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Controller Design and Implementation for a Powered Prosthetic Knee

by

Matthew David Rosa

A dissertation submitted in partial satisfaction of the

requirements for the degree of

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In

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of the

University of California, Berkeley

Committee in charge:

Professor Homayoon Kazerooni, Chair

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Spring 2012

Controller Design and Implementation for a Powered Prosthetic Knee

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Matthew David Rosa

## Abstract

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by

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Doctor of Philosophy in Engineering – Mechanical Engineering

University of California, Berkeley

Professor Homayoon Kazerooni, Chair

Powered prosthetic knees offer many improvements over passive devices; however, the added actuation is difficult to control due to the lack of input from the user. A means of controlling a powered prosthetic knee is proposed by predicting the type of swing behavior the knee must perform based on the position of the foot relative to the person during toe-off. The software to accomplish this is implemented via a finite state machine which activates specific knee angle reference generators for each state. The reference generators serve as the input of a nonlinear feedback controller to ensure accurate positioning of the knee joint.

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## 1. Introduction

Despite many advances in modern medicine the amputation of a limb is a sad reality for many people. The majority of these are amputations of the upper extremities. The most common being the loss of a finger. However there are a large percentage of lower-limb amputations of the foot, shank, or knee that result in a serious reduction of mobility and quality of life. Through the use of prosthetic devices doctors seek to restore this loss to reasonable success. Depending on the length of the residual limb different prosthetics are used. Prosthetics for below knee amputees consist of a replacement prosthetic foot and ankle. Additionally, a rod, called a pylon, of custom length is inserted between a custom socket and the ankle to account for the distance between the residual limb and the prosthetic foot. This is done to maintain equality of the length of both legs. Using these devices below knee amputees have much of their mobility restored, being able to walk, climb stairs, and even run. Indeed, in the most recent summer Olympic Games double-amputee sprinter Oscar Pistorius competed against able bodied athletes who were concerned that his prosthetic feet might actually be an unfair advantage in his favor [1].

Above knee amputees, however, have significant difficulties restoring mobility through prosthetic devices due to the nature of many prosthetic knees on the market. Like prosthetic feet and ankles, most prosthetic knees are passive devices, so they do not flex and extend under their own power. This means that many maneuvers that require minimal power at the knee, like walking on level ground, become very difficult while maneuvers that require significant power, like climbing a flight of stairs, become nearly impossible with passive prosthetic knees.

It is because of this fundamental difficulty of replacing such a pivotal joint with a passive device that recent attempts have been made to advance the technology of active, or powered prosthetic knees. Through the use of an actuated knee joint it may not only be possible to restore the ability to go up stairs but also improve the user's safety and ease of walking on level ground, helping to maintain healthy activity levels and improve the amputee's overall quality of life.

With the help of many peripheral advances in technology, like more powerful electric motors and higher capacity rechargeable batteries, the Human Engineering Laboratory set about the task of developing a powered prosthetic knee system that is both portable and effective. My colleagues and I divided the project into several pieces. My specific contribution, and the subject of this thesis, was the identification and experimentation of the type of sensor systems required for the safe control of the prototype in development. Also, once a prototype was designed and assembled, I worked to develop and implement a control system which I then examined its viability on several amputee subjects.

The chronological progression of my research atypically started by experimenting on the first-generation, tethered, proof-of-concept prototype developed by my colleagues. During this time in parallel processes, I developed the fundamentals of the control system while exploring the use of new sensors not originally included on the prototype to determine their necessity on the second-generation, battery-powered prototype. Then while the second-generation prototype was being built, I continued working on the control system's more advanced modes and features. When the second-generation prototype was designed, a different type of computer was selected to run the control software I had developed. Because of this decision, when the second-generation prototype was assembled I worked to transfer the control system code from the previous language of LabView to the C language used by the new computer. I then tested the full system on several amputees to validate its effectiveness.

In an attempt to dissect and present this work in a logical manner, this thesis presents its work organized by conceptual topic, from all stages of development at once, rather than

chronologically. It begins with a review of the literature to establish the current state of the art in prostheses for above knee amputees in section 2. It continues by presenting the goals and design criteria for all stages of development in section 3, highlighting how this establishes the foundation for the control system created. Section 4 examines the hardware of the first and second prototype. Section 5 describes the implementation of a finite state machine as the high level controller used in the powered prosthetic knee prototype's software. The state machine observes the sensors in section 4 and determines which reference generator, detailed in section 6, is to be implemented by the feedback controllers, discussed in section 7. In section 8, the finalized system is evaluated and conclusions drawn from the experiments are discussed.

## **2. Literature Review**

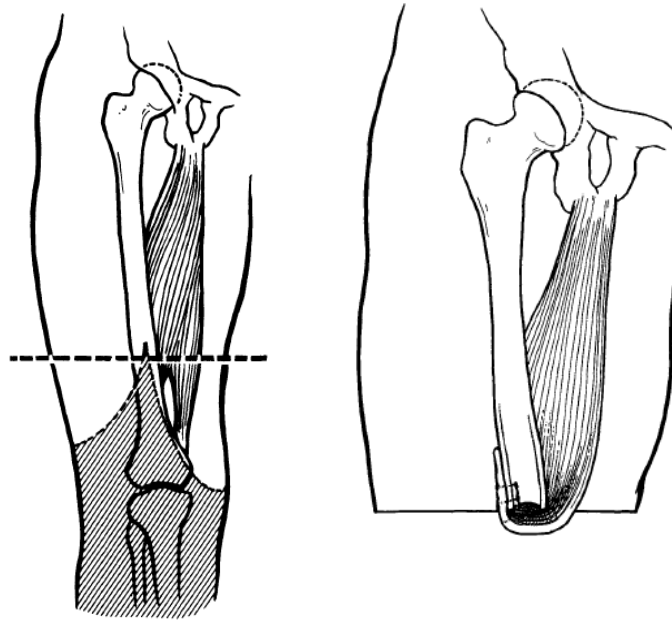
### **2.1 Amputation**

Lower limb amputation generally refers to two kinds of amputations, transtibial and transfemoral. Transtibial amputations are done below the knee joint. The name of this amputation refers to the tibia, the larger of the two bones in the shank of the leg (the other being the fibula). In the literature, these amputations are referred to as simply below knee (or BK) amputations. On the other hand, transfemoral amputations are those done above the knee joint. Similarly, the term transfemoral refers to the amputation of the femur bone, the largest and longest bone in the human body, located in the thigh. In the literature, these amputations are referred to as above knee (or AK).

The population of major lower extremity amputees (AK) in the United States is 300,000 to 400,000 [2][3]. 30,000 lower body amputations are performed each year in the United States for a variety of reasons. The first major contributor to this number is the aftermath of warfare. World War I and II greatly contributed to this number because of the sheer number of soldiers involved on the ground [4]. More recent wars in the Middle East and Bosnia-Herzegovina, although fought with smaller numbers, contribute to the number of amputated limbs because of the use of land mines and improvised explosives against U.S. forces [5][6].

A second cause for amputations in the past century is the increased use and speed of cars compared to other land-based transportation and the subsequent accidents the users have. These two sources combined account for most of the trauma-caused amputations performed [7].

The third most common and steadily growing reason for amputations is due to vascular diseases, especially as a result of diabetes [7][8]. The majority of these patients have widespread systemic manifestations of the disease that lead to a poor ability of the patient to heal a lower level amputation. Therefore, these patients have a higher frequency of transfemoral amputation as opposed to transtibial amputations [9]. A reference on the surgical procedure is given by Gottschalk [9].



**Figure 1: Amputation Process, The Femur Is Cut Along The Dotted Line And The Abductor Magnus Is Then Attached To The Lateral Femur.**

## **2.2 Rehabilitation**

After the amputation is complete, patients go through an extensive training program and rehabilitation to teach them to walk with a prosthetic in an efficient gait. This efficient gait is important so that the patient does not tire as easily from using the prosthesis during daily life and will therefore use the prosthesis more frequently [2]. When a patient chooses to use a prosthetic leg over a wheelchair, they reap many health benefits including cardiovascular health, wound healing, and muscular development.

The rehabilitation procedure often begins with the patient using a rigid prosthetic knee. This type of knee is often referred to as a stubby knee in the literature. This locked knee joint prosthetic provides a stable platform for the new amputee to begin learning to walk with a prosthetic knee [2]. Whether due to pre-operation muscular weakness or to post-operation bed rest during wound healing, this period of training is useful to redevelop musculature in the hip to account for the increased demand required there to operate even simple prosthetic knees.

## **2.3 Life with prosthesis**

The prescription of a prosthetic knee to an amputee is largely based on clinical experience and trial and error [10]. Once the patient regains muscular strength, coordination, and balance using the stubby knee, the patient is progressively given more and more advanced prosthetic knees, each requiring more time spent in physical therapy and training, until at last they reach a prosthetic knee that they will use in their daily lives [11]. This prosthetic knee is chosen in a compromise between the needs of the amputee and the amount of complexity in the knee that the patient can cope with. Generally, more advanced prosthetic knees can perform more complicated maneuvers such as going down stairs and slopes foot over foot. This can be accomplished by the C-leg, a computer controlled, passive prosthetic knee. These maneuvers require a large amount of confidence and familiarity with their prosthetic knee that may not be possible in all cases.

Despite efforts by doctors and prosthetists, only 46% of above-knee amputees are successful ambulators. To be a successful ambulator the amputee does not use the prosthetic merely for a cosmetic purpose or for transfers [12]. Transfers refer to the act of moving from one seated position to another, such as from a wheelchair to a bed or to the seat of their car. A successful ambulator is able to use the prosthetic device to ambulate in a bipedal fashion.

Because the difficulties of using prosthetic devices in the real world obstacle course of the workplace, in addition to the large number of unsuccessful ambulators, many patients have to settle on a lower paying job [2]. 57% of patients feel their reintegration to the work force is unsatisfactory whereas they rate their self-work, home mobility, and psychological adjustment as satisfactory [13]. Therefore, vocational rehabilitation is vital to improving the quality of life for amputees.

#### **2.4 The deviation of walking with a prosthetic and normal walking**

To be successfully ambulatory through the use of a prosthetic does not imply fully natural walking ability. There are many notable differences between normal walking behavior and those of an amputee using a prosthetic knee. Many of these arise directly from the prosthetic knee as opposed to the prosthetic foot used in tandem with it; however, even below knee amputees have been shown to have a gait deviating from normal walking when on level ground [14][15][16]. It has also been shown that the more proximal an amputation is, the more energy is required to walk with a prosthetic [17].

One such example of unnatural walking behavior is when the amputee vaults over their prosthetic knee [18]. During normal walking, the ankle and knee bends to cushion heel strike and reduces the motion of the person's center of gravity as it passes over the stance foot. Many amputees are not able to do this because of limitations in their prosthesis and so they pass over their planted prosthetic foot in an unnatural manner. When an amputee vaults over their prosthetic knee, the knee remains locked and their torso, increasing its potential energy, moves in an arc over their prosthetic foot. This change in height of the amputee's torso requires additional energy to be inputted by the other leg as it pushes off the ground. This extra push results in less time spent in stance on the prosthetic leg and therefore a visibly asymmetric gait. This additional energy is then absorbed by the other leg as it places the next footfall. This impact at the sound foot's heel-strike can contribute to soreness and fatigue, consequently reducing the amount of time the patient will likely walk.

Depending on the prosthetic knee and the experience of the amputee, additional muscular activity is required in the thigh during stance to maintain knee stability [18]. In essence, what the patient does is flex their thigh down against the knee joint in a manner that hyper-extends the knee, thereby keeping it at full extension even in the presence of destabilizing forces. This is described in more detail in section 2.6.1.2.. The amputee must do this because of the lack of muscular control over the prosthetic knee joint. When the knee flexes under load in stance, the leg will collapse in an unstable fashion because as the knee continues to bend, it becomes easier to bend further. It is because of this nature that when the knee bends slightly in stance, the patient experiences anxiety at the possibility of a fall and forces the knee straight with their thigh [18]. Performing this contributes to thigh muscle fatigue and reduces the amount of time the patient will walk in a day. This behavior also gives rise to the vaulting previously discussed.

Another example of unnatural walking behavior is limited to specific types of prosthetic knees. Some prosthetic knees, in an attempt to provide assistance during stance and swing, inhibit the motion of the user [18]. This inflexibility to the user's intent is often visible when walking at faster or slower speeds than the prosthetic is adjusted for.

Additional differences between normal walking and prosthetic use are the result of the shank of the prosthetic being a different mass and inertia than the sound leg of the amputee. The current trend in the prosthetic market is towards lighter and lighter devices. Many companies are moving to making components out of titanium and carbon fiber in pursuing this goal [19]. However, a study by S.A. Hale showed many interesting results. Even though they speculated that a prosthesis which had equal mass and rotational inertia of the amputee's original leg would require 3 times the amount of hip effort than a standard prosthetic knee, they found that despite varying shank mass from 35%, 75%, to 100% of the amputees' sound shank mass, there were no noticeable changes in the kinematic variables when walking [19]. Amputees were still able to maintain consistent swing time and walking speed. The change they were able to detect was the major hip musculature effort that occurs at the final stages of swing when the thigh and shank is decelerating just before heel strike. This musculature effort did increase with increased shank weight. However, when given a choice over the weight of the prosthetic, 66% preferred 75% total mass while only 33% chose the 35% total mass, the weight of the standard prosthetic. While no subject chose the 100% total mass, this does highlight that the trend to reduce the weight of the prosthetic as much as possible may not actually be the desired course [19].

The cause of this discrepancy may be due to the physics of the motion involved. If the swing-phase resistance was unchanged for each experiment, the increased mass of the shank would provide more inertial resistance to being moved as the thigh began to pass under the body during swing-phase. This would result in increased knee flexion during swing-phase as the mass of the shank increased. The increased flexion would therefore increase the amount of clearance the toe would have above the ground. In another study by Card et al, patients have expressed that an increase in toe clearance by just 1-2 cm from the standard that they are used to would reduce the risk of stubbing the toe and falling [20].

In addition to added toe clearance, patients are also interested in increased levels of functioning beyond that of simple ambulation [2]. This includes traversing slopes and stairs, even playing sports and riding bicycles. However, functional outcome is largely dependent on the strength, balance, and coordination of the individual and the extent of the amputation [2]. Because of how debilitating the above knee amputation is in addition to the side effects of the cause of the amputation, the desire for advanced function may not be attainable by most amputees.

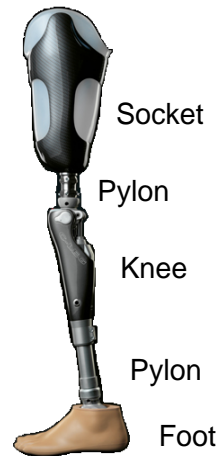
## **2.5 Prosthetic leg devices**

### **2.5.1 Prosthesis Prescription**

Prosthetic prescription is handled as an outpatient process. The patient is referred to a prosthetist who uses his clinical experience to make the proper selection of prosthetic for the patient. Many of the components of prosthetic legs are modular so that most prosthetic knees can be assembled with most prosthetic feet, resulting in an impossibly large number of combinations available. This selection process is largely based on trial and error, starting off simple and slowly working towards more advanced prosthetics, but on the whole it is unscientific and based on the clinicians' experience [10].

### **2.5.2 Prosthesis components**

Once the prosthetic is selected and fitted for the patient all of the modular parts must be adjusted so that they are properly aligned for use. It has been shown that varying the prosthetic alignment will affect energy consumption of transfemoral amputees during walking [21]. The energy consumption of patients was shown to increase at all speeds of walking and even more so at speeds faster or slower than self-selected walking speeds.



**Figure 2: Prosthet Leg Is Composed Of Modular Componentets.**

The prosthetic leg is made up of several modular components. The major components are a custom socket that is made for each amputee to form fit their residual limb, the prosthetic knee, and a prosthetic foot. When additional space is needed to separate components to add length to the prosthetic leg, custom length pylons are added between the modular parts. For example, on a tall patient with a very short residual limb, a pylon would be placed between the socket and the prosthetic knee so that the axis of rotation of the prosthetic knee is located at the same distance from the torso as the sound knee joint of the other leg. Additionally, a pylon would be placed beneath the prosthetic knee so that the shank of the prosthetic leg matches the length of the sound leg. With these two pylons in place, the patient would stand on level ground without his hip tilted left or right.

#### **2.5.2.1 Prosthetic Socket**

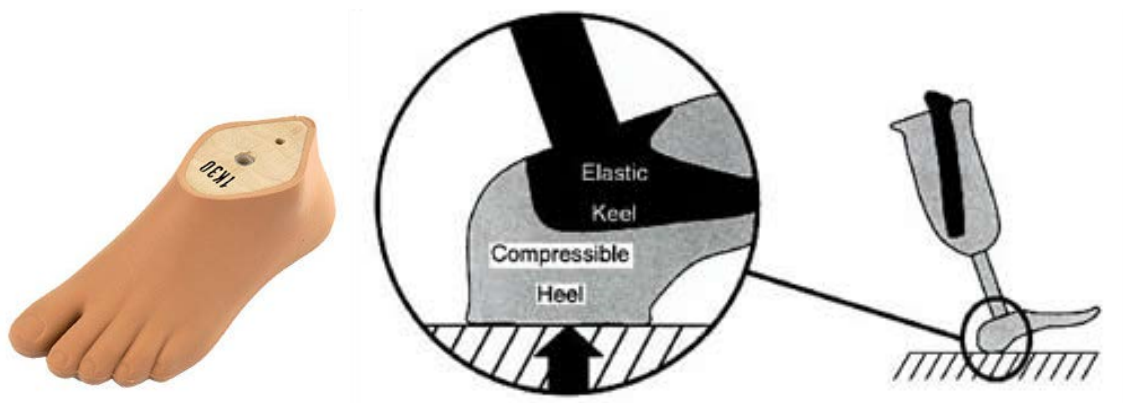
Careful consideration must be made so that the socket achieves satisfactory load transmission, stability, and efficient control for mobility. The two families of socket design vary from either distributing most of the load over specific load bearing areas or more uniformly distributing the load over the entire limb [22]. In an effort to take into account the underlying residual limb anatomy and the biomechanical principles involved, socket were redesigned, such as the quadrilateral transfemoral suction socket following World War II [24]. These designs provide a more effective distribution of loads around the residual limb so that the load-tolerant areas of the residual limb can chiefly take the load, while relief can be given to the sensitive areas. In the 1980s, the hydrostatic weight-bearing principle and the total surface bearing (TSB) concept were introduced [22]. Additional information of socket development can be found in Mak [22].

#### **2.5.2.2 Prosthetic Ankle-Foot**

The human ankle and foot is responsible for providing both flexibility and rigidity during stance to support and propel the body with each step. Studies show that the ankle is responsible for producing far more work than the knee and hip joints during walking. On average, ankle joint muscles produce 540% more work than they store during gait [25]. Power inputted at the ankle is responsible for foot motion, shock absorption, stance-phase stability, toe clearance, energy conservation and propulsion during level walking [26][27]. Because of the passive nature of most prosthetic devices, the lack of this power generation capability in the ankle is the largest

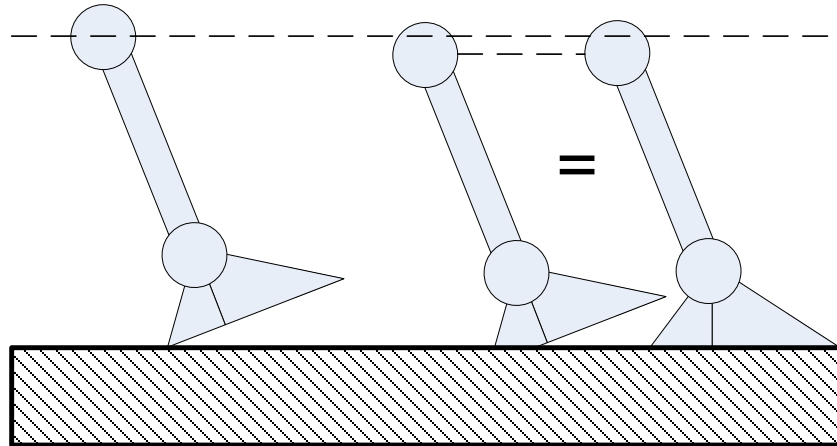
challenge in restoring normal gait to below knee amputees [28]. This difficulty is extended to above-knee amputees.

Thanks to the amazing dexterity of the human ankle and foot, people are able to traverse over a variety of terrains. There are many different types of prosthetic feet on the market. Most prosthetic feet seek to replicate this function through the use of passive components. Foot selection is very important towards the goal of ambulation. In a study of above knee prosthetics by Stein et al, it was shown that foot design strongly influences walking mechanics independent of the prosthetic knee chosen [29]. Foot selection is important because the foot is the point of contact with the ground and the amputee will need walking functionality to extend to all types of ground surfaces [30]. To this end, many prosthetic feet are designed both cosmetically and functionally to be the same size and shape as normal feet so that they can easily be worn with shoes. Because of their importance in prosthetic leg ambulation, several prosthetic foot designs are presented below.



**Figure 3: The SACH Foot. Comprised Of A Compressive Heel Surrounding An Elastic Keel.**

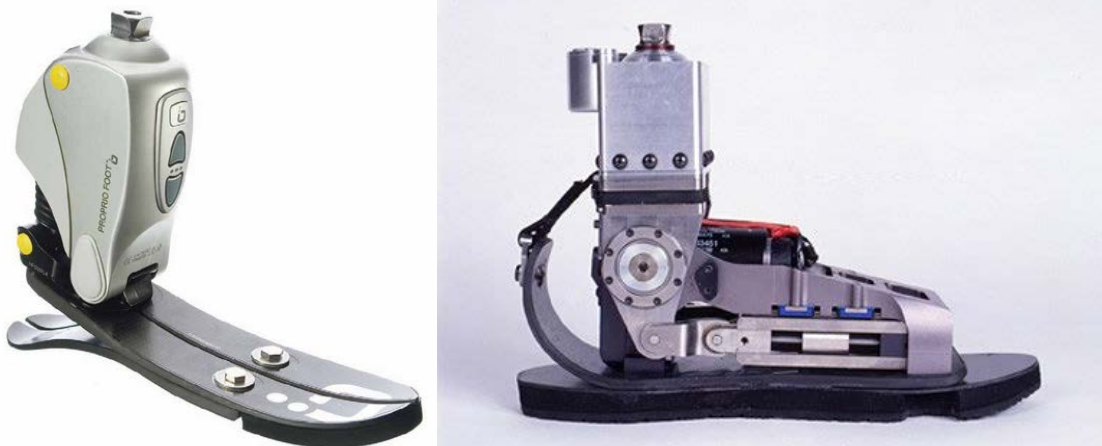
The SACH foot is the most common prosthetic foot used to restore ankle-foot functionality in prosthetic devices [31]. The acronym SACH stands for “solid ankle, cushioned heel” and describes a large number of prosthetic feet with the same basic design. The “solid ankle” refers to the fact that the foot itself is fixed relative to the shank of the prosthetic allowing no dorsiflexion or plantarflexion. The “cushioned heel” refers to a compressible wedge of material at the heel of the foot. This compressive member is responsible for shock absorption at heel-strike. It is also responsible for a “pseudo-plantarflexion” as the amputee begins to bear load with the prosthetic foot at the beginning of stance [31]. The concept of pseudo-plantarflexion is explained as follows. As load is applied to the prosthetic, the wedge compresses. The displacement of the point of contact between the heel and the ground when seen from the reference frame of the shank is similar to that seen in the plantarflexion of the human foot during normal walking. While the foot does not rotate at the fixed ankle, this pseudo-plantarflexion does replicate the motion and the benefits therein. Once planted flat on the ground, the rigid keel of the SACH foot provides stability during stance [31]. This type of prosthetic foot remains the industry standard and is popular in low income countries because it is robust and inexpensive.



**Figure 4: Visualization Of Pseudo-Plantarflexion. The Heel Compressing Under Load Is Equivalent To Plantarflexion.**

Another class of prosthetic feet is developed for the more athletic needs of some amputees. These prosthetic feet are called “energy-storing-and-returning” (or ESR) feet. ESR feet are able to store energy during stance and return it to the amputee at toe off through the use of elastic elements. There are many advances in design of ESR feet beyond the scope of this thesis, however, through the ability to return energy stored during stance, the prosthesis is able to conserve energy, resulting in enhanced gait efficiency in transtibial amputees at high speeds [32][33][34].

A new trend in bionic feet pushes the state of the art past simply passive devices and promises to restore even greater functionality from prosthetic feet. Most of these feet are on a purely research level, however [31]. Two notable exceptions are the Proprio foot from Ossur, a major manufacturer of prosthetic components, and the Powerfoot One from iWalk, a company that evolved from research conducted by the Media Laboratory of the Massachusetts Institute of Technology.



**Figure 5: The Proprio Foot and iWalk**

The Proprio foot is the first commercially available active prosthetic foot [31]. Its defining feature is an actuator at the ankle rather than passive elastic or rigid elements. The actuator in the Proprio foot is capable of dorsiflexing or plantarflexing when there are only minor forces acting on it. This means that the foot can only rotate relative to the shank under actuator



control during the swing-phase of walking. This means that the Proprio foot is not entirely active in the sense that there is an actuator at the ankle that provides the torques that would be seen in the natural foot during walking, specifically in stance-phase. Instead, the Proprio foot adapts to stairs and slopes by readjusting the angle of the ankle through the use of a stepper motor driven by a microprocessor that processes signals of integrated accelerometers and angular sensors [35]. Once the motor adjusts the angle for the proper angle of the terrain it remains there for the duration of stance, only readjusting each swing. In effect, the Proprio foot is an ESR foot in which the zero load angle of the ankle is adjusted during swing for the predicted angle of the next step. Indeed, the Proprio foot's passive elements are built off of the VariFlex line of ESR prosthetic feet, also developed by Ossur [35].

The Powerfoot One is a true active prosthetic capable of providing significant forces during the full cycle of walking. This is accomplished through the combination of traditionally elastic components of ESR feet in parallel with a high output force controllable actuator. This actuator is a series elastic actuator (SEA) at the ankle of the prosthetic foot. An SEA is an actuator that uses an elastic interface between the two sides of the actuator. For example a wheel attached to a motor is a traditional actuator. Instead of applying a torque directly between two components as is traditionally done with actuators, an SEA would be a wheel attached to a flexible axle which is then attached to the motor [36][37]. This compliance in series with the actuator is particularly important when dealing with interactions between humans and machinery.

## **2.6 Prosthetic knees**

The human knee joint is responsible for support of the body during stance and providing toe clearance during swing. The primary goal of prosthetic knee technology is to restore a natural gait to an above knee amputee. Prosthetic knees vary greatly in function and complexity. A number of authors have put forward different schemes to organize the current state of prosthetic knee technology; two are presented below [10][38][39].

### **2.6.1 Assistance based organization**

One method is to differentiate the prosthetic knees by the type of assistance they provide in the two stages of walking: swing and stance. This method provides more insight into the theoretical pros and cons of each means of assistance and is presented below in increasing complexity.

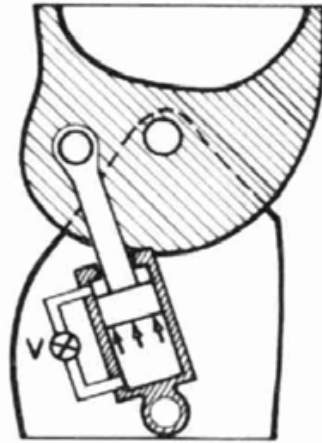
#### **2.6.1.1 Swing**

The simplest means of swing-phase assistance is that of a constantly fixed knee joint, or stubby. This method does not provide any means of obtaining toe clearance during swing, but it does provide the security of knowing that the prosthesis will always be ready to accept load for stance. The stubby is seldom used beyond early stages of rehabilitation [11].

Another means of swing-phase assistance is that of a simple rotary knee joint. The freedom provided by this joint allows the knee to bend during swing as the amputee swings their thigh forward. This flexion is induced by the inertial effects of the shank and foot resisting the forward motion of the knee joint. The amputee must accelerate and rotate the thigh quickly forward to initiate flexion and then decelerate to extend it. Because the nature of this flexion is indirect, it is difficult for the amputee to precisely control the knee joint during swing. This is seen in either the shank being swung forward too slowly or too quickly. If the shank swings too slowly, the knee joint will not be properly extended at the end of swing which is undesirable. If

the shank is swung too quickly, the knee joint impacts the mechanical limits of full extension well before the end of swing potentially causing pain and discomfort to the amputee.

To deal with this undesirable impact and to aid with user controllability a superior form of swing-phase assistance is achieved with the introduction of a rotary friction element. This friction resists the rotational motion of the knee joint reducing excessive heel rise and reduction of the end of swing impact previously discussed [10]. The friction provided in these systems is generally of constant amplitude; therefore, even when optimally configured, the prosthetic will only operate properly at the speed for which it was configured [43].



**Figure 6: Viscous Damping Diagram. Chambers Are Linked By A Variable Sized Orifice, V.**

The development of viscosity based knee prostheses during the past 60 years has overcome many of the limitations of constant friction designs [44]. In these prosthetic knees, a fluid is held within the two chambers of a hydraulic or pneumatic cylinder. Conceptually, these two chambers are connected by a fluid path with a constricting orifice that allows fluid to flow between the two chambers but are limited by the viscous effects of fluid flow figure 6. The housing of the cylinder is attached to one side of the rotary joint while the other side is attached to the piston rod. As the piston moves due to the rotation of the knee, the subsequent changing of volume of the two chambers initiates fluid flow. Because the resistive force of viscous damping is based on velocity rather than a constant value, it is able to provide appropriate forces over a greater range of walking speeds when compared to constant friction knees [39].

One means of viscosity based swing-phase control is accomplished through the use of pneumatic cylinders. Pneumatic control cylinders are generally filled with air, as opposed to other gases [45]. By nature of being a gas, the resistance provided by the cylinders is compressible. This compressibility adds compliance and comfort to the swing-phase motion of the shank, particularly cushioning the end of swing impact discussed earlier. Another advantage of pneumatic cylinders is that they are unaffected by changes in ambient temperature, so the knee prosthesis resistance is the same in a warm room as it will be after several hours of subzero outdoor winter activities [10].

Another means of viscosity based swing-phase control is accomplished through the use of hydraulic cylinders. Hydraulic control cylinders are generally filled with a silicone based oil [46], an incompressible fluid. By nature of being an incompressible fluid the resistance provided by the cylinders is unable to provide the springy compliance seen in pneumatic systems. Hydraulic damping can, however, provide higher resistive torques because of the significantly

higher viscosity of the fluid used. Hydraulic damping based systems are generally capable of performing at a larger range of speeds than pneumatic systems [38].

### 2.6.1.2 Stance

The simplest means of stance-phase assistance is that of a constantly fixed knee joint. As already stated, this method does not provide any means of obtaining toe clearance during swing, but it does provide the security of knowing that the prosthesis will always be ready to accept load at the beginning of stance[11]. Since the stubby is rigid, it is capable of providing large forces making it very stable.

Another means of stance-phase assistance is that of a simple rotary knee joint. The rotational knee joint, a simple hinge, is the first step in resuming traditional leg over leg locomotion and a natural looking gait during swing. During stance, however, the knee is very unstable. The knee must be forced into stability during stance by the amputee flexing his thigh down against the knee to counter the bending moment caused by heel-strike. This is called voluntary control of knee stability [38] and can be worked both ways to make the knee more stable, by applying a torque down at the thigh, or to buckle the knee, by applying a torque upwards at the thigh.

Voluntary control of knee stability can be further illustrated by deriving the equations that govern the interactions between of the residual stump of the amputee, the prosthesis, and the ground when statically stable and no buckling of the knee occurs.

To start, the hip joint of the amputee is modeled as a rotary joint in the sagittal plane. The interaction of the weight of the torso on the prosthetic is shown as a force directed towards the point of contact between the foot and the ground, P. The moment applied by the thigh against the prosthetic knee to resist buckling is labeled MH. The resulting loads seen at the prosthetic knee joint by the prosthetic shank are the same force, P, and a moment applied by the prosthetic knee about the knee joint, MK. The interaction of all loads on the prosthesis is given by the knee

stability equation:  $M_H = \frac{L}{h}(Pd - M_K)$  where, as shown in figure 7,

MH = the muscle moment about the hip joint.

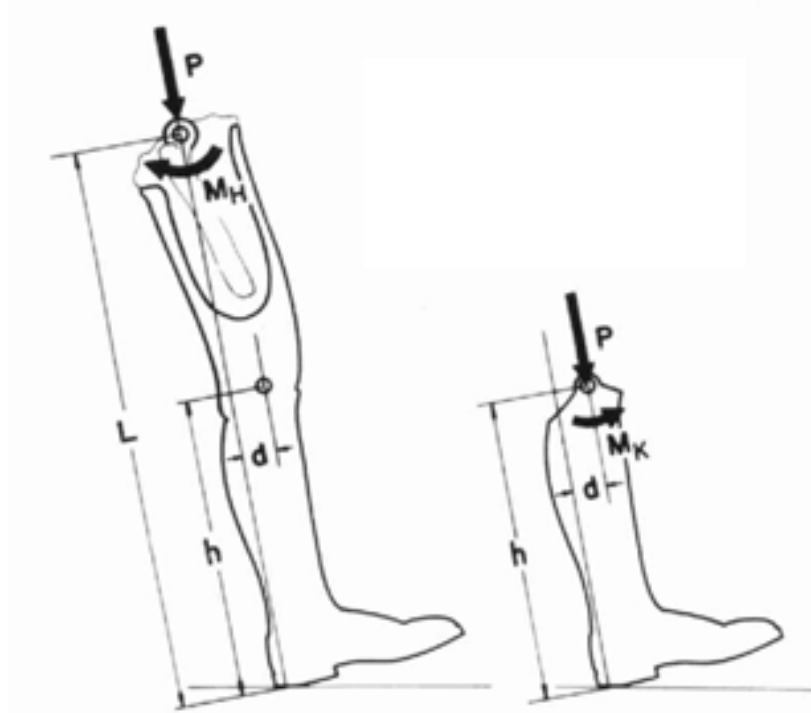
MK = the knee moment created by prosthetic knee ( $M_K \leq 0$  in simple rotational or four-bar knee).

L = the total length of the prosthesis from the hip to the bottom of the heel.

P = the force applied to the stance leg

h = the vertical height of the instantaneous center of knee rotation measured from the bottom of the heel

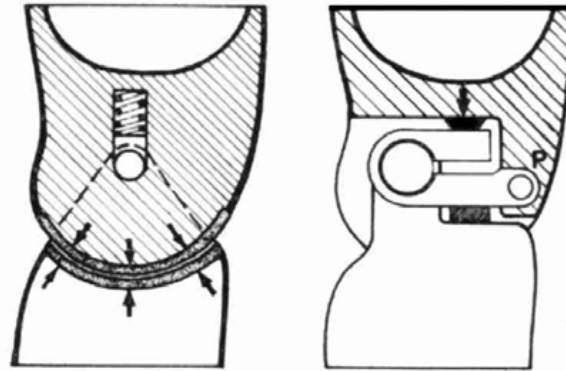
d = the distance forward from the hip/heel line to the instantaneous center of knee rotation.



**Figure 7: Voluntary Control of Knee Stability.**

The required moment about the hip to maintain a straight knee under voluntary knee control is found by applying the moment produced by the prosthetic knee's mechanical action to the stability equation. When the prosthetic knee is being flexed, the knee joint is unable to provide any resistance,  $M_k=0$ , resulting in the required thigh torque to be:  $M_H = \frac{L}{h}(Pd)$ .

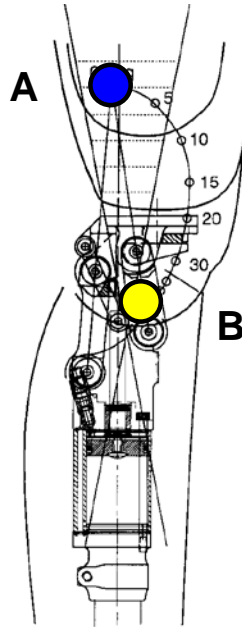
To reduce the need for the amputee to provide the thigh torques necessary for voluntary control of knee stability on rotary knee joint prosthetic knees, another means of stance-phase assistance is through the use of frictional brakes or weight activated locking clutches. Both of these systems rely on the application of the user's weight at the beginning of stance to provide assistance at the knee to the amputee. In the case of friction based prosthetic knees, the upper part of the prosthetic knee, the part that is then attached to the residual limb socket, is allowed to press down upon a friction pad on the lower part of the prosthetic knee, the part attached to the shank. This normal force between the two interfacing surfaces creates friction. This frictional interaction is designed through selection of materials so that static friction is sufficient to keep the knee joint at a constant angle of full extension once weight is applied to the prosthetic. In the knee stability equation, this is shown as the moment applied by the prosthetic knee being equal in magnitude to the moment applied through loading,  $M_K = Pd$ . This results in the muscular effort of the hip to equal zero,  $M_H=0$ .



**Figure 8: Friction Based and Clutch Based Locking Knees.**

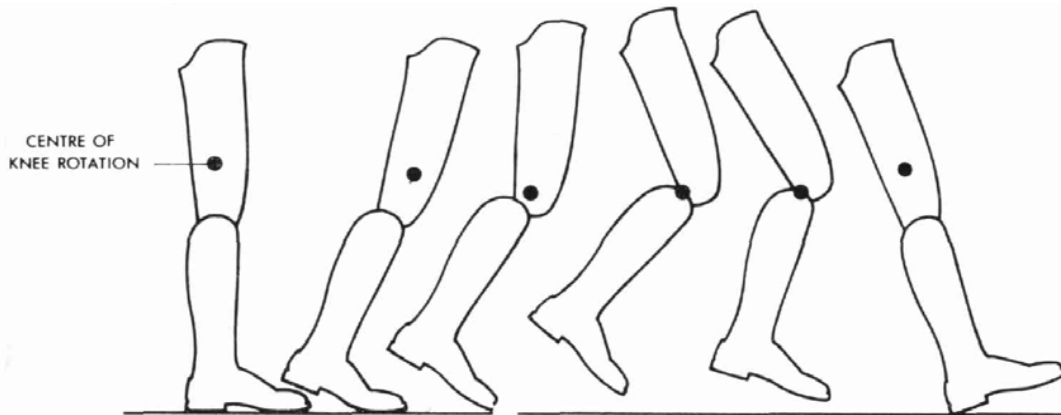
Weight activated clutch mechanism lock the knee joint into place when weight is applied at the beginning of stance. Like frictional breaks, when the weight is removed from locking prosthetic knees, the resistive torques they can apply drops to zero, allowing swing. These systems combine the rigid stability of the stubby with the swing-phase flexion of the rotational joint knee. However, these stance assistance systems still require some voluntary control of knee stability to be applied by the amputee during the early stages of stance, before the amputee's entire weight is applied to the prosthetic knee. Until weight is applied, the friction element may slip or the clutch may not activate, so amputees must still manually control the knee joint to be at full extension until their full weight is applied to the prosthetic.

To further increase the stability of a simple rotary joint prosthetic knee, the axis of the joint can be moved posterior to where it would naturally occur. In the previous discussion regarding voluntary control and the knee stability equation, the magnitude of the thigh moment required to maintain stability was dependent on two independent variables: the vertical height of the instantaneous center of knee rotation measured from the bottom of the heel,  $h$ , and the distance forward from the hip/heel line to the instantaneous center of knee rotation,  $d$ . Again, the equation for the thigh moment to maintain stability with no assistance by the prosthetic knee is: . By moving the knee joint of the prosthetic closer to the load line, reducing  $d$ , and closer to the thigh joint, increasing  $h$ , the magnitude of the required thigh moment is quickly reduced. Indeed if the rotational axis of the knee joint were moved to the other side of the load line,  $d < 0$ , the knee would naturally be in a stable configuration at the moment of heel-strike and the amputee would not need to apply any effort to keep the knee from buckling. Creating this hyper-stable knee joint in practice, however, causes the flexion of the knee to occur in an unnatural manner and makes the amputee's prosthesis stand out when walking or sitting.



**Figure 9: Motion Of Polycentric Prosthetic Knee's Instantaneous Center Of Rotation. Point A And B Are At Full Extension And 90 Degrees Flexed, Respectively.**

A family of prosthesis was designed using a four-bar linkage instead of a simple rotary joint to produce this concept of a hyper-stable knee during stance yet still appear to be rotating naturally when bent. Through the use of the four-bar linkage, the knee can be given a nontraditional rotational behavior shown in figure 9. In this figure for each angle of the thigh relative to the shank, the instantaneous center of rotation of the shank is shown as a point on the curve. Notably, when the knee is at full extension, the point by which the shank's motion rotates about is shown to be at point A, remarkably higher and closer to the load line than traditionally. As the knee then bends to ninety degrees, the position of the instantaneous center of rotation for the shank is shown to pass to a more natural position beneath the residual stump where the knee would normally be, point B. Through use of this design, the amputee is able to rely on a hyper-stable prosthetic knee yet still retain a natural looking prosthesis during use.



**Figure 10: Motion of Instantaneous Center of Rotation While Walking.**

The locking of the knee joint that occurs in friction and clutch based prosthetic knees during stance inhibits pre-swing flexion which is shown to be further required for more natural motion [10]. The development of viscosity based knee prostheses during the past 60 years has overcome many of the limitations of friction designs.

One means of viscosity based stance-phase control is accomplished through the use of pneumatic cylinders. Again, the resistance provided by the cylinders is compressible. This compressibility is undesired during stance-phase motion of the shank. Even if the damping resistance is increased to the point where no fluid is able to flow from one chamber of the cylinder to the other, the fluid is still a compressible gas. This means that the knee joint will then behave as if a rotary spring were placed at the knee joint [47]. Because the 'spring' is uncompressed at full knee extension, it will not provide any resistive force until the knee joint bends to compress the spring. This results in a leg that for angles near full extension has no effective resistive forces and therefore behaves like a simple rotary knee joint prosthetic. Because of this, pneumatic cylinders are seldom used for stance-phase control unless they are already present on the prosthetic knee for swing-phase control [40][41][42].

Another means of viscosity based swing-phase control is accomplished through the use of hydraulic cylinders. As previously stated, hydraulic control cylinders are generally filled with an incompressible fluid. By nature of being an incompressible fluid the resistance provided by the cylinders no longer results in the springy compliance seen in pneumatic systems. This allows the knee to flex while giving assistance prior to toe off, thereby promoting a more natural gait [38]. This is accomplished by the higher magnitude of resistive torques that hydraulic damping can provide. The knee under hydraulic damping can maintain enough torque so the knee does not buckle uncontrollably even under full weight. The user can thereby allow the knee to bend without the usual anxiety accompanied with knee flexion during stance [18]

## **2.6.2 Complexity Based Organization**

The second method authors have put forward to organize the current state of prosthetic knee technology is to separate prosthetic knees by the complexity of their control: mechanically passive, microprocessor controlled passive, and microprocessor controlled active [10]. This method provides little insight into the theoretical pros and cons of each type of control; rather, in conjunction with the conceptual knowledge discussed in the previous section, this organizational method provides an opportunity to describe the functionality of several example prosthetic knees and further establish the current state of the art in prosthetic knee technology. Examples are provided in a roughly chronological order of development to highlight the evolution of prosthetic knee technology. All of the prosthetic knees discussed are still sold and in use today [40].

### **2.6.2.1 Passive Prosthetics**

The first designs that make improvements to the peg leg used initially to substitute the natural knee since antiquity are the constant-friction and the friction-brake designs [39]. The prosthetic knees 3R22 and the 3R49 developed by Otto Bock, were designed based on research that took place in Europe after the end of World War I. These knees, as previously discussed, were not capable of walking at different speeds and/or on uneven surfaces.



**Figure 11: Otto Bock's 3R22 and 3R49.**

With pneumatic devices, speed-dependent swing-phase resistance can be provided by letting the compressed air in the cylinder escape through an orifice. However, they are not suitable for producing a firm assistance during stance because of the compressibility of air [47]. As a result of this and the advancement of hydraulic cylinders, there are no purely pneumatic prosthetic knees still on the market.

Hydraulic based prosthetic knees are able to provide sufficient assistance during stance but this amount of damping is too much during the swing-phase of walking. Creative means of engaging and disengaging this large amount of resistance must therefore be used. This is accomplished through the use of involuntary control signals derived from events occurring in the prosthesis itself as a result of walking. By tying a mechanism to activate or release the hydraulic stance or swing control based on these naturally occurring interactions, the prosthetic knee automatically provides the appropriate resistance. The primary involuntary control signals that have been used in prosthetic devices are heel and toe pressure, weight application, and hyperextension sensing. For example, during typical prosthetic walking, weight is applied to the prosthetic knee and an increase in heel pressure is seen at heel strike; this is an appropriate time to activate stance-phase control. At the end of stance, just before toe off, toe pressure increases and the knee joint is forced into hyperextension; this is an appropriate time to turn off stance-phase control. After that, weight is removed from the prosthetic; this is another appropriate time to turn off stance-phase control. As the knee bends and is swung forward for the next footfall, the knee is again forced into a moment of hyperextension, where stance-phase control could be activated. Various different combinations of these involuntary control signals are used today.



**Figure 12: Ossur's Mauch SNS and Otto Bock's 3R80.**



The Mauch SNS designed in the 1950s [10], currently the most prescribed prosthetic knee [3], uses a combination of hyperextension and motion sensing to activate and deactivate stance-phase control. The result is a prosthetic knee that through the normal action of walking activates hydraulic damping in stance-phase and releases it in swing-phase. At toe off, a hyperextension moment is applied to the knee, freeing the knee to flex. The knee joint is then able to flex freely until it begins to extend where it becomes weight bearing again [47]. The knee is able to extend freely in this weight bearing stage through the use of a one-way check valve, yet flexion is resisted by the hydraulic fluid being passed through the constricting orifice.

A secondary result of this design is that the prosthetic knee, through the application of a thigh torque similar to voluntary stability control, can be controlled directly by the user to deactivate stance-phase assistance prematurely. This is desirable for instances where the amputee wants to walk downstairs or downhill in a jackknifing fashion, a technique used with a standard prosthesis to descend these terrains [49]. This prolonged hyperextension moment from either natural walking or from stump extension pressure about the hip occurs at a point where there is no danger of buckling of the knee, so unlocking under these conditions is safe [47].

The 3R80 hydraulic knee by Otto Bock, alternatively, engages and disengages its stance-phase control resistance through detection of weight being applied to the prosthetic knee by the amputee. Additionally, the geometry of the joint allows several degrees of controlled knee flexion during heel strike to help absorb the shock of impact, thereby further resembling the kinematics of normal walking [10].

Additional prosthetic knees have been developed to make the most of all advancements in purely passive prosthetic devices by combining different forms of assistance in one prosthetic knee. Some may use a combination of pneumatic assistance during swing while using hydraulic damping or friction brakes during stance. Several also combine 4bar mechanisms with pneumatic or hydraulic cylinders.



**Figure 13: Endolite's ESK+, Ossur's Total Knee, and Otto Bock's 3R106.**

The Otto Bock 3R106, one of the few pneumatic devices still on the market, is a hybrid prosthetic knee combining a polycentric, four-bar knee joint for stance-phase hyper-stability and a double chamber pneumatic cylinder for swing assistance. The double chamber design of the pneumatics provides a simple means for having different levels of damping for flexion and extension motions resulting in more natural swing-phase behavior. The ESK+ knee by Endolite

is a hybrid prosthetic knee that combines the stance-phase stability of a locking rotary knee joint with the swing-phase assistance of a pneumatic control cylinder. The Total Knee from Ossur combines a polycentric knee with a hydraulic control cylinder for late stance flexion to improve walking performance.

### 2.6.2.2 Computer controlled prosthetic knees

Fluid based prosthetic knees offer many advantages over their predecessors already discussed. One key advantage is their ability to operate and provide assistance over a wide range of walking speeds because their resistive torques are based on viscous damping and therefore linearly scale with velocity. Further improvements in producing a more natural gait pattern are claimed to be possible by manufacturers if the magnitude of the damping is fine tuned for each speed of walking on a step by step process. This is the fundamental motive behind the development of microprocessor controlled adjustable prosthetic knees.



**Figure 14: Blatchford's IP, Otto Bock's C-leg, Ossur's Rheo Knee, and Endolite's Adaptive Microprocessor Controlled Prosthetic Knees**

The Intelligent Prosthesis (IP), developed by Blatchford in 1993 [39], is the first commercially available prosthetic knee controlled by a microprocessor [51]. The IP utilizes the microprocessor to fine tune the swing-phase pneumatic control cylinder for varied speeds of walking. Active management of this swing-phase behavior is claimed by the manufacturer to reduce energy expenditure and further increases the range of possible walking speeds compared to purely pneumatic assistance prosthetic knees.[51]

The C-leg, developed by Otto Bock in 1997 [39], is the first commercially available prosthetic knee to use a microprocessor to control the dynamics of a hydraulic control cylinder for both swing and stance-phases of walking [51]. The addition of computer adjusted stance-phase assistance further expands the functional domain of the prosthetic knee to include the descent of stairs, slopes, and uneven terrain in a safe, controlled manner by the amputee. This added function is a distinct advantage over the IP as both manufacturers claim to reduce the energy expenditure of amputees.

The Adaptive knee, developed by Endolite in 1998 [39], is a microprocessor controlled prosthetic knee that incorporates both an adjustable pneumatic control cylinder for swing-phase assistance and an adjustable hydraulic control cylinder for stance-phase assistance. This

prosthetic knee combines the effectiveness of the IP and Cleg in one system boasting the advantages provided by both [51].

The Rheo Knee, developed by Ossur in 2001 [39], uses a magneto-rheological brake system, not previously mentioned, to provide assistance similar to hydraulic control cylinders. To illustrate how this works, the Cleg microcontroller controls the size of an orifice in a hydraulic system, thereby modulating the damping seen as the constant viscosity fluid passes through it [52]. The Rheo Knee uses a magneto-rheological fluid whose viscosity is modulated by the electric current intensity flowing through it; this fluid's flow is then constricted by a constant sized orifice [53]. The two systems are therefore equivalent, though the Rheo Knee has less moving parts which is an advantage in terms of product life.

### **2.6.2.3 Active prosthetic**

The Power Knee, developed by Victom in 2007 [39], is the first active knee for unilateral above-knee amputees available at the market. It produces an effective flexion and extension of the knee that restores natural kinematics of gait by mirroring the kinematics of the sound leg through the use of an electric motor-driven ball-screw mechanism. This produces exceptional performance on leveled floor and even walking up and down stairs and slopes [39]. The required motors needed to substitute muscular functions results in a large and heavy prosthesis that severely limits the acceptance of this kind of prostheses.

In a study by Stein, it was shown that normal kinematics imposed upon the prosthetic knee does not necessarily produce normal hip kinematics [29]. It is possible that Stein's modified echo approach fails because it does not account for lack of ankle flexion of the prosthetic foot, yet the Power Knee may do this.

### **2.6.3 Studies of microcomputer controlled prosthetic knees**

Computer controlled, adjustable prosthetic knees require more effort to design and build which results in an increased price for the manufactured product. A corollary interest in separating prosthetic knee designs by complexity of control is to determine if the added complexity does indeed merit the increased cost to the amputee. To this aim, many studies have been conducted to evaluate the benefit of a microprocessor controlled prosthetic knee. Unfortunately, due to the recent release of the PowerKnee, there are no available studies on the advantages of powered prosthesis.

#### **2.6.3.1 Studies of the Intelligent Prosthesis**

Because the IP knee uses the microprocessor to control the swing-phase of gait, the advantages of using the IP over traditional prosthetic knees should occur primarily during the swing-phase of gait. Active management of swing-phase behavior is claimed by the manufacturer to reduce energy expenditure, provide the ability to walk at a greater range of walking speeds, and produce a more natural gait pattern beyond that of traditional pneumatic control prosthetic knees.

Energy expenditure, as measured by oxygen consumption rate in the literature, is important for amputees. The energy cost of ambulation is shown to be 65% above normal for amputees for level walking [54][55]. Again, current prosthetic knee technology seeks to reduce energy expenditure so that amputees will increase the amount of walking they will perform daily to increase their health. Several studies have compared the energy consumption of the microprocessor-controlled IP with nonmicroprocessor-controlled knees. One study, a survey given to 22 IP knee users by Datta et al, suggests that the microprocessor control most influenced metabolic energy expenditure, walking at varying speeds, and distance walking [56]. Taylor et al [57], in a study of a single subject, found that energy expenditure was reduced with the IP knee

but only at speeds in excess of 53m/min (1.975 mph). Buckley [58] and Datta [59], in a later study, showed similar results, that the IP knee significantly reduced the energy expenditure at walking speeds above or below (but not at) self-selected. These results makes conceptual sense because traditional prosthetic knees are adjusted by prosthetists to perform optimally at the amputee's self-selected walking speed; deviation out of this optimal zone would cause traditional prosthesis to perform poorly.

The influence of the IP compared to traditional prosthetic knees on gait and cognitive demand was also investigated. Datta [60], in yet another study, observed the walking speed and step symmetry in 10 amputees, showing no significant differences when comparing the IP to traditional prosthetic knees. The concentration required by amputees for ambulation with their prosthetic knees was investigated by Heller et al [61] by measuring the body sway of 10 subjects wearing an IP and a mechanical control knee. No significant differences were seen in the amount of total body sway measured while subjects walked and simultaneously performed simple or complex mental tasks while using the IP compared to the traditional prosthesis. These results do not support those reported by IP knee users [56].

### **2.6.3.2 Studies of the Cleg & Rheo Knee**

Because the C-leg and Rheo Knee utilizes a microprocessor to control the damping assistance for all phases of gait, the advantages of using them over traditional prosthetic knees should occur during the full cycle of gait. On the other hand, hydraulic damping based prosthetic knees are generally seen to perform well in stance while pneumatic control cylinders perform well in swing. So the C-leg and Rheo Knee can be expected to advance the achievements of hydraulic prosthetic knees, namely: to provide a more natural loading of the knee during stance, thus further reducing the vaulting phenomenon over a wider range of walking speeds.

In a study by Hefner et al [51], the functional ability, performance, and satisfaction of 17 transfemoral amputee subjects during the transition from an established, mechanical control prosthetic knee system to that of C-Leg were observed. The C-leg was shown to produce statistically significant improvements in the ability to descend stairs, time required to descend a slope, sound-side step length while descending a hill, preference, satisfaction, self-reported frustration with falling, and self-reported frequency of stumbles and falls. Hefner concluded that the C-leg was able to produce improvements in functional areas beyond level walking but traditional views of prosthetic knee effectiveness do not include this.

Similar to the IP, several studies have compared the energy consumption of the C-leg and Rheo Knee with nonmicroprocessor-controlled knees. Johansson et al [62] observed that when using the Rheo knee, metabolic rate decreases by 5% compared with the Mauch and by 3% compared with the C-leg. Additionally they found an enhanced smoothness of gait, a decrease in hip work production, a lower peak hip flexion moment at terminal stance, and a reduction in peak hip power generation at toe-off beyond that of the traditional Mauch SNS. Seymour et al [63] compared the C-leg to a number of traditional prosthetic knees: Mauch SNS, 3R80, Total Knee and others. There were observed statistically significant differences between the C-leg and the other prosthetic group for mean oxygen consumption at both normal and fast walking paces (7% reduction). The average normal pace on the treadmill was 49+/-15 m/min (1.83+/-0.6 mph) and the mean fast pace was 70+20 m/min (2.61+/-0.75 mph). Perry et al investigated the effect of the C-leg on a single bilateral amputee subject [11]. When wearing the C-Leg prostheses, the subject walked the farther and faster. Additionally, the rate of oxygen consumption during walking was shown to be greater with Mauch SNS prostheses than with the C-Leg prostheses.

A dissenting study by Orendurff et al [64] showed the self-selected walking speed of 18 amputees using the C-Leg was significantly faster than with the Mauch SNS (mean  $\pm$  SD = 1.31  $\pm$  0.12 m/s vs 1.22  $\pm$  0.10 m/s), but the oxygen cost did not significantly increase during ambulation at the faster walking speed. Additionally, Kaufman et al [3] found a small 2.3% decrease in the energy expenditure of walking with the C-leg over traditional prosthetic knees was not statistically significant. However, the 11 amputees reported qualitatively that the C-leg required less effort during ambulation. Therefore, Kaufman asserts that the improved energy efficiency could be considered clinically significant.

The influence of the C-leg compared to traditional prosthetic knees on gait, obstacle course evaluation, and quality of life was also investigated. In another study by Kaufman et al [65], 15 subjects were shown to have improvements in their gaits that were statistically significant. Also, balance improved significantly when using the microprocessor controlled knee. Kaufman was also able to show that the amputees demonstrated improvements in equilibrium by scoring them on a six conditioned sensory organization test. In the previously mentioned study by Seymour [63], the amputees were observed running an obstacle course. The results showed that the number of steps, total time, and the number of step-offs with hands-free using the C-leg compared to the traditional prostheses were significantly less. The total time while carrying the weighted basket using the C-leg was also significantly less compared to the traditional prostheses. They also showed scores on a quality of life index for subjects using the C-leg were above the mean for norms in the US [63].

### **3. Controller Design**

The first step to develop a successful knee control system is to develop and elaborate the design goals of the system. Design goals must be oriented according to research of the different aspects associated with the biomechanics of human walking both with and without prosthetics, as well as general mechatronics concepts. There is additional value in researching competing controller ideas and prosthetic knee evaluation guidelines. Once the goals are formulated, various proposed control schemes can be evaluated and the system with the best potential can be selected.

#### **3.1 Project Goals and Research**

##### **3.1.1 Initial goals**

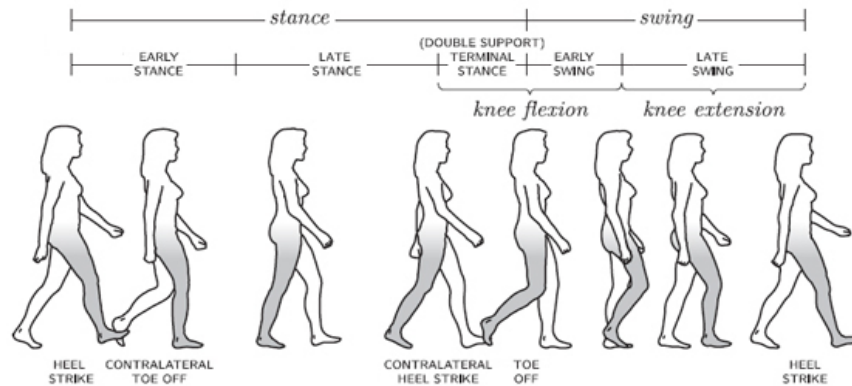
The primary goal of the knee control system is to control the actuation of the powered prosthetic knee prototype's main rotary joint so that it supports the user during the stance-phase of walking and bends the knee properly during the swing-phase of walking. The powered prosthetic knee prototype's motion needs to be smooth while simultaneously adaptable to the various maneuvers the amputee must perform daily; at this point in development, those maneuvers include walking on level ground, up and down inclined slopes, as well as up and down stairs. It is also important that each of these behaviors can be accomplished at varied speeds. Running is not a means of locomotion considered for this powered prosthetic knee prototype.

A secondary goal for the powered prosthetic knee prototype's controller is that it requires no direct input from the user. An example of an undesirable user input would be to implement a button that is pressed by the user to initiate the extension of the knee joint. This distinguishes the amputee using the powered prosthetic knee as being simply the wearer rather than the operator. Therefore, the knee must possess a level of perception and intelligence such that it behaves in a smart and intuitive manner. It must take a step forward when the user wishes to do so and only when wished to do so.

### 3.1.2 Walking Fundamentals:

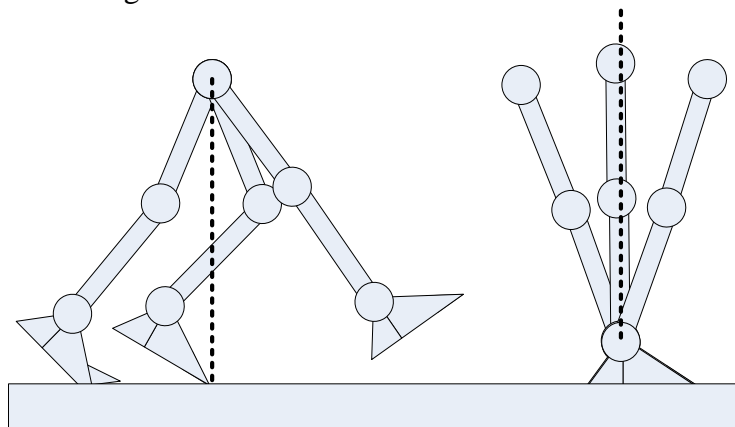
By looking closer at how typical walking locomotion occurs, other design goals may present themselves. For the sake of discussion some walking terms should first be described. Additional information can be found in Waters, Perry, and Rose where the following information originates [17][26][66].

Walking is the cyclical locomotion by which people traverse by foot. During sustained level ground walking, each leg is described as being in swing-phase or stance-phase while the foot of that leg is in the air or in contact with the ground, respectively.



**Figure 15 : Level Ground Walking Phases**

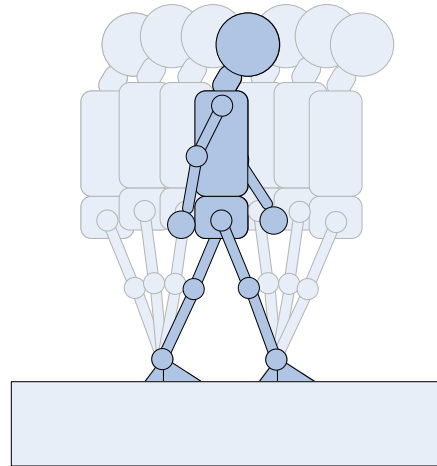
Level ground walking can be split into several stage, seen above. Starting at heel-strike of the near leg, the person enters a period where both feet are on the ground. The load of the torso is initially supported by the far leg as the cycle begins with the heel-strike of the near leg. Weight is transferred from the far leg to the near leg during this period called double-stance. Once all the weight has been transferred to the near leg, the far leg leaves the ground at a point called toe-off. During this period the far leg is in swing-phase. As the leg swings posterior to anterior, heel-strike repeats the cycle for the other leg. The amount of time for each leg in each phase as a percentage of a full cycle is 60% for stance-phase and 40% for swing-phase. This results in a period of time, 20% of the total cycle, where both legs are in the stance-phase. This is often referred to as the person being in double stance.



**Figure 16 : Early and Late Phases of swing and stance, respectively**

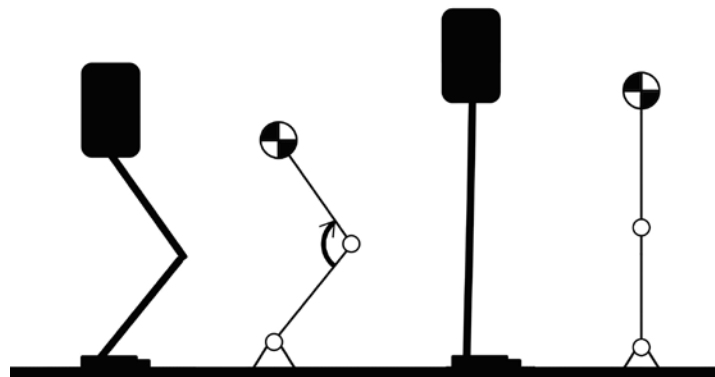
Additionally, each phase can be further split into early and late stages. Early-stance occurs at heel-strike and continues until the foot passes beneath the torso, with late-stance

beginning at this point and continuing until toe-off. Alternatively, early-swing occurs at toe-off and continues until the foot passes beneath the torso, with late-swing beginning at this point and continuing until heel-strike.



**Figure 17 : Pendulum Model**

There are many subtleties to the motion of the body during stance, but the overall behavior of the leg can be generalized by the inverted pendulum model [66]. This model states that at heel-strike, the knee joint becomes fixed and the load of the person's weight, minus that of the stance leg, pivots about the ankle like an inverted pendulum. Considering this model energetically, the forward kinetic energy of the body is transferred to potential energy when the center of gravity is raised. It will then be reconverted back to kinetic energy as the mass falls before the next footfall. The opposite leg impacts the ground at the moment of heel-strike and its respective ankle joint becomes the new pivot of a second inverted pendulum. While there is obviously a loss of energy due to the inelastic impact in this model, actual walking conserves energy through the elasticity of the plantarflexion and dorsiflexion muscles of the ankle. It has been shown that because of these muscles, the ankle joint acts like a spring, absorbing energy at heel-strike and releasing it at toe-off [25]. Therefore, a good deal of energy is conserved, thus making human stiff legged locomotion very efficient. The muscles only need to provide the slight energy lost from inefficiencies to continue walking. The muscles responsible for rotation of the thigh and flexion of the ankle are those that provide the most significant necessary forces [66].



**Figure 18 : Bent Knee and Supporting Torque Versus Straight Legged Stance**

Straight legged human locomotion is a significant improvement over a primate's bent knee locomotion where the knee joint must also provide a supporting torque, as illustrated above. It is important to note that while no mechanical work is done by the knee torque to maintain the bent knee stance, mechanical work requires the force to act across a distance, chemical energy is still being consumed to maintain muscle tension. This can be observed anecdotally by comparing the length of time a person can stand with a straight knee stance versus a bent knee angle of ninety degrees. Similarly, there are losses in electromechanical systems to maintain a torque with an electric motor even when no mechanical work is completed. Therefore, it is important that the control system for the powered prosthetic knee prototype capitalizes on the efficiency of human walking by maintaining a straight knee during the stance-phase. The muscles responsible for thigh rotation are still present in above knee amputees so this means of locomotion will still be viable even with the loss of ankle power. Indeed, this can be seen in the effectiveness of current passive prosthetic knee devices.

The inverted pendulum model does not describe what occurs for the leg in swing-phase and therefore is not a complete model. An alternative model of human walking is to describe the neuromuscular system as performing an energy optimization routine using a neural network, literally the brain. Evidence is given that a person's individual gait and manner of walking arise due to personal differences in the mass and inertia of that person's limbs as well as the strengths of their muscular-skeletal systems [66]. When learning to walk, through constant manipulation of the applied muscular torques, the neuromuscular system observes and evaluates different actions that produce locomotion. Additional evidence shows the resulting gait pattern is the one that reduces muscular expenditure to a local minimum for the specified walking speed. Ideally the best behavior for controlling the powered prosthetic knee prototype would mimic that of the user's original walking pattern because that pattern was the most efficient motion the user's body found in developing the gait. This is specifically useful for swing-phase motion controller development.

### **3.1.3 Prosthetic Devices:**

As discussed earlier, most prosthetic knees on the market are described as passive devices, meaning the knees rely on passive elements such as springs and dampers to provide torques at the knee joint rather than active elements like actuators and motors [10][38][39][40][41][42]. Because of this the wearer must provide all of the energy required to walk from their thigh muscles when they were previously able to use all of the muscles of the leg. The powered prosthetic knee prototype seeks to relieve some of this extra demand by knee actuation, specifically during swing where it is not present on a passive prosthetic knee.





**Figure 19 : C-leg and Adaptive Prosthetic Knees**

A largely successful method of providing torques during stance is through hydraulic damping [10][38][39], see section 2.6.1.2. Many prosthetic knees on the market do so [40][41][42]. The C-leg and Adaptive Knee, for example, are microprocessor controlled passive knees which use adjustable hydraulic damping to provide different levels of support during stance for level walking or the descent of slopes and stairs [67][68]. It would therefore be practical to use similar technology and control schemes during stance for the powered prosthetic knee prototype.



**Figure 20 : Power Knee with Sensor Module**

A competing powered prosthetic knee was developed by Victom and later sold and marketed by Ossur called the Power Knee [69]. The primary actuator is a ball-screw mechanism driven by an electric motor. The control system utilizes an inertial measurement sensor array placed on the non-amputated leg's ankle and pressure sensors integrated into an insole for the non-amputated shoe. While the user is walking, this signal is analyzed to generate a history of where the foot is moving through space. The system then replays the observed motion of the good leg at the powered joint, but half a step out of phase. On the surface, this appears to be a highly successful method of controlling the prosthetic joint as a variety of motions can be

replicated and terrain overcome so long as the un-amputated leg can perform it first. However, there is a difficulty present when it is not desired that the second step mirror the first step. An example would be when climbing a set of stairs with an odd number of steps. It also makes it impossible to lead with the prosthetic leg or be used for double above knee amputees. The prosthetic must always lag behind using this method of control. Additionally in a study by Flowers et al, mirroring the motion of the sound leg onto the prosthetic did not produce a natural gait for the amputee [29]. The fact that the Power Knee is being sold to amputees shows that the designers have solved this problem, but it may have been difficult. It is therefore desired that the powered prosthetic knee prototype operate on its own impetus rather than use a similar means of control.

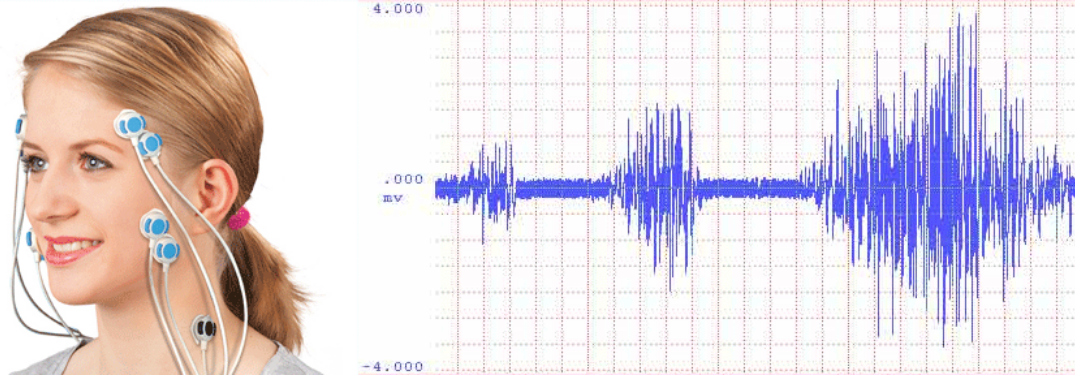
### 3.1.4 Exoskeleton Technology:

Exoskeletons are electromechanical devices worn by people that use sensors and actuators to supplement the power of a human [70]. The specific category of exoskeleton useful in the development of the knee prototype is lower limb exoskeletons. Investigating exoskeleton technology does not produce any direct goals for the development of the prosthetic prototype, however this technology provides several examples of control methods that could be used in the development of a prosthetic knee. The ExoClimber exoskeleton, also developed in the Human Engineering Laboratory, and the HAL5 exoskeleton are discussed below.



Figure 21 : HAL5 and ExoClimber Exoskeletons

First, work completed for the Berkeley exoskeleton project developed many mechanical aspects that are present in the prosthetic knee prototype, see section 4 and Lambrecht for more information [71]. From a control theory point of view, the exoskeleton uses a hierarchal finite state machine to control the high level behavior of the exoskeleton's powered joints by observing contact sensors in the foot. The development of this system is detailed in my masters' thesis [72]. The net output of this control scheme is a knee-joint that is hydraulically damped during stance and completely free during swing, similar to a passive prosthetic knee like the C-Leg. It may therefore be possible to use the same method for the powered prosthetic knee prototype.

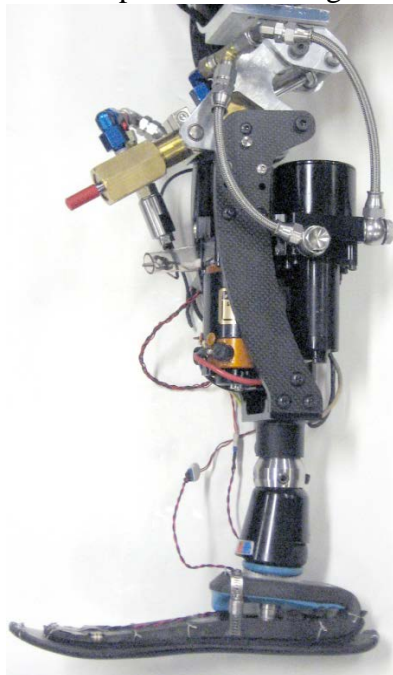


**Figure 22 : SEMG Sensors And Activation Signal Example**

The HAL5 control scheme is significantly different. The HAL5 measures the muscular activity of the user via surface electromyography (SEMG) sensors [73]. SEMG sensors are electrodes placed across a muscle on the surface of the skin that measure and detect the motor action potential caused by voluntary muscle contraction. This signal is highly complicated and nonlinear, depending on muscle length and duration of contraction [74]. But, it can be roughly equated to a measurement of the strength of the muscular contraction. The torque applied across the joint the muscle acts upon can be estimated using this measurement. The torque applied by the user can then be supplemented by an actuator on the exoskeleton at the same joint. It may therefore be possible to use a similar means of detecting muscle activity to control the torques applied by the prosthetic knee during swing.

### **3.1.5 First-Generation Prototype Hardware:**

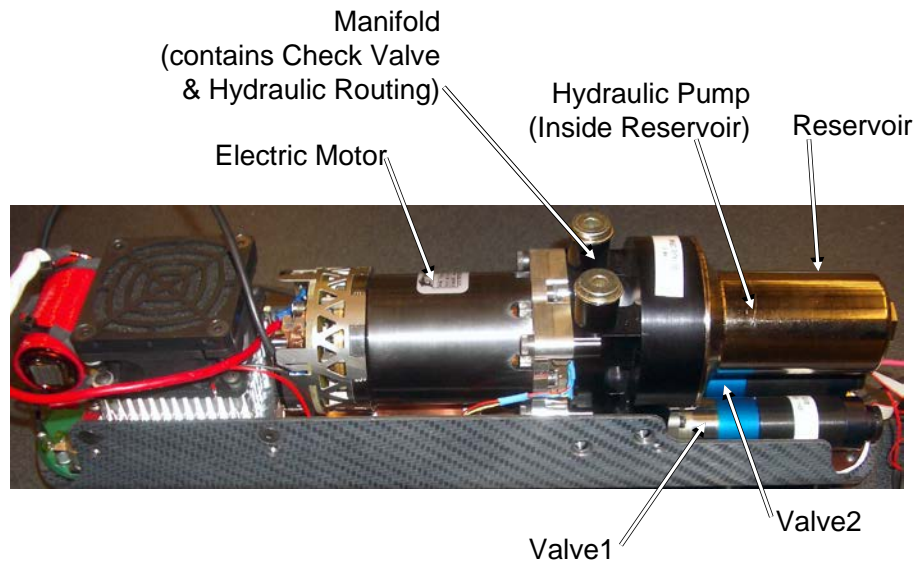
Previous work from colleagues in the HEL had been completed on this powered prosthetic knee project through the development of a first-generation prototype.



**Figure 23 : 1st Generation Prototype Knee**

As a majority of the sensors and actuators on the original prototype were used on the second-generation prototype, much of the topics mentioned in this section will be discussed

further in section 4. However, because a model of the system must first be defined to discuss the control theory, the important aspects are discussed now. This model of the system is used in section 3.2 to evaluate possible controllers.



**Figure 24 : ExoClimber Actuation System**

This first-generation prototype used an electro-hydraulic actuator, similar to that used in the Exoskeleton project, to power the rotation about the knee joint. This type of actuation system, from a controls theory point of view, is relatively simple to design for as the current applied through the motor is directly related to the force applied at the actuator. The system also utilizes a second DC rotary motor to servo the position of a hydraulic valve responsible for hydraulic damping and adding additional behavior to the hydraulics.

There are also many sensors integrated into the prototype as well. An optical rotary encoder is placed on the valve motor rotor to enable accurate servo positioning. A potentiometer is placed in the knee joint to measure the rotation of the shank relative to the thigh attachment. There are also two hydraulic pressure sensors placed in the hydraulic circuit to calculate forces acting on the hydraulic actuator during operation.

Additional sensors are located in the foot to give information of the pressures felt on the bottom of the foot during stance. These sensors are resistive based pressure sensors that measure the pressure acting across the area of the sensor. This data can be used to detect if the foot is on the ground and if the load is primarily on the heel or toe. This type of sensor is undesired for future prototypes because of the need to instrument the prosthetic foot. Because of the modular design of most lower-extremity prosthetics, it would be better to use sensors embedded directly in the prototype knee assembly so that all third party prosthetic feet can be used with the prototype [38][39][40][41][42].

### **3.1.6 Evaluation Criteria Goals:**

It is important to consider the way in which the powered prosthetic knee prototype's effectiveness will be evaluated so that once it is completed it stands the best chance at competing with other prosthetic knees on the market, both passive and active.

In order to compete with passive prosthetic knees the prototype must not hinder the momentary, subtle motions associated with everyday life [12][13]. For example, a person may only sit down or get up from a chair a handful of times throughout the day compared to the

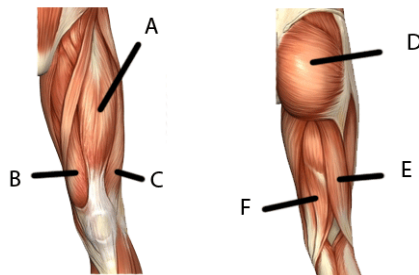
hundreds of steps he takes, but if he can not get out of a chair once he sits in it the device would be unusable.

Additionally, powered prosthetics have to compete with passive prosthetic knees for insurance money. Because of this, they simply can not replicate the existing benefits of passive prosthetic knees, no matter how vast the improvements are. They must provide something new that has a significant improvement on the quality of life for the patient [75]. To the insurance companies a prosthetic knee is a prosthetic knee. In order to stand out, it is argued that the powered prosthetic knee prototype being developed will reduce the energy needed for the amputee to walk and improve their comfort and safety so that the user will be encouraged to walk further and more often, therefore reducing co-morbidity. To support this claim, the prosthetic prototype will be evaluated on its energetic demands via a calorimetric measurement called VO<sub>2</sub> consumption [76]. This measurement will be performed to calculate the average energy use for a person using their current prosthetic knee and the powered prosthetic knee prototype for a constant speed on a treadmill, as was done in previously referenced studies of prosthetic knees [54]-[64]. It will therefore be important to assure that the prototype will perform especially well during steady state level ground walking.

To compete with other powered prosthetic knees, it is important to make improvements in battery life. This is accomplished by either using more batteries or using less power. Adding batteries increases weight and the size of the prototype. Both of which are unfavorable. It is therefore important to operate the prototype in the most efficient manner to increase battery life.

### 3.2 Goal and Proposal Evaluation:

A suitable control scheme must be developed that achieves the aforementioned objectives while capitalizing on the successes of the first-generation prototype; this would include utilizing all sensors and actuators already present.

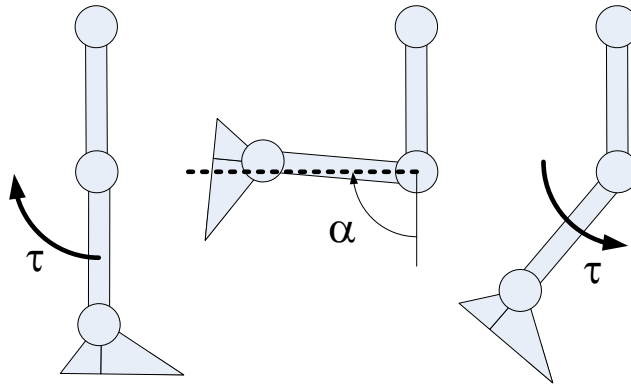


**Figure 25 : Thigh Muscles - A) Rectus Femoris, B) Vastus Medialis, C) Vastus Lateralis, D) Gluteus Maximus, E) Biceps Femoris, F) Semitendinosus.**

First we consider the control scheme suggested from research of the HAL5 exoskeleton [73]. Using SEMG sensors would be an exciting means of controlling the movement of the powered prosthetic knee prototype's actuated joint; however, it is unsure how successful a sensing method this would be on the residual limbs of amputees. The cause and extent of each amputation varies greatly from case to case. For example, the best option to control knee extension from a SEMG sensor would be to measure the activity of the either the vastus medialis or vastus lateralis muscle [77]. These muscles are the two of the four quadriceps muscles that only control knee extension and are accessible by the surface-based SEMG sensors on an able bodied person. After amputation, these muscles may not be present because they were part of the excised tissue. These muscles may also not be accessible to the sensors as the rectus femoris muscle covers much of these muscles in able bodied thighs. They also may no longer behave independent of other muscles as the common amputation practice is to suture the remaining

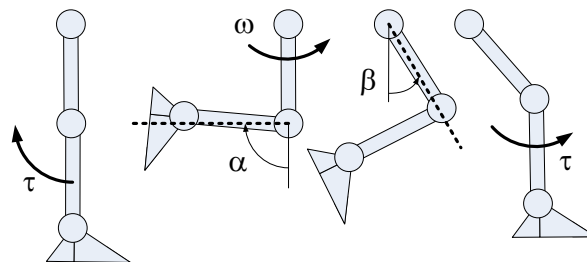
muscles together covering the residual femur [78]. Even if viable, depending on a suitable muscle to be present in the residual limb that the user could flex independently of the thigh muscles responsible for thigh motion, would make it so that the prototype could not be used by all amputees, significantly hindering a chance at penetrating the market.

The evaluation criteria of performing under steady state level ground conditions and the fact that we want the prototype to be able to be able to perform a variety of tasks dictates the need to create many different modes of operation, including a specific level ground walking mode for use during this steady state condition. The most straight forward means to do so is to implement a finite state machine for the high level control scheme which is responsible for detecting this walking condition and activating a walking mode [72]. The state machine is described in detail in section 5.



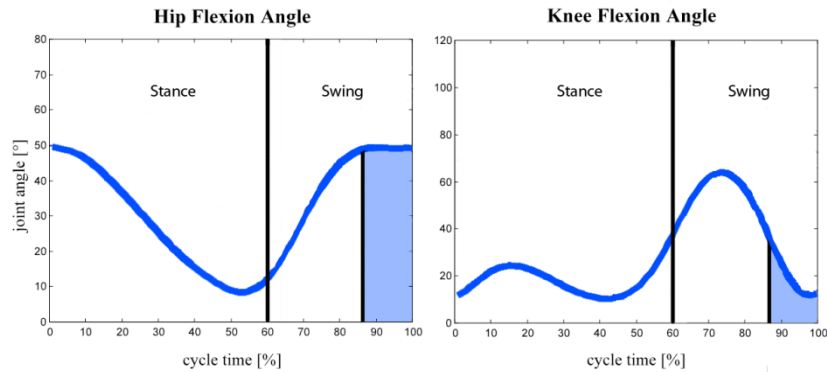
**Figure 26 : Initial Knee Control**

It is now important to develop a level ground walking mode that meets the previously stated goals. The original powered prosthetic knee prototype controller implemented only a very basic walking mode, but analyzing its operation provides a useful foundation to illustrate the process by which the final controller is developed. This original controller watched the foot pressure sensors to detect toe-off. Once toe-off was detected, the knee would bend by commanding a fixed current in the main actuator motor until the knee angle was less than a certain angle, seen above as alpha. Once this angle was reached, the knee was then extended by commanding a second current in the opposite direction until the knee was straight and the foot was on the ground. These two current command values and alpha were adjusted until the user was able to walk. This method, unfortunately, will only work well for the speed of walking the values are adjusted for. If the user walks slower, the knee will extend before it has swung beneath the body or if the user walks too fast, the knee will not extend fully when the user is placing his foot on the ground.



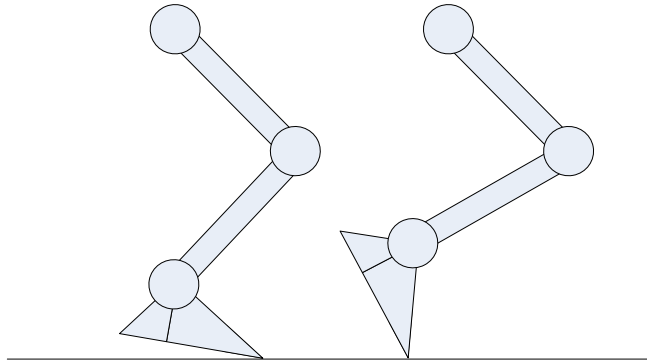
**Figure 27 : Modified Knee Control With Condition Beta. Torques are labeled  $\tau$ , while rotation is  $\omega$ .**

A simple solution for timing knee extension more accurately would be possible if the absolute thigh angle were available to the control system. The behavior of the powered prosthetic knee could then be to bend the knee to the angle  $\alpha$  and hold it there until the thigh passes beneath the body at angle  $\beta$ , as illustrated above. A means of measuring the absolute thigh angle is proposed via the combination of a rate moment gyroscope and an accelerometer [71]. This resolves the issue of the knee misfiring and extending too soon or too late. Unfortunately the knee bending and extension speeds for this method is still independent of the user's walking speed.



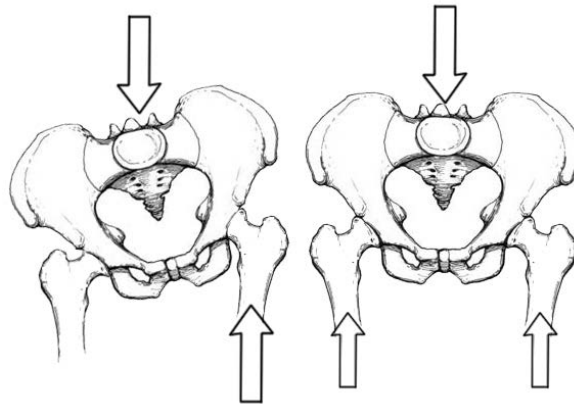
**Figure 28: CGA Data For Hip and Knee Angles. Area in Blue Shows Unusable Area.**

Another alternative means of controlling the knee angle is proposed by using Clinical Gait Analysis (CGA) data [79][71]. This source provides a reference for the knee angle and thigh angle as a function of time for level ground walking. The data is presented above. The angles are given as a percentage of the gait cycle, normalizing the data associated with the various speeds of walking for which the data was taken. Without knowing the speed of the user's walking it is not possible to use time to generate the desired values directly from this data. However, it might be possible to generate the knee angle as a function of the thigh angle by using time as the intermediate variable. Unfortunately, there are several points at the end of swing-phase where there is not a one to one correlation, shown above. So, this is not possible in this region. More importantly, an additional difficulty with this method is that for normal able-bodied walking the ankle also bends to help with toe clearance. Most prosthetic feet used in conjunction with prosthetic knees are also passive devices. They are typically spring-loaded at the ankle so the ankle can bend and flex during stance, but provide a restoring force to its original position [14]. This helps to cushion the impact of heel strike and provide a similar torque at toe-off to what the calf muscles provide an able-bodied person. The prosthetic foot is typically adjusted so that it is at approximately a right angle to the shank. This means that even if the method of bending the knee as the CGA data dictates were to work perfectly, the toe might still come into contact with the ground during swing-phase because the ankle will not bend as, illustrated below. The CGA data must therefore be modified to account for the lack of bending at the ankle by bending more at the knee. The amount of modification is not known explicitly, though it could be approximated by considering the geometric differences between the bending and non-bending ankle.



**Figure 29 : Different Amounts Of Knee Flexion For Mobile Ankle, Fixed Ankle.**

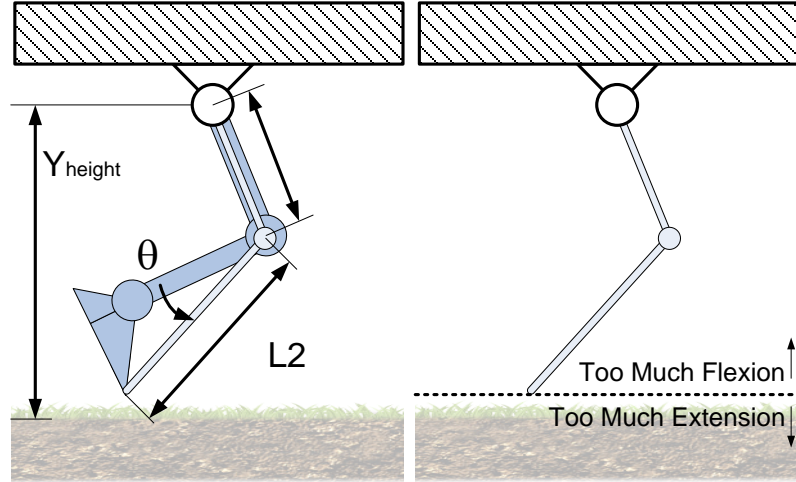
The complexities inherent in the above method necessitate the consideration of a simpler method. The Robotic Ground Avoidance method (RGA) is one such method developed in this thesis. It is based on the same concept of using the geometric constraints to find the desired knee angle in the CGA-modified method and is ultimately the method used for the powered prosthetic knee prototype.



**Figure 30 : Hip Adduction Seen During Single Stance Compared To Double Stance.**

The RGA method takes a robotics approach to the bending of the knee joint. To do so it assumes that the hip joint of the leg with the prosthesis maintains a constant distance from the ground while the opposite leg is in the stance-phase. This approximation is fairly reasonable due to the hip's abduction in the coronal plane combined with the pendulum model of walking [66].





**Figure 31 : Robotic Model Of Prosthetic Knee- L1 is the length of the thigh, L2 is the length from the knee joint to the toe, theta is the angle between the shank and L2, and Yheight is the vertical distance from the hip joint to the ground. Also shows Toe Location Compared to Buffer Line.**

An additional assumption is made that the ground is flat and parallel to this horizontal motion of the hip. Combining these assumptions and simplifying, the hip joint can be described as a rotary joint located a fixed distance ( $Y_{height}$ ) from the ground. Then, the leg can be seen as a two-link robotic arm and the ground becomes an obstacle to avoid collision with. With the lengths of each link known, it is possible to apply the robotics technique of inverse kinematics to find the thigh and knee angles required to position the end-effector of the arm at a desired position in X and Y coordinates [81].

The RGA method uses inverse kinematics to position the knee joint in a way that it prevents the toe from coming into contact with the ground. It accomplishes this by establishing a boundary line several inches above the ground as a buffer. This boundary line corresponds to the minimal amount of toe clearance required to safely avoid contact with the ground; if the toe travels any lower, the user runs the risk of stumbling and falling. Additionally, as described in our earlier goals, it is important that the swing behavior of the foot be the most efficient in motion and electrical power use. Therefore the boundary line also corresponds to the maximum amount of toe clearance desired; if the toe travels any higher, the system is creating excessive toe clearance that is not needed to safely walk. Combining these two cases results in the only desired leg configuration being those that result in the toe being exactly on the boundary line. Therefore, it is now possible to calculate the required knee angle if the thigh angle is known. The knee angle can be calculated directly from,

$$\theta_{knee} = \cos^{-1}\left(\frac{Y_{height} - Y_{buffer} - L_1 \cos(\theta_{thigh})}{L_2}\right) - \theta_{thigh}.$$

Derivation of this equation is shown in section 6.1.1. Once this reference angle is calculated, a position controller can be used to servo the knee joint to the desired angle, as detailed in section 7.

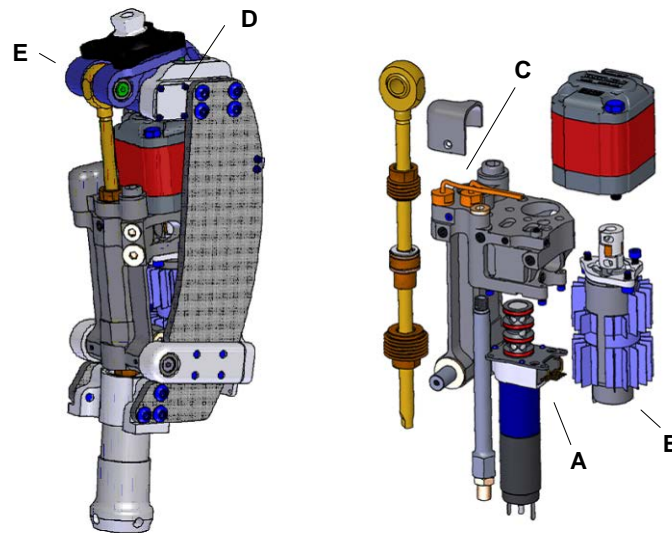
The RGA method also provides a theory for controlling the knee joint for types of locomotion where there does not exist CGA data. An example of this is the case of walking backwards on level ground.

### 3.3 Controller Design Conclusion

The RGA is used because it is the best proposed method for reaching the goals on level ground walking. While the RGA method is not used during the entirety of the swing-phase in the prototype, it establishes the foundation that for each mode in the finite state machine the control system uses a position based feedback controller to servo the knee joint to a desired angle calculated by a reference generator specific to that mode. The reference generators for each mode and their derivations are described in detail in section 6. The RGA, a novel method for controlling the knee joint of a powered prosthetic knee through the use of angular data sensed real-time from the residual sump, is a major contribution of my research to the current state-of-the-art in powered prosthetic knee technology.

#### 4. Prototype Hardware:

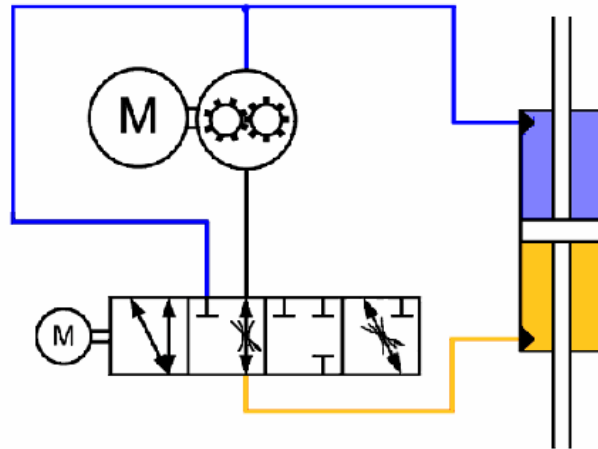
Previous work from colleagues in the HEL had been completed on this powered prosthetic knee project in the development of a tethered, first-generation, proof-of-concept prototype before the research presented in this thesis was conducted. Most of the controller design was completed on this first-generation prototype. A second-generation, untethered prototype was then created with several improvements to the mechanical system. Much of this work is detailed in Lambrecht [71].



**Figure 32 : Mechanical Model Of Prosthetic Knee With Exploded View. Sensors: A) Valve Encoder, B) Pump Encoder, C) Pressure Sensors, D) Magnetic Encoder, E) Rate Moment Gyroscope and Accelerometer Combination Sensor**

The prototypes use an electro-hydraulic actuator, similar to that used in the Exoskeletons, to power the rotation of the knee joint. This actuator uses an electric DC motor to turn a gear pump [72]. The pump creates fluid flow and pressure to drive a dual rod hydraulic linear actuator. Because the pump directly produces the pressures in the actuator, resulting in forces applied by the actuator, the hydraulic fluid reduces to the equivalent of a gearbox and transmission system between a rotary motor and linear actuator in a convenient package. This type of actuation system is therefore capable of capitalizing on the high mechanical utility of linear hydraulic actuators while maintaining the control advantages of using an electric DC motor. Additionally, the amplifier used to power the motor is capable of producing a desired current amplitude applied to the motor via an internal current controller. By using this type of amplifier, it is straightforward to provide a desired torque at the motor using the following

equation and the motor's torque constant:  $\tau_{motor} = K_t i_{motor}$ . This type of actuator is ultimately used on both prototypes because of aforementioned combined simplicities.



**Figure 33 : Hydraulic Circuit Diagram. Big M - Pump Motor. Small M - Valve Motor. Valve Positions From Left To Right: Open Both, Damped With Motor, Locked, Damped Without Motor**

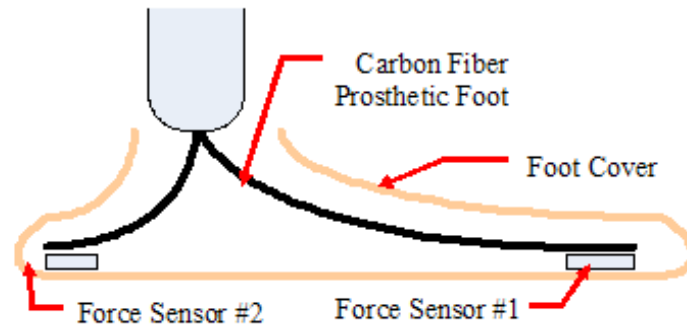
The first-generation and second-generation prototypes also utilize a second DC rotary motor to servo the position of a hydraulic valve responsible for providing additional behavior to the hydraulics. It is powered by a simple H-bridge circuit connected to a pulse-width-modulation output from the microcontroller. The valve is designed so that it can gradually open and close hydraulic paths resulting in two notable situations. The first situation connects the main motor driven pump to the circuit. It does so in a way that does not constrict the flow of fluid through the circuit. This results in the system described above where the pump's rotation directly correlates to the displacement of the linear actuator. In the second situation the motor driven pump is removed from the hydraulic circuit and the path of fluid from one end of the actuator is constricted through a small orifice whose size is controlled by the valve position. This results in the linear actuator being viscously damped by an amount determined by the position of the valve. This system of switching in and out the hydraulic pump via valves was originally developed for the Exoskeleton project and was modified by Lambrecht for the powered prosthetic knee prototype.

There are also many sensors integrated into the prototype. An optical rotary encoder is placed on the valve motor rotor to enable accurate servo positioning. The resolution of this sensor is  $1/14^{\text{th}}$  degree increments. A magnetic rotary encoder is placed at the knee joint to measure the rotation of the shank relative to the thigh attachment in  $1/16^{\text{th}}$  degree increments.

The previously mentioned RMG-accelerometer combination chip used to measure the absolute angle of the thigh joint is used in the first-generation and second-generation prototype. The first-generation has the combination chip attached directly to the thigh of the user. The second-generation has the combination chip integrated into the prototype knee at the point of contact between the thigh socket and the prototype knee.

There are also two hydraulic pressure sensors placed in the hydraulic circuit to measure the pressures. It is necessary to have two sensors because the type of sensors used only measure positive pressures accurately. By combining the signal it is possible to generate an accurate representation of the pressures acting on both sides of the cylinder and, therefore, the net force acting on it during operation.

It is preferential that accurate measurements of the magnitude of the forces acting on the prosthetic foot can be taken so that contact with the ground can be detected. A different method of sensing was implemented on each of the two prototypes.



**Figure 34 : First-generation Prototype Foot Sensors**

The first-generation prototype uses sensors located in the foot to give information on pressures felt by the bottom of the foot during stance. These sensors are resistive based pressure sensors that measure the pressure across the area of the sensor. As the pressure applied to the sensor increases, the resistive properties change and can be measured by our electronics. Two such sensors are used on the prosthetic foot; one is placed at the ball of the prosthetic foot, and the other is placed at the heel of the foot. By having the sensors in two separate sections it is possible to not only calculate forces on more of the foot, but to combine the two signals to differentiate if the load is primarily on the heel or ball of foot.

Additionally, data representing the forces applied to the foot during the swing-phase need to be obtained. The first-generation prototype did not have a direct means of measuring horizontal forces applied to the foot, only vertical forces from the pressure sensors. In order to infer the horizontal forces acting on the foot during swing, a dynamic model of the entire system of the user's residual limb attached to the prosthetic knee prototype was created.

The model, derived in appendix 1, is then used in combination with real-time joint angle data and inverse dynamics to observe disparities between the predicted behavior of the powered prosthetic knee prototype and its observed motion. The model-based prediction is accurate unless the foot strikes an obstacle, therefore, this can be used to detect a stumble with an obstacle.

In a study by Dumas et al, it was validated that using joint angles and inverse dynamics in such a way is equivalent to using direct measurements on the torques applied to the legs during level ground walking [80]. Using this method to detect a stumble in the case of a prosthetic leg, however, is a novel contribution to the current state-of-the-art in prosthetic devices that could be used in future prosthetic knees.

Unfortunately this method has its faults. It requires the model parameters to be calculated for each person, which may not be significant considering the prototype as a product that will only be used by one person. It will, however, require some sort of training period for each person and this added requirement for a specialist's time is undesired. Because of this, the second prototype is designed to directly measure the forces acting on the foot. It does so via three strain gauges built into the prosthetic knee's coupler with the shank. By combining the signal from the three strain gauges it is possible to measure the forces acting axially to the shank as well as moments acting in the coronal plane, removing the need to calculate the dynamic model. By combining the axial force and moment data the position of the load can still be determined as being primarily on the heel or the toe. Additional care has been taking to combine the two

vertical load sensors to cancel out effects seen due to moments in the sagittal plane. Using strain gauges in the powered prototype knee prototype has the added benefit of not requiring sensors to be placed on the prosthetic foot.

## 5. Finite State Machine:

Finite state machines (FSM), also known as finite state automation (FSA) or state machine (SM), are a means of producing the desired complex behaviors of a system through the use of a limited number of defined states. Each state dictates a simple behavior that the system enacts. Within each state, state transitions can occur based on changes in the circumstances of key variables. A set of rules are defined for each state that dictates the occurrence of these state transitions. Through the combination of the FSM and simple behaviors, complex behaviors can be produced. For example, a mobile robot can be programmed to navigate a maze through the use of a FSM and the simple behaviors of turn left, turn right, and move forward.

### 5.1 State Machine Theory:

The overall behavior of the state machine is codified in theory that I propose to be a means of predicting locomotion maneuvers called Origin Causality Prediction (OCP). The OCP method deduces the intended motion of the leg during swing-phase based on the location of the foot during toe-off. Because of the dynamic nature of human locomotion, at any point in time, each step and the torques required to complete it are somewhat tied to the previous step and the subsequent step. For example, before a person can take the first step from a standstill, weight must shift onto the opposite foot so that the other foot can come off the ground to take a step. If it can be reasonably assumed that the person will not perform a maneuver that would result in them falling over, each action gives clues to the next desired motion. A person won't pick his foot off the ground unless the other leg is ready to take the load. It is therefore possible to predict the location of the other leg and the desired maneuver of the prosthetic prototype by simply being aware of what the state of the prototype is at toe off.

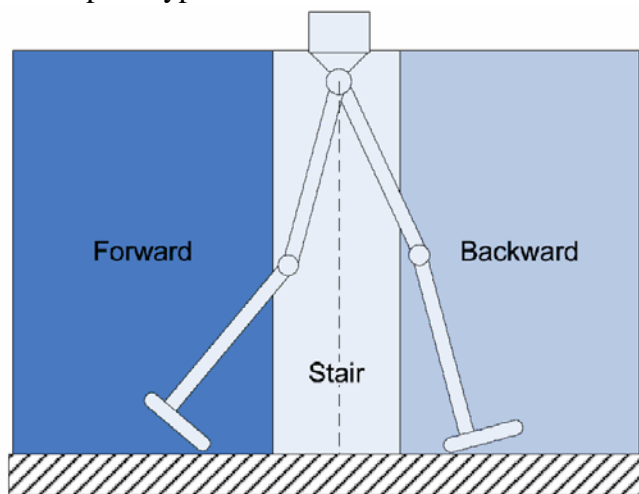


Figure 35 : OCP Transition Diagram From Stance To Swing

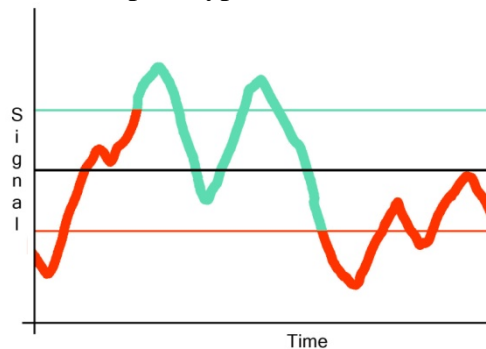
Conceptually, this means that if the foot of the prototype comes off the ground and it is behind the user, the other leg must be in front of the user and it must be in stance; therefore, the user is taking a step forward. Additionally, if the foot is in front of the user when it comes off the ground, the other foot must be behind and the person is taking a step backwards. If the foot comes off the ground directly beneath the user, this is an ambiguous stage where it is unsure what the person is trying to accomplish. The person can either be beginning to take a forward

step or a backwards step. There is also the possibility that the person is beginning to take a step up a stair. Ultimately, a combination of taking a stair step and a forward step is assumed for this case, while safety precautions are made should this be an incorrect prediction and a backwards step was desired.

A second OCP theory is proposed using similar reasoning for stance-phase. This theory states that based on the position of the knee entering stance-phase it is possible to predict the intended behavior of the knee during that stance-phase. It is argued that if the knee is already bent, that further flexion would result in the user falling to the ground. Therefore if, for any reason, the system enters stance while the knee is significantly bent, then the knee should straighten during stance-phase, powered by the hydraulic actuator. Additionally, if the knee is straight as the foot comes into contact with the ground, the knee should be allowed to bend while resistance is provided by hydraulic damping through the valve. This is similar to how a passive prosthetic knee provides resistance during stance.

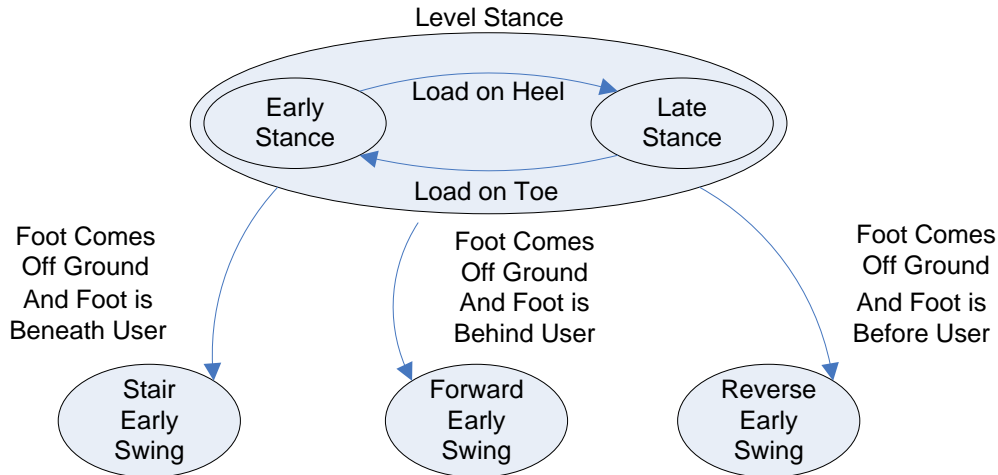
## 5.2 State Machine Details:

Implementing the OCP and the predicted maneuvers of forward, reverse, and stair swings through is accomplished through the creation of a finite state machine. The state machine is responsible for observing all of the sensors and determining which method of control should be implemented. A number of software flags were created to characterize the behavior of the sensor data to be used by the state machine to determine state transitions. The following discussion uses the sensors on the second prototype, but could easily be modified for the sensors present on the first-generation prototype. For instance, a flag was created to determine whether or not the foot was on the ground by looking at the signal from strain gauges. This could also be done with the pressure sensors of the first-generation prototype.



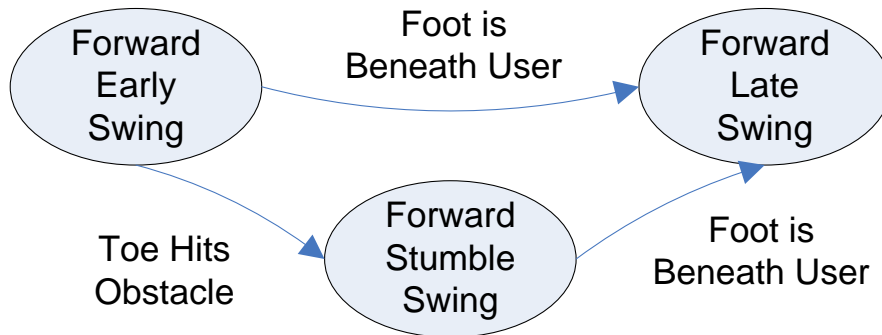
**Figure 36 : Deadband Threshold. Signal Starts Low (Red) Until Value Surpasses High Threshold (Green). Sensor Value Must Then Go Beneath Low Threshold To Turn Low Again**

If the vertical load is greater than some threshold value, the flag becomes active saying that the foot is now on the ground. When the value then goes below a similar threshold, the flag would return to saying the foot is not in contact with the ground. For this and many of the flags used, a minimum separation between the two thresholds is created that is larger than the amount of signal variation due to noise that is typically seen from that sensor. This reduced the likelihood of the flag switching between values due to noise in the signal.



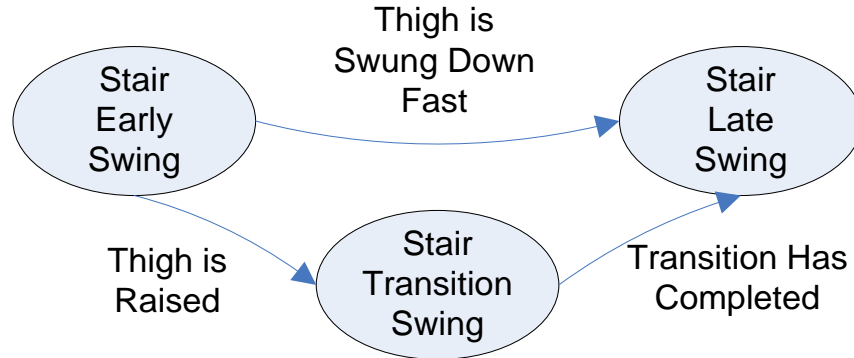
**Figure 37 : State Machine Transitions - Level Stance**

For the sake of simplicity, the description of the state machine will be discussed from the point of view as coming from stance-phase to swing-phase and coming from swing-phase to stance-phase, individually. This is to reduce the clutter seen in the figures in the space between the two major areas, stance and swing, as each swing-phase states can transition to both stance-phase states and each stance-phase state can transition to the three swing-phase states as proposed by the OCP theory. Additionally the special stance state of descent-stance and all of its subsequent states will be discussed at the end as they do not behave as the other states do.



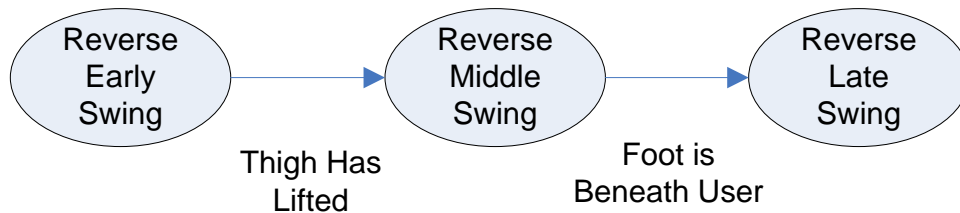
**Figure 38 : State Machine Transitions - Forward Early Swing**

From early-stance or late-stance, if the foot comes off the ground behind the user, the user is assumed to be taking a step forward and the state will transition to early-forward-swing. From the early-forward-swing state there are two possible state transitions, other than to a stance-phase state. If the foot detects that it has struck an obstacle, the state will transition to the stumble-swing state. If the foot passes unencumbered beneath the user's torso, the state will transition to the late-forward-swing state.



**Figure 39 : State Machine Transitions - Stair Early Swing**

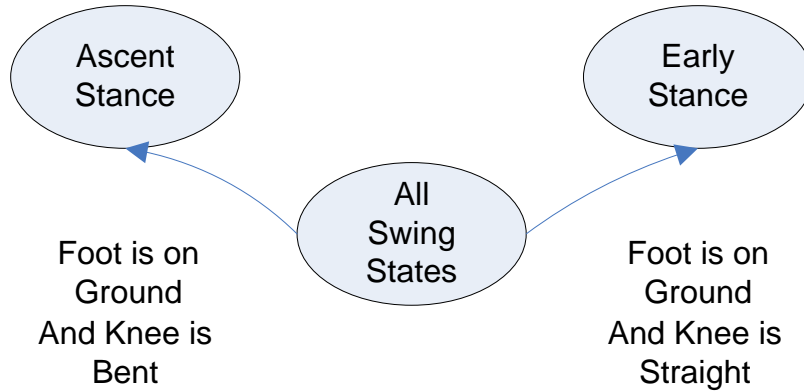
From early-stance or late-stance, if the foot comes off the ground beneath the user, the user is assumed to be taking either a stair step or a small step forward and the state will transition to early-stair-swing. If the system is in early-stair-swing, the state will move to stair-transition-swing if the thigh angle is raised past a certain point. This thigh angle threshold is set so that once the thigh has passed this angle, the knee joint could extend without the toe coming into contact with a stair, if it were present. Once the Stair-Step-Transition maneuver (discussed in section 6.2.2) completes itself, the state will transition to Stair-Late-Swing. This transition can also occur from Stair-Early-Swing if the thigh joint is rotated downward quickly. This secondary path is implemented as a safety precaution. If the thigh is rotated downward before this angle is reached, the person may have taken a small step and still requires the leg to be ready to support it in stance.



**Figure 40 : State Machine Transitions - Reverse Early Swing**

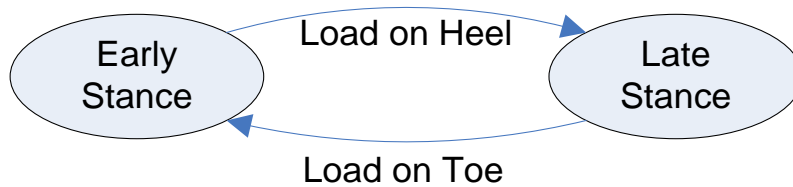
From early-stance or late-stance, if the foot comes off the ground in front of the user, the user is assumed to be taking a backwards step and the state will transition to early-reverse-swing. If the system is in early-reverse-swing, it will move to middle-reverse-swing if the thigh is raised past a certain angle. This angle corresponds to the amount of height the thigh must lift to transition from the heel being the closest point of contact to the toe being closest. For more information see section 6.3.1. Once in middle-reverse-swing, it will stay there until the foot passes beneath the user, where it will transition to late-reverse-swing.





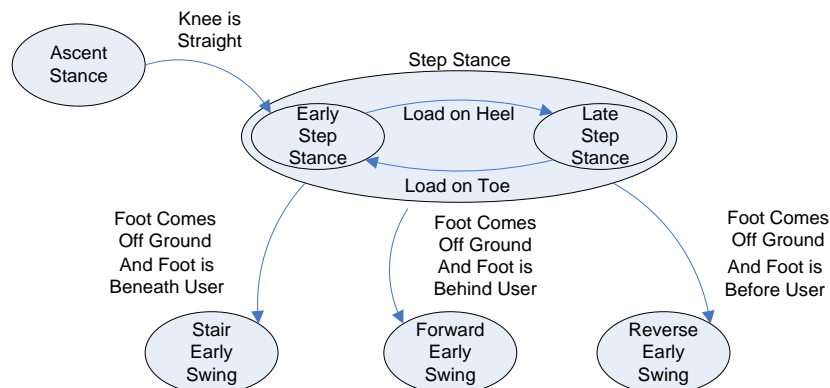
**Figure 41 : State Machine Transitions - Swing States**

From any of the above swing states, if the foot comes into contact with the ground there are two paths the state machine can transition to, ascent-stance and early-stance. If the knee is straight, it is assumed that the person has just completed heel-strike on level ground and the state transitions to early-stance. On the other hand, if the knee is bent, the user is coming into contact with ground that is higher than it was at toe-off and the state transitions to ascent-stance. This transition also occurs when the user plants the prototype's foot directly beneath him as the result of a stumble on level ground. In this case there would be no difference in height between toe off and contact, but extension assistance is still desired because the knee is flexed.



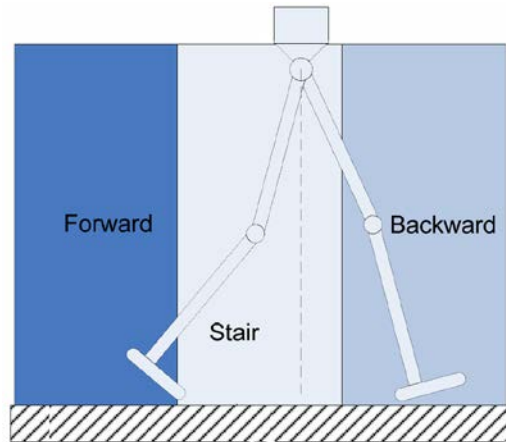
**Figure 42 : State Machine Transitions - Early Stance and Late Stance**

Following the state transition from swing-phase to early-stance, the state may then transition to a state called late-stance when the loads seen on the foot sensors change from being primarily on the heel to that of being primarily on the toe. The state can then transition back to early-stance if the load regresses again to being primarily on the heel. This cycle can repeat indefinitely until the foot comes off the ground.



**Figure 43 : State Machine Transitions - Ascent Stance To Step Stance**

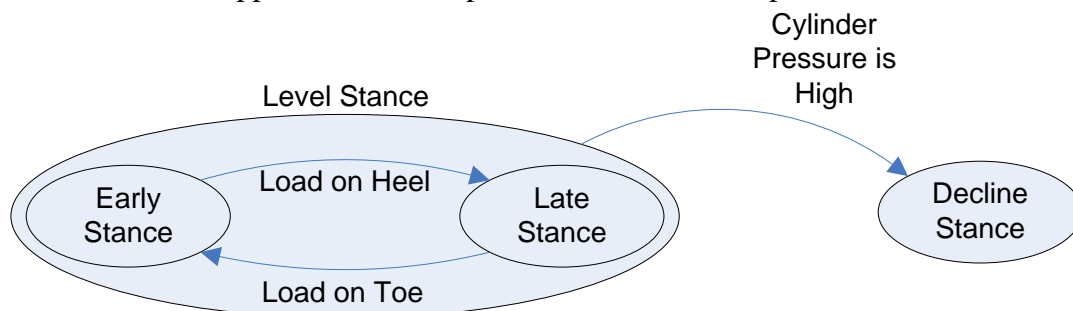
Following the state transition from swing-phase to ascent stance, the state will then transition to a state called early-step-stance when the knee has become straight. This state is essentially the same state as early-stance with the same transition path to late-step-stance. However, the transition thresholds that govern the behavior of which swing-phase will be chosen as the foot comes off the ground for these two states is adjusted such that it is easier to transition to the stair-swing state.



**Figure 44 : OCP Transition Diagram From Step Stance To Swing**

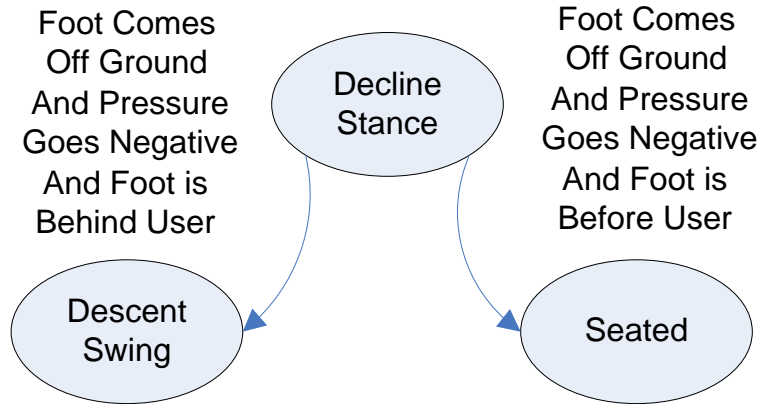
The reason for this change is to make it possible to walk foot over foot up a set of stairs. The second step up a set of stairs is difficult to differentiate from a normal forward step as the foot come off of the ground behind the user rather than beneath as done on the first step. By remembering that a step has just previously been taken, the current step can be reasonably assumed to be another stair step. Even if this assumption is incorrect and a forward step is desired, the maneuver is still capable of being performed because the knee will still bend to avoid toe contact with the ground during swing. This is due to the amount of flexion required for stair stepping being greater than that required for level ground.

Additionally, after transitioning from ascent-stance to early and late-step-stance, if the state transitions to the early-stair-swing state again, then the reference generator used for control of the knee in this state is told to avoid two stairs. The reason for this change is because the second stair step immediately after a single stair step is assumed to be traveling over a total of two stairs, the stair the opposite foot is on plus the next stair to step over.



**Figure 45: State Machine Transitions - Level Stance To Decline Stance**

When the foot is in either of the level stance states, the system will transition to the decline-stance state if the cylinder pressure is seen to go to a high value.

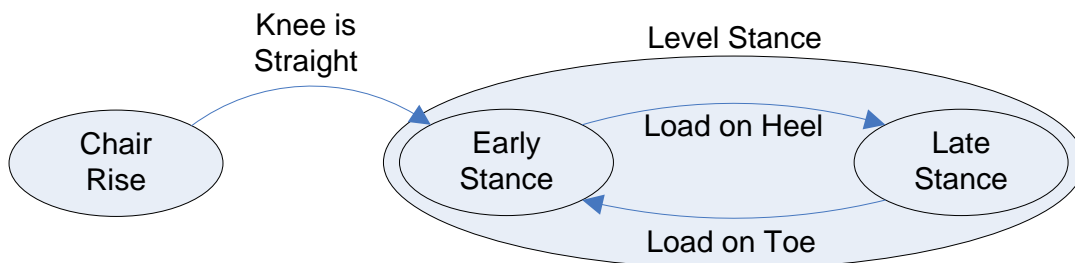


**Figure 46: State Machine Transitions - Decline Stance**

Once in decline-stance, if the foot comes off the ground and is behind the user, the system assumes that the user just descended a slope or stair and the state moves to the descent-swing state. The descent-swing state can then transition to any of the stance-phase states using the same OCP transitions the previous swing states used.

It is very difficult to lift the foot after entering decline-stance and have it be positioned in front of the user because of the position the knee must be in before the pressures seen can pass the required threshold. The knee is likely significantly bent and therefore using the dynamic stability assumption, the person must be walking backwards down a slope or performing some sort of lunge. This is a highly unlikely maneuver for any prosthetic knee to perform well in; therefore, it can be assumed that the user would not attempt this maneuver. It must then be assumed that the user is imparting an outside force to maintain stability. It is likely that the person is in a seated position and because their weight no longer needs support from the legs, their feet have come off the ground. So, if the foot comes off the ground and it is in front of the user, the system moves to the seated state.

If the system is in the seated state, it will transition to the chair-rise state if the foot is seen on the ground and beneath the user. The chair rise state is similar to ascent stance in that it provides extension assistance to the flexed knee joint. Therefore, being beneath the user is the important qualification to initiate chair-rise. If the user's weight is not situated directly over the feet when chair-rise is initiated, they would not be able to maintain balance. If the foot comes into contact with the ground and is not beneath the user, the system assumes that they are not wishing to rise from their chair. Once in chair-rise, the system will move to early-stance if the knee is straight.



**Figure 47: State Machine Transitions - Chair Rise**

### 5.3 State Machine Conclusion

The overall behavior of the state machine is codified in theory that I propose to be a means of predicting locomotion maneuvers called Origin Causality Prediction (OCP). The OCP method deduces the intended motion of the leg during swing-phase based on the location of the foot during toe-off. The OCP method also deduces the intended behavior of the leg during stance-phase based on the location of the foot-fall. This method of controlling the FMS based off of positional information is a novel technique and is a major contribution of my research towards the development of the state-of-the-art in prosthetic knee control technology.

## 6. Reference generators:

During each state previously discussed, there are two tasks that are performed. For each state a knee angle reference generator and valve angle reference generator must be activated. These reference angle generators then feed the desired reference angle to the feedback controllers discussed in section 7 to enact a desired maneuver. The passive states of Level-Stance, Stair-Stance, and Seated require no powered actuation and therefore no knee angle reference generator is described for these states.

### 6.1 Forward walking ref generators:

#### 6.1.1 Horizontal-Toe-Tracking

Horizontal-Toe-Tracking is the reference generator active for Early-Forward-Swing and Late-Reverse-Swing states. This reference generator commands the knee to an angle using the previously validated assumptions that the thigh joint acts as a fixed pivot joint a distance of  $Y_{ground}$  above the ground. We want the toe to track a horizontal buffer line, a distance of  $Y_{buffer}$  above the ground.

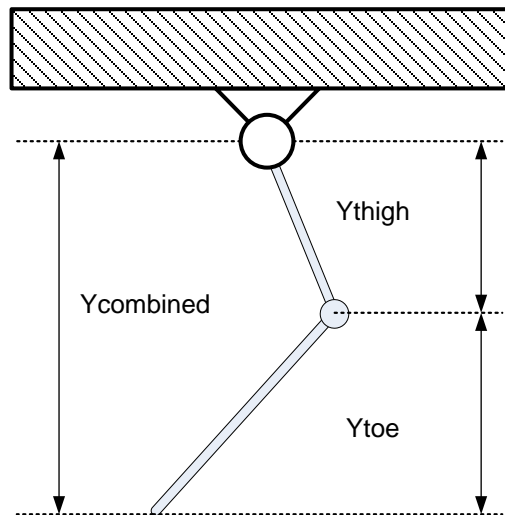
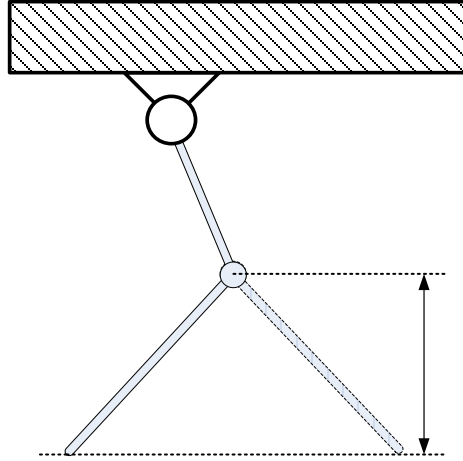


Figure 48 : Vertical Components Of Thigh and Shank

As can be seen above, the combined vertical components of the thigh and shank must equal the vertical distance to the boundary layer. With the vertical component of the thigh known, this leaves the thigh angle calculated directly from,

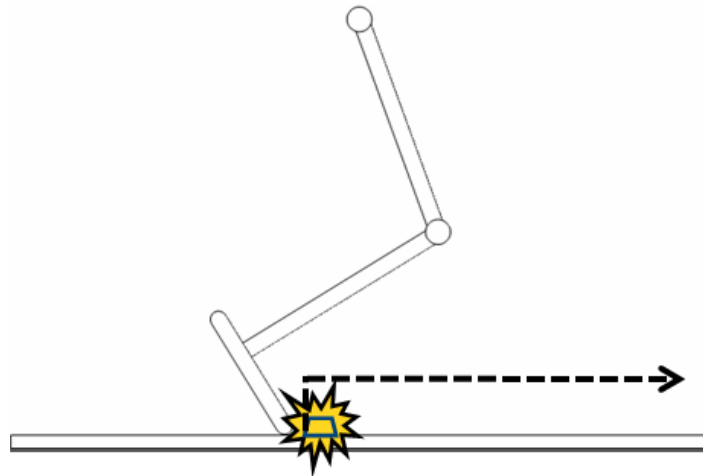
$Y_{height} - Y_{buffer} = L_1 \cos(\theta_{thigh}) + L_2 \cos(\theta_{thigh} + \theta_{knee})$ . This equation has two solutions; however, one of the solutions corresponds to the toe being in front of the knee, which does not occur during the early part of the swing-phase.



**Figure 49 : Two Knee Angle Solutions.**

Therefore it is possible to find a one-to-one equation for the desired knee angle given the thigh angle at any time, independent of walking speed. Using inverse kinematics, the equation of the knee angle required to position the toe on this line for a given thigh angle is:

$$\theta_{knee} = \cos^{-1}\left(\frac{Y_{height} - Y_{buffer} - L_1 \cos(\theta_{thigh})}{L_2}\right) - \theta_{thigh}.$$



**Figure 50 : Stumble Recovery. Buffer Line Is Raised After Contact Is Detected**

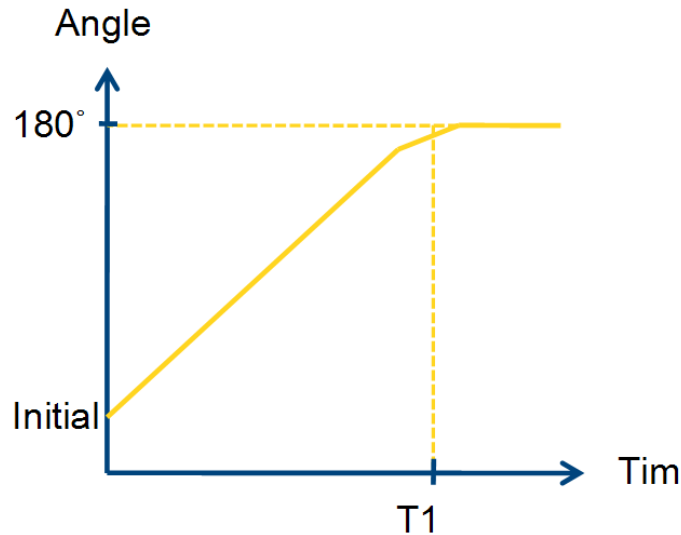
### 6.1.2 Stumble-tracking

This reference generator becomes active when an obstacle is detected in front of the toe during the early-swing state and the system is in the forward stumble swing state. The reference angle is calculated using the same equation for toe-tracking, however, the buffer line is raised by several inches so that the toe rises above the obstacle.

### 6.1.3 Speed-Based-Knee-Extension

This reference generator is responsible for extending the knee in a smooth fashion during the late forward swing state. The speed of this extension is designed such that if the user is walking quickly, the extension will occur at a rapid pace. The reference generator accomplishes this by first saving the knee angle and the rotational speed of the thigh angle as the state machine enters the forward-late-swing state. Conceptually, the equation governing this motion is split into

two parts. The first part is a linear interpolation equation between the saved, starting angle and final knee angle, a straight knee, as a function of time, occurring over one second. This is used in conjunction with the second part, a time controller that governs the passage of time in the first equation. The time controller uses the saved rotational speed of the thigh angle upon entering the forward-late-swing state in a linear fashion to amplify or decrease the speed of the extension in the first equation.



**Figure 51 : Knee Extension Function Plot**

The reason this reference generator is split into two parts is to facilitate the implementation of two safety precautions. First, the linear interpolation is smoothed out at the end of the prescribed motion to avoid a jarring impact as the knee slams to full extension. By doing this smoothing before taking into consideration the speed of walking, the smoothness is relative to the speed of walking and therefore the knee performs better when operated at high walking speeds. This is because at high walking speeds it is more important that the knee be fully extended before heel-strike than to reduce impact. Additionally, a precaution is implemented easily for slow walking speeds via the secondary time controller. If the prototype enters the forward-late-swing state while moving at a very slow pace, the time controller will implement a minimum speed extension so that the user will not have to wait for the knee to extend.

## **6.2 Stair Reference Generators**

### **6.2.1 Vertical-Toe-Tracking**

Vertical-Toe-Tracking is the reference generator active during the Stair-Early-Swing state. This reference generator commands the knee to an angle using RGA method and assumptions used in horizontal toe tracking. The desired trajectory, however, is to track a vertical buffer line, a starting at the point on the ground where the toe left it and extending vertically. The vertical line is displaced by Xoffset, the width of a stair, when the person is taking a second step up two stairs. Using inverse kinematics, the equation of the knee angle required to position the toe on this line for a given thigh angle is:

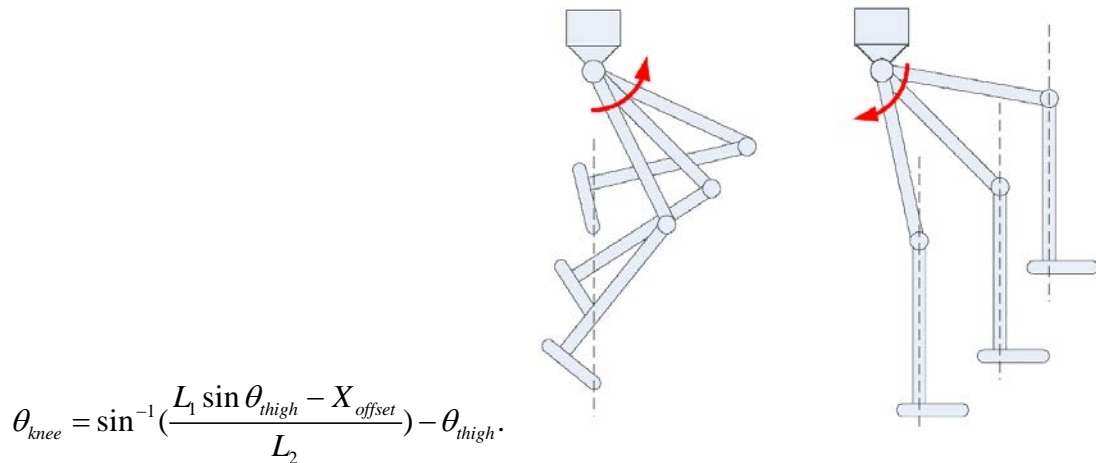


Figure 52 : Vertical Tracking And Absolute Angle Tracking

### 6.2.2 Stair-Step-Transition

Stair-Step-Transition is the reference generator used during the Stair-Middle-Swing state. This reference generator creates a smoothing motion between the previous Vertical-Toe-Tracking reference and the subsequent Absolute-Angle-Tracking reference. It does so in a similar fashion to the Speed-Based-Knee-Extension, by interpolating between the two references as a function of time. It does so with smoothing gradients applied on both sides of the linear interpolation as shown above. It should be noted, though, that rather than interpolating between two fixed angles in Speed-Based-Knee-Extension (the saved angle and straight), the Stair-Step-Transition continues to update the final reference angle (the one generated by Absolute-Angle-Tracking, 6.2.3) so that once the transition finishes the knee is at the desired angle, even if the user continued to move during the transition.

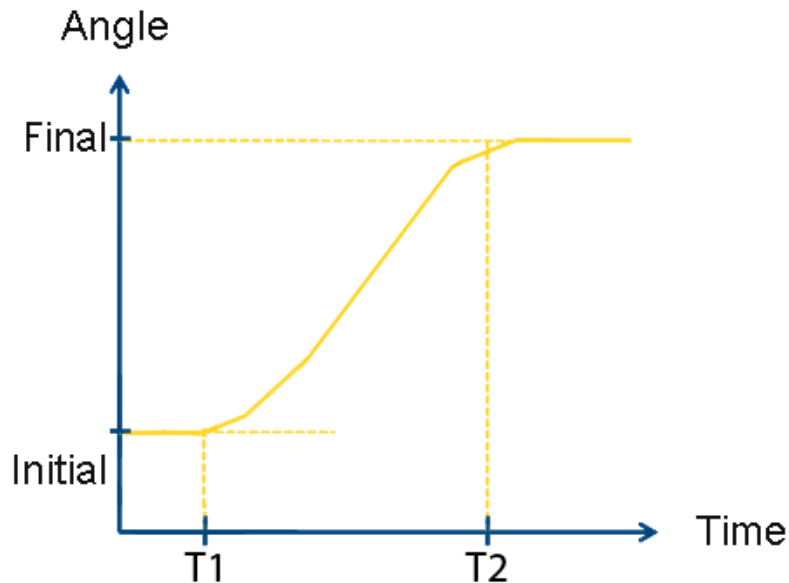


Figure 53 : Double Smooth Transition Plot

### 6.2.3 Absolute-Angle-Tracking

This reference generator commands the knee to an absolute angle using a fairly simple equation during the Stair-Late-Swing state. The knee joint's absolute angle is calculated by combining the relative angle between the shank and the thigh with the absolute angle of the thigh relative to ground. The equation of the knee angle required to maintain an absolute knee angle for a given thigh angle is:  $\theta_{knee} = \theta_{thigh} - \theta_{absolute}$ .

### 6.2.4 Fixed-Speed-Knee-Extension

This reference generator is similar to Speed-Based-Knee-Extension in that it is responsible for extending the knee in a smooth fashion. However, the speed of this extension is done at a fixed rate rather than as a function of the user's walking speed. The rest of the equation remains the same. The Fixed-Speed-Knee-Extension reference generator is active during Reverse-Late-Swing, Ascent-Stance, and Chair-Rise states.

## 6.3 Reverse Walking Reference Generators

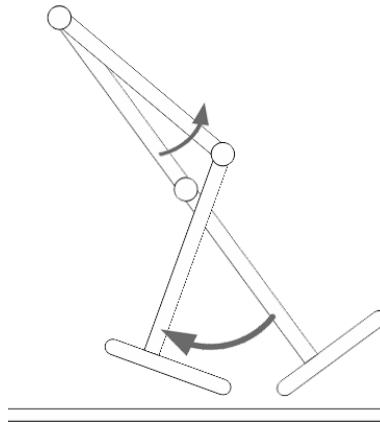


Figure 54 : Heel-to-Toe Transition

### 6.3.1 Heel-to-Toe-Transition

This is the only transition reference generator that does not explicitly use time in any way. Upon entering the reverse-early-swing state, the knee angle and thigh angles are both recorded. Additionally, the point at which the prototype would be upon entering the subsequent reverse-middle-swing state is calculated. This point corresponds to the link between the knee joint and the toe being purely vertical. For this point, the knee joint and thigh angles are calculated and stored using the same method of inverse kinematics used in the Horizontal-Toe-Tracking reference generator. Once these two data points are stored, a linear interpolation is created between the two and the equation is solved generating the knee angle as a function of thigh angle.

### 6.4 Valve Position Reference

The valve reference generators set the desired angle for the hydraulic valve. These desired angles are constant values corresponding to the different cases shown in section 4. The specific cases are shown in table 1.

Table 1

Valve Position	State
No Damping	Seated



Low Damping	Late Stance
Medium Damping	Descent Stance
High Damping	Early Stance
Open Full with Pump	All Swing States + Ascent Stance

## 6.5 Conclusion

The reference generators discussed operate based on the robotics approach of inverse kinematics. Successful application of this approach towards the control of a powered prosthetic knee is a novel contribution to the current state-of-the-art technology.

## 7. Feedback position controllers:

### 7.1 Introduction

The reference generators in section 6 produce the desired knee angle for each maneuver the prototype completes. It is the feedback controller that is responsible for ensuring that the motor is driven in such a way that the knee is positioned to the reference angle. To accomplish this a non-linearity compensated feedback controller is designed and implemented. The necessity and motivation for using this specific controller is established in modeling the system that is to be controlled. The derivation of this model and the design of the controller to enable accurate servo positioning of the prosthetic knee angle are a further contribution to the state-of-the-art in the control theory of powered prosthetic knee devices.

### 7.2 Model Derivation

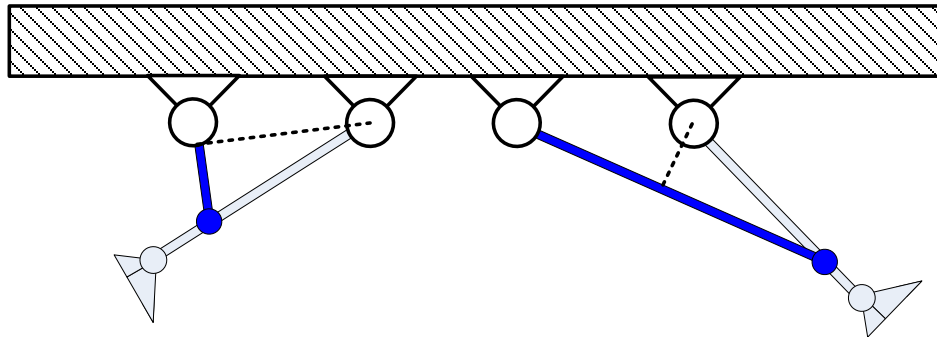
The idea of controlling a prosthetic knee joint attached to a person is an inherently complicated problem because of the person's presence within the system. Any torque that is applied across the knee joint to rotate the shank is also applied to the thigh coupler and subsequently enacts forces on the user's thigh. It is therefore important to consider how the thigh of the user reacts to forces applied from the prototype.

From a systems point of view, the person has a number of sensors, actuators, and feedback controllers to position the thigh at their desired angle. Without going too much into detail of how the body controls its motor functions it is possible to split the behavior into two noteworthy parts, conscious and reflexive. The conscious portion is the high level behavior of the muscle or joint. It is roughly equivalent to performing as a position controller (hold my arm out at this angle) or a force controller (push against the door with a specified amount of force to open it). The reflexive behavior is the low level behavior responsible for the behavior of the individual muscles to elicit the behavior needed from the high level system. This system is what holds the muscles at the desired lengths to create the proper tension in the muscle. This is what a doctor test when he strikes your knee just below the kneecap. The muscle stretches due to the impact of the hammer and adjusts to compensate. However, because the stretch of the muscle was only momentary, the muscle is seen to over-react and the knee kicks forward. This is called the patellar reflex. Because of these combinations of independent feedback controllers, it is difficult to properly model the system of the human controlled thigh joint. Indeed, accurately modeling the human motor system in the sense of control theory is the subject of much research in the field of haptics. Therefore it is necessary to make some assumptions to simplify this part of the system.

It has been observed that if the person can apply forces in parallel to an actuated system that the person will act in a stabilizing manner. It is therefore possible to simplify the muscular-actuation system by ignoring the direct effect of the muscular actuation. The muscles therefore reduce to a set of springs and dampers acting in parallel. The resting angle due to the springs forces are equal to the intended angle from the person. Unfortunately, even this system, as simple as it is, still has too much variability due to the fact that the muscles' stiffness is specific to each person. Because of this and the above mentioned difficulties, it is decided to forego modeling the thigh of the person. To still account for the behaviors of the thigh, the design of the controller will be adjusted to be stable given a larger amplitude disturbance.

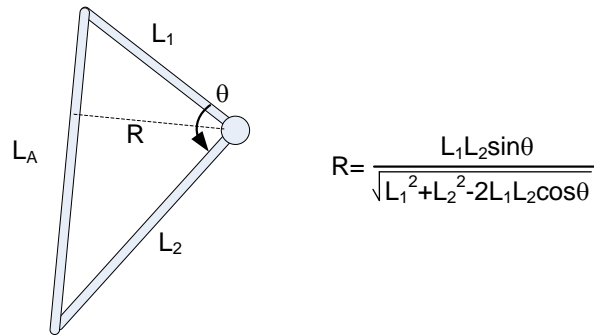
The development of the position controller now treats the thigh joint as an object tied directly to ground. This simplifies things greatly; however, there are still many complexities in the remaining system. Specifically, the remaining complexities are the nonlinear application of torque from the actuator and the torques due to gravity. Each of these must be dealt with.

The hydraulic system does not directly apply torques from the motor. As discussed earlier, the electric motor is attached to a hydraulic pump. This pump pushes fluid through a manifold to a linear hydraulic actuator. The manifold and the cylinder of the actuator can be assumed to be sufficiently stiff and leakage is significantly low such that the torque applied at the motor is linearly related to the force applied by the actuator.

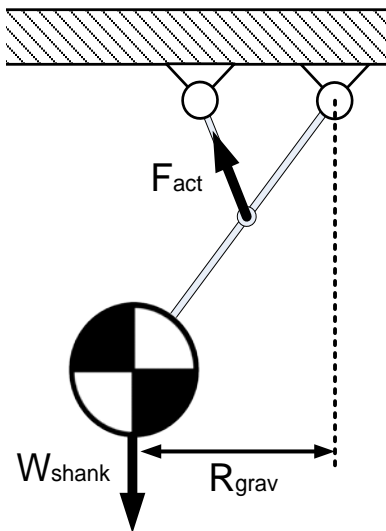


**Figure 55 : Exaggerated Modification Of Moment Arm As Knee Join Expands**

This actuator operates across the knee joint as shown above, and therefore does not apply a torque directly. The actuator applies a force across a moment arm that changes length as the knee joint angle changes. This moment arm is easily calculated using the known geometry of the knee joint and where the linear actuator connects to the shank and thigh coupler. This equation is shown below. The moment applied is force multiplied by this moment arm length. This equation can easily be converted to find the required force given the desired moment and the angle of the knee joint. It is this equation that is used to circumvent the problem caused by the nonlinearity as a form of feedback linearization.

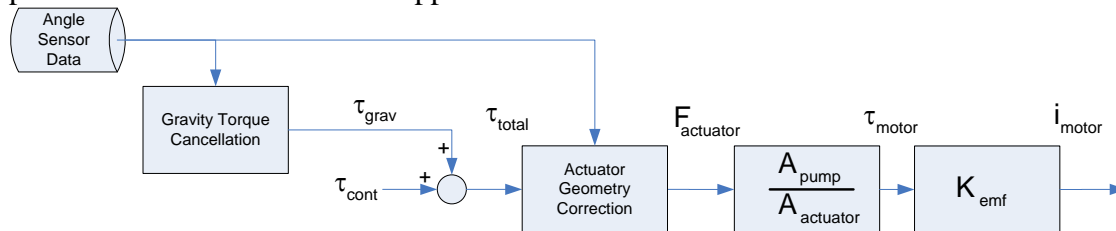


**Figure 56: Moment Arm Calculation.**



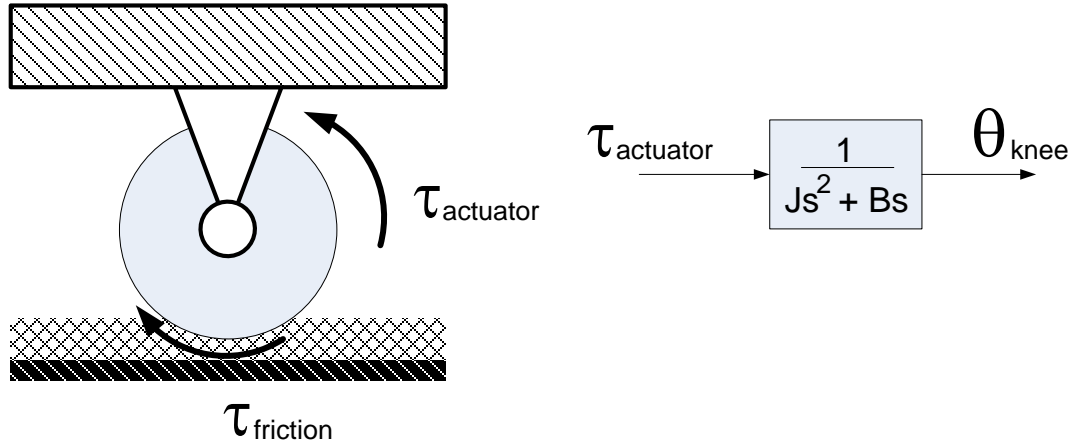
**Figure 57 : Gravity Torque**

An additional nonlinearity is that this system is operating in the vertical plane and therefore the elements are acted upon by gravity and do not move purely on their own. The gravity forces are seen to be applied at the center of gravity of each link in the leg system. The resulting torques applied to the shank change as a function of the cosine of the absolute knee angle. They are therefore nonlinear in nature and must be accounted for in the controller design. This is accomplished by using the explicit equations to solve for the nonlinear torques and remove their effects by applying a counteracting torque in the opposite direction. It is important to note that this counteracting torque must also be fed through the earlier discussed nonlinear compensator to calculate the force applied at the actuator.



**Figure 58 : Linearizing Compensators Block Diagram**

The nonlinear compensators are diagramed above showing the desired torque from the yet to be discussed feedback controller outputting the desired force to be applied from the actuator. Additional function blocks can be added to calculate the desired current to be applied at the motor. These blocks are simply linear multipliers derived from the pump displacement, hydraulic actuator area, and motor torque constant.

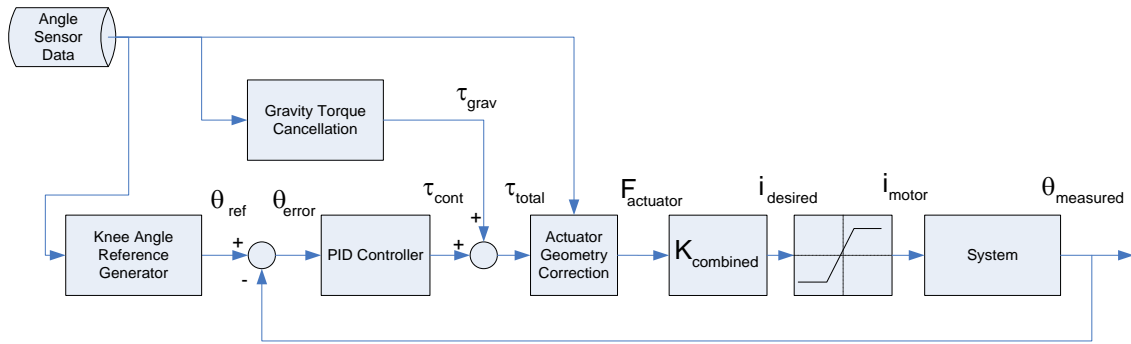


**Figure 59 : Linear Rotational Inertia Model And Transfer Function Block Diagram**

It is important to then come up with a model of how the remaining system behaves. If we demand a controller torque,  $T_c$ , from the controller, we see that after applying the gravity compensation torque and geometric correction, the knee system simply becomes that of a rotational inertia being acted upon by the controller torque. We make the addition that there is a viscous friction torque also acting on the system. The magnitude of the viscous term is unknown; however, it is included for the sake of determining what type of feedback controller to use. The transfer function for a viscous damped rotary inertia acted upon by an outside torque with the output being the knee theta is shown above.

A linear system of this type, a two pole system with one at the origin, is easily controlled by a PD controller. A simple root locus can verify this claim. As was done yielding satisfactory results on the ExoClimber project [72], an integrator term is added to the controller to overcome un-modeled static friction. This integrator is limited to a maximum value through a saturation function. The magnitude of the maximum value is adjusted on the working prototype to a value such that it does not overcome and exceed the static frictional effects. It is set to a value which is slightly less than the value needed to overcome it. This does not remove the static friction entirely, but serves to reduce the magnitude perceived and therefore reduces the loss of accuracy typically seen due to the static friction nonlinearity [82][83].

An additional nonlinearity is added to the system to ensure the safety of the motor. To protect the motor, the current applied through it is limited to its maximum sustained current, via a software saturation function. As seen in similar systems, a saturation nonlinearity of this type does not result in instability, but causes delays in settling time. The complete system model is shown below.



**Figure 60 : Complete Controller Block Diagram**

A noteworthy consequence of using relatively simple feedback controller is that a simpler microcomputer can be used to run the software. This results in smaller electronics and reduces the overall size of the powered prosthetic knee prototype.

A simple PID controller is used for the valve servo controller. This is possible because of the linear nature of the valve-motor system.

## 8. Conclusion

As stated previously this powered prosthetic knee prototype was not developed to be just an academic endeavor, but also to be a product that could eventually be marketed and sold to the above knee amputee population. To this end, the technology developed through this research was pitched to prosthetic knee manufacturers and evaluated on amputee test users. In total, the prosthetic knee prototype was tested by three above knee amputees.

Most of the development, however, was not performed on amputees as we did not have the resources available. The initial development was conducted using an adaptor so that an able-bodied person could wear and test the prototype as if they were amputated above the knee. Once development was completed, the prototype was first tested by an amputee at a facility owned by Otto Bock in Wisconsin. The second amputee testing was completed in Otto Bock's headquarters in Duderstadt, Germany. After these two meetings to test the overall viability of the system, we conducted several experiments using a treadmill with a local amputee to verify and support several design decisions.

### 8.1 Observational Conclusions

First we evaluate the prototype's successes in the goals stated in section 3.1, via amputee use observation. To establish a priority for the evaluation, the initial goals are the most important goals discussed and any of the subsequent goals should be interpreted as secondary goals that are primarily useful for driving development rather than to evaluate failure or success.

As a whole, the prototype is very successful, particularly on level ground as desired. Through testing with amputees it succeeds at providing support during stance on level ground via energetically-passive hydraulic damping. It also provides sufficient flexion of the knee joint on level ground to maintain toe clearance during early swing and extension of the knee joint during late swing to ensure a straight leg at the moment of heel strike. This is done smoothly through powered actuation of the knee joint. Walking on level ground mimics able-bodied locomotion and can therefore be performed at various different speeds from a slow crawl of 0.5 mph to a maximum tested speed of 3 mph. At some faster speed the actuator will no longer be able to flex and extend the knee joint fast enough to maintain the walking motion; however, testing at these various speeds shows that the principles behind the maneuver is sound.

Also, through the implementation of the OCP theories and the finite state machine, the knee is able to perform varied maneuvers like walking backwards or climbing up stairs without the need of any direct input from the user. The user is even able to get up and down from a seated position without difficulty.

There are, however, several difficulties that result from the disparities of developing through working with an able-bodied person yet testing on an amputee that present themselves when initially attempting to perform maneuvers other than walking. Many of these difficulties arise from the perhaps misguided design philosophy of trying to make the knee behave as much as possible like the user's original knee. The reasoning behind this assertion is that amputees who have used passive prosthetic knees for an extended period of time no longer use their residual limbs in the same way as they would before amputation. This is the result of their relearning how to walk with their passive prosthetic knees and ankles. While this may seem obvious, it means that they no longer perform maneuvers in the traditional manner. Their intuition is no longer the same. For instance, the concept of a person having a dominant foot may be rewritten through the trauma of amputation. This can be illustrated by considering a person's right-handedness after an upper limb amputation on the right arm. After the amputation, the person will understandably learn to reach for objects with his left hand first rather than what was his previously dominant arm. The similar concept of a person's dominant foot, the foot he leads with when taking the first step, may also become the un-amputated leg and this would be perfectly reasonable. Therefore because many of the controller design decisions were based on the intention of restoring pre-amputation motion it can no longer be claimed that the use of the prototype is entirely intuitive to a post-amputation user.

One such difficulty occurs when the amputee takes a step but does not firmly plant the foot when it is placed on the ground. For example, when a very short forward step is completed, the state machine will have transitioned from early to late forward swing, ending at early stance when the foot is placed on the ground. If the user does not firmly place their foot on the ground and subsequently loses contact, the state machine will see that the foot has just come off the ground and is in front of the user (the conditions required to transition to early reverse swing). In this state the knee will try to bend, expecting the thigh to swing posterior for a reverse step. This is not only undesired but potentially dangerous as the knee joint will flex at a moment where it is needed to be fully extended.

A similar occurrence happens during standing where the amputee will shift all of their weight off of the prosthetic leg to reduce discomfort from the skin pressures induced by the socket on the residual limb. The result is that the prototype sees the foot come off the ground and assumes the person wants to take a step, erroneously.

Another example of difficulties arisen from the differences between development testing and amputee testing was when stair climbing was attempted. First, the amount of mass and rotational inertia differences between development testing and that of the amputee were very large. During the development, the un-amputated tester still has the entirety of their leg to resist the torques applied by the prosthetic knee actuator. It was in this setting that the gains for the feedback controller were adjusted for stability. Therefore, it should be obvious that once the amputee wore the prototype, with their reduced mass and inertia, that the feedback controller would provide too much corrective torques and the joint would have difficulties with stability. This was originally accounted for; however, the hidden feedback loop between thigh angle and knee angle due to the reference generators in section 6 resulted in far more significant instabilities than were expected for the specific case of ascent swing.

A second instance of stair climbing difficulty occurs due to a fundamental difference of intuition between development testers and actual amputees. Because the amputees whom tested the prototype were amputees of several years, they had already fully adjusted to ascending stairs with passive prosthetic knees. The most common method of doing so was by completing repeated, single steps with their good leg. Each step would begin with them planting their good leg on the stair and stepping up through extension of their knee while simultaneously abducting their amputated leg slightly and rotating it laterally to avoid contacting the stair with their prosthetic foot. This is the way many able bodied people will approach the task of climbing stairs if told to artificially maintain one of their knees at full extension. When using the powered prosthetic knee prototype and told to ascend stairs step over step rather than one at a time, they would continue much in the same fashion they did with their passive prosthetic. They would first lead with their good leg, swing it up, and then place it on the step. Then, rather than through a combination of extending the planted leg's knee while simultaneously raising the opposing thigh, as done in able bodied stair climbing, they would extend the knee fully, stop, and then raise the opposing thigh. Because the trajectory of the reference generator for stair climbing was meant to clear this step as the thigh rotated, this resulted in very unnatural motion when combined with the instability previously discussed.

Alternatively, the reverse level ground walking maneuver was performed by amputees with almost as much relative ease as they adopted the forward level ground walking maneuver, much to their excitement. This maneuver was designed to mimic the pre-amputated motion of a reverse step so it would be expected that the amputee would have similar difficulty, but they did not. The reason for this disparity is unknown. It may be due to the fact that taking a step backwards is a maneuver infrequently performed, especially for above-knee amputees. Because of this they may not have formed such a strong muscle memory for walking backwards with their prosthetic and could therefore easily adopt the maneuver required for the powered prosthetic knee prototype.

Another instance of the amputee not being able to easily revert to normal locomotion patterns occurred, during the in-house testing on the treadmill, when we were attempting to show that the prototype reduced the phenomena of "hip-hike", a co-morbidity where the amputee adducts the thigh of the stance leg at the hip rather than abducting it as done in normal human walking. Hip-hike is a mechanism by which the amputee can compensate for the knee joint not bending enough and create additional toe clearance. However, when the amputee uses the prototype and sufficient toe clearance is created through the actuated flexion of the knee joint, the hip of the user will still "hike" during stance. Even when the variables of the controller are exaggerated to create excessive toe clearance, the motion of the hip did not change. The amputees could be coached to drop their hips during testing, but the behavior regressed when attention was not maintained.

A second in-house experiment was conducted to verify that the use of our prototype during level ground walking would reduce energetic consumption and therefore increase the amount of walking that a person could complete daily with the prototype over passive devices. To test this claim, the amputee was instructed to walk on a treadmill for 10 minutes with the prosthetic prototype knee, his own prosthetic foot, and his own shoes. This was followed by 10 minutes with their own prosthetic knee, foot, and shoe. Both of these experiments were completed with the amputee hooked up to a VO<sub>2</sub> machine to measure calorimetric costs. The following week, the amputee conducted a similar test, this time with his prosthetic knee first followed by the prototype knee. This was done in an attempt to remove any procedural bias. The

results of this test, unfortunately, show that there is no noticeable reduction in VO<sub>2</sub> costs using the prototype knee over the amputee's original prosthetic knee. This likely means that any benefit resulting from power being inputted at the knee is negated by the added weight of the prototype. It has been shown that adding weight to the legs of able bodied individuals increases the energetic demands of walking [19]. This result can therefore be reasonably applied to amputees and the weight of their prosthetics. It is thus argued that because the prototype is significantly heavier than the user's passive prosthetic knee, while the net result of using the prototype is no reduction of energetic costs, then it is possible to claim that the concept of reducing energy due to the use of a powered prosthetic device is inherently sound. To obtain an observable reduction in calorimetric costs all that would be required would be the reduction of the weight of the device while maintaining functionality.

The claim that the prototype would be safer through the implementation of a stumble recovery program was not experimentally tested on an amputee. To do so would require the user to walk on a treadmill while an experimenter purposefully tried to trip them. This test is inherently unsafe, so tests were completed using the adaptor on an able bodied person to simply verify that the stumble recovery mode would be properly triggered when a stumble was encountered. One test was completed using a stationary obstacle on the ground that would strike the foot of the prosthetic prototype during swing and resulted in the proper activation of stumble recovery.

Another verification of proper activation occurred by accident while testing robustness of the controller through repeated walking on the treadmill with the prosthetic prototype. The parameters used in early forward swing-phase were improperly adjusted so that they did not provide sufficient toe clearance resulting in the foot scuffing the ground during swing. If this amount of frictional force between foot and ground became significant, it would also trigger stumble recovery and create additional clearance as desired.

## **8.2 Numerical Verification**

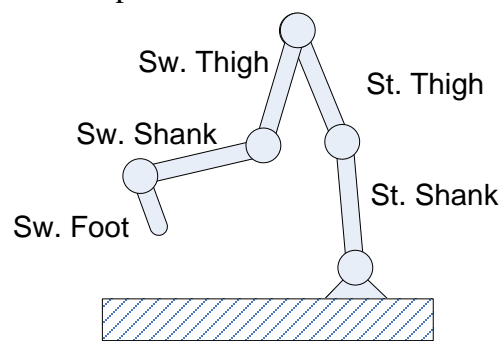
Outside of experimental observation that the FSM and reference generators produce a prosthetic knee that can restore natural walking behavior and extend functionality beyond passive prosthetics to maneuvers that only an active prosthetic can accomplish, numerical validation is also possible.

In a study by Riener et al on human gait kinematics and mechanics, the motions of the legs of people walking at self-selected speeds were observed and recorded. This data provides a foundation on which to derive and understand how the fundamental physical interaction of muscles and joints produce the motions seen in typical walking on level ground [79]. Specifically, this is done by measuring the relative angles of the foot to the shank, the shank to the thigh, and the thigh to the torso. The data presents these relative joint angles, normalized against time, as a percentage of the gait cycle. The first 60% of the data encompasses stance while the final 40% denotes swing. The motion of the opposite leg during this motion is easily found by using the original data offset by 50% of the gait cycle. The combination of these two signals results in two periods of double stance, from 0%-10% and 50%-60%.

It is possible to use the recorded angle data in combination with knowledge of human geometry and average limb length to use a robotics approach to determine the position of each joint in the sagittal plane during the walking cycle. Of particular importance to the numerical verification of the prosthetic knee system, this can be expanded to observe the motion of the toe in swing-phase as it passes under the body.



A means of validating the control of the powered prosthetic knee is proposed by comparing the positional motion of the toe during swing for two cases. The first case uses the entire CGA data to determine the position of the toe during swing of normal walking. The second case replicates the behavior of the powered prosthetic knee and observes the position of the prosthetic toe during swing by replacing the joint angle trajectories for the ankle and knee joints to that of the powered prosthetic knee. This is accomplished by replacing the human ankle with the fixed ankle of a typical prosthesis. Then by using the FSM and reference generators, the desired knee angle the powered prosthetic knee would actuate towards is calculated. The simulation then assumes perfect control, as in the prosthetic knee precisely actuates the powered prosthetic knee to the desired angle. Once the two prosthetic joint trajectories are known, the same approach is used to calculate the position of the toe during swing for the prosthesis. It is therefore possible to simulate the toe trajectory for the powered prosthetic knee and compare that to the toe trajectory of an able bodied person.



**Figure 61: Robotic Linkage Representing Legs During Walking.**

The fundamental robotics concepts of forward kinematics and rigid body transformations are utilized to calculate the positional data of the joints in human walking. The person walking is modeled as a series of rigid links of known length connected by rotational axis emanating from the foot of the stance leg, shown in the figure above. The foot of the stance leg is assumed to be planted fully on the ground and is, therefore, modeled as fixed to the ground; the ankle acts as the first pivot and the shank as the first real link. The shank is then attached to the thigh by the knee joint. The thigh of the stance leg is then attached to the hip bone of the person by the hip joint of the stance leg. The swing leg is similarly attached from the hip bone of the person to the thigh of the swing leg by the hip joint on the swing side of the body. Similarly, this chain continues down to the foot on the swing side. The foot during swing-phase is, of course, free and is modeled as the final rigid link with the point of concern being the toe at the end.

The results of this verification are shown below. First, it is worth noting that the trajectory of the toe during normal walking is observed to travel just a few centimeters above the ground during the first half of the swing-phase. This validates the previous assertions made in section 3.2 that formed the basis for the hypothesis that the FSM and reference generators would provide accurate flexion of the knee joint by following this trajectory.

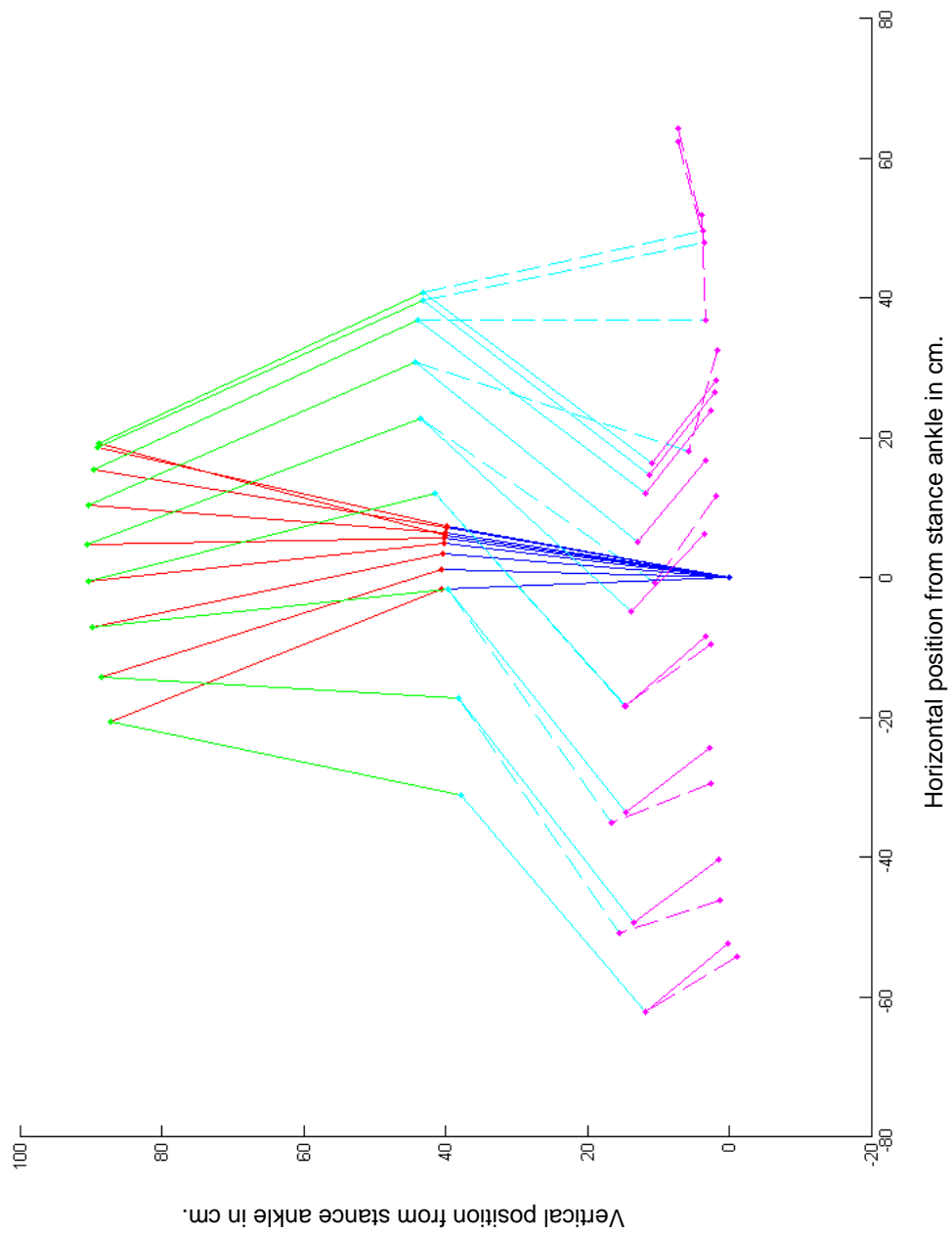
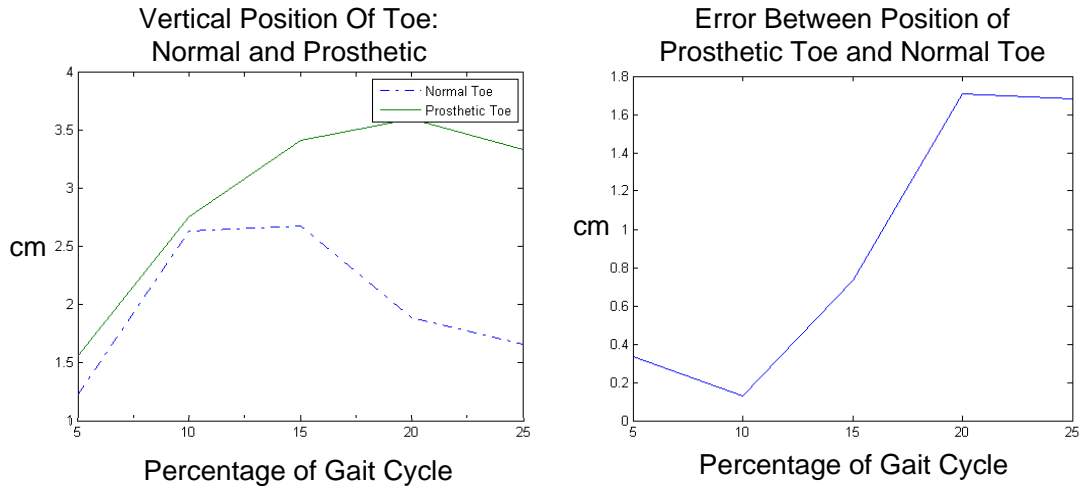


Figure 62: Position of limbs during walking. Dashed lines represent normal walking behavior, while solid lines represent the position of the powered prosthetic.

Figure 62



**Figure 63: Difference Between Normal And Prosthetic Toes.**

Second, the vertical position of the toe on the powered prosthetic knee is seen to mimic very closely the vertical position of the toe for normal walking even though the horizontal position of the toe differs. The toe on the powered prosthetic knee is seen to rise above the desired height due to the motion of the hip joint in the simulation. As discussed earlier, the FSM and reference generators assume that the hip joint does not change its distance from the ground and bends the knee joint so that contact does not occur. Because the hip moves vertically as it moves forward, the position of the toe mirrors this.

Closer inspection of the difference shows that the powered prosthetic knee does have error. But, this error is on the side of caution, as the knee creates additional toe clearance as the toe passes under the body where the danger of toe stubbing is most significant. The maximum error is also small, a mere 1.7 centimeters. This result verifies that the FSM and reference generators does produce swing-phase kinetics that avoids toe contact with the ground, mimicking normal motion, by only using the measured angle of the thigh to control the knee flexion of the powered prosthetic knee.

### **8.3 Experimental verification**

Now that it has been demonstrated that the reference generators provide adequate clearance during the forward-swing-phase of normal walking, the next logical step in verifying the success of the system is to investigate the performance of the feedback controller under use.

#### **8.3.1 Introduction and experiment protocol**

It is the feedback controller's role to actuate the motor in such a way that the knee is positioned to the desired reference angle. To this aim, the feedback controller is evaluated by close inspection of experimental data, specifically in regards to the error between the desired knee angle and the measured knee angle. This analysis must go beyond merely noting that the error is maintained within acceptable margins but also examine the nature of the system's response. The assumptions made during the design of the feedback controller can be validated through analyzing the system for telltale characteristics of expected and unexpected behavior. It is also vital to observe the performance of the system under normal walking conditions, rather than on a test bench, to see it in proper context. With this final verification it is possible to connect the theoretical success seen in the simulation to that of the anecdotal successes seen in observation of the prosthetic knee under use by amputees.

The system is walked on a treadmill at specified, constant speeds varying from a slow walk of 0.96 mph to a brisk walk of 1.75 mph to get a diverse view of how the prosthetic knee performs under the normal load variations of daily walking. Each test is performed for 20 seconds for an average of 35 steps at the set speed. Many system variables are recorded during the experiments by collecting internal variable data directly from the digital signal processor of the prosthetic knee system. The speed of the treadmill is then increased to the next speed value and the system is recorded for another 20 seconds after steady state walking is achieved. The most relevant, recorded variables are the knee angle reference and the measured knee angle. Traditionally, the signal of the applied torque, in this case the signal of the knee torque measured in the hydraulics, is also used in control theory analysis. However, by itself the torque signal does not provide useful data because of the addition of the torques applied by the feedback linearization.

With both the reference signal and the measured angle signal available, it is possible to evaluate the effectiveness and stability of the feedback controller by observing the closed-loop system dynamics and looking at the transient response of the relative error of the knee joint. Applying control theory concepts to this experimental data makes it possible to analytically evaluate the controller design decisions, verify assumptions, and validate the system model derived in earlier chapters.

### **8.3.2 Presentation of Experimental Data**

Each experiment results in a collection of angular positions over time and therefore over many separate steps. The data from the experiments is displayed graphically below. For each experiment the raw data for the knee angle and reference knee angles are processed so that each individual swing maneuver is identified and isolated. The reference and measured knee angles are then regrouped so that they all start at zero seconds thereby producing a locus of trajectories for the 35 number of steps per experiment. The loci of measured knee angles are presented in Figure 64 through Figure 68. These values are then used to calculate the mean and standard deviations for the trajectory of the knee angle and reference angle. The conditioned data is presented in two different graphs to avoid confusion, see Figure 69 through Figure 73. First, the mean values of the knee angle and reference angle are plotted versus time. The data is presented a second time to illuminate the magnitude of variation in the experimental data. It is presented in graphical form with a solid line for the mean values and a shaded area encompassing plus and minus one standard deviation. Additionally a graphical representation of the error between the two signals is similarly presented in the first subplot of Figure 74 through Figure 78, with a mean line and a shaded area encompassing plus and minus one standard deviation. An example of one individual step is presented in the second subplot of Figure 74 through Figure 78, to highlight possible details lost when looking at the mean values.

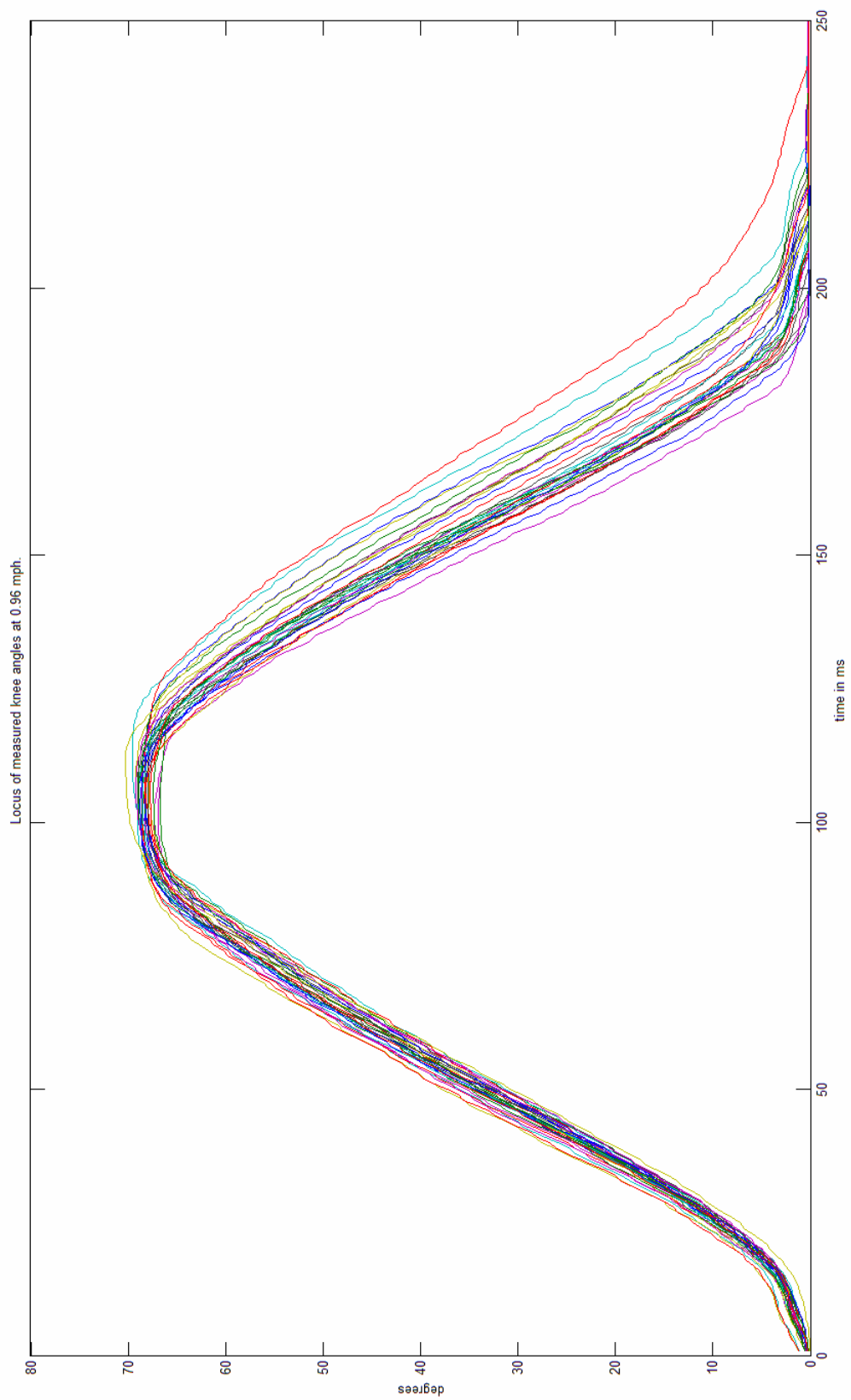
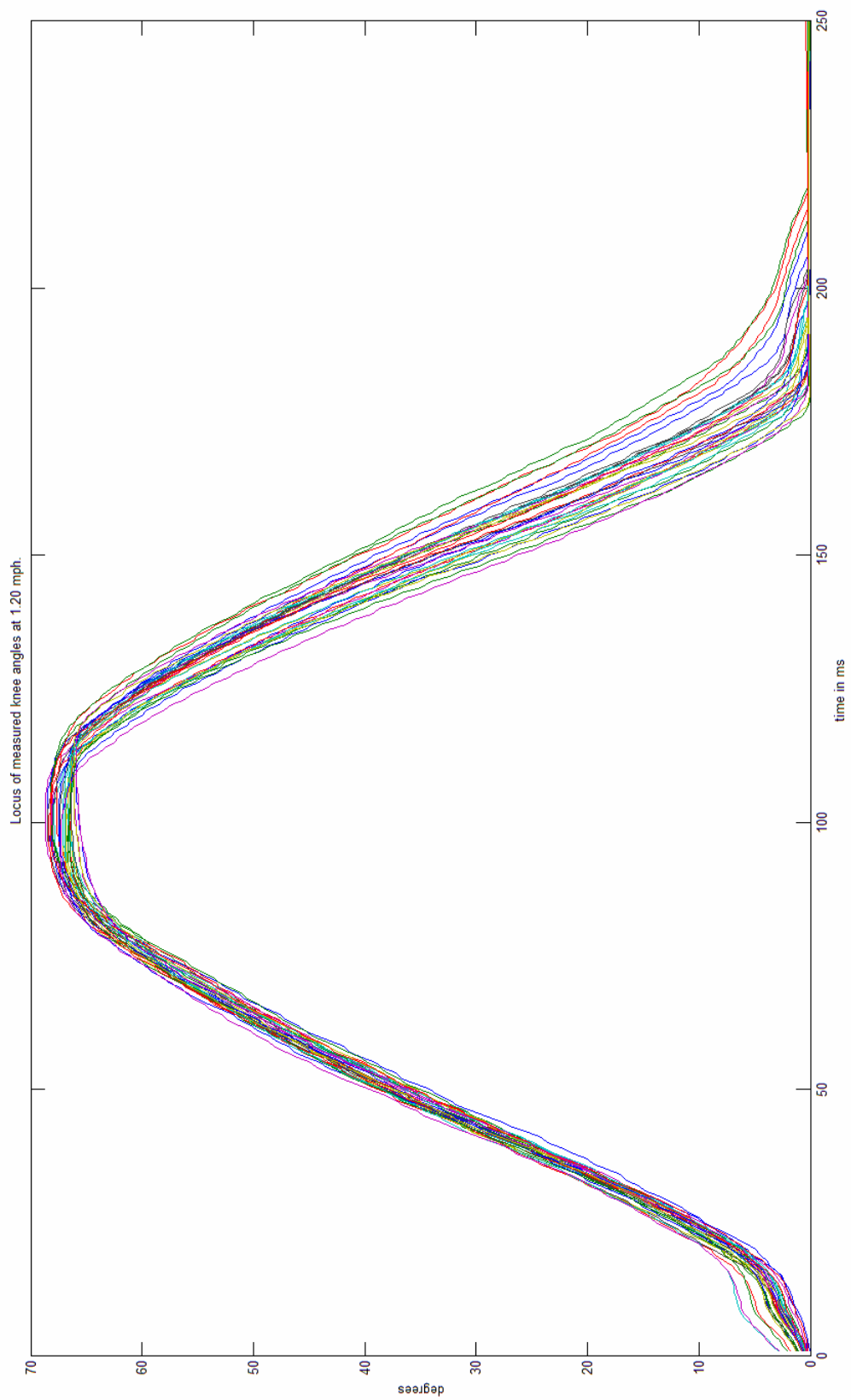
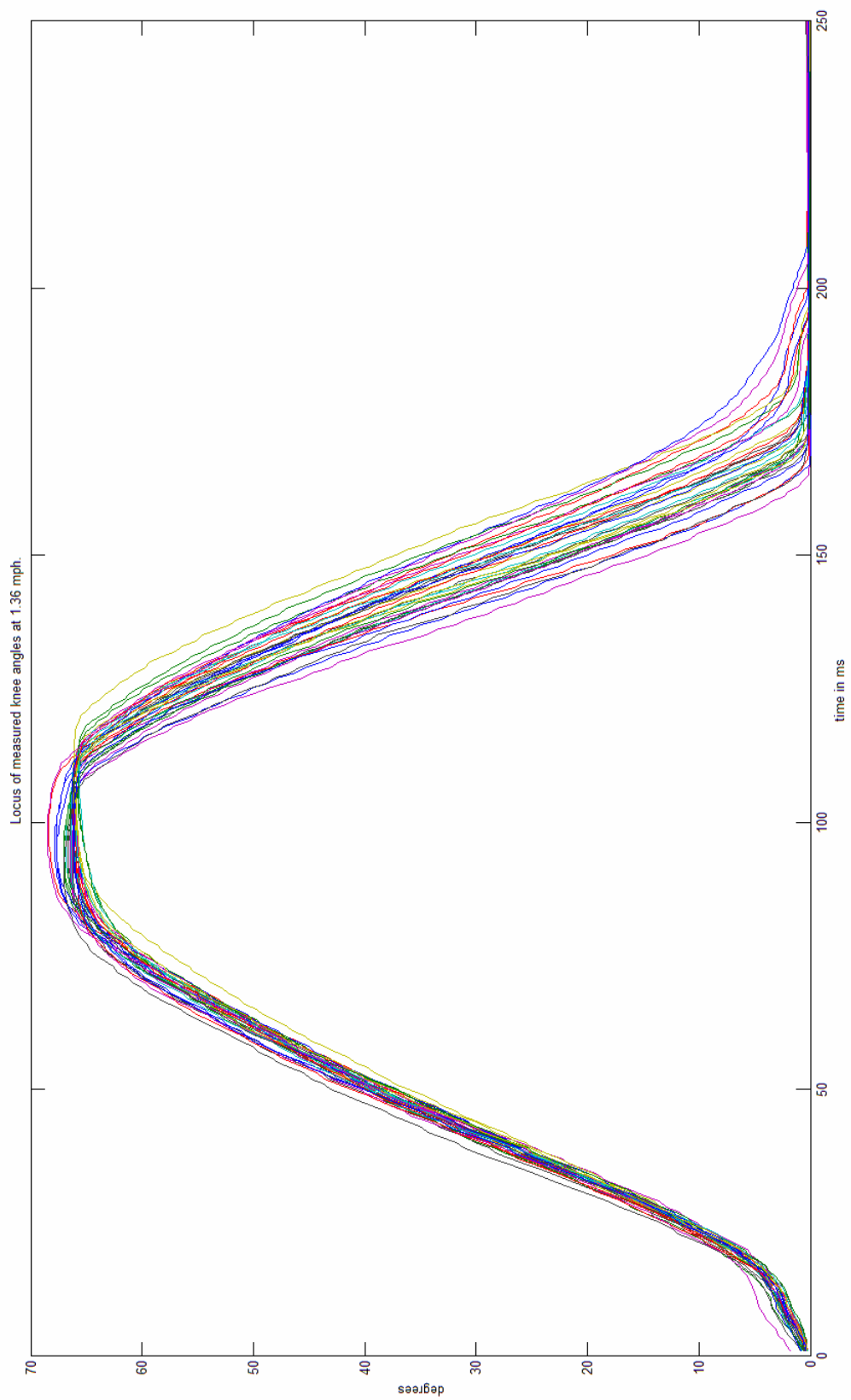


Figure 64: Locus at 0.96 mph



**Figure 65: Locus at 1.20 mph**



**Figure 66: Locus at 1.36 mph**

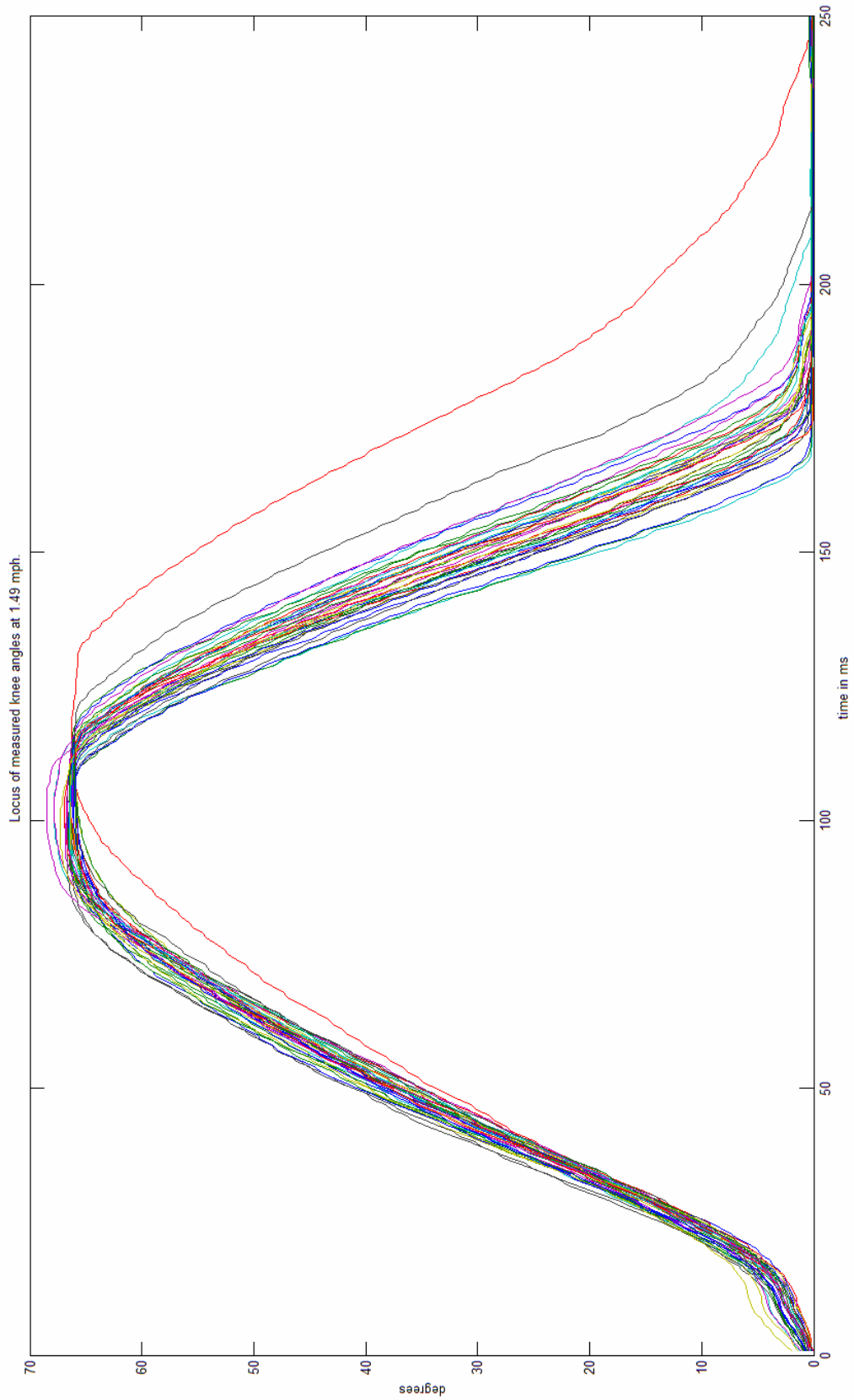
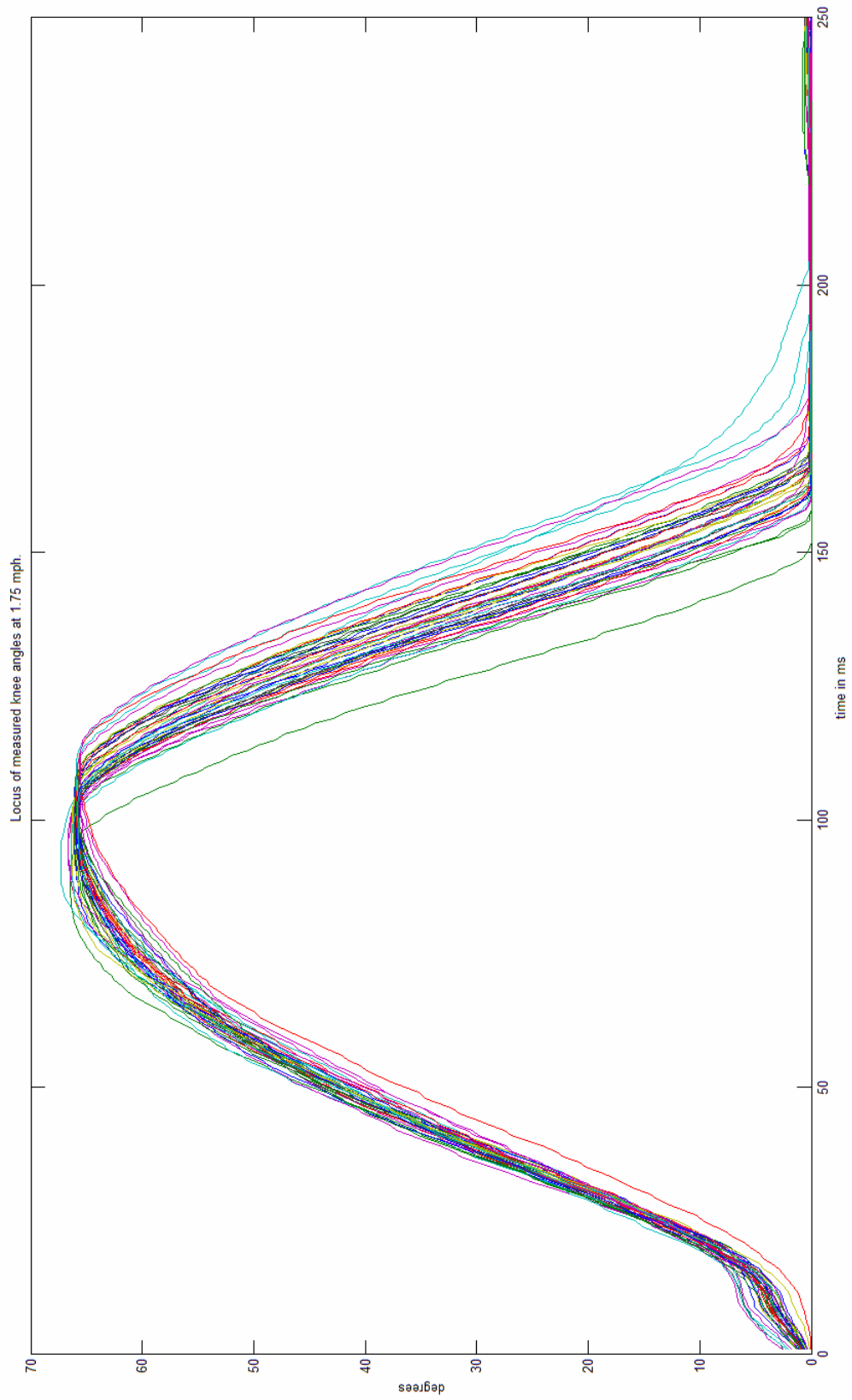
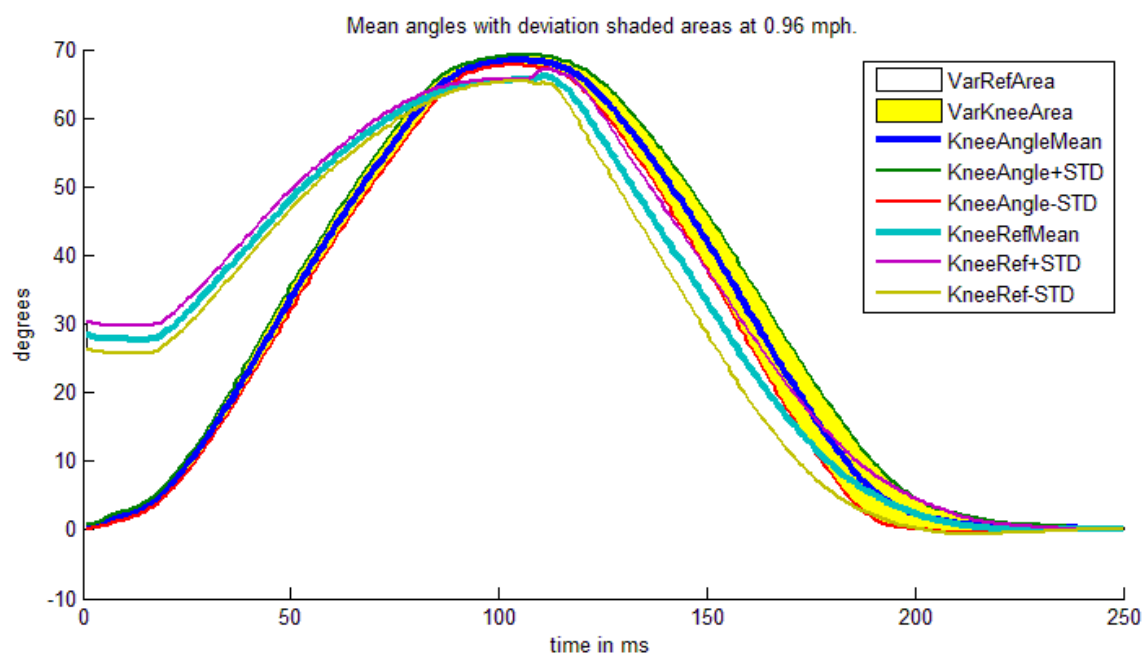
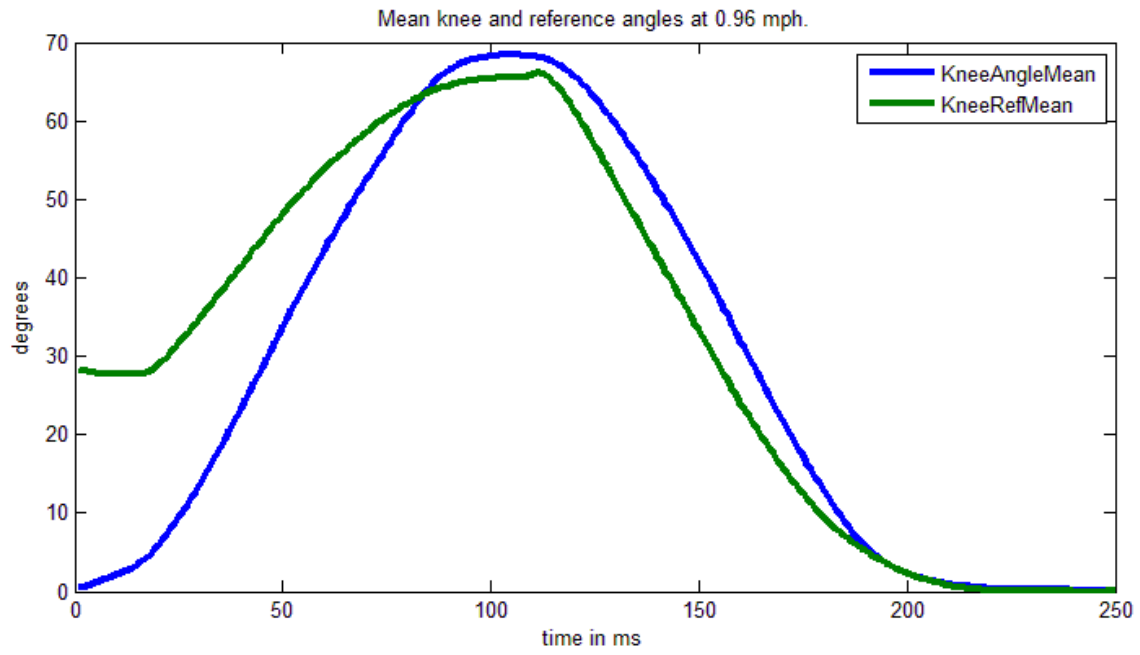


Figure 67: Locus at 1.49 mph

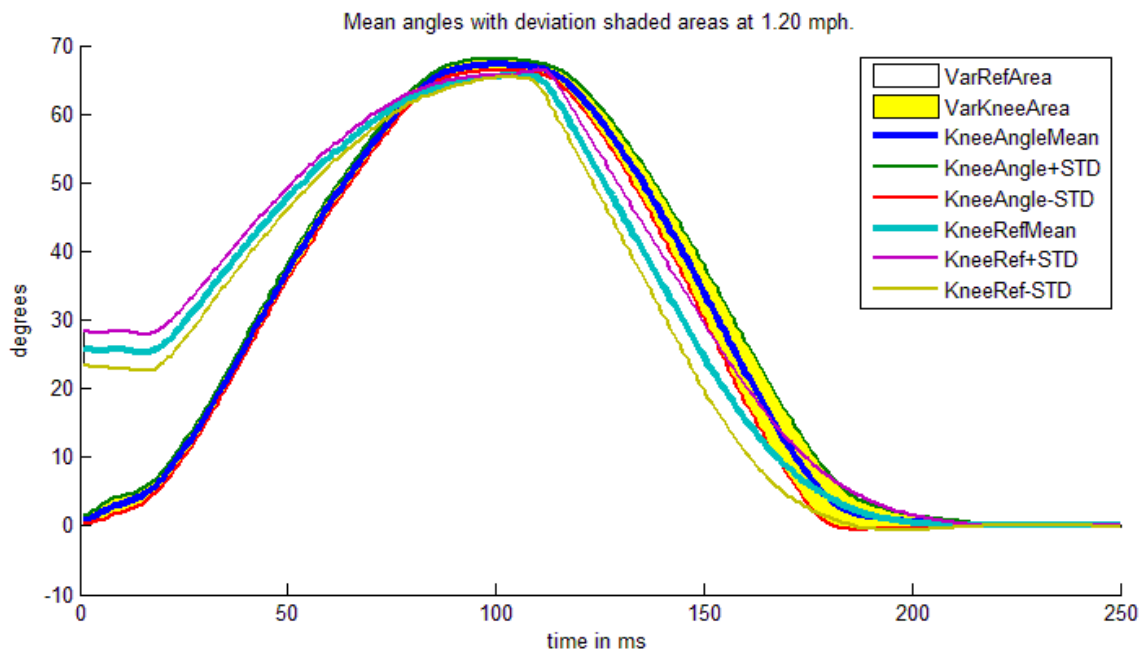
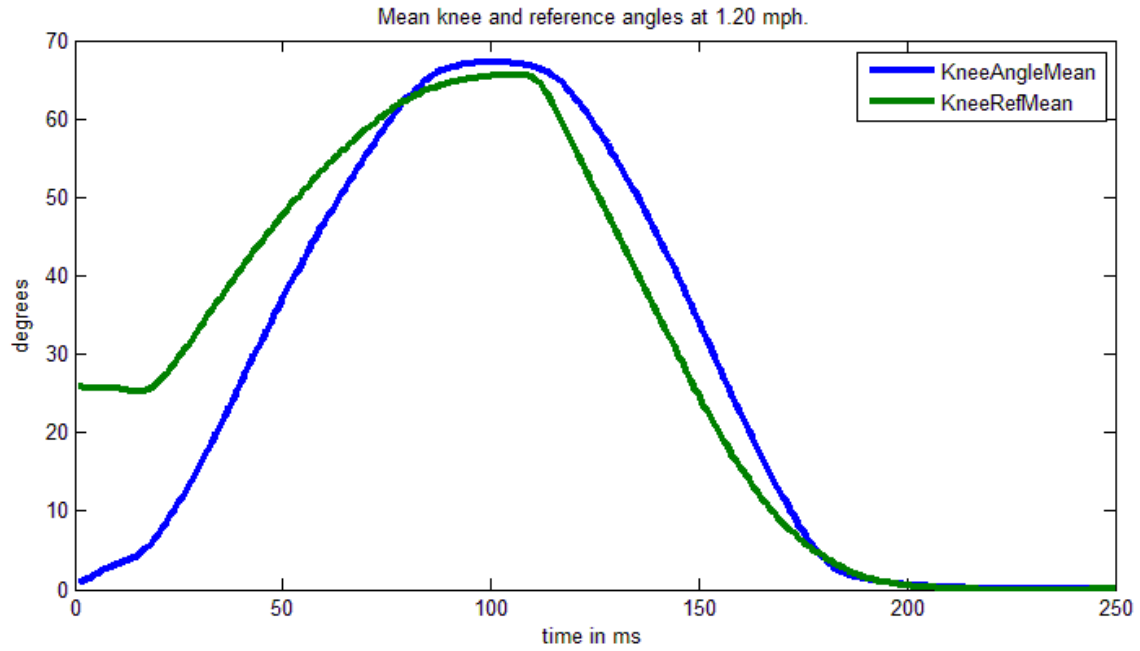




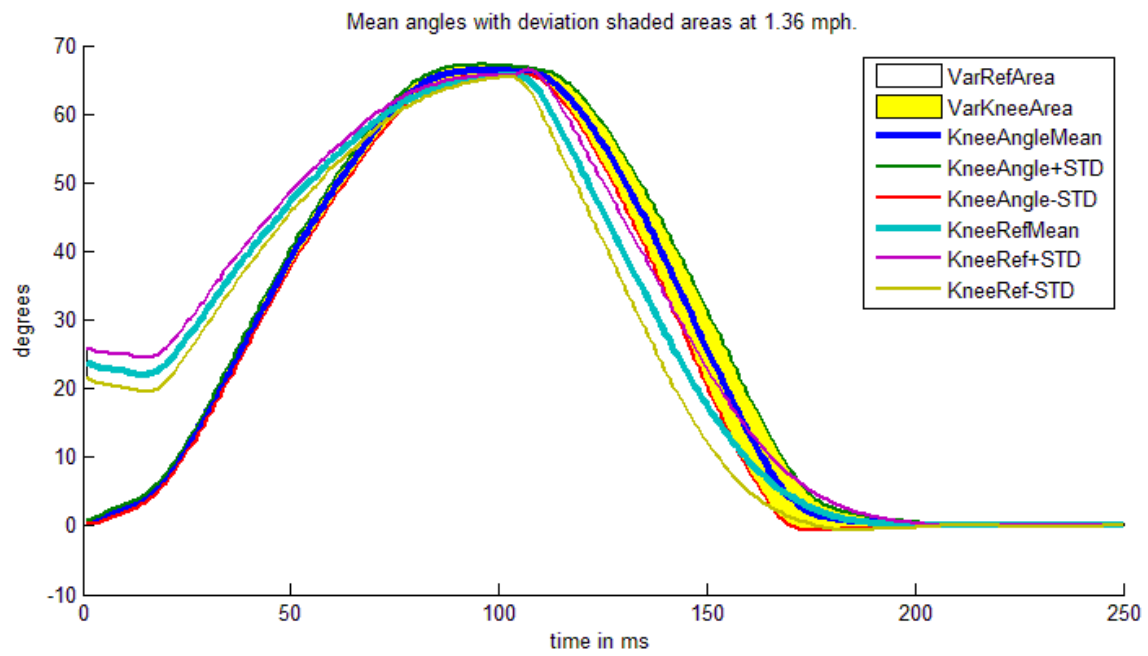
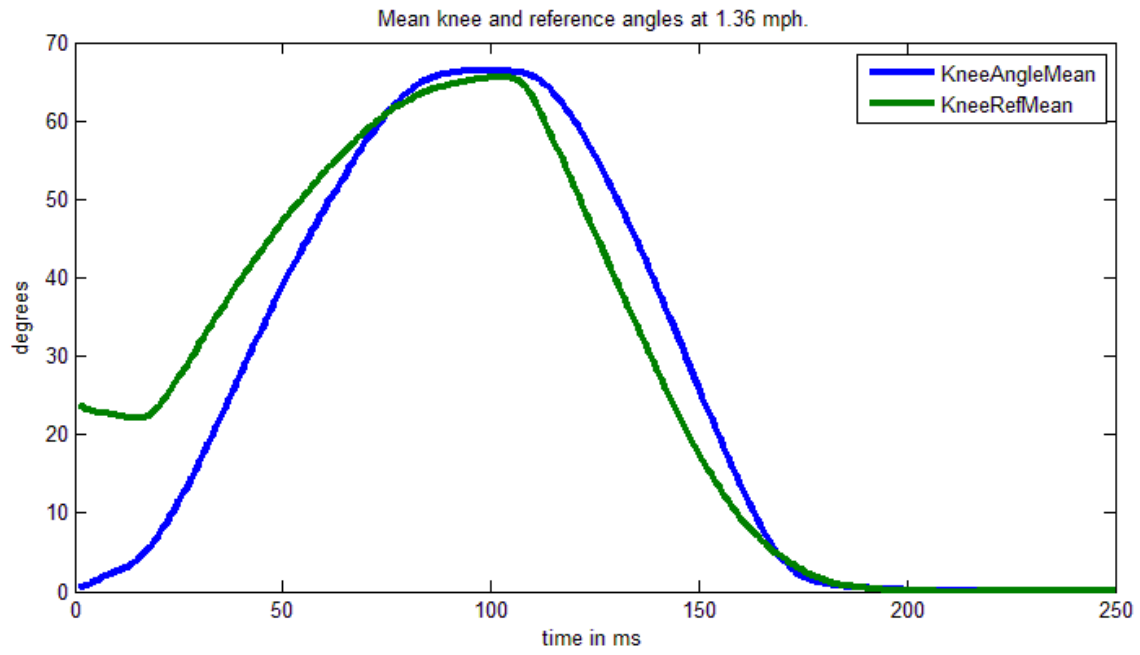
**Figure 68: Locus at 1.75 mph**



**Figure 69: Mean Knee and Reference Angles at 0.96mph**



**Figure 70: Mean Knee and Reference Angles at 1.20mph**



**Figure 71: Mean Knee and Reference Angles at 1.36mph**

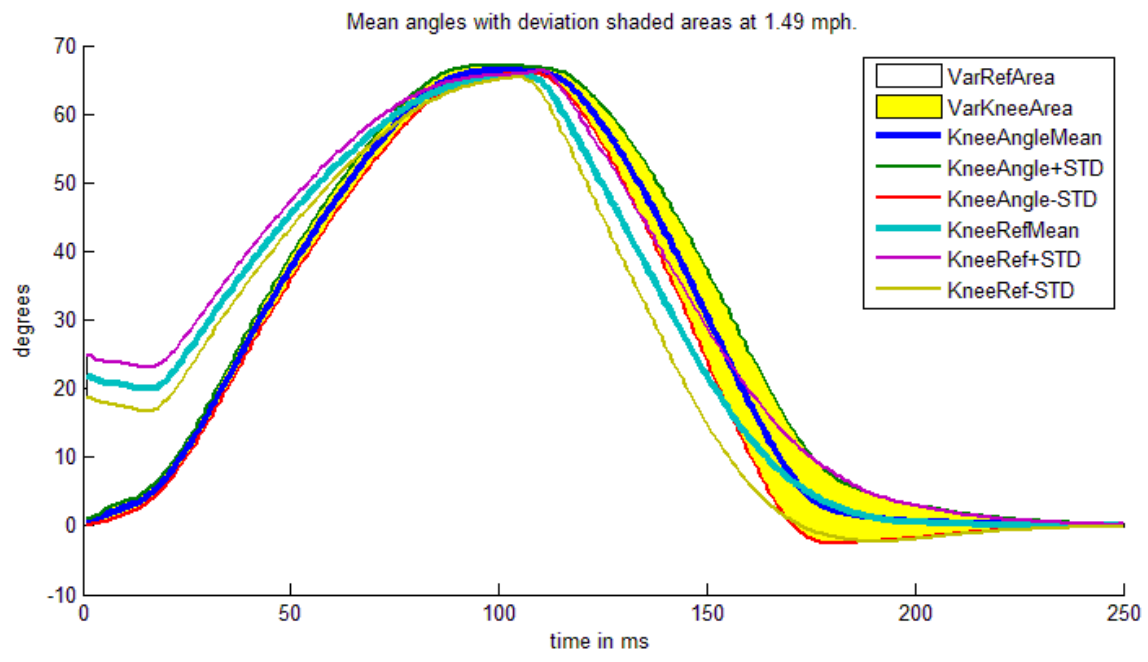
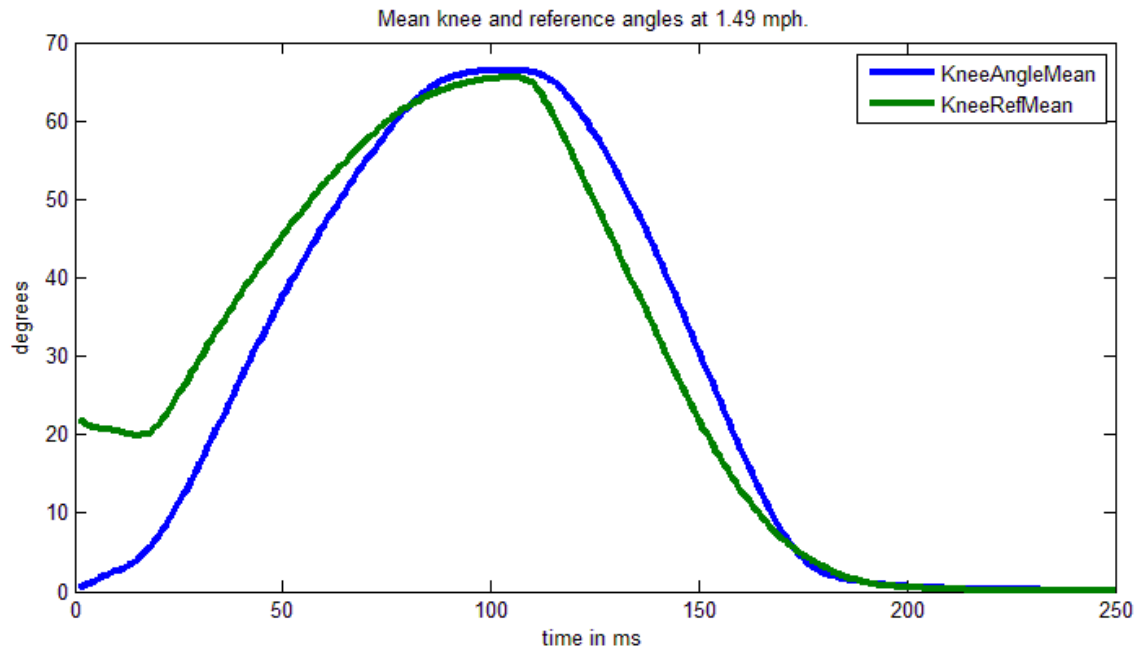


Figure 72: Mean Knee and Reference Angles at 1.49mph

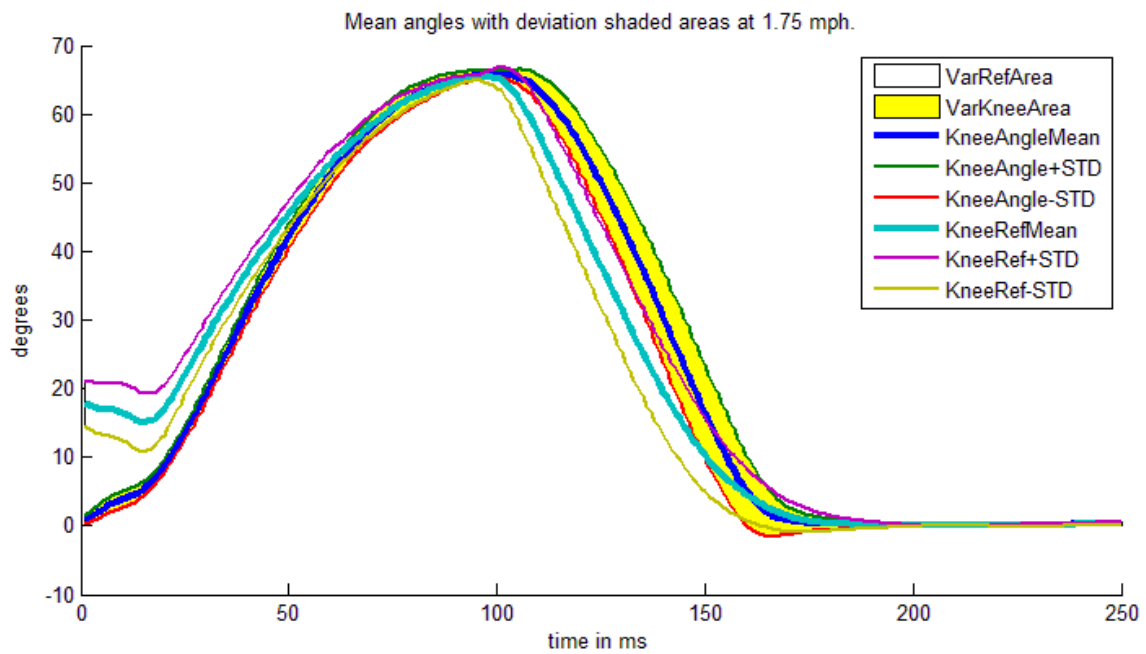
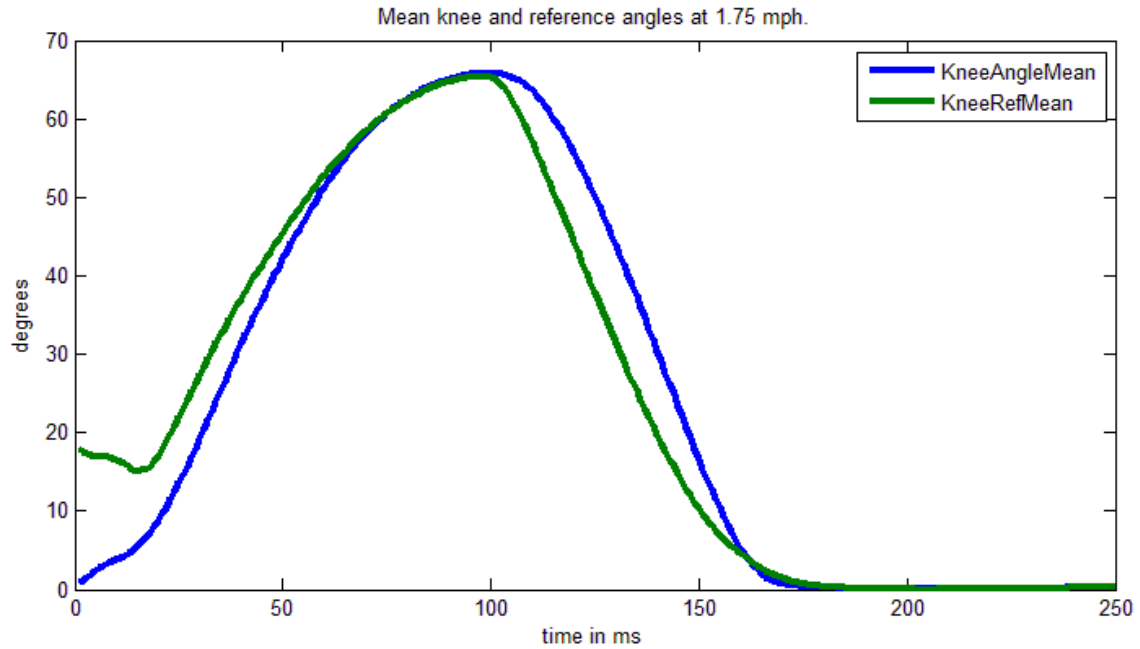


Figure 73: Mean Knee and Reference Angles at 1.75mph

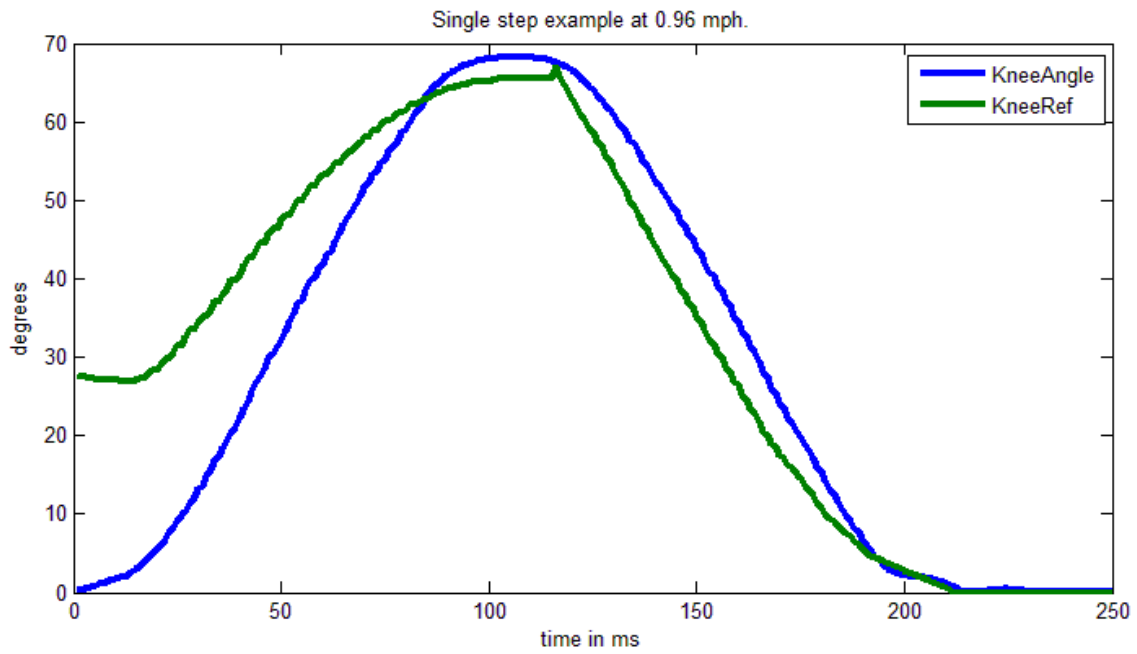
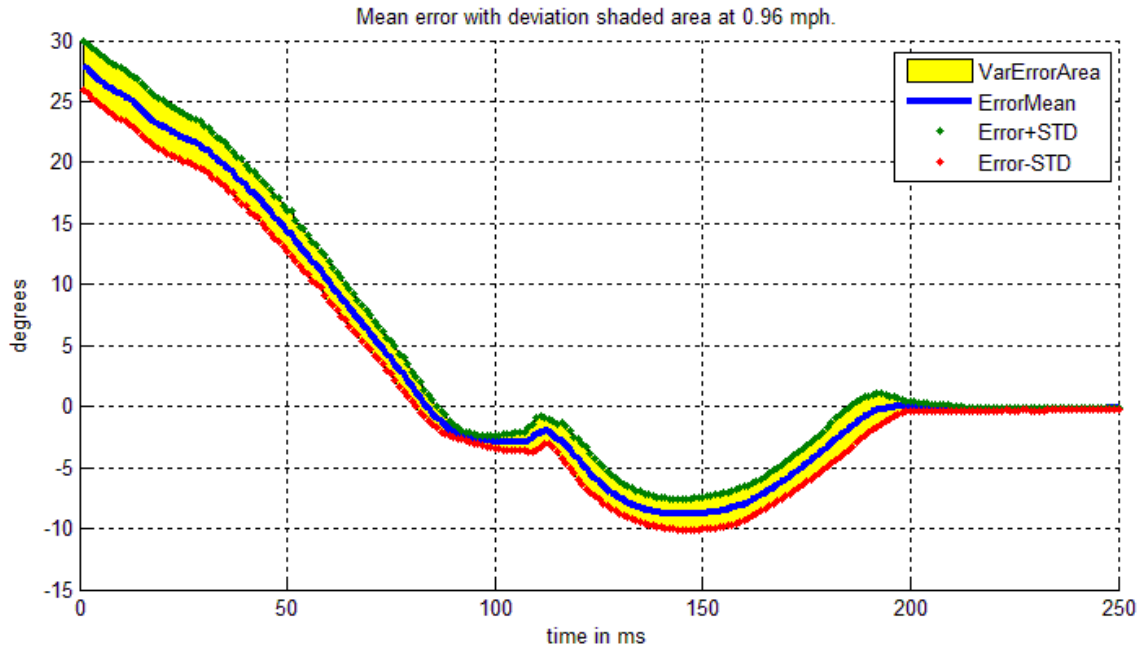
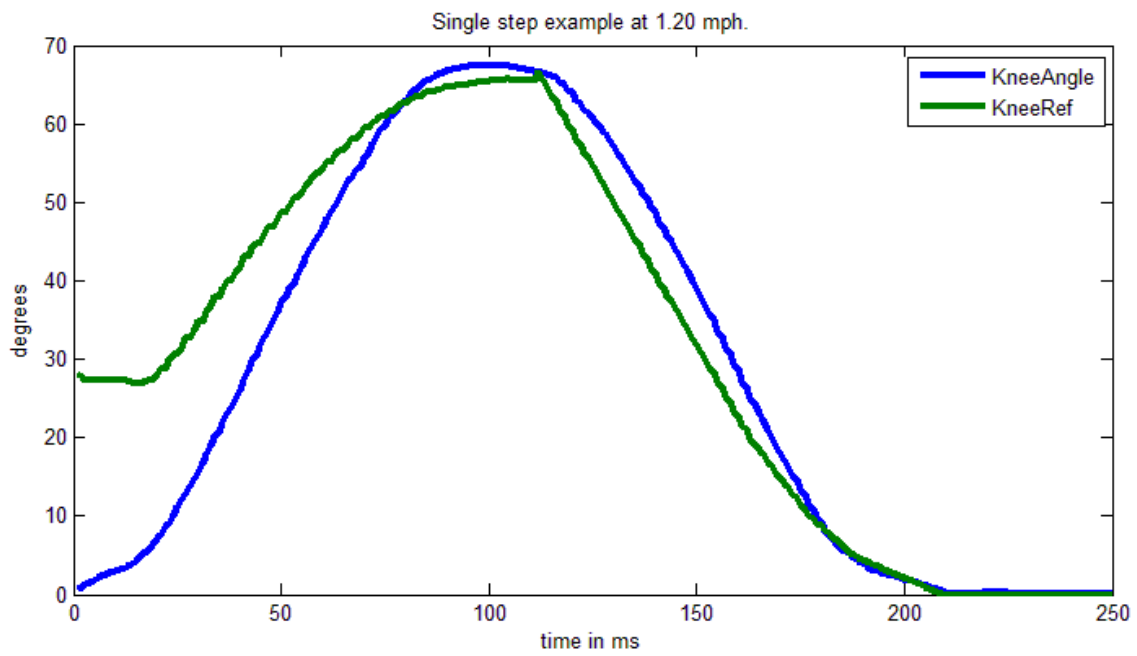
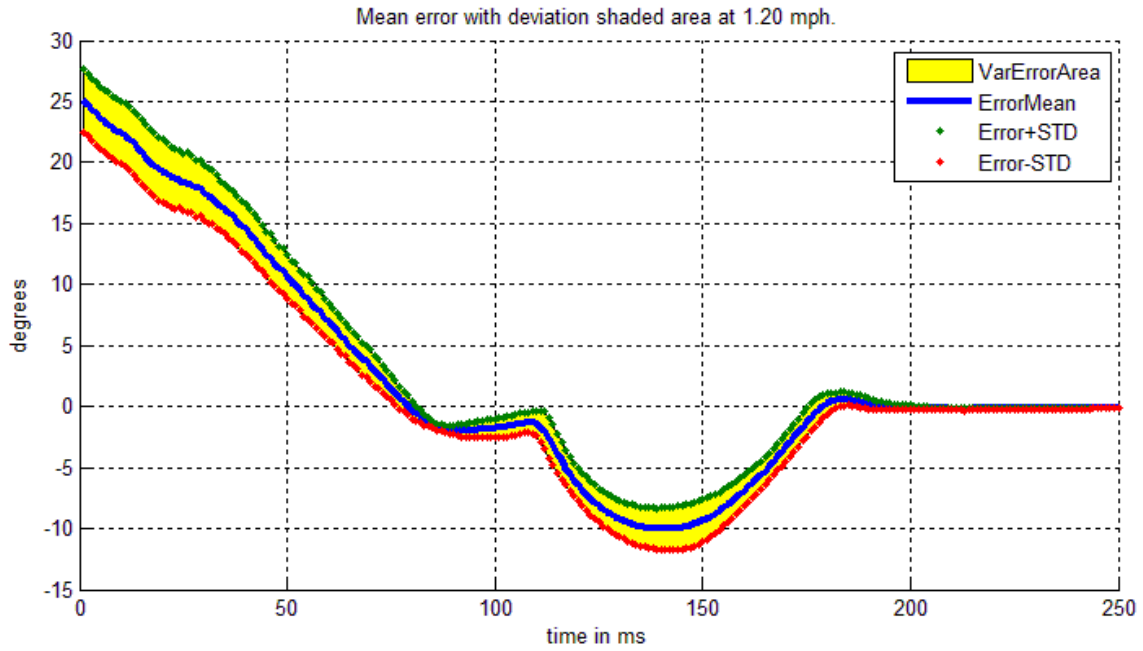


Figure 74: Mean Error and Example at 0.96mph



**Figure 75: Mean Error and Example at 1.20mph**



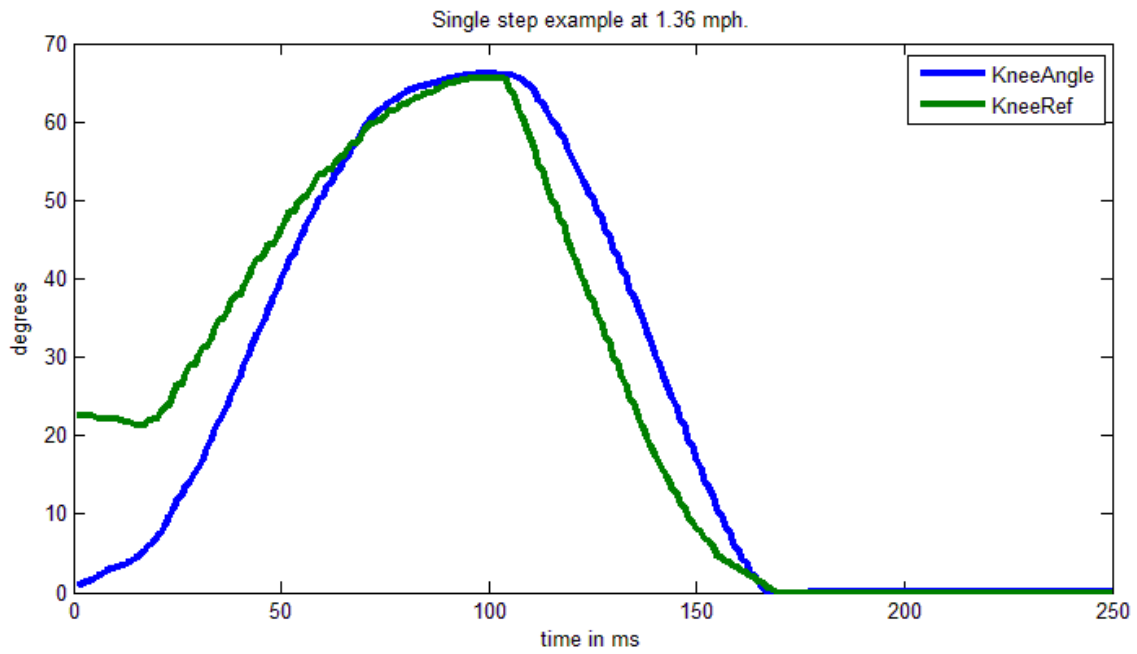
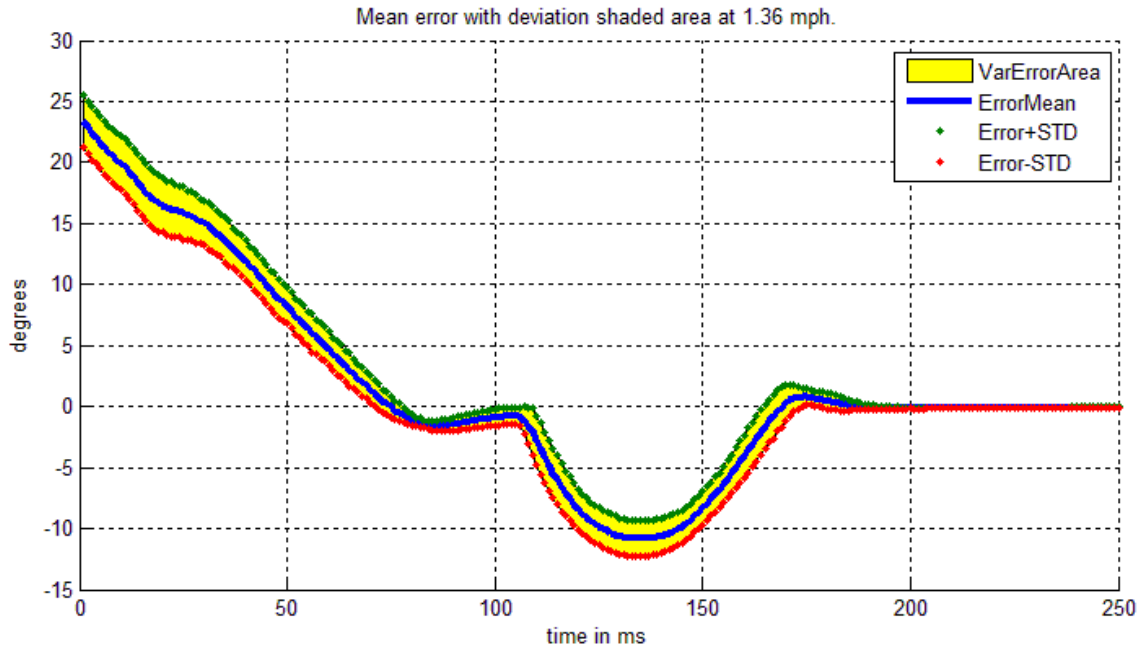


Figure 76: Mean Error and Example at 1.36mph

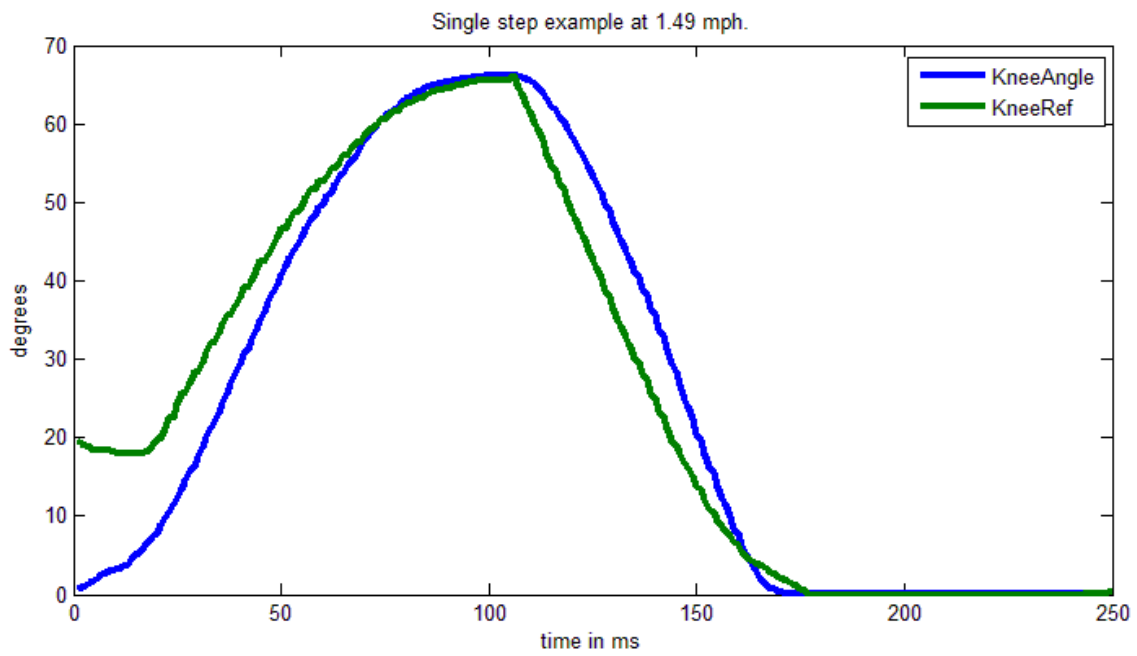
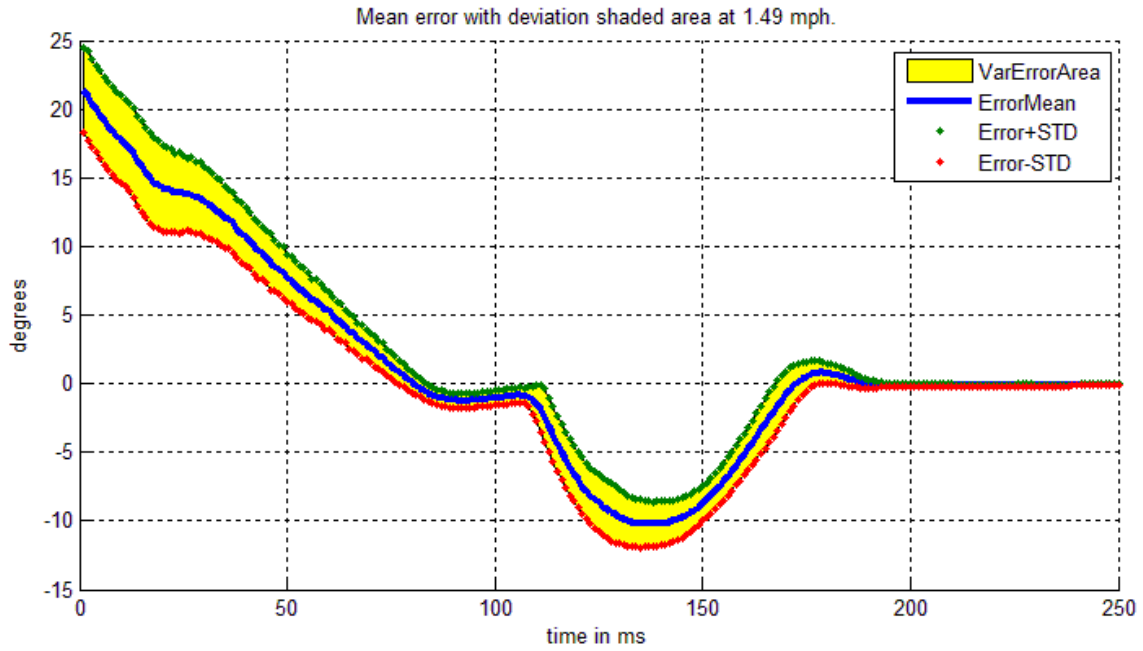
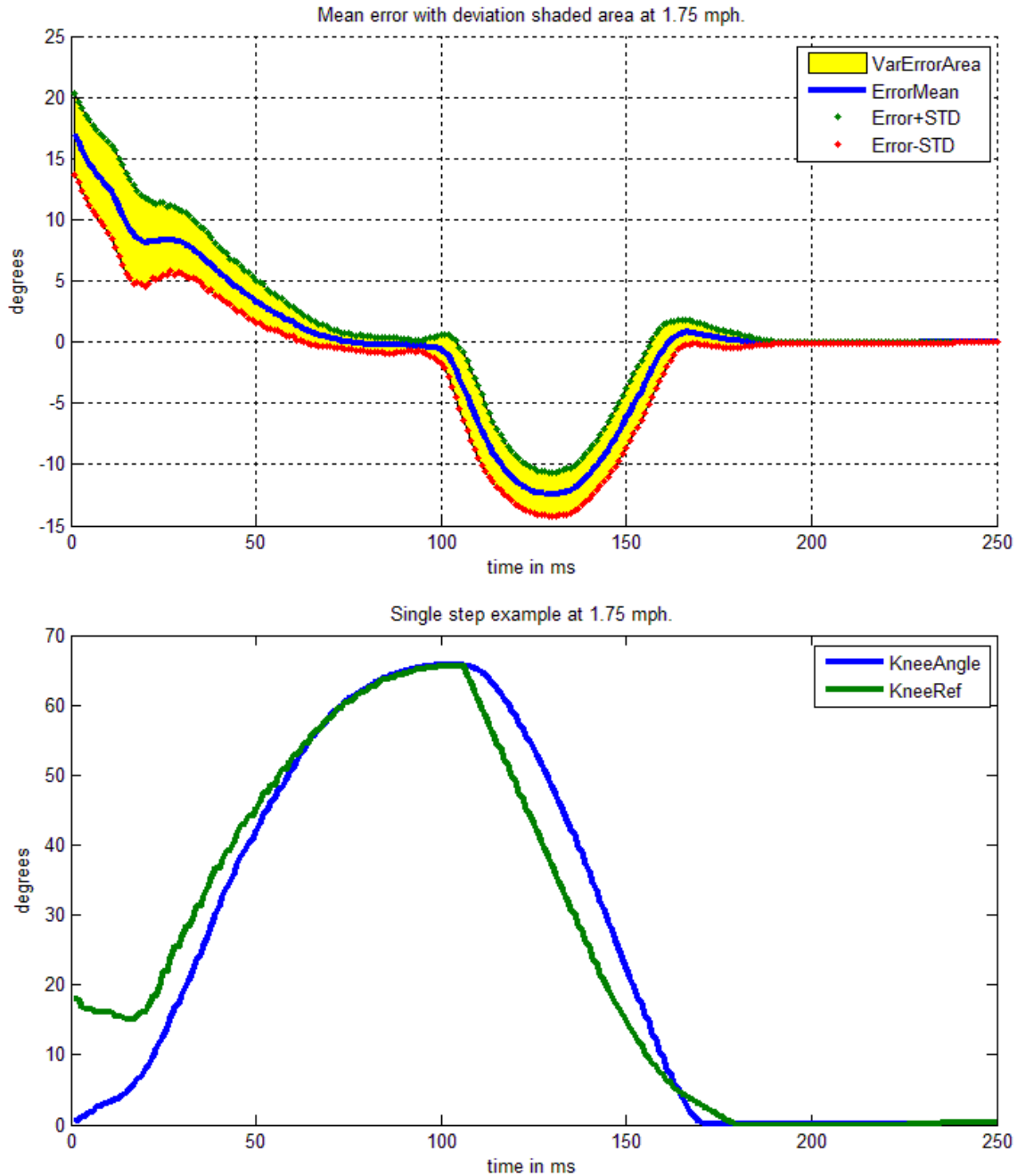


Figure 77: Mean Error and Example at 1.49mph



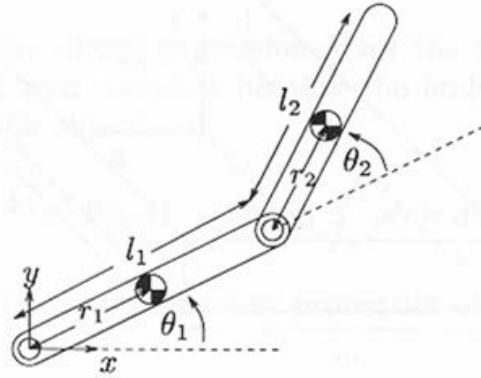
**Figure 78: Mean Error and Example at 1.75mph**

With this presentation of data available it is now possible to look closely at the reference and error signals to quantitatively discuss the successes and failures of the feedback controller.

### 8.3.3 Experimental Support for the trivialization of un-modeled dynamics

The idea of controlling a prosthetic knee joint attached to a person is an inherently complicated problem because of the person's presence within the system and their conscious and reflexive forces of unknown magnitude applied at the thigh. Therefore it is necessary to make some assumptions, detailed earlier, to simplify this part of the system and develop a simpler model. These forces from the user, the subsequent motions of the thigh and the resulting

interaction with the prosthetic knee, were trivialized into an additional source of system disturbance called the un-modeled dynamic noise that required further demands on the feedback controller. This assumption is possibly dangerous in nonlinear systems due to the lack of the superposition property and the non-Gaussian nature of this un-modeled dynamic noise. Through observing the experimental data it is possible to find supporting evidence that this is a safe design decision.



$$\begin{bmatrix} \alpha + 2\beta c_2 & \delta + \beta c_2 \\ \delta + \beta c_2 & \delta \end{bmatrix} \begin{bmatrix} \ddot{\theta}_1 \\ \ddot{\theta}_2 \end{bmatrix} + \begin{bmatrix} -\beta s_2 \dot{\theta}_2 & -\beta s_2 (\dot{\theta}_1 + \dot{\theta}_2) \\ \beta s_2 \dot{\theta}_1 & 0 \end{bmatrix} \begin{bmatrix} \dot{\theta}_1 \\ \dot{\theta}_2 \end{bmatrix} = \begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix} + T_{\text{gravity}}$$

where

$$\begin{aligned} \alpha &= I_{z1} + I_{z2} + m_1 r_1^2 + m_2 (l_1^2 + r_2^2) \\ \beta &= m_2 l_1 r_2 \\ \delta &= I_{z2} + m_2 r_2^2 \end{aligned}$$

**Figure 79: Two-Link Lagrangian Dynamic Model and Equation of Motion**

First, it is beneficial to consider the more complete dynamics of a 2 link model for the prosthetic knee. It is possible to get a better idea of what to expect from the un-modeled dynamic noise when under different circumstances by doing so. Figure 79 shows the relation between the angular trajectories, their derivatives, and the torques applied at the joints as derived via Lagrangian dynamics (see appendix 1). This model can be reduced to the simple 1 link model used in chapter 7 by setting theta1 to be a constant value.

When compared to the dynamics of the one link system (theta1, etc) it is possible to identify the terms corresponding to the un-modeled dynamic noise,

$$\text{One link system: } \delta \ddot{\theta}_2 = \tau_2 + T_{2\text{gravity}}$$

$$\text{Two link system: } (\delta + \beta c_2) \ddot{\theta}_1 + \delta \ddot{\theta}_2 + \beta s_2 \dot{\theta}_1^2 = \tau_2 + T_{2\text{gravity}}$$

$$\text{"un-modeled dynamic noise"} = (\delta + \beta c_2) \ddot{\theta}_1 + \beta s_2 \dot{\theta}_1^2.$$

The noise term is therefore shown to depend on the speed of both joints and the acceleration of the thigh. If the trajectories of the knee and thigh angles are assumed the same regardless of walking speed, then finding the new values are done easily via a time transformation.

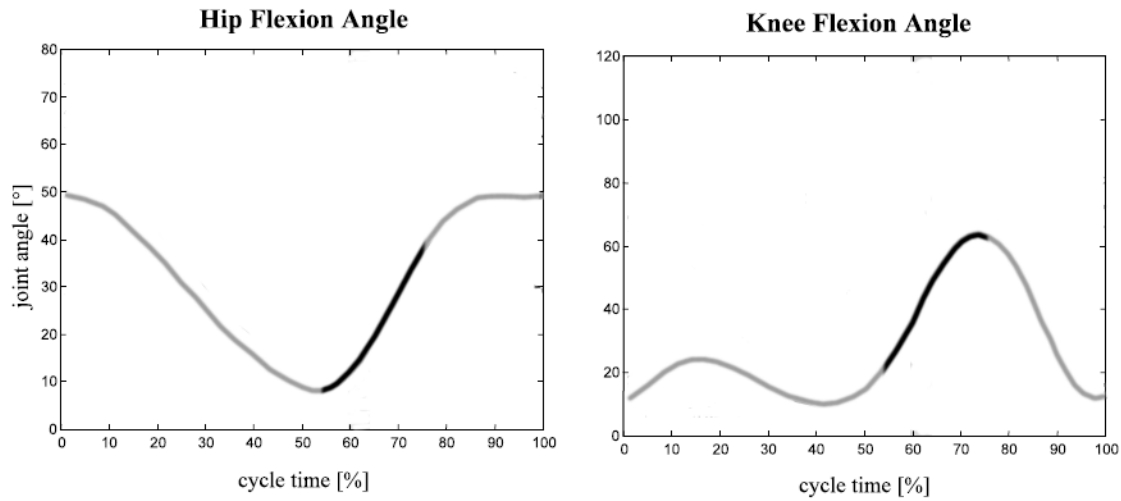
$$\begin{aligned}\theta_{fast}(t) &= \theta_{slow}(\omega t), \\ \dot{\theta}_{fast}(t) &= \omega \dot{\theta}_{slow}(\omega t), \\ \ddot{\theta}_{fast}(t) &= \omega^2 \ddot{\theta}_{slow}(\omega t).\end{aligned}$$

This equation in conjunction with the previous equation shows that the magnitude of the un-modeled dynamic effect is proportional to the square of the time scale factor,

$$(\delta + \beta c_2) \ddot{\theta}_{1fast} + \beta s_2 \dot{\theta}_{1fast}^2 = \omega^2 \left( (\delta + \beta c_2) \ddot{\theta}_{1slow} + \beta s_2 \dot{\theta}_{1slow}^2 \right) \Rightarrow \eta_{fast} = \omega^2 \eta_{slow}.$$

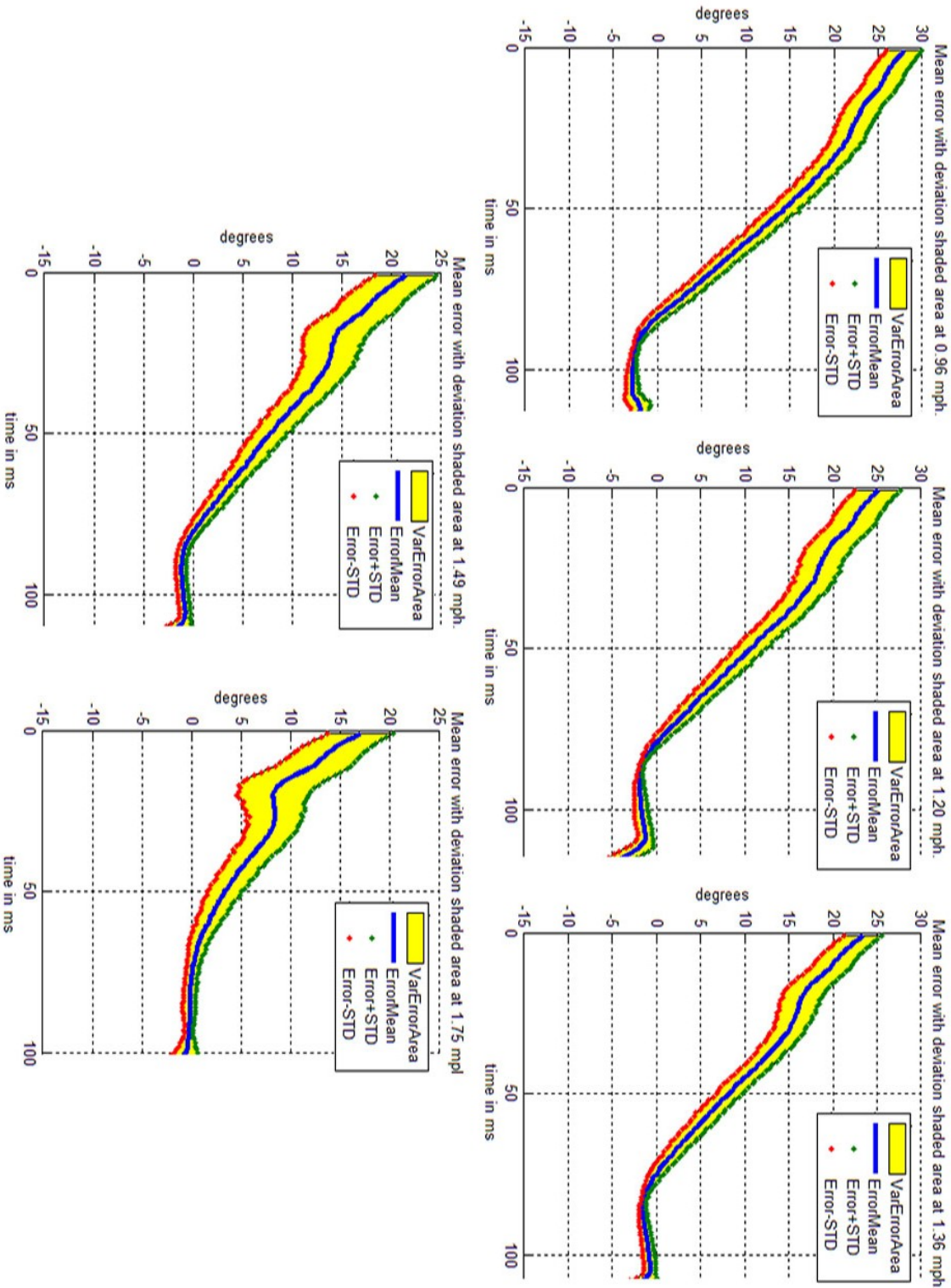
So, for example, if the person is walking 4 times faster, then the magnitude of the ignored noise is 16 times larger.

Again, it is asserted that the un-modeled dynamics are trivial enough to be manageable by the feedback controller. If it is not, then there will be some effect in the behavior of the system related to the square of the speed.



**Figure 80: Highlighted Trajectory of Hip and Knee Angles During Forward-Early-Swing**

The period of time in the walking cycle where the most significant amount of thigh motion, and therefore the region of walking behavior where it is most likely that the effects of the un-modeled dynamic noise will be visible as the speed changes, is during the phase of forward-early-swing, as seen in Figure 80. If the feedback controller is able to adequately compensate for the thigh motion and the assumption that these effects are relatively small holds true, there will be no noticeable changes in the error seen from speed to speed. If, however, these forces can't be compensated by the controller, there will be a noticeable increase in error relative to the square of the speed increase.

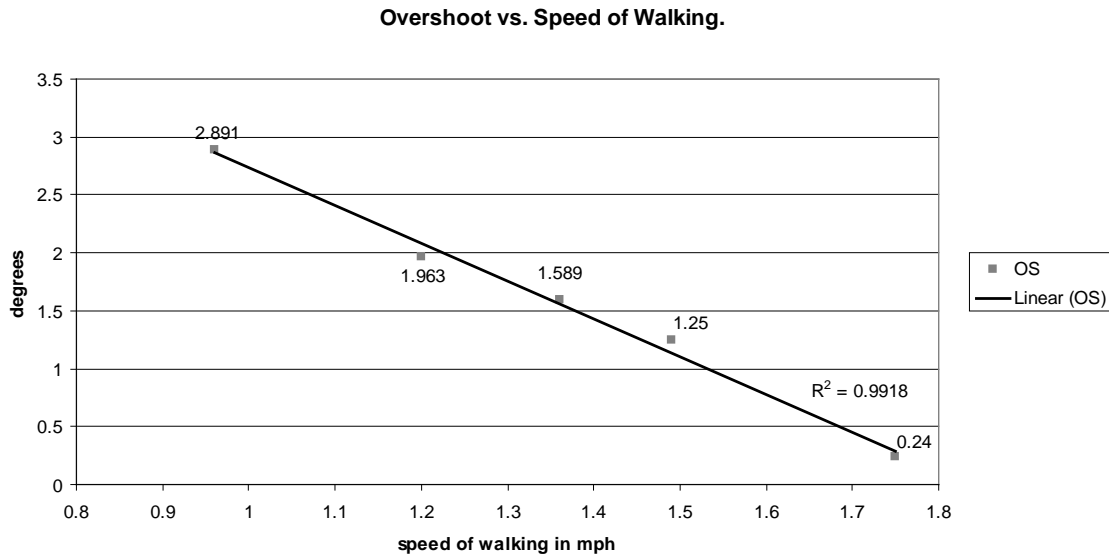


**Figure 81: Mean Error For All Speeds During Forward-Early-Swing**

Looking at the data during the phase of control called flexion tracking, where the reference signal is generated via the robotic avoidance method; there are two noticeable differences when one compares the results across varied speeds. First, the initial error is more

significant at lower speeds. This is a consequence of the longer steps seen when walking at faster speeds, not necessarily a failure of the feedback controller. As the steps become longer, the angle the knee needs to bend at toe off to gain clearance from the ground is smaller.

On the other hand, the convergence of the error is noticeably better as the speed increases. It converges to zero error faster and there is also less overshoot as the speed increases. The values of overshoot are seen in Figure 82. On the other hand, the size of the overshoot is relatively small for all speeds when compared to the 65 degrees of overall flexion of the knee joint during this phase of walking, measuring just 4.4%. When related to the speed of walking, the trend of overshoot values is seen to be linear, not quadratic.



**Figure 82: Presentation of Data Comparing Overshoot as Speed of Walking Varies**

These results appear to neither entirely support nor discredit the decision to cluster inertial effects into an un-modeled dynamic noise as an acceptable decision. The system does not behave as is expected for either case. Something is missing from the system model that causes the error to decrease as the speed of walking increases. That being said, the overshoot is relatively small for all speeds which does support the decision. At the very least, the unintended consequences of this decision are positive for the speeds of walking seen in the experiment, yielding better performance while not impacting stability.

### **8.3.4 Experimental support for feedback linearization**

In addition to the unknown external forces created by the user, there are other nonlinearities present in the system. Two remaining complexities of the system are the nonlinear application of torque from the actuator and the torques due to gravity.

As discussed earlier, the hydraulic system does not directly apply torques from the motor to the knee joint. The electric motor is attached to a hydraulic pump which pushes fluid through a manifold to a linear hydraulic actuator. The actuator applies a force across a moment arm to apply a torque at the knee. The length of the moment arm changes as the knee joint angle changes. It is proposed that by using knowledge of the joint geometry it is possible to invert the calculations to remove the effect of this nonlinearity.

The second nonlinearity is that this system is acted upon by gravity and therefore does not move purely on its own. It is proposed that this effect is negated by using the explicit

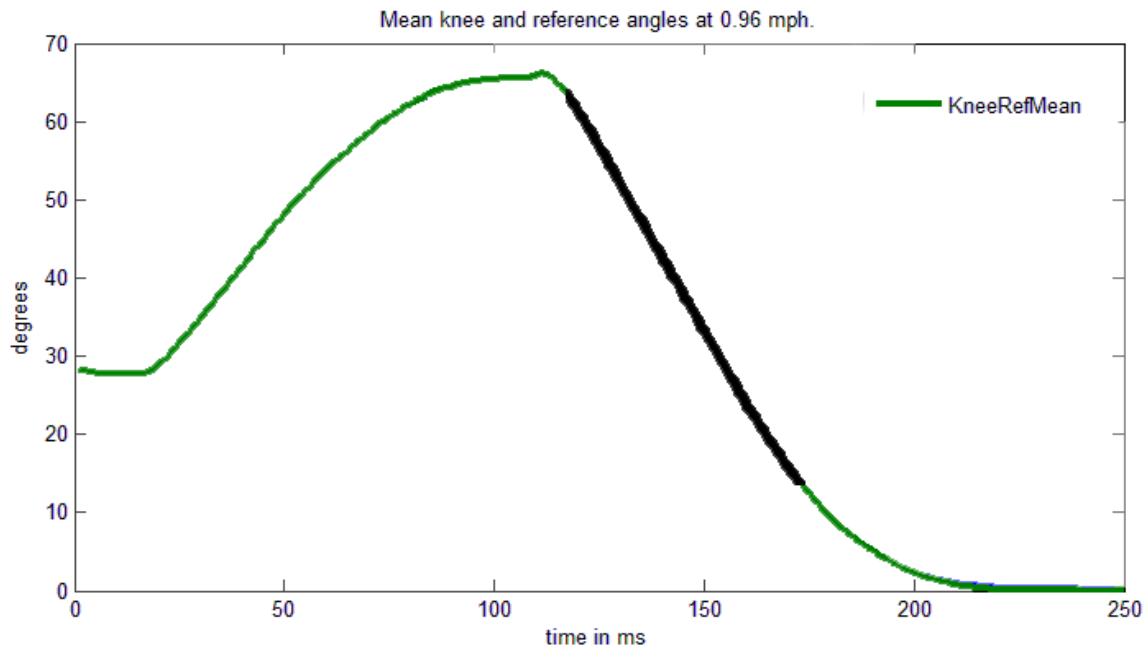
equations to solve for the nonlinear gravity-torques and remove their effects by applying a counteracting torque in the opposite direction.

These two feedback linearization methods are used to reduce the complexity of the nonlinear system to that of a linear form. The model derived in chapter 7 asserts that after applying the gravity compensation torque and geometric correction, the knee system reduces to behaving like that of a rotational inertia acted upon by a controller torque and a viscous friction torque. The experimental data is investigated to support this claim.

Once the system is reduced to a linear system, a linear-control-theory based feedback controller is implemented. A linear system like the reduced system, a two pole system with one at the origin, is easily controlled by a PD controller. So, one such feedback controller was implemented. The validity of using feedback linearization on the prosthetic knee system is evaluated through the analysis of experimental data.

If the system does reduce properly to a linear system it will behave linearly under linear-control in accordance with linear-control-theory. It is therefore possible to verify the linearization attempts by performing linear analysis on the system and comparing the result to those seen in the experimental data. One such hallmark of linear behavior is to calculate the steady-state-errors in response to standard inputs (see app 2 for derivation). For the case of a viscous, rotational inertia system under PD control when a ramp input is applied, analysis shows that there will always be a constant, non-zero steady-state-error.

Again, using the data presented, consider the phase of walking during late swing. In this phase there is a period of near linear ramp input during the extension of the knee joint after the leg has passed from posterior to anterior. The reference generator used in this phase is the speed-based-knee-extension. The reference signal for this period is highlighted below in Figure 83.



**Figure 83: Highlight of Knee Reference During Late Swing**

Because the feedback controller is given a ramp reference during this late phase of swing, it is expected that a constant steady-state-error response from the system will be present. If, however, the system has an ever increasing error to the ramp, then the argument should be made



that not only does the knowledge of the system prove to be inadequate and the nonlinearization fail, but so does the feedback controller as a whole. After all, the system would be unstable. If, however, the steady-state-error approaches zero, then our nonlinearization design rationale is also invalid, regardless of the fact that the system performs better than expected. For the understanding of the system to be accurate and the validation of the design decisions to be possible, there must be a constant, non-zero steady-state-error.

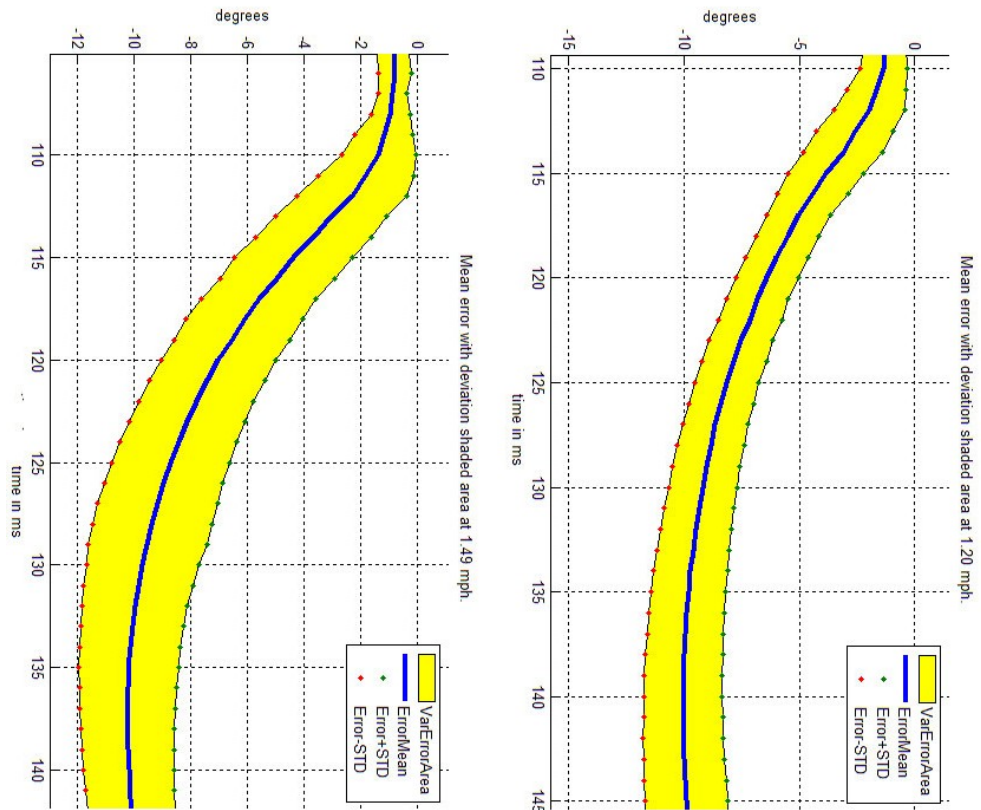
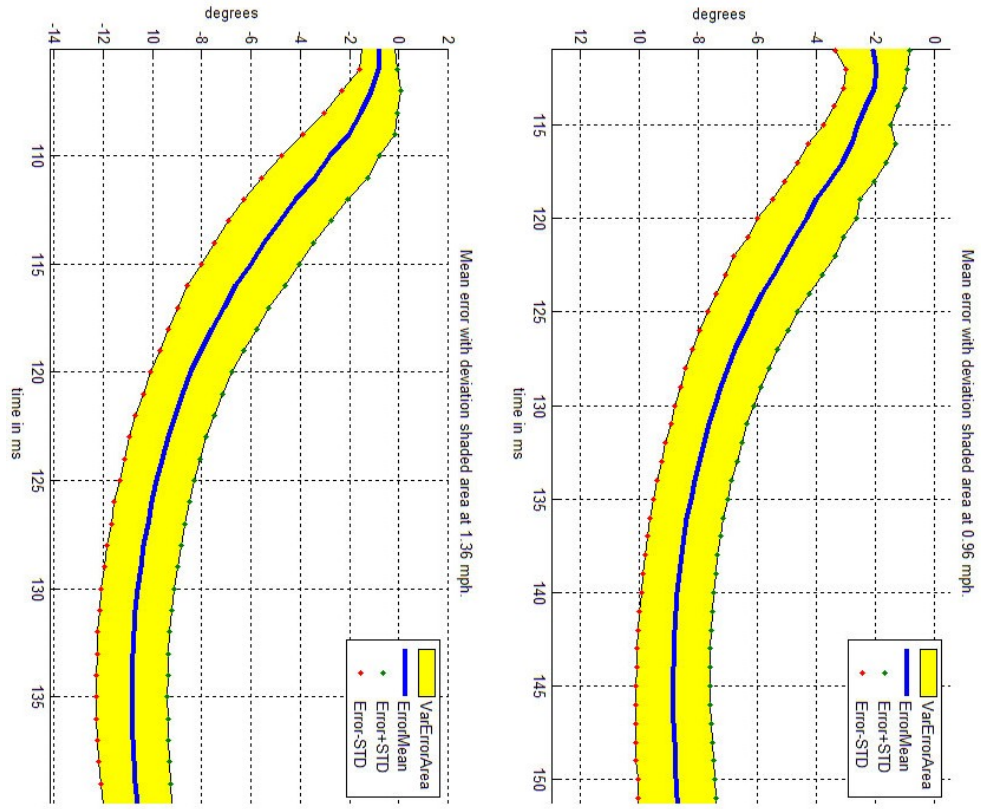
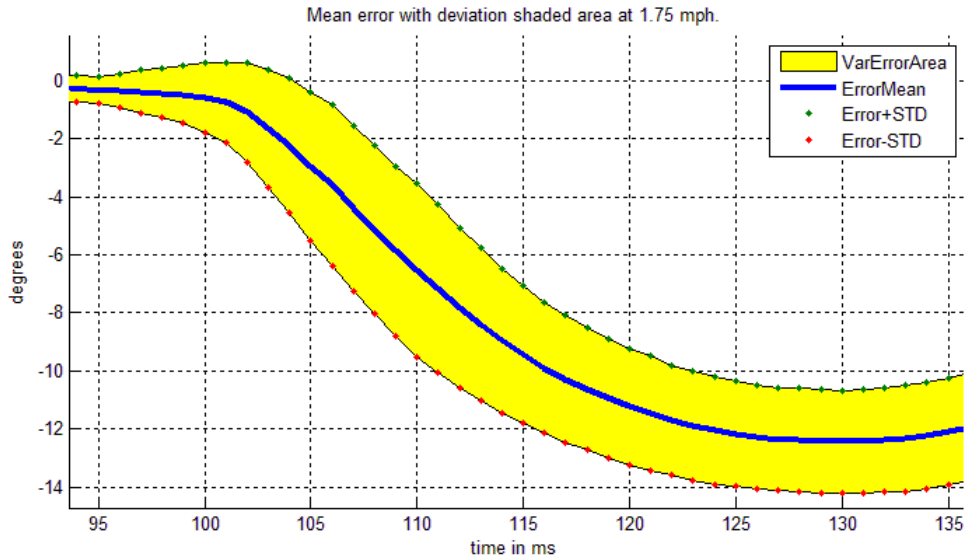
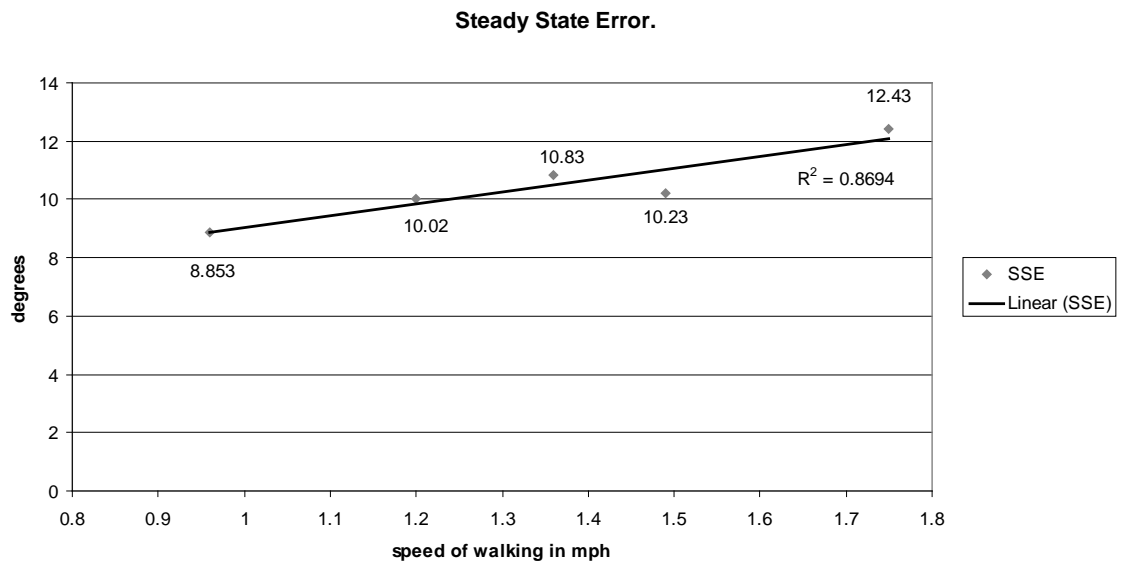


Figure 84: Mean Error During Late Swing for 0.96mph, 1.20mph, 1.36mph And 1.49mph



**Figure 85: Mean Error During Late Swing for 1.75mph**

Shown above is the experimental data for the phase of control called extension tracking, where the reference signal is approximately a ramp input for a large portion of the trajectory. Use this to determine whether or not there is a steady-state-error present in keeping with the concepts of linear-control-theory. Indeed, it is seen that in Figure 84 and Figure 85 there is a steady-state-error for each ramp input. It is also seen that as the speed of the ramp increases, the magnitude of the measured steady-state-error also increases as expected. The relation of the steady-state-error to speed of walking is shown in Figure 86. The relation is approximately linear with  $R^2=0.869$ . This data supports the validity of the nonlinear cancellation design decisions.

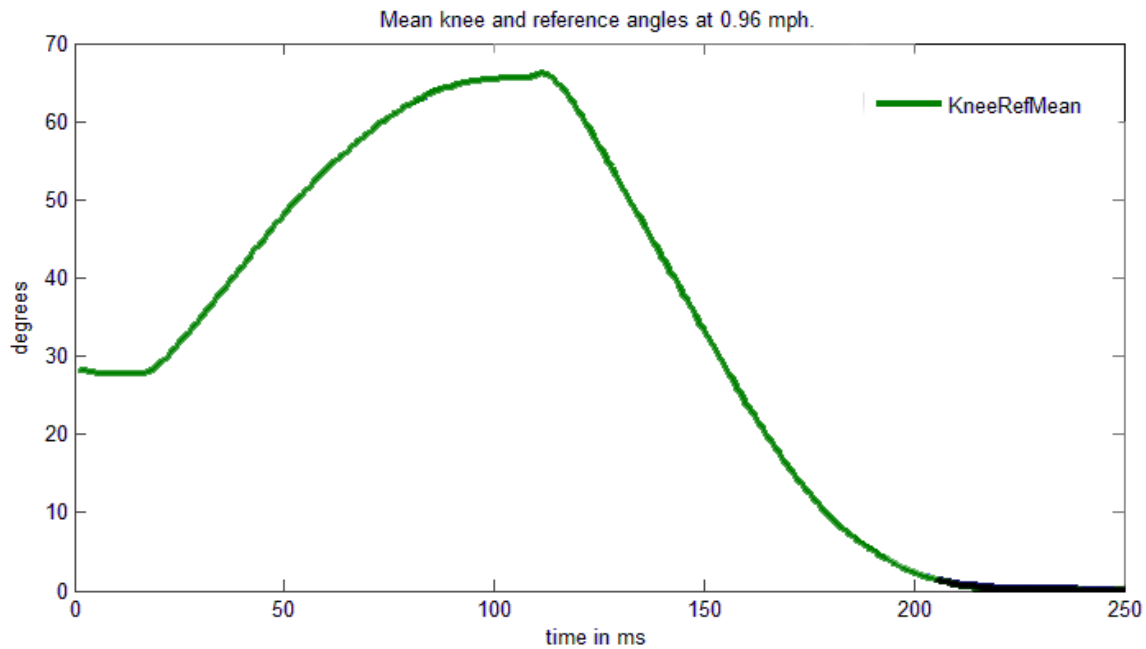


**Figure 86: Presentation of Data Comparing Steady State Error as Speed of Walking Varies**

### 8.3.5 Experimental support for jitter free cancellation of static-friction

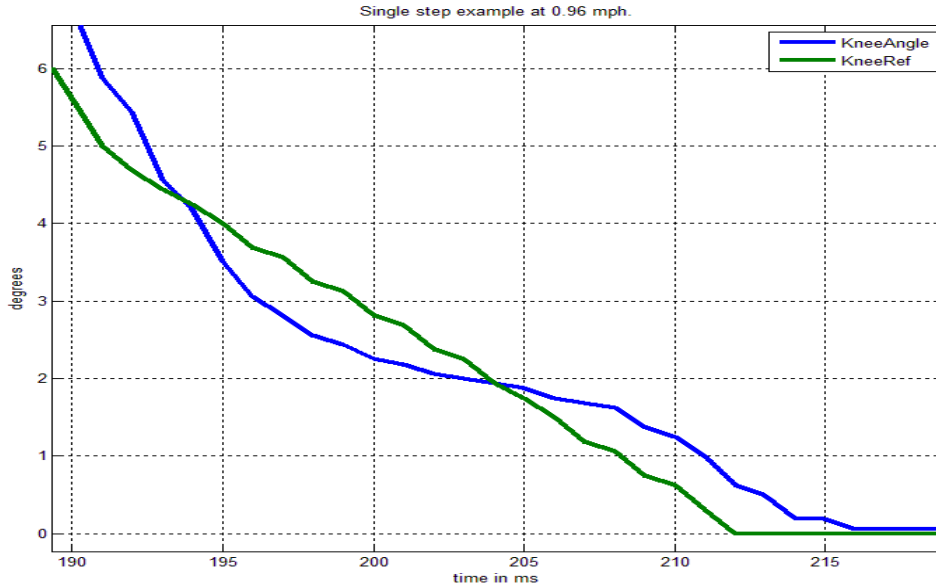
An additional nonlinear complexity in the system is the presence of a significant amount of static-friction aggregated along the power application path. Starting from the motor, through the pump, and continuing then to the linear actuator, this friction quickly builds and begins to pose a problem for control. The static-friction affects the system as a whole by making it difficult to start a position move from standstill. Additionally, it makes it more difficult to accurately move to a final stationary position. The reduction of this effect in the prosthetic knee is attempted by the implementation of the saturated integrator in the feedback controller, described in section 7.2.

Static-friction has been observed to reduce accuracy when performing a servo manipulation to a set position [82][83]. Based on this understanding, if the intended cancellation of the static-friction by the saturated integrator method does not work properly when performing a servo manipulation of the knee to a stationary angle, there will be a steady-state-error at a stationary reference signal. Alternatively, if the system does counter this static-friction properly and assuming all other assumptions are valid, then there will be no steady-state-error when positioning to a stationary reference knee angle.



**Figure 87: Highlight of Knee Reference During Forward-Late-Swing**

During normal walking motions, the best time to observe and measure the steady-state-error is at the final moments of the phase of walking called forward-late-swing. The key moment happens at the termination of the speed-based-knee-extension. This is the only motion throughout walking where the desired positional datum occurs, at full extension and zero velocity. This reference period is highlighted above in Figure 87.



**Figure 88: Close View of 0.96mph Example During Extension Tracking**

The experimental data for a sample step during the final moments of forward-late-swing when the reference is a static value of full extension is shown above. These results are typical for each step and at all speeds of walking. It is possible to use this information to determine the presence of a steady-state-error at the point where static-friction has its most noticeable effects. As seen in Figure 78, and for all walking speeds, there is very little error at the low angular velocities during the end of swing. Looking at the data, it is possible to observe the smoothness of the motion as the knee joint approaches the desired extension value. If the saturated integrator method fails to work, there will be jerky motion. There should be a starting and stopping of motion as the feedback controller overcomes and succumbs to the static friction over and over as it follows the reference trajectory. There should also be a point when the error that the feedback controller sees is not great enough to overcome static-friction resulting in the knee angle remaining stuck a small distance from the desired value. Because of this, one would also expect the knee joint to fall short of the desired value and have a significant, constant steady-state-error. This is not present in the data. Therefore, it is possible to conclude that the inclusion of the static-friction canceling integrator is successfully reducing the effect of static-friction on the system.

### **8.3.6 Experimental support for the Saturated Integrator Method.**

A consequence of utilizing the saturated integrator method is that it technically adds an integrator to the feedback controller. This is not desired. The design decision was made that by limiting the maximum torque that the integrator is capable of producing to a value slightly less than the static friction torque then traditional integrator effects will not be noticeable when the system is viewed as a whole.

It is therefore possible to verify the absence of integrator effects by performing linear analysis on the system and comparing the result to those seen in the experimental data. As was performed for the previous feedback linearization method, the steady-state-error in response to a standard input is calculated (see app 2 for derivation). For the case of a viscous, rotational inertia system under PID control when a ramp input is applied, analysis shows that there will always be zero steady-state-error.

Again, consider the phase of walking during forward-late-swing where there is a period of near linear ramp input during the extension of the knee joint after the leg has passed from posterior to anterior.

Because the integrator is limited in value, one expects that since the feedback controller is given a ramp reference, that a constant steady-state-error response from the system will be present. If, however, the system has an ever increasing error to the ramp, then the argument should be made that the knowledge of the system dynamics proves to be inadequate. If, however, the steady-state-error approaches zero, then our design rationale is also invalid, regardless of the fact that the system performs better than expected. For the understanding of the system to be accurate and the validation of the design decisions to be possible, there must be a constant, non-zero steady-state-error.

Observe the same data as in section 8.3.4 to determine whether or not there is a steady-state-error present in keeping with the concepts of linear-control-theory. Indeed, it is seen that in Figure 84 and Figure 85 there is a steady-state-error for each ramp input. This data supports the validity of the saturated integrator method.

## **9. Future Work**

Several possible avenues of future work to improve the performance of the prosthetic knee system present themselves in chapter 8. The two main possible paths are in removing false positives and improving upon instances of undesired results.

### **9.1 False Positives**

Several problems with false positives are observed when an amputee uses the prosthetic knee system. Future work will be completed to address these issues. One such instance occurs if the user does not firmly place their foot on the ground at the end of swing and subsequently loses contact with the floor. The state machine will consequently transition to a swing-phase, prematurely. A potential solution to this problem would be to implement a delay after foot contact where during that period if the foot comes off the ground, then the state machine will not transition out of stance. However, utilizing this delay might make the prosthetic knee system behave improperly when it is actually intended to transition to swing-phase quickly.

A similar false positive happens during standing still, when the amputee shifts all of their weight off of the prosthetic leg to reduce discomfort from the skin pressures induced by the socket on the residual limb. Consequently, the prosthetic system observes the foot coming off the ground and assumes the person wants to take a step, erroneously. A simple correction for this false positive would be to modify the ground sensing thresholds such that it requires much less vertical loading to initiate the transition from stance to swing while the foot is directly beneath the user.

Another example of difficulties arisen from the differences between development testing and amputee testing occurs when stair climbing is attempted. The mass and rotational inertia differences between development testing and that of the amputee are very large. This causes far more significant instabilities than were expected for the specific case of ascent swing. Future work will be conducted around the performance of the feedback controller in this situation. A simple readjustment of controller gains, specifically increasing the derivative gain, will restore stability. Although, it should be noted that this restored stability will be at the cost of reduced speed in ascending stairs and taking steps. This trade-off will need to be considered and finding the correct balance will take time.

Also, because the way the amputee naturally attempts to climb stairs, more difficulties arose when they used the prosthetic knee system. This resulted in very unnatural motion when

combined with the instability previously discussed. One solution for this problem would be to have a more extensive training period to restore early thigh rotation to stair stepping. Alternatively, the stair ascent reference generator could be modified so that it does not operate purely off of thigh angle. Future work will be completed to incorporate the accelerometer signal into the feedback controller. A possible way to do this would be to integrate the accelerometer signal twice to calculate an approximate value for the vertical displacement achieved as the amputee attempts to step up the stair. This signal would have significant drift and other unreliable characteristics; however, the extension of the planted knee would raise their torso up one step, a distance significant enough that it could be observed real-time. This displacement could then be used to modify the thigh joint location in the RGA method.

This would also create the obvious distinction between a stair step and a small forward step that was a difficulty present in the OCP theory. A stair step would occur when the foot comes off the ground and the torso has been raised vertically; otherwise, a small forward step is intended by the user.

## **9.2 Undesired Results**

Several instances of undesired results are observed when the prosthetic knee system is tested. Future work will be completed to address these issues. One such example occurs when attempting to show that the prosthetic knee system reduces the phenomena of hip-hike. The amputees could be coached to drop their hips during testing, but the behavior regressed when their attention was not maintained. Because of this, it should be possible to maintain proper hip motion without conscious attention through extended use of the prototype. If the amputees were able to take the prosthetic knee system home to train and live with the system, future work could be completed to observe whether or not these maneuvers would improve over time.

The results of section 8.3.3 appear to neither entirely support nor discredit the decision to cluster inertial effects into an un-modeled dynamic noise as acceptable. The system does not behave as is expected. Something is missing from the system model that causes the behavior seen. Because of this and the difficulties encountered with friction in 8.3.5, we suspect that the underlying cause for the irregularities seen is an incomplete understanding of the true nonlinear behavior of friction in the system. Future work will be conducted in completing the system identification for the friction along the prosthetic knee drive train so a better feedback controller can be designed.

With the successes seen in chapter 8 combined with the future work proposed herein, amputees using the powered prosthetic knee system will not only be able to restore the ability to go up stairs, but also improve their safety and ease of walking on level ground. The powered prosthetic knee system will help amputees to maintain healthy activity levels and improve their overall quality of life.

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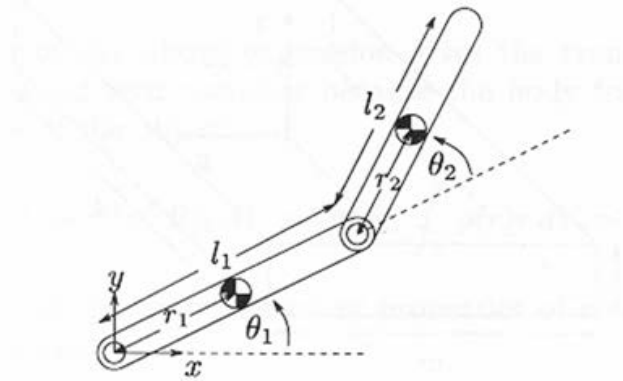
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## Appendix 1:

Developing this system is derived in two parts: First, the dynamic equations of motion are formulated using an inertial frame of reference at the thigh joint. Second, the equations' constants are identified using experimental data.



$$\begin{bmatrix} \alpha + 2\beta c_2 & \delta + \beta c_2 \\ \delta + \beta c_2 & \delta \end{bmatrix} \begin{bmatrix} \ddot{\theta}_1 \\ \ddot{\theta}_2 \end{bmatrix} + \begin{bmatrix} -\beta s_2 \dot{\theta}_2 & -\beta s_2 (\dot{\theta}_1 + \dot{\theta}_2) \\ \beta s_2 \dot{\theta}_1 & 0 \end{bmatrix} \begin{bmatrix} \dot{\theta}_1 \\ \dot{\theta}_2 \end{bmatrix} = \begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix} + \mathbf{T}_{\text{gravity}}$$

where

$$\alpha = \mathcal{I}_{z1} + \mathcal{I}_{z2} + m_1 r_1^2 + m_2 (l_1^2 + r_2^2)$$

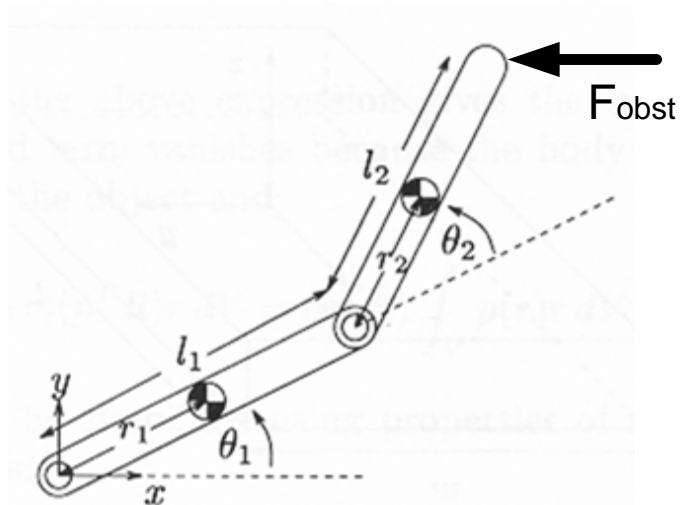
$$\beta = m_2 l_1 r_2$$

$$\delta = \mathcal{I}_{z2} + m_2 r_2^2$$

**Figure 89 : Inertial Leg Model And Equations Of Motion**

To begin the derivation of the equations, the thigh joint is assumed to be moving at a constant speed and can therefore be calculated as being stationary. The two links of the system, the thigh and shank, were modeled as uniform prismatic bars with an unknown mass and inertia. The initial conditions for this system are considered to be the instant after toe-off and the final conditions are at heel-strike. During this intervening period the system is assumed to operate primarily on its own, with the only significant torques being applied at the knee joint by the prototype. While there is obviously a torque applied at the thigh joint during swing, it is relatively small and therefore can be ignored for the sake of calculation [84].

The equations governing the motion of this two-link system are derived using Lagrangian dynamics and are shown above. It would then be possible to obtain the values of the masses and inertias completely from measuring the physical components in the prototype. However, since the equations will be evaluated with real-time sensor data, it would be better if the recorded data was used to calculate the required variables. Comparing the number of equations to the number of unknowns, it is possible to use algebraic methods to calculate the masses and inertias of the links required to produce the observed recorded motions.



**Figure 90 : Modified Inertial Model with Obstacle Interaction Force**

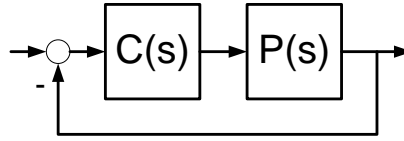
Once these values are known, the model is complete and can therefore be used in conjunction with a second dynamic model, shown above. The notable difference in this model is that there is now an unknown horizontal force acting on the toe. Using the values from the previous model in conjunction with real-time data during operation, it is possible to detect if an opposing force is applied at the toe or if it is behaving as if it is in free pendulum motion. This force can be assumed to be arising from the interaction of the toe with a stationary obstacle, for example a bump in the ground.

Unfortunately this method has its faults. It requires the mass and inertia for the thigh to be calculated for each person, which may not be significant considering the prototype as a product that will only be used by one person. It will however require some sort of training period for each person and this added requirement for a specialist's time is undesired.



## Appendix 2:

To validate earlier assumptions on the behavior of the system, the steady state error of the system under the effect of a ramp input must be analytically found. The steady state error is discovered by applying the final value theorem to the error signal of the feedback controller. The result of this analysis is that the steady state error is dependent on the number of pure integrations in the forward path, also known as the system type.



**Figure 91: Block Diagram of Simple Feedback Controller**

Consider, in general terms, a simple linear feedback controller configured with a unity feedback loop; controller transfer function,  $C(s)$ ; and plant transfer function,  $P(s)$ . To apply the final value theorem the system must be stable under feedback control. One method of checking this is to determine that there are no poles in the right half side of the imaginary plane. To accomplish this, consider the controller and plant separately.

The controller in the prosthetic knee system is either a PID or PD feedback controller. Each controller is designed to increase the stability of the closed loop system. Therefore, certain characteristics can be noted about the system even though we maintain general terms. One such characteristic is that all poles and zeros originating from the feedback controller are purposely placed in the left half of the imaginary plane.

If the controller is a PD controller, then it can be seen that the controller contributes one stable zero in the left half of the imaginary plane. Alternatively, if the controller is a PID controller, then it contributes two stable zeros and one pole at the origin. The plant itself contributes no zeros but it does contain two poles. One pole is at the origin. The other can be assumed to be located in the left half of the imaginary plane, because it is originating from frictional effects. The corresponding equations are shown below.

$$C_{PID}(s) = K_{gain} \left[ \frac{K_p s + K_I + K_D s^2}{s} \right]$$

$$C_{PD}(s) = K_{gain} [K_p + K_D s]$$

$$P(s) = \frac{1}{s(s+a)}$$

When the PID controller and the plant are combined, the open-loop transfer function has a total of two left-half-plane zeros, two poles at the origin, and one left-half-plane pole. Drawing a root locus for any such system yields stable behavior under all possible feedback gain magnitudes greater than zero.

When the PD controller and the plant are combined, the open-loop transfer function has one left-half-plane zeros, one pole at the origin, and one left-half-plane pole. Drawing a root locus for any such system, also, yields stable behavior under all possible feedback gain magnitudes greater than zero. It is therefore acceptable to apply the final value theorem to this system to obtain the steady state error for a ramp input.

The open-loop transfer function,  $G(s)=C(s)P(s)$ , can further be written as  $G(s) = N(s) / s^n D(s)$ . In this general form, the open-loop transfer function is composed of  $N(s)$ ,

the numerator polynomial in  $s$ ;  $D(s)$ , the factored denominator polynomial in  $s$ ; and  $s^n$ , the powers of  $s$  that can be factored out of the denominator polynomial. The number  $n$  is used to describe the “type” of the system. The PD controlled plant is a type 1 system while the PID controlled plant is a type 2 system.

The error transfer function relating the reference signal to the error signal is found through block diagram reduction to be  $\frac{E(s)}{R(s)} = \frac{1}{1+G(s)}$ . Combining this equation with the eqn#X

yields 
$$\frac{E(s)}{R(s)} = \frac{s^n D(s)}{s^n D(s) + N(s)}.$$

Applying the final value theorem to the error signal gives the steady state error,

$$e_{ss} = \lim_{x \rightarrow 0} (sE(s)) = \lim_{x \rightarrow 0} \left( sR(s) \frac{E(s)}{R(s)} \right).$$

For a ramp input,  $R(s)$  is  $\frac{1}{s^2}$ . Therefore for a type 1

system, by taking the limit as  $s$  approaches zero, the steady state error is found to be  $\frac{D(0)}{N(0)}$ ,

which is a finite value. For a type 2 system, taking the same limit finds the steady state error to be  $\frac{0}{N(0)}$ , which is zero.

This analysis concludes that if the system described earlier is controlled by a PID feedback controller, then there should be no steady state error for a ramp input. Also if the system is controlled by a PD feedback controller, then there should be a finite error.