BIOMECHANICAL ANALYSIS OF LOWER LIMB PROSTHESIS

A Thesis

by

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MASTER OF SCIENCE

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ABSTRACT

This thesis presents a case study of the initial analysis of self-contained powered transfemoral (TF) prosthesis (AMPRO II). We analyze how the prosthesis influences the biomechanics of TF amputee walking gait. TF amputees have problems with increased energy expenditure and gait asymmetry, which leads to problems with their intact leg problems, such as osteoporosis and scoliosis. In order to assess the effectiveness of AMPRO II, we must analyze how it addresses these issues. This study will compare the amputee's energy expenditure, kinematic gait symmetry (joint angles) and kinetic gait symmetry (joint reaction forces, and moments) while wearing the participant's own microprocessor-controlled TF prosthesis, and AMPRO II. Using AMPRO II enhanced the kinetic symmetry for the hip and knee flexion moment and enhanced kinematic symmetry for the knee and ankle angles. However, using AMPRO II led to increased energy expenditure and decreased symmetry in the hip angle and ankle moment. The findings from this study will lead to an understanding of how AMPRO II affects TF amputees and provide vital information that can be used in the future to improve the functionality of AMPRO II and future iterations of the device development.

DEDICATION

Through my entire graduate school experience, God has been there guiding, providing and encouraging me. I would like to dedicate my thesis to Him. I would also like to dedicate this to my mother for her continued encouragement and support. Lastly, I would like to dedicate this to my wonderful grandmother, Elizabeth Williams, who is the motivation for my research.

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NOMENCLATURE

AMBER	A&M Bipedal Experimental Robot
AMPRO	A&M Prosthetics
COM	Center of Mass
COP	Center of Pressure
TF	Transfemoral
TT	Transtibial
IMU	Inertial Measurement Unit
GRF	Ground Reaction Force

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INTRODUCTION AND LITERATURE REVIEW

The Amputee Coalition of America estimates that there are 2 million amputees within the United States and of those, 86% have lower limb amputations (Program & America, 2015). Lower limb amputations have a large effect on mobility and this population is continually growing, with approximately 185,000 new lower extremity amputations each year (Program & America, 2015). Of these lower limb amputations, approximately 45% are transfemoral (TF, above the knee) amputees. TF amputees have a unique set of problems in regards to their walking gait and balance.

During flat ground walking, most humans follow a certain pattern called the gait cycle (Figure 1). The walking gait has a stance phase, from heel strike to toe off, and swing phase, from toe-off to heel strike. The gait cycle is composed of seven events: initial contact, load response, heel off, opposite initial contact, toe off, feet adjacent, tibia vertical, and next initial contact.

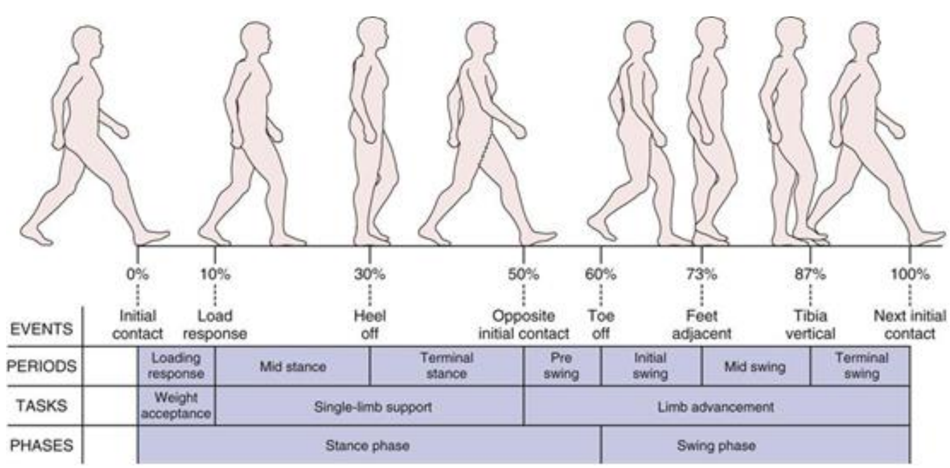

Figure 1: Gait cycle (Neuman, 2010)

During these phases, certain strategies are typically used in healthy intact people (Kishner & Laborde, 2015). Some of these strategies are knee flexion (bending of the knee) in the stance phase, dorsiflexion at initial contact and plantar flexion at terminal stance (Kishner & Laborde, 2015). Due to unilateral TF amputees missing their knee and ankle, they are unable to implement some of those strategies on the side of amputation. This yields less efficient walking, increased asymmetries, and increased energy expenditure. Energy expenditure for unilateral below the knee amputees is 20-30% higher than intact subjects (Herbert, Engsberg, Tedford, & Grimston, 1994), and there is an additional 25% increase in energy expenditure for TF amputees (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011).

The prosthetic limb typically performs approximately half of the work of normal muscle (Kishner & Laborde, 2015). This leaves the amputee to compensate using their intact limb. Many passive prosthetics restrict knee flexion during stance phase to increase stability. However, this leads to a less efficient walking gait. Due to TF amputees' concern about putting weight on the prosthetic, this can sometimes lead to a longer stance phase in the intact limb and prolonged knee extension (Kishner & Laborde, 2015). The compensations by the intact leg and restrictions of prosthetics can lead to asymmetries in walking and strength that result in problems with TF amputees (Gailey, Allen, Castles, Kucharick, & Roeder, 2008; Lloyd, Stanhope, Davis, & Royer, 2010). Some of the complications that unilateral amputees have after long-term prosthetic use are osteoarthritis, osteopenia (low bone mass), scoliosis and a greater risk for back problems (Gailey, Allen, Castles, Kucharick, & Roeder, 2008). Attempts to resolve the problems described above include a variety of methods, such as changing the socket linear material, prosthetic fitting, physical therapy, and prosthetic knee and ankle mechanisms.

Review of Lower Limb Prosthetics

Prosthetic mechanisms fall into three categories: passive, microprocessor-controlled, and powered. Each of these prostheses could have different effects on gait symmetry, balance, and energy expenditure. Passive prostheses (Figure 2) are the most common and least expensive out the prosthetic types. Passive prostheses have simple mechanisms for movement and are lightweight compared to non-passive devices. However, they do require the users to use their own muscle to maintain stability when standing and tend to lock the knee joint (in order to support body weight) during stance or swing phases. Due to the simple mechanisms passive prostheses, joint angles tend not to mimic normal walking.

Figure 2: Polycentric passive knee

Microprocessor knees (Figure 3) have become more prevalent in recent years. Microprocessor knees have onboard sensors to detect movement and timing. They also have varying stiffness during swing and stance phases, allowing a more natural gait. However, microprocessor knees do not provide any power into the system.

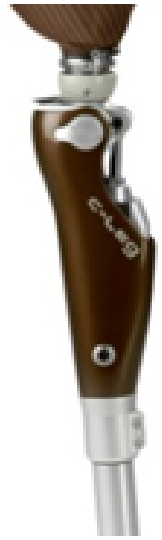

Figure 3: C-Leg (Microprocessor knee)

Powered prostheses are the least prevalent among the prosthetic types. There is currently only one powered knee on the market, Ossür power knee (www.ossur.com, 2014) (Figure 4) and one powered ankle, iWalk BiOM (Figure 5). However, several powered prostheses have been developed for research purposes (Martinez-Villalpando, Weber, Elliot, & Herr, 2008; Sup, Bohara, & Goldfarb, 2008; Fite, Mitchell, Sup, & Goldfarb, 2007; Kapti & Yucenur, 2006). Powered knee and ankle systems benefit the user by putting power into the systems and adding more variability during the stance and swing phases. This can allow for knee flexion in the stance phase, which helps to optimize walking. Powered prostheses utilize motors to provide actuation at the joints. Powered ankles use active plantar flexion in terminal stance period to aid in push off. Amputees tend to use less energy while using powered prostheses (Martinez-

Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011).

Figure 4: Ossür power knee

Figure 5: BiOM powered ankle

Even with the increasing number TF and TT amputees, passive prostheses still dominate the prosthetic market. This is more likely due to their ease of use and low cost of production. Currently, the powered prosthetics are expensive and heavier than most passive prostheses due to the required actuators. The added weight can increase the energy expenditure if there is no ankle actuation and could prove to be adversely effective (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011). However, there are studies where the powered device is heavier, compared to passive prostheses but still have reduced energy expenditure (Au, Weber, & Herr, 2009).

To the best of our knowledge, no studies have investigated if the powered prosthetics result to a more symmetric gait. Commercially available powered prosthetic devices have limited studies showing the benefits of powered prosthetics, which have the potential to resolve the problems that TF amputees face. If they are not better than what is available, then we need to find ways to improve their functionality. In the following sections, we will assess how the current prosthetic mechanisms impact gait symmetry, balance and energy expenditure.

Gait Symmetry and Balance

Kaufman et al. (Kaufman, Frittol, & Frigo, 2012) showed the difference in gait symmetry when using a passive mechanical knee and a microprocessor knee. In this study, there were no significant differences in the hip, knee, and ankle kinematic symmetry between the passive mechanical knee and microprocessor knee. However, there was a significant difference in kinetic symmetry. The microprocessor knee showed an increased symmetry in joint angles in the hip, knee and ankle joints.

Kaufman et al. (Kaufman, et al., 2007) also studied balance to determine the effects of using a microprocessor knee and a passive mechanical knee. In order to measure balance, postural sway was measured by an equilibrium score, ranging from zero to 100, with a score of 100 being the least postural sway. This was measured for six different Sensory Organization Tests (SOT) to assess the three sensory components of balance (visual, somatosensory, and vestibular inputs) under a variety of altered visual and surface support conditions (Kaufman, et al., 2007). The equilibrium score was higher for the microprocessor knees indicating enhanced balance compared to passive knees. In this same study, it was also found that using the microprocessor controlled prosthesis resulted in a more normal walking pattern compared to when using a passive knee. Microprocessor knees resulted in knee flexion during loading and the knee moment changed from an internal flexion moment when using the mechanical prosthesis to an internal extension moment (Kaufman, et al., 2007). This means that the knee is extending during loading and helping the user stay upright.

A prosthetic developed by Michael Goldfarb group at the Vanderbilt University was tested for stability enhancement (Lawson, Varol, & Goldfarb, 2011). In this study, they successfully obtained optimal support at different slopes while standing. This allowed the prosthetic to bear an even amount of weight during standing. The ability of the prosthetic to bear some of the weight possibly reduced the overall postural sway, although this parameter was not measured.

In the same group, Sup et al. (Sup, Bohara, & Goldfarb, 2008) developed a powered TF prosthesis that controls the knee and ankle simultaneously. The control resulted in similar joint angles to the average trajectory for the normal gait. The joint torques were also close to the average torques for a normal gait for walking. However, this study did not provide any measures for similarity or symmetry.

Energy Expenditure/ Cost of Transport

According to the study by Göktepe et al. (Göktepe, Cakir, Yilmaz, & Yazicioglu, 2010), higher amputation levels required more energy to walk. This showed that TF amputees use more energy than TT amputees do. In that study, the average oxygen consumption for TF amputees using a passive mechanical knee with a walking speed at 3 km/h was 10.08± 2.21 mL/min/kg.

In the study by Schmalz et al. (Schmalz, Blumentritt, & Jarasch, 2002), unilateral TF amputees using a mechanically controlled hydraulic knee at 3.4 km/h had an average O_2 consumption rate of 15.1±1.1 mL/min/kg. When using a C-leg (a microprocessor controlled hydraulic knee) (Ottobock, Duderstadt, Germany) in the same study, the average O_2 rate was found to be 14.2±1.2 mL/min/kg (Schmalz, Blumentritt, & Jarasch, 2002). In this study, the amputee using the microprocessor knee had a lower O_2 consumption rate than did the amputee with the passive knee. The significance of these differences dwindles at higher walking speeds. This study, which used a variety of passive feet, found that the different passive feet did not have a significant impact on O_2 consumption rate. Comparing the two studies by Göktepe et al (Göktepe, Cakir, Yilmaz, & Yazicioglu, 2010) and Schmalz (Schmalz, Blumentritt, & Jarasch, 2002), it seems that using passive knees (Göktepe et al.) results in a lower O_2 consumption rate than the microprocessor controller knee (Schmalz, Blumentritt, & Jarasch, 2002). This is possibly due to the length of time walked in the study, for Göktepe it was five minutes (Göktepe, Cakir, Yilmaz, & Yazicioglu, 2010), while in the study by Schmalz, the test was fifteen minutes (Schmalz, Blumentritt, & Jarasch, 2002).

A study done by Martinez (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011) assessed the metabolic cost of walking with antagonistic active knee prosthesis in comparison to a variable damping prosthetic knee

(C-leg). The antagonistic knee, developed at MIT, incorporated two unidirectional series elastic actuators. In this study, the average resting metabolic cost was 1.06 W/kg (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011). The metabolic power associated with the C-leg was 6.50 W/kg and the metabolic cost when using the active knee was 6.13 W/kg (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011). There was an overall reduction of 6.8% when using the active knee (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011). This is despite the fact that the overall mass of the artificial limb (knee prosthesis with subject's own foot and shoe) was 3.6 kg when using the active prosthesis and 2.6 kg when using the C-Leg. This shows that the active knee still reduces the amount of energy used during walking despite being heavier. The weight usually has adverse effects on metabolic cost.

Summary

Microprocessor knees result in lower energy use during walking compared to passive knees (Schmalz, Blumentritt, & Jarasch, 2002). Using the powered knee in the study by Martinez-Villalpando et al. (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011) resulted in lower energy use during walking even though it was heavier than the microprocessor knee. This shows that the device being powered helps to reduce the amount of energy the amputee has to use. In both studies by Kaufman (Kaufman, et al., 2007) and (Kaufman, Frittol, & Frigo, 2012), the microprocessor knee was shown to allow for better balance and kinetic symmetry than passive knees. While microprocessor knees improved balance and kinetic symmetry, they did not show any significant improvement in kinematic symmetry. Powered knees showed tactics that could allow for better weight distribution and possibly better gait. Additional research is required to assess the effectiveness of powered knees in improving balance, symmetry and energy expenditure.

As seen above, there is potential for powered prosthetics to improve the movement of unilateral amputees. However, research in powered prosthetics often does not analyze how they affect symmetry and energy expenditure. These two areas are important to show the effectiveness of a prosthetic for TF amputees. Recently, we have developed an adaptable control algorithm for our custom-designed powered TF prosthesis (AMPRO II). However, the device and controller have not been tested in terms of biomechanical outcomes (e.g., gait symmetry, joint mechanics and energy expenditure). Therefore, it is important to investigate whether the new system can enhance biomechanical outcomes of TF amputee gait.

The aim of this work is to determine how AMPRO II impacts the biomechanical outcomes of TF amputees' gait during normal flat ground walking. Based on the preliminary data and simulation results, control adaptability and minimal calibration have positive effects on several biomechanical outcomes (e.g., gait symmetry index and energy expenditure). A secondary aim is to assess what possible improvements can be made to improve biomechanical outcomes of TF amputees' gait.

DEVICE OVERVIEW AND TARGET USER

The following two sections will give an overview of the device and participant information.

Device Overview

AMPRO II (Figure 6) utilizes brushless DC motors for both the ankle and the knee joints. The battery for the device is housed in a waist belt worn by the participant. AMPRO II has a height of 380 mm and mass of 4.6 kg. AMPRO II controls the knee and the ankle simultaneously and utilizes motion sensors in the inertial measurement units (IMUs) to receive feedback from the intact leg and force sensors in the prosthetic to create the gait for the knee and ankle. For the knee, AMPRO II uses a human inspired control that attempts to create the optimal walking trajectory based the parameters of the user (Ames, 2014). The controller uses feedback from the IMUs to estimate step progression. For the ankle, AMPRO II implements a flat foot walking control, so the foot remains flat through the gait cycle.

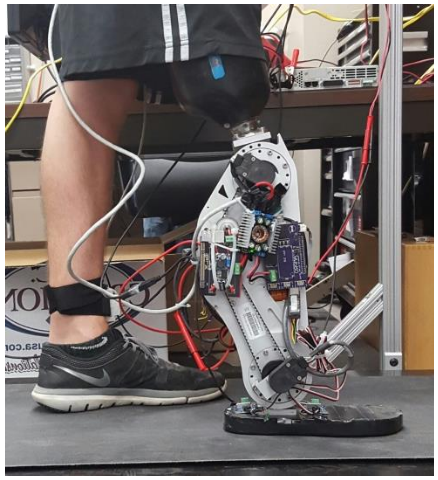

Figure 6: AMPRO II

Target User

The target user for this study was a male unilateral TF amputee. The participant was 21 years of age and 54.43 kg (without prosthetic). At the time of thee study the participant was using a Genium microprocessor controlled knee and a low profile Triton foot by Ottobock. The total mass of his prosthetic knee, ankle and socket is 4.08 kg. He has been using this device for approximately one year. Amputation was due to lower limb cancer. The participant was able to walk, stand and sit without getting short of breath (with his current prosthetic) for at least 5 minutes.

METHODS

In order to measure energy expenditure, kinetic and kinematic data and symmetry, the methods below were used.

Study Protocol

The study for AMPRO II required the participant to be involved in 17 sessions. In the first session, motion capture data were collected while the subject was using his microprocessor prosthetic knee and passive ankle. In the second session, the participant underwent peak VO2 (maximum volume of oxygen) testing in order to measure their energy expenditure during walking with their current prosthesis. The participant walked for 5 minutes at a self-selected walking speed. For the third through the fifteenth sessions, the participant practiced walking with AMPRO II. On the sixteenth visit, the participant underwent motion capture while using AMPRO II. In the last session, the participant underwent peak VO2 testing at a self-selected walking speed while using AMPRO II. This study protocol was approved by the TEXAS A&M IRB (IRB number: IRB2015-0607F).

Energy Expenditure

The energy expenditure was measured by using a device (TrueOne ® 2400, Parvo Medics, Sandy, UT) that measures the amount of oxygen uptake while walking. The apparatus used restricted breathing to just the mouth (Figure 7). The participant was asked to walk on a treadmill at a self-selected speed for 5 minutes. The participant was allowed to self-select the speed in order to measure the energy consumption during normal gait. The VO2 was measured at the beginning of the study with the participant's microprocessor controlled prosthetic knee and at the end of the study with AMPRO II.

Figure 7: Subject participating in peak VO2 test

Motion Capture Data

The kinematics of the participants walking was captured using a Qualisys motion capture system (Oqus 210c, Qualisys, Goteborg, Sweden) and the kinetic data was collected using Bertec force platform (FP9090-15-2000, Bertec Corporation, Columbus, OH). Markers were placed on the lateral joint rotation centers in the lower extremity, heel and tip of the foot, and upper back (Figure 8). Data were collected, while the participant walked along a set path with an embedded force plate, six times (three times with the right foot striking the force plate and three times with the left foot striking the force plate).

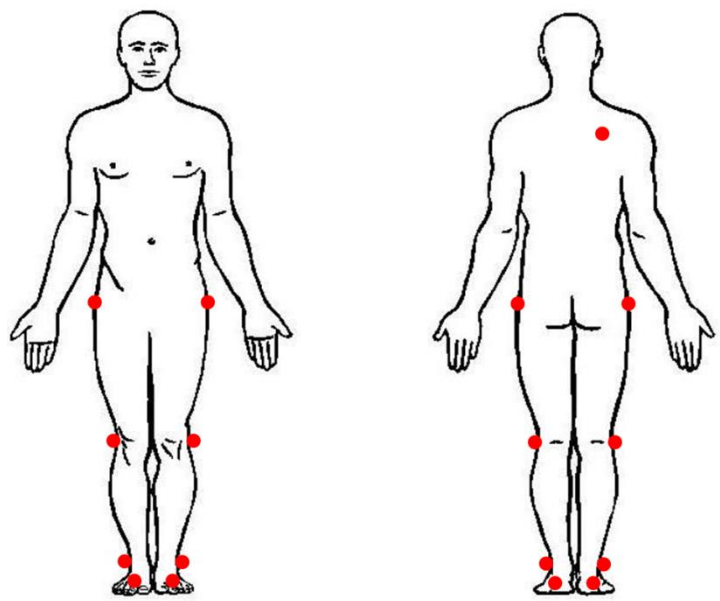

Figure 8: Motion capture marker placement

The motion capture and force plate data was filtered using a second-order Butterworth filter with a cut-off frequency of 10 Hz and 5Hz respectively. The gait speed (v) was measured by tracking the distance of the marker on the shoulder divided by the time walked time. The step length was measured by measuring the distance from heel strike to opposite heel strike. The segment mass (m), segment COM location, and segment moment of inertia (I) were estimated using COM, and radii of gyration data (r) with information from a cadaver study (David, 2009). Upon completion, the COM linear acceleration (a) and angular acceleration ($\ddot{\theta}$) were calculated by differentiating the position segment COM and segment angles, respectively, twice.

Kinematic Data

The kinematic data includes lower limb joint angles and angular velocities during flat ground walking for the amputee. To calculate the kinematic data, the filtered motion capture data were used to obtain segment angles as seen in Figure 9. To define the segment angles we used the endpoints of the trunk, thigh, shank and foot segments relative to a horizontal line.

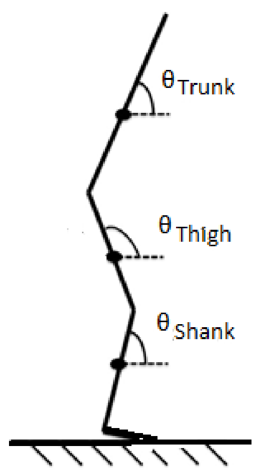

Figure 9: Definition of joint angles and segment angles (Pereira, et al., 2014)

Joint Angles

For this study, we are only looked at lower limb joint angles in the sagittal plane (hip flexion angle, knee flexion angle and ankle flexion angle). In order to calculate the joint angles we used the trunk, thigh, shank, and foot segment angles, seen in Figure 9. Equations 1-3 were used to calculate the joint angles. Zero degrees for the flexion angles represents when the participant is standing straight up.

$$\text{Hip Flexion Angle} = \theta_{thigh}-\theta_{trunk} \quad (1)$$

$$\text{Knee Flexion Angle} = \theta_{thigh}-\theta_{shank} \quad (2)$$

$$\text{Ankle Flexion Angle} = \theta_{foot}-\theta_{shank}-90° \quad (3)$$

These angles were obtained for the microprocessor knee and AMPRO II to assess the relative range of motion for the joint.

Kinetic Data

The force plate data was the sole kinetic input to the data analysis. Inverse dynamics was used to calculate the joint moments.

Joint Moments

A simplified link segment model was used in order to estimate the joint moments (Figure 10). Newton-Euler equations were used iteratively to solve the equations of motion. The following assumptions were made while solving the equations of motions:

- Each segment has a fixed mass located as a part mass at the segment's COM.
- The location of COM remains fixed within the segment.

- Joints are considered frictionless pin joints.
- The moment of inertia of each segment is fixed.

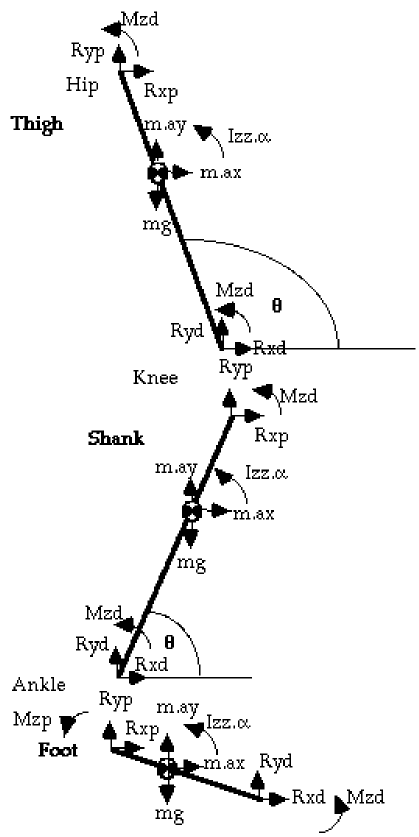

Figure 10: Link segment model for inverse dynamics (Kirtley)

The coordinate system was defined as y to the top, x is the direction of movement, and z is to the left. In order to solve for the moment about the COM for each segment, the joint reaction forces (*R*) was be solved by summing the forces (*F*), in the x and y directions (Equations 4 -7). This was done for the foot, shank and thigh segments respectively. The joint reaction force for the foot segment was calculated first, because R_{yd}, the force at the distal end (d), was known. This gave us only one joint reaction force at the distal end to solve. For the foot, R_{yd} was considered the GRF in the y direction and acts at the COP. This was repeated for the shank and thigh segments as well. Both a_x and $a_y$are the COM accelerations the x and y directions, respectively.

$$\sum F_x = R_{xp} + R_{xd} = ma_x \tag{4}$$

$$\rightarrow \; R_{xp} = ma_x - R_{xd} \tag{5}$$

$$\sum F_{y=} \; R_{yp} + R_{yd} - mg = ma_y \tag{6}$$

$$\rightarrow \; R_{yp} = m_{ay} + mg - R_{yd} \tag{7}$$

After this, the joint moments were calculated by summing the moments (M) about the COM for each segment (Equations 8 and 9). I is the moment of inertia for the segment and $\ddot{\theta}$ is the segment angular acceleration.

$$\begin{aligned}\sum M_{COM} = M_{zd} + M_{zp} + R_{xp}(y_p - y_{COM}) - R_{yp}(x_{COM} - x_p) \\ - Rx_d(y_{COM} - y_d) + R_{yd}(x_d - x_{COM}) = I\ddot{\theta}\end{aligned} \tag{8}$$

$$\begin{aligned}\rightarrow \; M_{zp} = I\ddot{\theta} - M_{zd} - R_{xp}(y_p - y_{COM}) + R_{yp}(x_{COM} - x_p) \\ + Rx_d(y_{com} - y_d) - R_{yd}(x_d - x_{COM})\end{aligned} \tag{9}$$

The hip and knee joint moments were calculated so a positive moment indicates flexion and a negative moment indicates extension. For the ankle, a negative moment indicates dorsiflexion and a positive moment indicates plantarflexion.

Symmetry Index

The joint angles, and joint moments for the right and left legs were compared using the symmetry index (Equation 6). These were analyzed for one-step for both legs through stance and swing phase for a TF amputee using a microprocessor knee and AMPRO II.

There are several methods that can be used to calculate the symmetry index (Nigg, Vienneau, Maurer, & Nigg, 2013). We used a symmetry index developed by Robinson et al., which is one of the most commonly used methods (Nigg, Vienneau, Maurer, & Nigg, 2013). Gait symmetry is quantified at discrete time points using the symmetry index (SI). The closer the value is to zero, the more symmetric the gait is. The symmetry index is described as follows:

$$\mathrm{SI}=\frac{|x_I - x_P|}{\left(\frac{1}{2}\right)(x_I + x_P)} \times 100 \tag{10}$$

where *SI* is the symmetry index, x_I is the variable recorded for the intact leg and x_P is the variable recorded for the prosthetic leg.

RESULTS AND DISCUSSION

In order to assess the effectiveness of using AMPRO II, the kinetic and kinematic differences have been investigated while the participant was using AMPRO II and a Genium microprocessor knee and Triton low profile foot were investigated. The effects on the intact legs while using each device were observed as well for both the kinetics and kinematics. The average speed for the motion capture walking trial with the Genium microprocessor knee was 4.6 km/h. The average speed for the motion capture walking trials with AMPRO II was 3.13 km/h. The average step length while using the Genium microprocessor knee was 0.78 m. The average step length while using AMPRO II was 0.55 m. We attempted to match the gait speed and step length. However, the participant could not do so comfortably.

Energy Expenditure

While participating in the peak VO2 test the subject self-selected a walking speed of 0.58 meters per second. There was a significant difference between peak VO2 results for the AMPRO II and microprocessor-controlled knees. The peak VO2 for the session using the Genium microprocessor controlled knee was 14.2 ml/kg/min, while the peak VO2 for the session using AMPRO II was 17.2 ml/kg/min. The energy expenditure was greater for AMPRO II more than likely due the heavier weight (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011). The effects of the weight could possibly be reduced if active plantarflexion and dorsiflexion was included. AMPRO II has flat foot walking implemented, which does not assist in push-off.

Ground Reaction Forces

The GRF values while using AMPRO II and the microprocessor knee were similar to results in a study by Nolan et al. (Nolan, Dudzinski, Lees, Lake, & Wychowanski, 2003) for both devices. The first of the two peaks indicates load acceptance and the second peak indicates push-off (Nolan, Dudzinski, Lees, Lake, & Wychowanski, 2003). The SI for the GRF was two points lower for the microprocessor knee (Table 1) than for AMPRO II. However, the peak forces were closer to the intact leg when using AMPRO II (Figure 12). For AMPRO II, there was a shift in the GRF (Figure 12). This shift was not seen in the microprocessor leg (Figure 11). The shift in the GRF plot for AMPRO II indicates that there was a delayed load acceptance. The delayed load acceptance could be due to the flat foot controller. However, even with the flat foot control imposed on AMPRO II the peak force at push off was similar to the intact leg, despite not having active plantarflexion and dorsiflexion. Possibly, the powered knee was giving assistance during push-off causing to the GRF to be more symmetric during terminal stance. Even though using the microprocessor knee yielded a more symmetric GRF, the max GRF values were closer using AMPRO II. The smaller difference in max GRF is a benefit for amputees, since they tend to have greater loading on their intact leg.

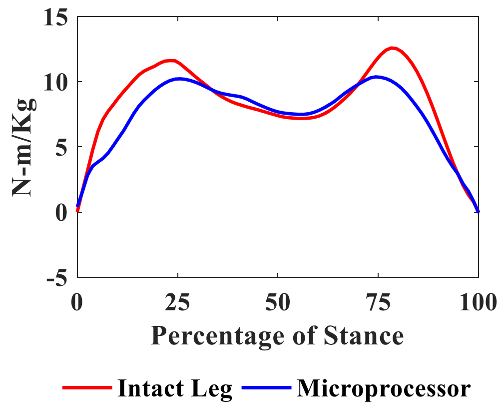

Figure 11: GRF for intact leg and microprocessor knee

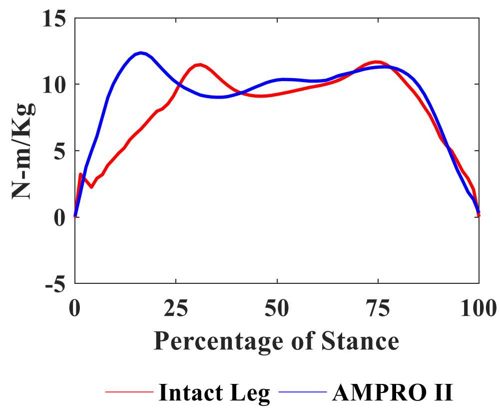

Figure 12: GRF for intact leg and AMPRO II

Table 1: Max GRF and GRF SI while using microprocessor knee and AMPRO II

	GRF SI	Max GRF Difference (N/kg)
Microprocessor	11.51	1.91
AMPRO II	17.13	0.6

Hip Kinematics and Kinetics

When observing the average SI for the hip over a gait cycle for the microprocessor knee and AMPRO II (Table 3), the hip angle had approximately a 22 percent increase in SI when using AMPRO II. This possibly due to the heavier weight of the device compared to the microprocessor knee. The intact leg while using the microprocessor knee had a greater hip flexion angle at the beginning of stance and the end of the swing phase (Figure 13). Increased hip flexion is common in transfemoral amputee gait (Kishner & Laborde, 2015). While using AMPRO II there is a decreased flexion angle during stance (Figure 17).

There was an increased hip moment in the residual limb while using the microprocessor knee and AMPRO II compared to the intact leg (Figure 15 and Figure 17). The hip joint moment indicated a greater hip flexion moment in the residual limb during the beginning of stance phase when using AMPRO II and the microprocessor knee (Figure 15 and Figure 17). The increased hip moment is due to the soft tissue in the residual limb not being design for high loads (Gailey, Allen, Castles, Kucharick, & Roeder, 2008). There was also a greater hip flexion moment in the intact leg while using the microprocessor knee (Figure 18). This could be a result of speed differences and not related to the differences in prosthesis (Kwon, Son, & Lee, 2015). The hip moment SI, while using AMPRO II, was lower than when using the microprocessor knee (Table 2). At the last 20 percent of the stance phase, the hip moments were almost perfectly symmetric (Figure 17). This is more than likely due to the powered knee offering assistance during push-off.

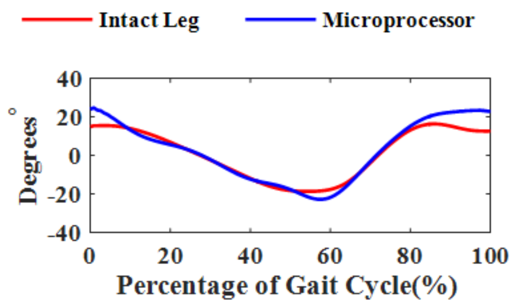

Figure 13: Hip angles while using microprocessor knee

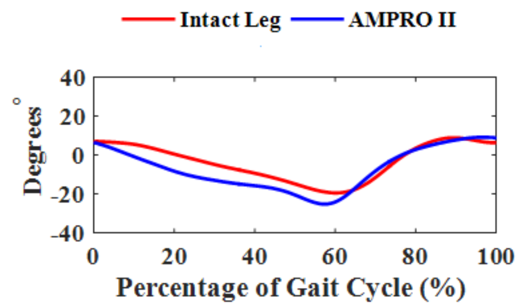

Figure 14: Hip angles while using AMPRO II

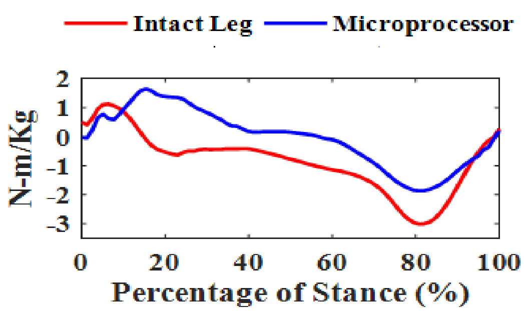

Figure 15: Hip moments while using microprocessor knee

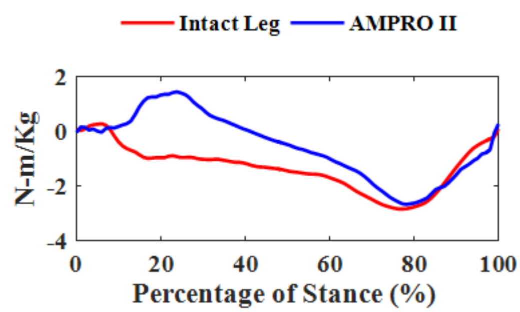

Figure 16: Hip moments while using AMPRO II

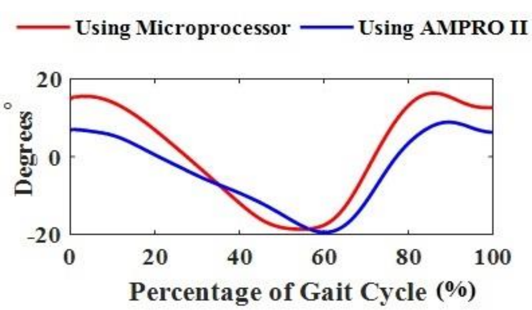

Figure 17: Hip angles of intact legs while using microprocessor knee and AMPRO II

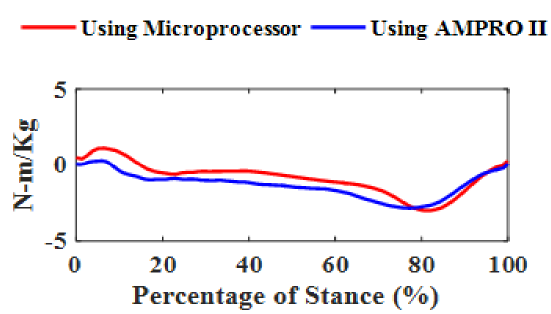

Figure 18: Hip moments of intact legs while using microprocessor knee and AMPRO II

Table 2: Average hip SI while using microprocessor knee and AMPRO II

	Hip Angle SI	Hip Moment SI
Microprocessor	40.10	76.66
AMPRO II	51.10	68.49

Knee Kinematics and Kinetics

There was an increased knee flexion angle in the swing phase while using AMPRO II (Figure 20) in the intact leg when compared to the residual limb. This was also seen while using the microprocessor knee as well (Figure 19). Knee flexion was increased during the stance phase for AMPRO II (Figure 20). Increasing knee flexion during stance can lead to more efficient walking (Kishner & Laborde, 2015). This knee flexion also occurs at the same point of the gait cycle as the intact leg while using AMPRO II. While using the microprocessor knee, the knee flexion of the prosthetic leg during stance was delayed (Figure 19). However, while using AMPRO II, there was not adequate knee extension during terminal stance for the intact leg or prosthetic leg while AMPRO II (Figure 20). This leads to an abnormal walking gait. When comparing the intact legs while using AMPRO II and the microprocessor knee, the maximum flexion angles were the approximately equal (Figure 23). The main difference was the lack of extension in the knee angle before swing phase. Despite the lack of knee extension during stance, there was still a lower SI for the knee angle (Table 3).

While using the microprocessor knee and AMPRO II the knee moments were lower than the intact leg (Figure 21 and Figure 22). This usually happens with TF amputee gait due to the stiffness of knees in order to maintain stability. However, while using AMPRO II, at the end of the stance phase the knee moment values were closer to the moment in the intact knee (Figure 24). The overall magnitude of the moment for the intact knee was higher while using AMPRO II as well (Figure 21). This could have been due to the weight of AMPRO II being greater than the weight of the microprocessor knee. The knee moments, while using AMPRO II, had about a 30 percent decrease in the SI (Table 3), indicating that the knee moments were more symmetric while using AMPRO II. This was due the powered knee helping to propel the participant forward at the end of stance phase.

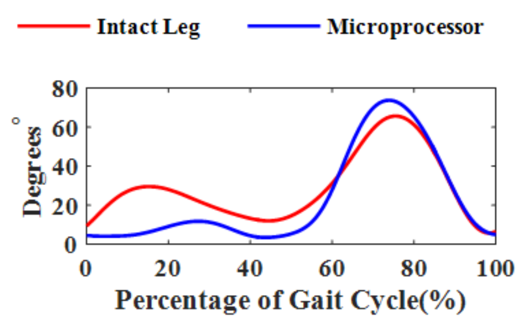

Figure 19: Knee angles while using microprocessor knee

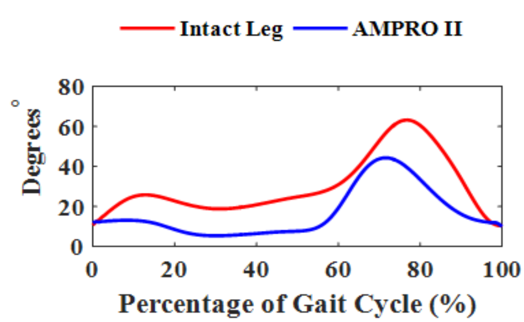

Figure 20: Knee angles while using AMPRO II

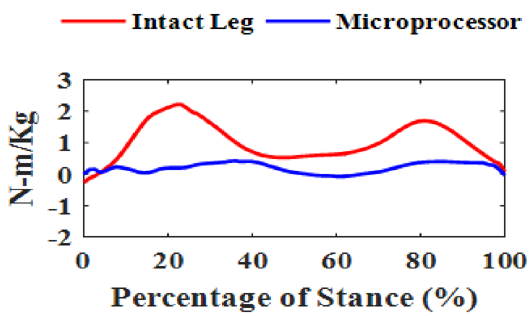

Figure 21: Knee moments while using microprocessor knee

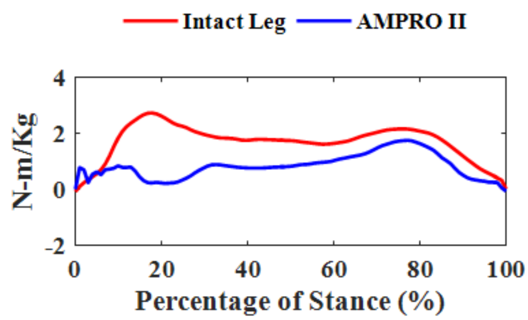

Figure 22: Knee moments while using AMPRO II

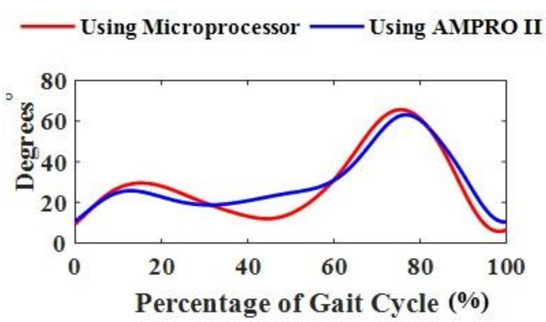

Figure 23: Knee angles of intact legs while using microprocessor knee and AMPRO II

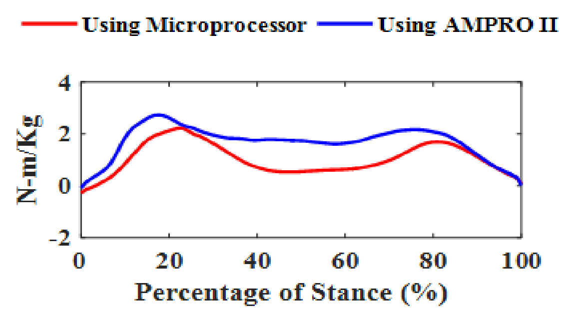

Figure 24: Knee moments of intact legs while using microprocessor knee and AMPRO II

Table 3: Average knee SI for microprocessor knee and AMPRO II

	Knee Angle SI	Knee Moment SI
Microprocessor	58.42	93.57
AMPRO II	54.69	65.23

Ankle Kinematics and Kinetics

Even though the ankle in AMPRO II utilized flat foot walking, the SI for the ankle angle was slightly lower than the SI for the microprocessor knee (approximately 8 percent decrease) (Table 4). This is partly due to the flexion about the ankle that is allowed during the stance phase. Even though there is no active plantarflexion or dorsiflexion due to the flat foot constraint, there is still angle variability during stance. This is not seen in the passive ankle angle (Figure 25). The passive ankle is relatively stiff during waking and allows for little to plantarflexion or dorsiflexion angle.

The greater ankle angle symmetry did not translate to greater ankle moment symmetry, while using AMPRO II. AMPRO II's ankle had a very low moment in the prosthetic ankle, which was very asymmetric (Figure 28). The average SI for the microprocessor ankle moment at the ankle was less than half of the ankle moment SI while using AMPRO II (Table 4). The lack of active plantar and dorsiflexion, due to the flat foot control, was more than likely the reason there was less symmetry in the ankle joint moment for AMPRO II. The human foot does not stay flat during stance during normal flat ground walking (Figure 1). Plantarflexion and dorsiflexion help the gait to be more efficient, and aides during push off and weight acceptance (Kishner & Laborde, 2015).

Using the passive foot allowed for a much more symmetric ankle moment, despite having an asymmetric ankle angle (Figure 27). This leads a possible reason for the lack of kinetic symmetry in the ankle, while using AMPRO II, could be due to the foot not having a shoe, padding or springs for shock absorption or to aid during push off.

It was seen in one study, that shoes have a major effect on the ankle and hip flexion moments and GRF (Shakoor, et al., 2010). During the beginning of stance phase for AMPRO II, there were variations in magnitude and a shift in load acceptance at the beginning of stance in the GRF (Figure 12). This could have also been due to lack of shoe, and shock absorption, as well as a result of the foot flat condition.

When comparing the intact legs while using both devices, the intact leg while using AMPRO II has an increased dorsiflexion moment compared to the microprocessor (Figure 30). The peak plantarflexion angle was also reduced. This could have been due to the reduced speed and not due to the differences in prosthesis (Kwon, Son, & Lee, 2015). Increasing dorsiflexion helps to increase stability during the being of stance phase. Increased dorsiflexion often happens when there is reduced knee extension, which was observed in the intact leg while using AMPRO II (Figure 20**)**.

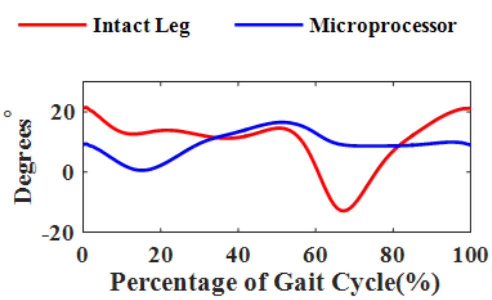

Figure 25: Ankle angles while using microprocessor knee

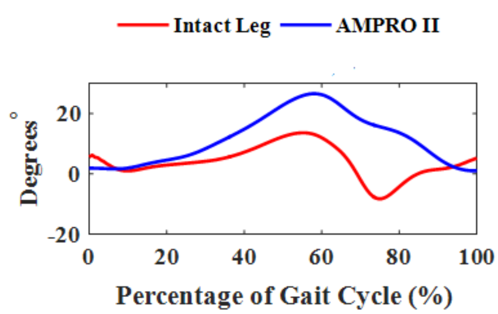

Figure 26: Ankle angles while using AMPRO II

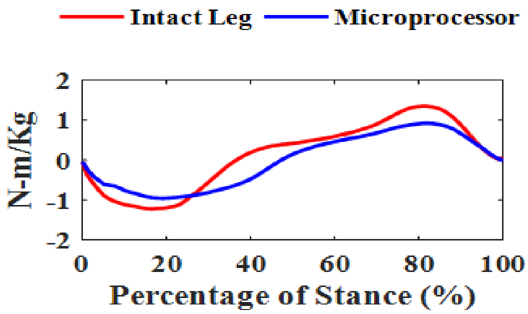

Figure 27: Ankle moments while using microprocessor knee

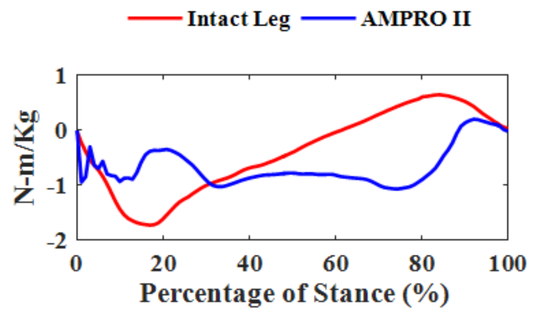

Figure 28: Ankle moments while using AMPRO II

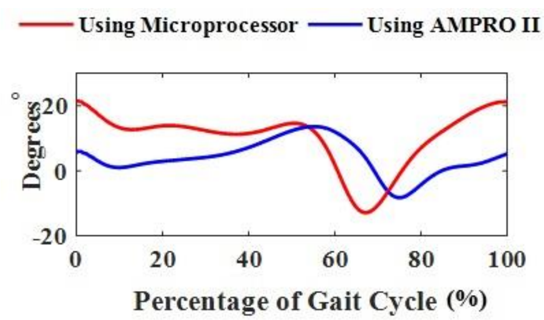

Figure 29: Ankle angles of intact leg while using microprocessor knee and AMPRO II

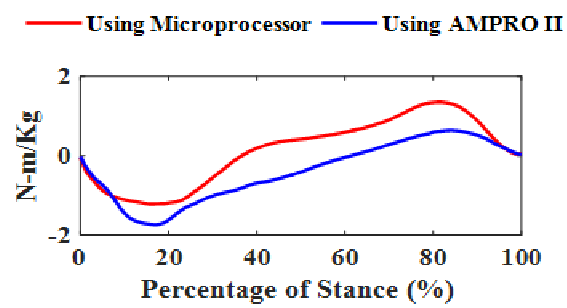

Figure 30: Ankle moments of intact leg while using microprocessor knee and AMPRO II

Table 4: Average ankle SI while using the microprocessor knee and AMPRO II

	Ankle Angle SI	Ankle Moment SI
Microprocessor	64.89	38.66
AMPRO II	59.98	71.13

Participant Feedback

On the post visit survey, the participant was asked how AMPRO II felt in comparison to his microprocessor knee; he said similar. When asked if he experienced any soreness or discomfort while using AMPRO II he said he experienced very mild discomfort. He mentioned that the device being powered was very helpful. Despite the symmetry differences and weight differences in AMPRO II and the microprocessor knee, the participant still felt that the device was similar to his microprocessor device.

Summary of Results

Although the gait was more symmetric for the knee and ankle angles (Table 3 and Table 4), and knee and hip moments (Table 2 and Table 3) while using AMPRO II, there were still abnormalities in gait: increased hip flexion moment (Figure 14), negligible plantar flexion (Figure 26), and prolonged knee extension (Figure 20) AMPRO II performed better in gait symmetry for the hip (Table 2) and the knee moments (Table 3) compared to the microprocessor knee. AMPRO II had a significantly better symmetry index for the knee moment (30% decrease in SI) due to the knee actually being powered. AMPRO II had a very asymmetric ankle moment. It had increased energy expenditure a low self -selected speed.

For the kinematics, the ankle and knee angles have the highest SI (i.e., greatest asymmetry) for both devices. Other studies have indicated that the knee and ankle that have the greatest SI for kinematics (Kaufman, Frittol, & Frigo, 2012). This varies greatly between users. This is not surprising, because the knee and ankle of the prostheses greatly differ from the intact leg.

Although AMPRO II had a greater SI for the ankle moment, hip angle, and energy expenditure it performed better in most areas. The improvements in kinematics were not as great as the improvements in symmetry for the kinetics. This indicates that AMPRO II shows promise if improvements are made, due to the fact that most long-term problems for transfemoral are caused by increased moments at the intact joints.

CONCLUSIONS

General benefits and potential changes from this study can help improve future work. This is true despite only using one participant and having differences in speed and step length (for motion capture) Analyzing the results shows us that there are some benefits using AMPRO II, despite using the flat foot condition at the ankle. AMPRO II helped in even distribution of loads between the intact and prosthetic leg. It also yielded more symmetric moments for the hip and knee during stance. This shows that the powered knee does help with symmetry at the end of stance. While AMPRO II does not cause perfect symmetry, it does reduce asymmetry in joint moments when compared to the microprocessor knee.

Although there are benefits, we also observed some areas that could use improvement. Utilizing a different foot and multi-contact control, would aid in push off. Lightening the device will more than likely lead to increased kinematic and kinetic symmetry and decreased energy expenditure. Significantly improved gait symmetry is expected with multi-contact walking and a lighter device. A lighter powered prosthesis will likely yield more symmetric gaits (Martinez-Villalpando, Mooney, Elliott, & Herr, Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee, 2011).

Another area of focus in future studies could be the practice and preparation time. The participant only had 13 one to two hour sessions to practice with the device. It is also possible that with longer or continuous training sessions the user would grow more accustomed to the device and have more symmetric walking (Sjödahl, Jarnlo, Söderberg, & Persson, 2002). This study has shown us what is not optimized in AMPRO II, and what needs to be improved. It has also shown us some features that are beneficial to TF amputees. There will be a new powered prosthetic designed that will address the issues that we have found in AMPRO II.

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