Variable Damping Controller for a Prosthetic Knee during Swing Extension

by

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Submitted to the Department of Mechanical Engineering in partial fulfillment of the requirements for the degree of Bachelor of Science at the MASSACHUSETTS INSTITUTE OF TECHNOLOGY

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Abstract

Transfemoral amputees exhibit both increased metabolic consumption and gait asymmetry during level ground walking. A variable damping control strategy has been developed for swing extension in order to improve gait symmetry and reduce energy expenditure during level ground walking. Preliminary biomechanical studies suggest that the knee utilizes a variable damping control during swing extension. This thesis proposes a biologically inspired variable damping control strategy which can be simplified into a piecewise function with respect to the knee angle. The variable damping profile of the knee during swing extension has been modeled as an initial linear increase with respect to knee angle followed by a quadratic increase at the end of swing. A damping controller based on this proposed piecewise function has been implemented in a biomimetic, active, knee prosthesis (AAAKP) developed at MIT’s Biomechatronics Lab. Preliminary studies on a unilateral, transfemoral amputee have shown that the AAAKP with the proposed damping control strategy is able to more closely emulate the damping profile of the unaffected leg, when compared to a conventional knee prosthesis (Otto Bock C-Leg®). This Initial study suggests that the proposed variable damping strategy for swing extension is able to more accurately emulate the joint mechanics of the unaffected knee. This work is intended to improve prosthetic knee behavior in order to reduce metabolic consumption and improve gait symmetry in transfemoral amputees during level ground walking.

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Chapter 1

Introduction

The consequences and effects of amputation differ for each patient, but are related to the location of the amputation. With regards to lower limb amputees, as the height of the amputation increases so do the difficulties. Lower limb amputees experience exhibit pathological gaits which are often asymmetric and less stable than non-amputees [1]. Pathological gaits of transfemoral amputees result in an increased oxygen consumption of nearly 50%, when compared to non-amputees [2]. This increased oxygen consumption represents the increased metabolic consumption experienced by transfemoral amputees while walking.

The current prostheses on the market range from simple four-bar linkages to motorized electro-mechanical devices. Although variable damper knees, such as the Otto Bock C-Leg® and the Ossur RHEO®, shown in Figure 1-1, have reduced the metabolic consumption of transfemoral amputees during level ground walking as compared to a mechanically passive, hydraulic-based Mauch knee prosthesis [3], neither one of these prostheses are able to participate in net-positive power movements, since they are both quasi-passive devices. MIT’s Biomechatronics Lab is developing an active agonist-antagonist knee prosthesis (AAAKP) in order to further reduce the metabolic costs associated with walking and net-positive power movements [4, 5, 6]. The AAAKP uses two series elastic actuators in order to balance the transfer of energy between flexion and extension; the design and implementation of the AAAKP is further discussed in Section 2.2.
The AAAKP is being developed in order to explore the advantages and consequences of having an active knee prosthesis, capable of providing both positive and negative power at the knee joint. During level ground walking, the knee is a net negative power joint, however it does require positive power during certain stages of the gait cycle. All passive and quasi-passive prosthesis are only able to provide dissipative torques, which precludes positive-power movements resulting in asymmetric gaits and increased metabolic consumption during level ground walking [3]. The AAAKP utilizes its dual actuators to provide the amputee with positive power during stance phase, allowing the knee to more closely follow the kinetics of an unaffected knee. Along with more accurately replicating the knee trajectory during level ground walking an active knee also makes it possible for the prosthesis to participate in movements which require substantial positive torques such as standing up, stair climbing and running. The AAAKP can also provide variable impedance control for movements that require greater dissipative torques, such as sitting down. Unlike completely passive prostheses, quasi-passive and active prostheses are often capable of replicating variable damping control. Variable damping allows the knee to emulate the kinematics of the knee during certain stages of the gait cycles, including swing extension. As an
active knee, the AAAKP is also able to implement a variable damping control [7].

The major advantage of the AAAKP over quasi-passive prostheses, is the AAAKP’s ability to participate in net-positive power movements. However, in order to be an effective prosthesis, the AAAKP must also allow a user to efficiently walk on level ground, a mostly negative power activity for the knee. An important phase of the level ground walking gait is swing extension, the period between maximum swing flexion and heel strike. Swing extension is important for both gait stability and symmetry as discussed in Section 2.1.

The focus of this research is on swing extension and the control strategy implemented in a prosthetic knee to effectively emulate the kinematics and kinetics experienced during swing extension. Current prosthetic knees on the market approach swing extension in different manners. Due to the inherent fluid dynamics, hydraulic knees exhibit a resistive torque during swing extension that is proportional to the velocity squared. Quasi-passive prostheses, such as the RHEO®, are able to implement more sophisticated resistive torques. One control strategy developed at MIT implements a virtual damper with a constant high damping coefficient just at the end of swing extension [8].

In order to implement an effective swing extension in the AAAKP, a preliminary biomechanical analysis, discussed in Section 3.1, was performed on two able-bodied subjects. A Vicon motion capture system was used to determine both the knee kinematics and kinetics during level ground walking. The results of this preliminary study were used to develop a piecewise damping function with respect to knee angle. The knee initially experiences a linear increasing damping trajectory until approximately 15-20 degrees, where the damping at the knee begins to increase quadratically. A parameterized function was developed in order to customize the damping profile for each subject. Another preliminary study has also been completed with the Vicon motion capture system, examining the resulting kinematics and kinetics while using this variable damping control strategy.

Current quasi-passive knees, such as the Otto-Bock C-Leg® and the Ossur RHEO®, implement variable damping control during swing extension. Although these pros-
theses are able to effectively emulate the knee kinematics during swing extension, the light weights of these devices results in lower dissipative torques at the knee. With a weight closer to that of a biological lower leg, the AAAKP produces both damping and torque profiles at the knee which more closely emulate those of a biological knee. Section 4 describes the effects of the AAAKP on both the users affected and unaffected sides.
Chapter 2

Background

The mechanics of walking can be analyzed as a repetitive series of movements known as the gait cycle, which follows the trajectory of one leg from heel strike to heel strike. Swing extension comprises nearly a quarter of the gait cycle and is critical for stability and symmetry. When designing prostheses it is necessary to design devices which are capable of mimicking each stage of the gait cycle. The AAAKP has been designed to replicate an ideal model, which efficiently replicates the gait cycle during level-ground walking. The Agonist-Antagonist Active Knee Prosthesis emulates the gait cycle by utilizing dual series elastic actuators to provide active torque to the knee joint during both flexion and extension.

2.1 Gait Cycle

The human gait cycle is a complex process involving motions in the sagittal, coronal, and transverse planes. Each joint has multiple degrees of freedom including linear motions, rotations, and complex combinations of the two. A simple model of the knee represents all of these motions as just a simple pin joint in the sagittal plane, which divides the body into the left and right sides. This simplification is the basis for the AAAKP design, which controls motion in the sagittal plane. A graphical representation of the gait cycle is shown in Figure 2-1 from the point of view of the sagittal plane. The gait cycle traditionally begins with the heel strike of one foot. As
Figure 2-1: One gait cycle from the sagittal plane. The cycle begins and ends with the heel strike of one foot. Specifically, the knee begins in stance flexion, moves to stance extension, swing flexion and finally ending with swing extension, after which the heel strikes and the cycle begins again [9].

The focus of this research will be on swing extension and the control strategy implemented in a prosthetic knee to effectively emulate the kinematics and kinetics experienced during swing extension. Swing extension accounts for nearly a quarter of the gait cycle and effects the stability of the gait. An ineffective swing extension may lead to discomfort at the socket or tripping. If the knee swings forward too fast, the amputee may suffer from terminal impact and an asymmetric gait, but if the knee...
does not extend fast enough during swing, the toe of the prosthetic foot is in danger of tripping, causing the amputee to stumble or fall.

The swing extension of one side also has an effect on the opposite side. As the affected side is in swing extension, Figure 2-1 shows that the unaffected side will be in stance extension. Improving the swing extension of the prosthesis may not only benefit the affected side but possibly also the unaffected side in stance. One pathological gait that transfemoral amputees often exhibit is vaulting, in which the unaffected ankle over plantar-flexes in order to ensure the affected foot will not trip [10]. While improving swing extension of the prosthesis may improve the gait of the affected side, its effects on the unaffected side will also be investigated. Examining the kinematics and kinetics of the unaffected side while the affected side in swing extension, will give insight into the effects of the variable damping control on the unaffected side.

### 2.2 Agonist-Antagonist Active Knee Prosthesis

The Agonist Antagonist Active Knee Prosthesis (AAAKP) is a powered knee prosthesis being developed at MIT’s Biomechatronics Lab. The AAAKP is designed as a biomimetic device, which uses dual series elastic actuators to provide both positive and negative torques at the knee joint [4, 5]. It uses an agonist-antagonist design with series elastic elements, similar to the agonist-antagonist muscles found in the leg along with the elastic tendons. The AAAKP uses two separate series elastic linear actuators; one to extend and the other to flex the knee. The two separate SEAs differ by both their motors and elastic stiffnesses, which have been optimized to increase the efficiency of the device [7]. The AAAKP is shown in Figure 2-2.

The controller hardware uses an Atmega 328 to simultaneous run two 600 Hz motor controllers, rated at 24 volts and 15 amps continuous [7]. The controller relies on three encoders to determine the relative positions of the three critical elements: the knee carriage, which determines knee angle, the flexion actuator, and the extension actuator. Knowing the positions of these three elements along with the stiffness of the
Figure 2-2: The Agonist-Antagonist Active Knee Prosthesis (AAAKP) is an energy efficient knee prosthesis being developed at MIT's Biomechatronics Lab. The AAAKP uses two series elastic actuators to emulate the agonist-antagonist muscle structure around the knee joint.

series elastic elements allow for the mechanical torque seen by the knee to be known at all times. The knee carriages position is tracked through an optical, linear encoder connected directly to the knee carriage, and each actuator has an optical, rotary encoder on each motor. The controllers run according to a 600 Hz state machine running on a custom operating system [7]. There is also a USB port which allows for the sensors to be constantly read and recorded along with an on-board SD card, thus providing all of the necessary measurements for positions, currents, and commanded PWM.
Chapter 3

Variable Damping Profile

In order to determine the correct control strategy for a prosthetic knee during swing extension, a preliminary biomechanical analysis was performed on two able-bodied subjects. The study used a motion capture system to record both the kinematics and joint torques of the subjects as they walked on level ground. The results of this study were used to develop a variable damping control strategy for the knee during swing extension. The damping profile was characterized as a piecewise function with respect to knee angle. This parameterized damping profile has been implemented and tested with the AAAKP.

3.1 Preliminary Biomechanical Study

In order to explore the possible control strategies during swing extension, a biomechanical study was performed on two able-bodied subjects. A motion capture system (Vicon, Oxford Metrics) along with floor-embedded force plates were used to record the kinematics of two able-bodied subjects walking at different speeds. An inverse dynamics simulation was also performed in order to determine the sagittal plane moments at the hip, knee and ankle joints. The inverse dynamic simulation accounts for the kinematics of the body along with the force vectors recorded by the force plates in order to determine the joint moments. The averaged results of the kinematics and inverse dynamic simulation for the knee during swing extension are shown in figure
Figure 3-1: A motion capture system and floor-embedded force plates were used in a biomechanical analysis to determine the kinematics of the knee during swing extension of level ground walking. The knee angle and velocity are directly determined through the kinematics, while the knee moment is inferred through an inverse dynamics simulation. Note: a positive knee moment acts in flexion.
The biomechanics of the unaffected subjects suggest similar kinematics with a slight variation in knee moment during swing extension. Subject 2 experiences a higher peak knee velocity during swing and consequently must apply a higher resistive knee moment to stop the knee swing before heel strike. The data from the two able-bodied subjects were then used to develop an appropriate control strategy for the AAAKP during swing extension.

The architecture of the AAAKP was taken into consideration when determining the appropriate controls during swing extension. The series elastic actuators can provide both positive and negative moments about the knee, and through a high-gain controller the actuators can also mimic a position controller. As can be seen in Figure 3-1, the knee provides an almost exclusively positive, resistive moment during swing extension. This negative power portion of the gait cycle can be effectively modeled as a variable damper, as shown by the biomechanical analysis. Therefore, using Equation 3.1, the effective damping coefficient of the knee throughout swing extension was calculated.

\[
B_{\text{knee}} = \frac{\tau_{\text{knee}}}{\dot{\Theta}_{\text{knee}}} \tag{3.1}
\]

As seen in Equation 3.1, the effective damping coefficient is the ratio between the knee moment and knee velocity. The results from these preliminary able-bodies studies suggest that the damping profiles of different subjects are similar and follow a certain trajectory shape, which are shown in Figure 3-2. Compared to a static position based controller, a variable damping controller provides a more flexible control strategy that can accommodate a wider range of swing speeds. Another benefit to using a virtual damper is its stability; since a damper can only provide negative power, the system remains predictable and safe. If the knee does not correctly predict its current state during torque control, the knee may become unstable, but a variable damping controller only dissipates energy. Figure 3-2 shows the damping coefficient for each subject during swing extension.
Figure 3-2: The effective damping coefficient of the knee during swing extension is calculated as the ratio between the knee moment and knee rotational velocity.

### 3.2 Variable Damping Implementation

The controls for the AAAKP are managed by a finite state machine. An on-board control board handles power management, signals and logic in order to control the state machine of the knee. Each phase of the gait cycle is comprised of at least one state, including swing extension. There are three fundamental steps required to implement a variable damping control: determine the correct damping coefficient for the current state of the knee, calculate the required torque to model the virtual damper, and finally, impart the required torque on the knee joint.

The damping trajectory was determined by emulating the measured damping trajectories of able-bodied subjects, as discussed in Section 3.1. The results of the preliminary study suggest that the damping trajectory, as a function of knee angle, during swing extension is qualitatively consistent among different subjects and can be divided into two piecewise functions: an initial, linearly increasing function, fol-
lowed by a quadratically increasing function at the end of swing. Figure 3-3 shows the resulting damping profiles of the able-bodied subjects along with the proposed piecewise damping profile.

![Figure 3-3: A comparison of two damping profiles from able bodied subjects along with a proposed damping profile for the AAAKP, which is comprised of an initial, linear function and followed by a quadratic function, both with respect to the knee angle.](image)

The preliminary study showed that while the damping profiles are qualitatively similar, they are not quantitatively identical for everyone. Therefore, it is necessary to implement a parameterized control strategy that can be adapted for each user. The linear and quadratic damping trajectory of the knee can be described by four variables: $B_{beg}$, $\theta_{trans}$, $B_{trans}$ and $B_{end}$ as shown in Figure 3-4. Parameterizing the damping profile allows a user to efficiently tune the behavior of swing extension while maintaining the qualitative shape observed in able-bodied subjects.
Figure 3-4: The parameterized, proposed damping profile for the AAARKP. \( B_{\text{beg}} \) determines the initial damping coefficient at the beginning of swing extension. \( \theta_{\text{trans}} \) and \( B_{\text{trans}} \) determine the angle at which the function changes from linear to quadratic, and also the damping magnitude at which this transition occurs. Finally, \( B_{\text{end}} \) determines of the steepness of the quadratic function.

The linear portion of the damping profile is defined by the beginning damping magnitude \( B_{\text{beg}} \) at a fixed initial angle, and also both the angle \( \theta_{\text{trans}} \) and magnitude \( B_{\text{trans}} \) at which the damping profile transitions between linear and into quadratic. This transition angle and magnitude is also used to determine the vertex of the quadratic function. Finally, the steepness of the quadratic function is defined by \( B_{\text{end}} \).

Equation 3.2 shows the piecewise function used by the state machine to determine the correct damping coefficient at each angle of swing extension. The smoothest transition between the linear and quadratic zones of damping would involve transitioning at the point of the parabola where the derivative is equal to the slope of the linear trajectory. However, the transition is implemented at the vertex of the parabola in order to save computational time. The friction and lag in the system, along with the gradual slope of the linear trajectory mitigates this suboptimal transition.
\[ B_{\text{knee}}(\Theta_{\text{knee}}) = \begin{cases} \frac{B_{\text{beg}} - B_{\text{trans}}}{\Theta_{\text{trans}}} (\Theta_{\text{knee}} - \Theta_{\text{trans}}) + B_{\text{trans}} & : \Theta_{\text{knee}} > \Theta_{\text{trans}} \\ \frac{B_{\text{end}} - B_{\text{trans}}}{(5-\Theta_{\text{trans}})^2} (\Theta_{\text{knee}} - \Theta_{\text{trans}})^2 + B_{\text{trans}} & : \Theta_{\text{knee}} \leq \Theta_{\text{trans}} \end{cases} \] (3.2)

The knee is initially loaded with a set of average parameters, but it is necessary to tune these parameters in order to match the gait of the subject. Subjects are generally sensitive to two factors, the speed at which the knee initially swings and also the end of swing extension, when the knee comes to rest in preparation for heel strike. The initial swing speed of the knee is critical for the stability and comfort of the subject. When the damping is too large, the knee is slow and the subject feels as if he must wait for the knee. Conversely, when the knee swings too fast, the subject experiences an uncomfortable terminal impact. These parameters are tuned and set during the initial fitting session. The inputs of a trained prosthetist along with the comments of the subject are translated into parameter changes. This process is repeated until the subject feels comfortable and the swing is natural. Once the tuning is complete, the state machine has a complete set of parameters to generate the correct damping profile.

In order to calculate the appropriate force to emulate a variable damper, the state machine must also know the angular velocity of the knee. The position of the knee is measure via an optical encoder, which is then differentiated to determine velocity. Due to the resolution of the encoder, a moving, weighted average filter is applied to the velocity in order to reduce noise. The high sampling rate of the controller (600 Hz) allows for the velocity to be greatly filtered without incurring problematic lag (0.03 seconds). Having determined the required damping coefficient and knee velocity, the product of these two values yields the resistive torque needed at the knee.

Torque is applied to the knee through the two series elastic actuators and monitored by various internal sensors. Two DC motors drive a series of elastic carriages which are capable of exerting both positive (in flexion) and negative (in extension) torques. Quadrature optical encoders on each of the motors sense the position of the elastic elements. The disconnected, elastic elements exert a force on the knee.
joint when they come into contact with the knee carriage. This makes it possible to implement, zero torque control, single-side torque, and co-contraction. The torque control is achieved through a high-gain position controller of the elastic actuators. The optical linear encoder on the knee carriage is used to determine its relative position to both of the actuators. The locations of each of the carriages, along with the elastic elements of known stiffness permit an effective torque control of the knee via position control.
Chapter 4

Results and Discussion

Three factors used to evaluate the effectiveness of a prosthesis are comfort, gait symmetry and metabolic consumption. Comfort is difficult to quantitatively measure and was not investigated in this study. Qualitative comments from the subject were used to maximize the comfort of the prosthesis. Gait symmetry refers to the symmetry between the affected and unaffected sides. If a prosthesis is not able fully replicate a natural gait, then an asymmetric gait is likely to occur. The gait of unilateral transfemoral amputees is often pathologically asymmetric in order to compensate for the affected side. Stance flexion of the affected side is typically non-existent or substantially reduced, while the unaffected side often undergoes vaulting, in order to ensure ground clearance of the prosthetic foot [1]. Along with gait asymmetry, transfemoral amputees exhibit increased levels of oxygen intake while walking [2]. The increased energy expenditure associated with walking can substantially inhibit the mobility of an amputee. As the metabolic consumption of the amputee decreases he or she should be able to walk further and longer. The AAAKP has been shown to reduce the metabolic consumption of transfemoral amputee during level ground walking [11]. In order to assess the symmetry of the subjects gait while using the AAAKP and also investigate the possible causes of the metabolic reduction, a motion capture system and inverse dynamics model was used to analyze the gait of a subject while wearing the AAAKP.

This preliminary study was conducted on one subject. Data was taken while the
subject was using the AAAKP, and also while the subject used his own prosthesis, Otto Bock’s C-Leg. Multiple trials were combined in order to determine averages and standard deviations of joint kinematics and torques. Since the motion capture system records the entire body, comparisons were made between the different protheses and also between the affected and unaffected sides while using each prosthesis.

4.1 Gait Symmetry

Gait symmetry is externally apparent through joint kinematics. The motion capture system was used to investigate the joint kinematics of both the knee and hip during swing extension. Comparisons were made between the affected and unaffected sides during swing extension in order to determine the effects of the subject using either the AAAKP or C-Leg. The kinematic results of the knee angle trajectories during swing extension are shown in Figure 4-1.

The knee trajectories during swing extension, shown in Figure 4-1, suggest that the AAAKP does not significantly alter the joint kinematics during swing extension when compared to the C-Leg. Both the AAAKP and the C-Leg hyper flex and begin swing extension at a higher angle, but follow the unaffected trajectory and closely emulate the unaffected swing trajectory.

Unlike the knee angle profiles, the hip angle profile of the affected side is offset from the unaffected hip angle profile during swing extension. Both the AAAKP and the C-Leg show a similar significant deviation in hip angle during swing extension. The hip angle trajectories of the affected side are similarly asymmetric with respect to the unaffected sides; the affected side experiences an increased hip flexion. The hip angle trajectories during swing extension are shown in Figure 6-1, in the Appendix.

Although the knee and hip joint kinematics suggest that the AAAKP and C-Leg are equivalently performing, examining the effective knee damping coefficient during swing extension reveals an increase in kinetic symmetry while using the AAAKP. Figure 4-2 shows the effective damping coefficients of both the affected and unaffected sides during swing extension for both the AAAKP and C-Leg.
Figure 4-1: The knee angle trajectories of both the affected and unaffected sides during swing extension while the subject uses AAAKP and the C-Leg. The knee angle trajectories suggest that there is not a significant difference in swing symmetry while using the AAAKP or C-Leg. Note: The x-axis represents the percent of the gait cycle completed during swing extension.
Figure 4-2: The effective damping coefficient of both the affected and unaffected knee during swing extension while the subject uses the AAKP and C-Leg. The results show that while the subject uses the AAKP, the effective damping coefficient of the affected side more closely resembles the unaffected side. The biomimetic and greater damping provided the AAKP may be a result of its heavier weight.

Figure 4-2 suggests that the AAKP is able to more accurately emulate the damping profile experienced by the unaffected side during swing extension, while the
C-Leg shows a significant deviation between the affected and unaffected damping trajectories. Since both the AAAKP and C-Leg produce similar knee kinematics, the difference in effective damping coefficients is due to the differences in dissipative knee torques, which are shown in Figure 6-2, in the Appendix. Since the AAAKP more closely resembles the weight of the unaffected limb than the C-Leg, the AAAKP requires a greater damping torque which is more symmetric to the unaffected damping torque.

The increased damping provided by the AAAKP also translates into a more symmetric dissipative power seen at the knee during swing extension. Figure 4-3 shows that the power dissipated by the knee during swing extension is more symmetrical between the affected and unaffected sides when the subject uses the AAAKP. The power profiles of the two legs converge, whereas while the subject uses the C-Leg, the affected side dissipates significantly less power than the unaffected side. Although the lightness of the C-Leg explains why it does not have to dissipate as much power, it does not account for the discrepancy between the power dissipation of the unaffected side while the subject uses each of the prostheses. The increased knee power of the unaffected side may be a result of how each prosthesis approaches stance. Since the affected side is in stance while the unaffected side is in swing, the different characteristics of each prosthesis may result in this power discrepancy [7].

Comparisons of the symmetry between the affected and unaffected sides during swing extension suggest that both the AAAKP and the C-Leg can emulate the kinematics of an unaffected knee, but the moments and consequently the damping and power dissipated at the knee is different for each knee. The heavier AAAKP is able to more closely emulate the torques seen at the knee during swing extension, which may have metabolic consumption consequences.
Figure 4-3: The power dissipated by both the affected and unaffected knee during swing extension while the subject uses the AAAKP and C-Leg. The results show that while the subject uses the AAAKP, the power dissipated by both the affected and unaffected knees more closely match.
4.2 Metabolic Consumption Considerations

4.2.1 Affected Side

Recent studies suggest that use of the AAAKP may reduce the metabolic consumption during level ground walking when compared to the subject’s traditional, passive prosthesis [11]. One possible hypotheses is that use of the AAAKP allows the user’s affected side to undergo a more natural swing, which reduces the power consumed by the hip on the affected side. Since the AAAKP appears to more closely emulate the torque, damping, and power profiles of the unaffected side during swing extension, as seen in Figures 6-2, 4-2, and 4-3, the hip power of the swinging side may be reduced. The more symmetric swing of the AAAKP, may lead to the hip undergoing less work and thus reducing the metabolic consumption of the subject. Figure 4-4 shows the power profiles of the affected side hip while it is undergoing swing extension.

Figure 4-4 shows that the hip power profiles of the affected side during swing extension are not significantly different between the AAAKP and C-Leg. This preliminary study suggests that the swing extension of the AAAKP does not significantly reduce the hip power of the swinging affected side, and thus is not likely the source of the metabolic consumption reduction. Although the AAAKP is able to assist the user during initial swing extension, the increased weight of the AAAKP may negate any of the advantages offered by the active system.
Figure 4-4: Hip power profiles of the affected side during swing extension, while the subject uses the AAAKP and C-Leg. The preliminary study does not suggest a significant difference in hip power when using the AAAKP or C-Leg.

4.2.2 Unaffected Side

Although the AAAKP may not be reducing the metabolic consumption of the subject by improving the biomechanics of the affected hip, it is also possible that the unaffected side experiences different power profiles while the affected side is undergoing swing extension. For example, the unaffected stance side may expend less energy in order to ensure that the affected knee fully swings without tripping. The three joints of focus on the unaffected side are the hip, ankle and knee.
Figure 4-5: The hip and ankle power profiles of the unaffected side, while the affected side is undergoing swing extension. These preliminary results do not suggest a significant difference between the AAAKP and the C-Leg.
The hip and ankle power profiles of the unaffected side, while the affected side is swinging, are shown in Figure 4-5, but they do not suggest a significant difference between use of the AAKP or the C-Leg. Neither the hip nor the ankle of the unaffected side is significantly altered by using the AAKP or C-Leg during swing extension. The results of the ankle are especially interesting since this suggests that the swing extension of the AAKP does not significantly reduce the ankle power associate with vaulting, an exaggerated plantar-flexion during stance to ensure the affected side will safely complete the swing without tripping. However, more studies will be required to investigate the AAKP’s effect on vaulting, since the subject used in this study does not excessively vault.

Finally, this preliminary study also examined the effects of the AAKP and C-Leg swing extension on the unaffected knee. The AAKP has been shown to reduce the power associated with stance flexion on the unaffected side [7], therefore, similar results may be observed while the affected side undergoes swing extension. Figure 4-6 shows the knee power profile of the unaffected side while the affected side undergoes swing extension. Unlike the hip and ankle power profiles of the unaffected side, use of the AAKP may reduce the power seen at the unaffected knee during the swing of the affected side. The reduction is not significant in this preliminary study, but the average power is consistently lower when using the AAKP.
Figure 4-6: The knee power profile of the unaffected side while the affected side undergoes swing extension. Although the AAAKP and C-Leg are not significantly different, their consistent mean difference suggests that further investigate is required.

These preliminary results suggest that the AAAKP may not have a significant metabolic advantage over the C-Leg during swing extension. Both the AAAKP and the C-Leg are able to emulate an unaffected swing extension, and may not incur a significant metabolic increase. One source of a metabolic decrease may be the effects on the unaffected knee in stance while the AAAKP is in swing extension, but further research is required. The majority of the metabolic decrease may be a result of the affected side in stance, where the subject is forced to trust the prosthesis to a greater extent [7].
Chapter 5

Conclusion

Preliminary biomechanical studies show that the knee can be modeled as undergoing a variable damping control during swing extension. The variable damping profiles can be simplified into a piecewise function with respect to the knee angle. The knee initially experiences a linear increase in damping until approximately 15-20 degrees, where the damping begins to follow a quadratic increase. A variable, parameterized, damping controller has been implemented on the AAAKP, an active knee prosthesis. Preliminary studies have shown that the AAAKP is able to more closely emulate the damping profile of the unaffected leg, when compared to Otto Bocks C-Leg. Initial studies have been conducted on the effects of the swing extension controller on both the symmetry of the gait and the metabolic consumption of the subject. The results suggest that the AAAKP is able to more closely emulate the kinetics of the knee during swing extension. Metabolic advantages may also partially arise from advantages seen at the unaffected knee during the swing extension of the affected side. However, the differences between the AAAKP and C-Leg observed in this preliminary study are not statistically significant. A larger population study is required to further study the effects and demonstrate a significant improvement over conventional control strategies.

Although the damping profile of the AAAKP in swing extension closely follows that of the unaffected side, the parameters were achieved through a qualitative analysis of the subject’s gait. Experience and communication between the prosthetist,
subject and researcher are all necessary in order to effectively implement a correct and symmetric damping profile. This process can be mostly automated, which will both make the tuning process more accurate and efficient. The internal sensors of the knee can be used to automatically tune the four parameters to produce a swing extension that has minimal terminal impact and accurate kinetics. However, the use of external sensors, such as accelerometers on the unaffected limb, could lead to even more accurate tuning that matches the kinematics of the affected side with those of the unaffected side.

The variable damping trajectory developed for the AAAKP, does not rely on the active architecture of the AAAKP. The same control strategy for swing extension can be applied to other variable damping prostheses, even if they do not have active elements. Applying the parameterized variable damping profile to a quasi-passive prosthesis, such as the C-Leg, could serve in future studies as a method to evaluate the metabolic ramifications of the control strategy.
Bibliography


Chapter 6
Appendix
Figure 6-1: The hip angle trajectories of both the affected and unaffected sides during swing extension while the subject uses AAKP and the C-Leg. Although the hip trajectories between the affected and unaffected sides are significantly different, the results suggest that there is not a significant difference in swing symmetry while using the AAKP or C-Leg.
Figure 6-2: The knee moment profile of both the affected and unaffected sides during swing extension while the subject uses AAAKP and the C-Leg.