase 2 (Damper)	0.09	0.11	0.18	0.09	0.11	0.09	0.09
Phase 3 (Damper)	0.13	0.37	0.46	0.13	0.37	0.13	0.13

Table 4.2: Sensitivity of optimization cost to the complexity of components. Cost is reported as C^*, which is equal to C normalized to the maximum theoretical cost during the corresponding phase (i.e., the cost during that phase when no components are used to model the knee). For a given component, Z designates a mathematical representation with just a zero-order term, F designates a mathematical representation with just a first-order term, and S designates a mathematical representation with just a second-order term. Combinations of Z, F, and S designate mathematical representations in which more than one term are included. (For example, ZF for a damper indicates a mathematical representation of $M_{dmp} = -sgn(\dot{\theta})b_0 - b_1\dot{\theta}$)

4.2.7 Effects of inertial properties and cadence on optimal MMC

Using the simplest representation of each spring and damper, MMC were then determined for various masses of the segments of the prosthetic leg. Because the stump of a transfemoral amputee typically weighs significantly more than the socket, designers of prosthetics have little control over the mass of the upper leg. Thus, for practical significance, various upper leg masses were prescribed, and the effects of lower leg and foot mass on MMC were determined. In total, four upper leg masses, seven lower leg masses, and seven foot masses were considered for a total of 196 inertial configurations. Finally, the effect of walking cadence (slow, natural, and fast cadence) on MMC and corresponding C was calculated.

4.3 Results

4.3.1 Sensitivity of cost to spring and damper complexity

Table 4.2 shows the results of the sensitivity analysis examining the effect of component complexity on cost. For each phase, cost is minimal for the most complex component (ZFS). However, during phase 1, cost can be reduced to within 1.0% of its minimum value by using a first-order (F) spring in the mechanical model. In addition, for both phase 2 and phase 3, cost can alternatively be reduced to its minimum value by using a zero-order (Z) damper in the model. Thus, a simple mechanical model consisting of a first-order spring in phase 1, a zero-order damper in phase 2, and a zero-order damper in phase 3 can reproduce M_{req} over the gait cycle to a nearly equivalent accuracy as the most complex mechanical model considered.

Figure 4-4 illustrates the behavior of the simple mechanical model over the gait cycle for a prosthetic leg with a typical inertial configuration. M_{mod} accurately reproduces M_{req} ($R^2 = 0.90$). Note that the difference between M_{mod} and M_{req} around 45% of the gait cycle is a consequence of an E_{diss} to E_{gen} ratio of less than one during phase 1, meaning that a single passive mechanical component cannot perfectly reproduce M_{req} during the phase.

4.3.2 Effects of inertial properties on MMC

Figure 4-5 illustrates the effects of upper leg, lower leg, and foot mass on optimal MMC. The graphs illustrate numerous results. Optimal k_1 during phase 1 is relatively insensitive to changes in mass of the prosthetic leg, optimal b_0 during phase 2 is moderately sensitive, and optimal b_0 during phase 3 is highly sensitive. Specifically, k_1 during phase 1, b_0 during phase 2, and b_0 during phase 3 vary by up to 5.6%, 36%, and 330%, respectively, relative to their minimum values.

For optimal k_0 during phase 1, k_1 generally increases with upper leg, lower leg, and foot mass, with some exceptions for foot mass at higher upper leg masses ($m_{ul}^* = 75\%$ and 100%). Upper leg mass has the greatest influence on k_1. As m_{ul}^* varies between 25% and 100%, k_1 increases by up to 4.2%. However, as m_{ll}^* and m_f^* vary, k_1 increases by no more than 2.5% and 1.0%, respectively.

For optimal b_0 during phase 2, b_0 generally decreases with upper leg and lower leg mass, but varies inconsistently with foot mass. Lower leg mass has the greatest influence on b_0. As m_{ll}^* varies between 25% and 100%, b_0 decreases by up to 27%. On the other hand, as m_{ul}^* varies, b_0 decreases by no more than 8.0%, and as m_f^* varies, b_0 changes by no more than 2.7% from its minimum to maximum values.

Finally, for optimal b_0 during phase 3, b_0 consistently increases with upper leg, lower leg, and foot mass. Foot mass has the greatest influence on b_0, but both upper leg and foot mass have a large influence as well. Specifically, as m_f^* increases from 25% to 100%, b_0 increases by up to 180%. In comparison, as m_{ll}^* and m_{ul}^* increase, b_0 increases by up to 134% and 45%, respectively.

In summary, k_1 during phase 1 is relatively insensitive to the mass of the prosthetic leg, whereas b_0 during phase 3 is highly sensitive. Upper leg mass has the greatest influence on k_1 during phase 1, and lower leg mass has the greatest influence on b_0 during phase 2. Upper leg, lower leg, and foot mass all have a large influence on b_0 during phase 3, but foot mass has the greatest influence.

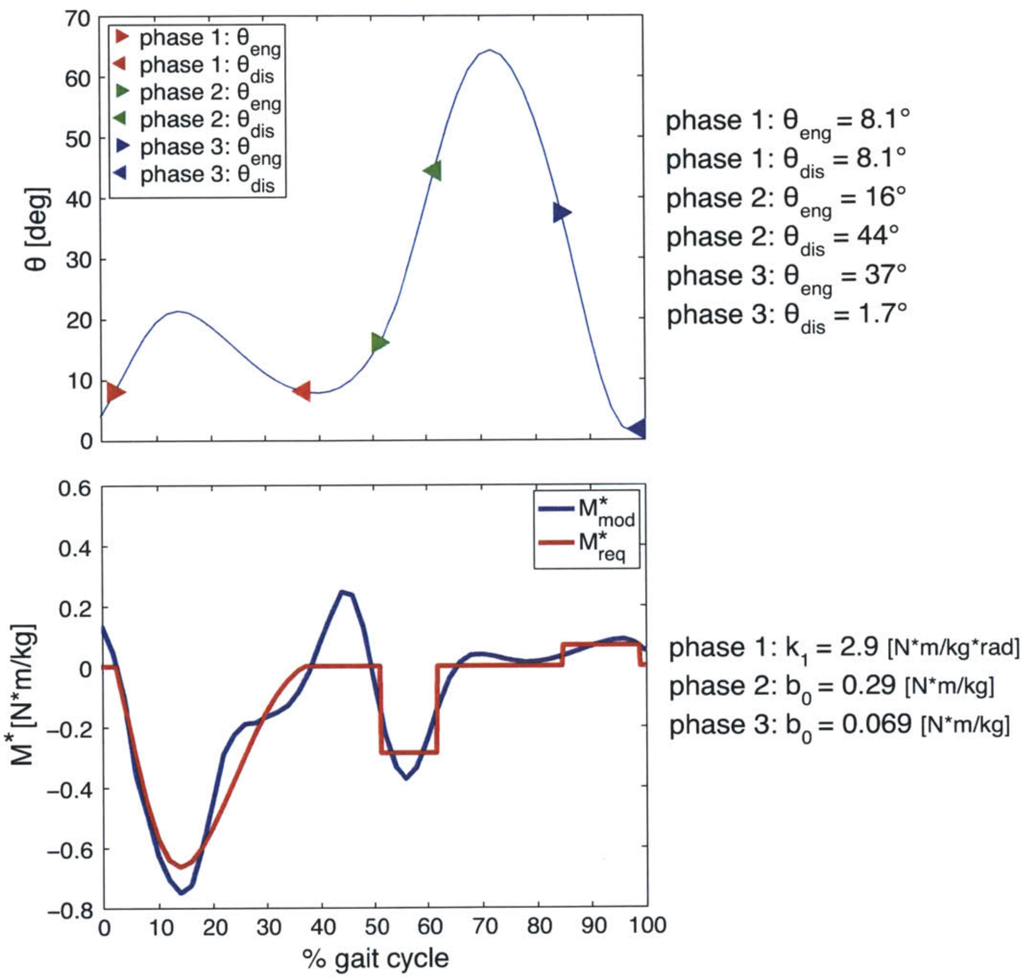

Figure 4-4: Behavior of simplified mechanical model over the gait cycle. The model was optimized to reproduce M_{req} of a prosthetic leg with a typical inertial configuration (mass of upper leg = 50% of able-bodied value, and masses of lower leg and foot = 33% of able-bodied values) walking at a natural cadence. Top: Illustration of optimized engagement and disengagement angles of the clutch for each component. θ_{eng} and θ_{dis} are the engagement and disengagement angles for a given phase. Bottom: Comparison of required moment for normative kinematics and the moment produced by the mechanical model ($R^2 = 0.90$). M^* is moment normalized to body mass, M^*_{req} is the required moment normalized to body mass, and M_{mod} is the moment produced by the model normalized to body mass. k_1 and b_0 are the linear spring coefficient and constant-damping coefficient for a given phase.

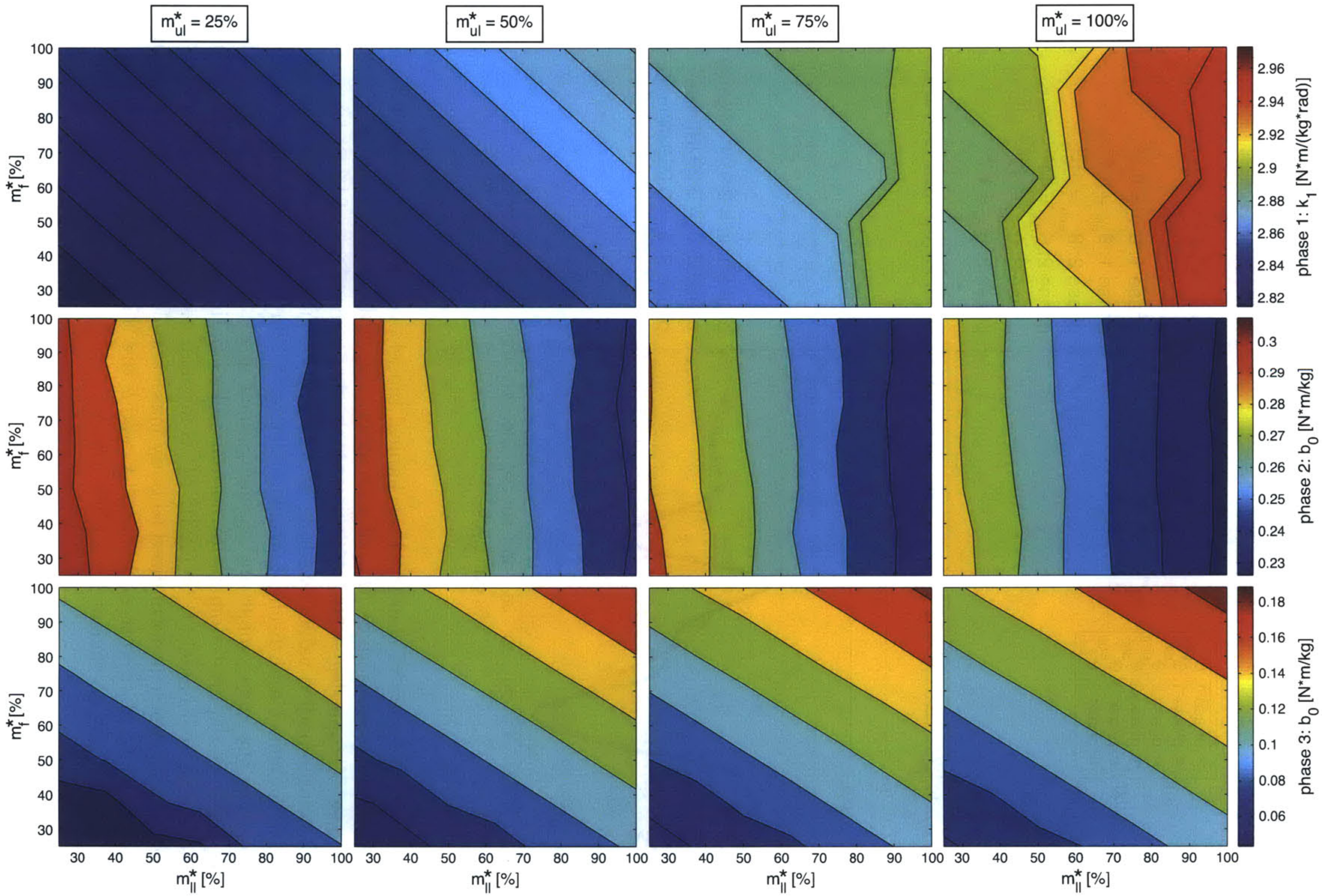

Figure 4-5: The effects of upper leg, lower leg, and foot mass on optimal MMC. m^*_{ul}, m^*_{ll}, and m^*_f are upper leg mass, lower leg mass, and foot mass normalized to body mass.

		Cadence		
	Parameter	Slow	Natural	Fast
Phase 1	$k_1[\frac{N*m}{kg*rad}]$	3.79	2.86	3.24
Phase 2	$b_0[\frac{N*m}{kg}]$	0.28	0.29	0.42
Phase 3	$b_0[\frac{N*m}{kg}]$	0.039	0.069	0.078
	C^*	0.093	0.085	0.057

Table 4.3: The effects of walking cadence on optimal MMC and corresponding cost for a prosthesis with a typical inertial configuration (upper leg mass = 50% of able-bodied value, and lower leg and foot mass = 33% of able-bodied values). C^* is equal to C normalized to the maximum theoretical cost over the gait cycle (i.e., the cost over the gait cycle when no components are used to model the knee).

4.3.3 Effects of cadence on MMC

Table 4.3.3 shows the effects of walking cadence on optimal MMC for a prosthesis with a typical inertial configuration. All parameters change significantly across cadences, with b_0 during phase 3 varying the most. Specifically, k_1 during phase 1, b_0 during phase 2, and b_0 during phase 3 vary by 33%, 50%, and 100%, respectively, relative to their minimum values. In addition, cost varies by 63% relative to its minimum value. However, from an absolute perspective, cost is low ($C^* < 0.10$) at all cadences, indicating that M_{mod} consistently reproduces M_{req}.

4.4 Discussion

4.4.1 Comparison to previous work

The present study designed a mechanical model and determined MMC to best reproduce M_{req} for a prosthetic leg across a wide range of inertial configurations. In addition, the effects of walking cadence on MMC were examined for a prosthetic leg with a typical inertial configuration. Because previous studies focused on determining MMC for an able-bodied leg, the only results of the present study that can be directly compared to those of previous studies are the MMC of the prosthetic leg with an able-bodied inertial configuration (i.e., upper leg, lower leg, and foot mass = 100% of able-bodied values) walking at a natural cadence.

Comparisons between the results are favorable. For the able-bodied inertial configuration, the present study modeled the knee as a linear spring with a spring coefficient of 2.86 $[\frac{N*m}{kg*rad}]$ during phase 1 (top-right of Figure 4-5). Shamaei et al[8] modeled the knee as a linear spring with a spring coefficient of 2.92 $[\frac{N*m}{kg*rad}]$ during the weight acceptance phase, which corresponds closely with phase 1. The spring coefficients of the present study and Shamaei et al are nearly equivalent, differing by only 2.1%. Sup et al[2] modeled the moment-time relationship of the knee with a linear spring, cubic spring, and linear damper over four different kinematically and kinetically defined regions of the gait cycle. The linear spring stiffness was determined to be 2.89 $[\frac{N*m}{kg*rad}]$ during the first region, which corresponds closely with phase 1. Again, the spring coefficients of our study and Sup et al are nearly equivalent, differing by approximately 1.0%. On the other hand, Martinez-Villalpando and Herr[3] modeled the moment-time relationship of the knee during stance as two linear springs with clutches and partially overlapped engagement. The spring coefficient was 1.95 $[\frac{N*m}{kg*rad}]$ for the first spring, which acted during part of stance that corresponded roughly with phase 1. The large (34%) difference between the spring coefficient of the present study and that of Martinez-Villalpando and Herr is likely because the present study derived M_{req} based on averages across many subjects[16], whereas the latter study derived M_{req} from measurement of a single subject.

4.4.2 Analysis of major findings

The results supported the hypothesis that a simple mechanical model of the knee could accurately reproduce M_{req} for a prosthetic leg, as a model consisting of a first-order spring, two zero-order dampers, and three clutches was shown to do so. In addition, the model was able to reproduce M_{req} nearly as accurately as a complex mechanical model consisting of nonlinear springs and dampers. Physically, this result indicates that a prosthetic knee with a torsional spring, constant-friction rotary dampers, and mechanical clutches may allow an above-knee amputee with limited musculotendon function to walk with normative kinematics. Since springs and constant-friction dampers (e.g., friction pads) are inexpensive and mechanical clutches (e.g., contact clutches and friction clutches) can be easily implemented, the knee may potentially be fabricated for a low cost.

The study also validated the hypothesis that the mass of a prosthetic leg would have a significant effect on optimal MMC. All studies found in the literature investigating MMC optimized the coefficients to reproduce M_{req} for a leg with *able-bodied* inertial properties. The present study indicates that alterations in mass that are typical of a prosthetic leg have a significant effect on M_{req}, altering peak magnitudes during stance and swing by 40-60% for walking at a natural cadence. In turn, the effects on M_{req} cause optimal MMC to change significantly, altering coefficients by up to 180%. This result strongly suggests that designers should estimate the effect of mass on MMC before selecting components for a prosthetic knee. This conclusion is supported by the study of Sup et al[2], which found significant differences between theoretically optimized MMC and user-preferred component coefficients. The authors proposed that the inertial difference between able-bodied humans and amputees was a primary cause of the discrepancy.

The present study also computed relationships between upper leg mass, lower leg mass, foot mass, and optimal MMC in detail. The results were reported in the form of parametric illustrations that may be useful for designers of prostheses that wish to select components for a prosthetic knee. Each segment of the prosthetic leg was found to be the principal inertial determinant for a particular coefficient–the upper leg for k_1 during phase 1, the lower leg for b_0 during phase 2, and the foot for b_0 during phase 3. However, k_1 did not change significantly (no more than 5.6%) over the entire range of inertial configurations, and all three segments were found to have a large influence on b_0 during phase 3. The results suggest that alterations in prosthetic mass may not require significant adjustments in spring components used during weight-bearing in the prosthetic knee, but they may necessitate adjustment of damping components used during late stance and swing. In addition, because MMC were calculated based on knee moments normalized to body mass, component coefficients must be scaled in proportion to body mass.

Finally, the present study supported the hypothesis that walking cadence would have a significant effect on optimal MMC. From slow to fast cadences, optimal MMC varied up to 100%. This result agrees with common knowledge that in a prosthetic knee, components need to be customized or dynamically adjusted to the walking cadence of the user[25, 26]. To reduce the need for adjustment, future analysis should focus on identifying single components that may not perform optimally at any given cadence, but perform sufficiently well across multiple cadences. For instance, the present study found viscous dampers (i.e., first-order dampers) to reproduce M_{req} less accurately than constant-friction dampers for walking at a natural cadence; however, viscous dampers are well-known in the prosthetics community for allowing amputees to walk comfortably at multiple cadences[25, 27]. The variation of optimal constant-friction damping coefficients and optimal viscous damping coefficients across cadences and their ability to accurately reproduce M_{req} at all cadences can be compared.

4.4.3 Limitations of study

There are a few notable limitations of the present study. One limitation is the assumption that if a difference exists between M_{mod} and M_{req}, the neuromuscular control system of the amputee can provide compensatory moments (e.g., with hip musculature or residual knee musculature) to produce normative kinematics. The idea that amputees possess a robust neuromuscular control system is supported in the literature. For instance, above-knee amputees have been able to walk using prosthetic knees with only constant-friction damping[28], and Selles et al[29] demonstrated that transtibial amputees respond to significant mass perturbations by altering joint kinetics to maintain kinematics. However, there are limits to such control. For example, transfemoral amputees using state-of-the-art prostheses still do not express certain normative kinematic features, such as stance flexion[22]. If the neuromuscular control system were not able to provide normative kinematics at all points in the gait cycle, the moment produced by the model would need to be modified to minimize accumulated kinematic error. Determining the extent of the modification would require forward dynamic simulation, which has well-known challenges and limitations[30, 31].

Another limitation of the study is the restricted range of mechanical models examined. Although power analysis was able to divide gait into three phases and determine the component (i.e., spring or damper) that best replicated the energy characteristics of each phase, only one component was allowed to engage during each interval. In addition, only up-to-quadratic mathematical representations of each component were considered. Eliminating these constraints would lead to a significant increase in the complexity of optimization and physical implementation, but the results may justify the effort. For instance, if an additional spring were engaged during both phase 1 and phase 3, energy stored in the spring during phase 3 could supplement energy released by the original spring in phase 1, allowing M_{mod} to more closely replicate M_{req} at 45% of the gait cycle (Figure 4-4). Since the moment produced by the knee at this time is a flexor moment during mid- to late stance[32], this addition would presumably allow an amputee to initiate flexion more easily prior to swing.

A final limitation of the study was the process used to perform the inverse dynamics simulation. During the process, kinematic and kinetic data were combined from multiple sources in the literature due to the lack of complete data sets available. Several kinematic estimates were made, and certain kinetic parameters (e.g., ground reaction force for a prosthetic leg during double-support) were estimated based on simplified assumptions. These estimates may have led to error in the calculation of M_{req}. A full discussion of the process and a comparison of moments to literature values are available in Narang and Winter (2013a)[11] (i.e., Chapter 3).

4.5 Conclusion

The present study aimed to determine optimal stiffness, damping, and engagement parameters for a low-cost, passive prosthetic knee through the design and optimization of a mechanical model. Inverse dynamics was used to determine the knee moment required for a prosthetic leg to walk with normative kinematics. A simple mechanical model of the knee was designed and optimized to accurately replicate the required moment. Mass of the prosthetic leg was found to have a significant effect on the required moment, and upper leg mass, lower leg mass, foot mass, and walking cadence were each found to have a significant influence on the optimal coefficients of the components in the model, particularly the damping coefficients.

In contrast to previous studies, the present work used power analysis to select components for the mechanical model, used sensitivity analysis to identify the simplest mathematical representation of each component, optimized the coefficients of each component to reproduce the moment required for a lightweight prosthetic leg (rather than an able-bodied leg) to walk with normative kinematics, and determined the effects of walking cadence on optimal coefficients. In addition, the study computed the effects of the mass of each segment of the prosthetic leg on optimal coefficients and reported the results in a parametric illustration that can be used by designers of prostheses. Future work should focus primarily on comparing the performance of constant-friction and viscous dampers across multiple cadences, as well as reducing restrictions on the engagement of components in the model to more accurately reproduce the moments required for normative kinematics.

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Chapter 5

Conclusion

5.1 Summary of thesis

The primary goal of the thesis was to identify detailed design requirements for a high-performance, low-cost, passive prosthetic knee for amputees in India. The introduction section described the general design requirements for the knee. The project partner, BMVSS, presented three initial requirements: the knee should allow normal gait on flat ground, provide stability on uneven terrain, and cost less than \$100 to manufacture. Interactions with other members of the prosthetics, rehabilitation, and academic communities across India enabled the identification of additional general requirements, including that the knee should be compliant with international standards for strength testing and easy for technicians to fit and align.

Chapter 2 presented the results of a structured survey of transfemoral amputees conducted at the BMVSS limb fitment center in Jaipur, India. The survey documented the demographics of subjects, whether various activities were difficult for them to perform, and whether the ability to do additional activities would significantly improve their lives. A metric was defined called the Potential Impact of Design Improvement (PIDI) score, which quantified the importance of enabling amputees to perform certain activities with newly developed prostheses. Subjects were generally young, male, and lived in villages. They felt that floor-sitting postures (i.e., cross-legged sitting, squatting, and kneeling) and walking on wet mud were the most difficult activities among those considered. Calculation of the PIDI score determined that newly developed prostheses should enable amputees to perform these activities, as well as walking fast, carrying heavy objects, walking in water, and riding a bicycle.

Chapters 3 and 4 focused on improving the baseline performance of prostheses–specifically, reducing the metabolic energy expenditure of amputees and allowing them to walk with normative gait kinematics. In Chapter 3, a 2-dimensional inverse dynamics simulation was conducted to determine how the inertial properties of a prosthetic leg and walking cadence affected the total hip work (i.e., integral of the absolute value of hip power) required to walk with normative kinematics. Total hip work was used as a measure of muscular effort at the hip, and changes in its value were used to estimate changes in metabolic energy expenditure. Decreasing the masses of the leg segments reduced total hip work to a nearly equivalent extent as decreasing both masses and moments of inertia (about centers of mass). Decreasing lower leg and foot mass reduced total hip work by up to 22%, and foot mass had up to a threefold greater effect than lower leg mass. Across all cadences, minimizing the masses of the leg segments minimized total hip work.

Chapter 4 determined how inertial properties of the prosthetic leg and walking cadence affected optimal coefficients of components in a prosthetic knee. Analysis of knee power was used to select components for the prosthetic knee, and a 2-dimensional inverse dynamics simulation was conducted to determine how inertial properties affected the knee moments required to walk with normative kinematics. The coefficients of the components were then optimized to produce the required moments. A simple model consisting of a first-order spring, two zero-order dampers, and three clutches was found to accurately reproduce required moments. Decreasing prosthetic mass affected required moments by up to 60%, and altering the mass of any of the leg segments had a notable effect on optimal coefficients, particularly for the dampers. Walking cadence also affected optimal coefficients by up to 100%.

5.2 Novel research contributions

The research presented in this thesis includes numerous methodologies, results, and analyses that may be relevant to the biomechanics and prosthetics communities. Whether these contributions are the first of their kind can only be determined through an exhaustive search of the literature; however, based on an understanding of the literature as presented in the previous chapters, the following contributions are considered candidates for novel work:

Chapter 2

- Quantification of the ability of transfemoral amputees in India to perform a large number (20+) of activities
- Quantification of the ability of transfemoral amputees to perform these activities using BMVSS above-knee prostheses
- Quantification of the potential for additional abilities to significantly improve the lives of transfemoral amputees in India
- Computation of a PIDI score to quantify importance of enabling amputees to perform certain activities with newly developed prostheses

Chapter 3

- Methodology to combine kinematic and kinetic parameters from multiple incomplete data sets

- Methodology to estimate the ground reaction force (GRF) acting on amputees by using fundamental equations of multi-rigid-body systems to modify the normative GRF acting on able-bodied humans

- Use of inverse dynamics to quantify the effect of prosthetic leg inertial properties on total hip work during both swing and stance

- Comparison of the effects of altering masses of the segments of a prosthetic leg on total hip work with the effects of altering moments of inertia (about centers of mass)

- Parametric quantification of the effects of a wide range of upper leg, lower leg, and foot masses on total hip work

- Use of inverse dynamics to quantify the effect of prosthetic leg inertial properties on total hip work at multiple cadences

Chapter 4

- Use of inverse dynamics to quantify the effect of prosthetic leg mass on the knee moment required for walking with normative kinematics

- Division of the gait cycle into energetic phases to select components for a prosthetic knee

- Optimization of coefficients of components to produce the knee moment required for a prosthetic leg with reduced inertial properties (rather than able-bodied inertial properties) to walk with normative kinematics

- Use of sensitivity analysis to quantify the effect of the mathematical order of components on their ability to produce the knee moment required for walking with normative kinematics

- Parametric investigation of the effects of upper leg, lower leg, and foot mass on optimal coefficients of components

In addition, although the contributions may not explicitly be considered novel research, it is a hope of the present work that illustrations in Chapter 3 and Chapter 4 will be particularly useful to designers of prostheses. Specifically, after some refinement, the parametric illustration of the effects of upper leg, lower leg, and foot mass on total hip work (Figure 3-4) and optimal coefficient values (Figure 4-5) could be used to reduce energy expenditure and select components for a prosthetic knee based on the needs and inertial properties of target users.

5.3 Recommendations for future work

Although the research in this thesis constitutes a useful framework for the design of a low-cost, passive prosthetic knee, significant potential still exists for improvement and extension of the work. The following is a list of recommendations for future research based on the work presented in each chapter:

Chapter 2

- *Increase sample size* – A larger sample size of subjects should be obtained to enable comparisons of strata within the population (e.g., subjects using manual-locking knee, subjects using four-bar knee, etc.)

- *Expand subject selection criteria* – Although constituting a small (3.6%)[1] percentage of the amputee population, bilateral above-knee amputees should be interviewed to understand their specific abilities and needs

- *Interview a more diverse population* – Since all subjects were male and lived in North, Northwest, or Central India, a more diverse population should be interviewed to identify the effects of gender and region on the needs and abilities of amputees

- *Ask questions on a multi-level scale* – Although subjects were generally unresponsive to a multi-level Likert scale, a means should be designed to effectively implement such a scale in order to provide greater resolution to the results (e.g., in Figure 2-6)

- *Expand the PIDI score* – To prevent overweighting activities that few subjects have performed (e.g., walking on snow), the PIDI score should include an additional factor equal to the percentage of amputees that have performed a given activity

Chapter 3

- *Conduct gait analysis* – A gait analysis study should be conducted to generate a complete kinematic and kinetic data set for walking at multiple cadences, which would minimize the effects of estimation error

- *Scale model to Indian anthropometric ratios* – The anthropometric ratios of Indian people may differ significantly from those of Western populations[2, 3] and should be applied to the model before using the results to specify inertial properties and components for a prosthesis

- *Refine estimate of GRF during double-support* – The estimate of GRF on the leg during double-support using a linear transition assumption should be improved with more accurate transition assumptions[4]

- *Further investigate relationship of hip work, muscular effort, and metabolic energy expenditure* – Further empirical and theoretical studies should be conducted to rigorously elucidate the relationship of hip work, various measures of muscular effort[5], and metabolic energy expenditure

- *Model the effects of passive ankle joint* – Because prosthetic ankle joints for the developing world are typically unpowered, the effects of using passive ankles on hip work should be examined

- *Perform 3-dimensional inverse dynamics* – Although the vast majority of joint work is in the sagittal (forward-back) plane, hip work in the frontal (left-right) plane has been found to be significant[6] and should be quantified with a 3-dimensional inverse dynamics simulation

- *Investigate amputee mass preference* – Empirical and theoretical studies should be conducted to identify why some amputees prefer heavier limbs[7, 8], and the results should be applied to the specification of inertial properties for prostheses

Chapter 4

- *Investigate neuromuscular compensation* – Empirical and theoretical studies should be conducted to determine the extent to which the neuromuscular system of an amputee can compensate for the difference in the moment produced by a prosthetic knee and the moment required for normative kinematics

- *Eliminate engagement constraints in the simulation* – The results of engaging an additional spring during both swing and stance should be tested, as the prosthetic knee may more accurately reproduce the moment required for normative kinematics

- *Identify components that work across multiple cadences* – Components should be identified that may not be optimal at any single cadence, but provide the best overall performance across multiple cadences

- *Examine sensitivity of engagement angles to inertial properties* – Although optimal engagement angles were calculated for a typical prosthesis, the sensitivity of engagement angles to inertial properties and walking cadence should also be examined

- *Apply techniques to other motions* – The simulation and optimization techniques presented in Chapter 3 and Chapter 4 should be applied to determine optimal inertial properties and components for additional motions, such as those investigated in Chapter 2 (e.g., carrying heavy loads)

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Appendix A

Consent Form and Survey

The consent form and survey as administered in January 2013 are included below.

Interview Consent Form

Before interview:

My name is Yashraj Narang. I am a researcher at M.I.T., a university in the U.S. I have come here to understand how people in India use prosthetic legs. I will be interviewing people at BMVSS, and I will be using the information from the interviews to design a new prosthetic leg for people in India.

The interview will last 10-20 minutes. I first want to ask you some questions, and I then want to take a video of you walking on a surface that you are experienced and comfortable walking on. If you do not want to answer any question or walk, you do not have to, and you can stop the interview at any time.

Sometime before you leave BMVSS, I may ask if I can visit you at your home and/or workplace for a few hours at a later time. During the visit, I want to watch you use your prosthetic leg throughout the day, write notes about my observations, and take video. You do not have to accept the request, and you can ask me to leave at any time.

You will not receive money for the interview, and what you say or do will not affect your treatment here in any way.

Is there anything you do not understand?

Would you like to participate in the interview?

After interview:

Would it be acceptable to share the results of the interview in publications and presentations to other researchers? *[Note: I will ask about each category of information, as listed at the bottom of the page.]*

The project will be completed by June 1, 2015. Please contact me at [email address redacted] or [phone number redacted] at any time with any questions or concerns. If you feel that you have been treated unfairly, or if you have questions regarding your rights, please contact my interview supervisor at [email address redacted] or [phone number redacted].

For researcher use only:

Subject consents to...

Verbal questionnaire	Yes/No
Walking exercise	Yes/No
Home visit	Yes/No
Sharing of questionnaire responses	Yes/No
Sharing of video of walking exercise	Yes/No
Sharing of notes of home visit	Yes/No
Sharing of video of home visit	Yes/No

Interview Questionnaire

Location: Date: Translator:

Background Information

Gender:

Age:

Lives in:

Unilateral/bilateral:

Cause of amputation:

years wearing most recent prosthetic leg:

Occupation before amputation:

after amputation:

after prosthetic leg:

Location of workplace:

Salary:

Usage History

What different types of prosthetic knees have you used?

Why did you change from one type to the next?

Do you ever go barefoot with your most recent prosthetic leg? If so, when?

Do you ever use a cane, crutches, or wheelchair with your most recent prosthetic leg? If so, when?

Activity Survey (Verbal)

1) For each of the following activities, are you able to do it easily while wearing your most recent prosthetic leg? (Researcher marks 'N' for not easy, 'D' for do not know)
2) For each of the activities marked with 'N', if you were able to do it easily with a different prosthetic leg, would your life be significantly improved? (Researcher marks 'I' for improved)
3) For each of the activities marked with both 'N' and 'I', why are you not able to do it easily, and why would your life be significantly improved?

- Walk on flat ground
- Walk fast
- Walk up/down stairs
- Walk on dirt
- Walk on wet mud
- Walk on rocks
- Walk on sand
- Walk on grass
- Walk on snow
- Walk through water
- Walk up/down hills
- Ride a bicycle

- Drive a motorcycle
- Drive a car
- Carry heavy objects
- Stand for a long time
- Sit in a chair for a long time
- Go from sitting in a chair to standing,

  and from standing to sitting in a chair
- Squat/use an Indian toilet
- Kneel
- Lie down
- Sit cross-legged

4) Are there any other activities that you are not able to do easily with your most recent prosthetic leg, but that if you were to be able to do it easily with a different prosthetic leg, your life would be significantly improved?

Stability Survey

1) How often do you fall with your current prosthetic leg?
2) What causes you to fall?

Video Segment

1) What surfaces in this facility do you have experience with and feel very comfortable walking on?
2) Could you show me how you walk on [name of surface]?

Additional Comments

1) Do you have any additional questions, comments, or suggestions for me?