

Volitional Control of a Powered Prosthetic Ankle

by

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A Thesis Presented in Partial Fulfillment
of the Requirements for the Degree
Master of Science

Approved June 2012 by the
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ARIZONA STATE UNIVERSITY

August 2012

ABSTRACT

Approximately 1.7 million people in the United States are living with limb loss and are in need of more sophisticated devices that better mimic human function. In the Human Machine Integration Laboratory, a powered, transtibial prosthetic ankle was designed and build that allows a person to regain ankle function with improved ankle kinematics and kinetics. The ankle allows a person to walk normally and up and down stairs, but volitional control is still an issue. This research tackled the problem of giving the user more control over the prosthetic ankle using a force/torque circuit. When the user presses against a force/torque sensor located inside the socket the prosthetic foot plantar flexes or moves downward. This will help the user add additional push-off force when walking up slopes or stairs. It also gives the user a sense of control over the device.

DEDICATION

To my family and friends who supported me through this entire project

ACKNOWLEDGMENTS

First and foremost I would like to thank Dr. Thomas Sugar for offering me the opportunity of pursuing my Master's degree and being able to work on this interesting project. In addition, I would like to thank Dr. Stephen Helms-Tillery for his continuous support during this project. His dedication and availability have been invaluable to me. Furthermore, I would like to thank Dr. Marco Santello for his support and for serving on my committee.

I would also like to thank Mark Werner of Arise Prosthetics for his ideas and professional help on assembling our prosthetic ankle. Finally, I would like to thank my lab mate Thierry Flaven for his valuable support and ideas.

TABLE OF CONTENTS

	Page
LIST OF TABLES	vi
LIST OF FIGURES	vii
LIST OF EQUATIONS.....	viii
 CHAPTER	
INTRODUCTION	1
LITERATURE REVIEW	5
EMG Systems.....	5
EMG Control of Upper Limb Prosthetics	6
EMG Control of Lower Limb Prosthetics	14
Peripheral nerve interface	14
Innervated nerve system from Northwestern	17
EMG in the socket	19
Uses of Force Sensing Resistors in lower limb prosthetics	21
METHODS	23
Background on Force Sensing Resistors.....	23
Background on Voltage Divider Circuit.....	25
Background on calibration	27
Background on Wheatstone bridge	30
Calculation of socket forces when sitting and standing.....	31
Calculation of socket forces when walking.....	33
Tibia Based Controller	36

CHAPTER	Page
RESULTS.....	45
FSR system for sitting	45
FSR system for standing	48
FSR system for walking.....	48
CONCLUSIONS.....	53
FUTURE WORK	54
BIBLIOGRAPHY	55

LIST OF TABLES

Table	Page
1. FORCE SENSING RESISTOR FOR THE AMPUTEE	45
2. FORCE SENSING RESISTOR FOR THE ABLE-BODIED	46

LIST OF FIGURES

Figure	Page
1. SPARKy	2
2. EXOSKELETON	15
3. FORCE SENSING RESISTOR.....	24
4. VOLTAGE DIVIDER DIAGRAM	25
5. CALIBRATION CURVES	29
6. EXPERIMENTAL CALIBRATION CURVES.....	29
7. WHEATSTONE BRIDGE SCHEMATIC	30
8. GAIT PERCENT VS. ANGLE	37
9. “BEAN GRAPH”	39
10. GAIT PERCENT VS. POLAR ANGLE	39
11. GAIT % VS. STRIDE LENGTH VS. POLAR RADIUS.....	41
12. GAIT % VS. STRIDE LENGTH VS. POLAR RADIUS.....	41
13. AMPUTEE SITTING	46
14. AMPUTEE MOVING FOOT TO GROUND	47
15. FSR WALKING OF ABLE-BODIED SUBJECT	49
16. GYROSCOPE WALKING OF ABLE-BODIED SUBJECT	49
17. FSR WALKING OF ABLE-BODIED SUBJECT	50
18. GYROSCOPE WALKING OF ABLE-BODIED SUBJECT	50
19. FSR WALKING OF AMPUTEE.....	51
20. FSR WALKING OF AMPUTEE.....	52

LIST OF EQUATIONS

EQUATION	Page
(1)	9
(2)	26
(3)	31
(4)	31

INTRODUCTION

Control systems have lagged behind the advances in mechanical components of active prosthetics since the beginning of their invention. In the Human Machine Integration Laboratory, at Arizona State University, where this research is being conducted, this is still the case. Thus, most prosthetic research today focuses on control systems which are devices that manage, command, and regulate the behavior of active prostheses.

In 1997, the Department of Defense asked researchers to develop an active prosthetic ankle for service personnel who want to return to active duty. At Arizona State University in 2007, Dr. Thomas Sugar and graduate students: Dr. Matthew Holgate, Dr. Kevin Hollander and Lt. Colonel Joseph Hitt, PhD, accepted the challenge and created a prosthetic known as SPARKy. SPARKy stands for Spring Ankle with Regenerative Kinetics, figure 1.



Figure 1: SPARKY. Spring Ankle with Regenerative Kinetics.

The actuator, called the robotic tendon, stores and releases energy during the gait cycle mimicking human behavior. In this system, the motor drives a screw, which stretches a pair of metal springs [1]. The advantage of this system is that the springs lower the peak power requirement and energy density needed during the gait cycle. Because energy is stored in the springs, a lightweight motor is used to adjust the position of a pair of uniquely tuned springs that provide most of the power required for gait [2]. Thus, less energy is required from the amputee as compared to wearing only a passive energy return ankle. The researchers found that the spring and motor combination was able to amplify the motor power by three-fold [1]. This significant finding allows SPARKy to be downsized from a 6 or 7 kg motor system to a 1 kg system [3]. This is a significant weight savings

for those who wear a prosthetic ankle. The device is built to take advantage of the functional mechanics of gait. A gait cycle is the natural motion of walking, starting with the heel strike of one foot and ending with the heel strike of the same foot.

Advances of the robotic tendon were formulated to make the device more compact and to increase energy efficiency. First, the actuator tube shape was changed from square to circular. This allows the springs to be mounted parallel to the socket. Second, the lead screw was custom-made with the most efficient threads and pitch. The pitch of the screw is the distance from the crest of one thread to the next crest. The end of the custom-made screw was designed to attach directly to the motor shaft, thus eliminating the gear box [3].

Since the mechanical aspects of SPARKy have been optimized, a very robust controller needed to be designed. Dr. Kevin Hollander determined the power profile needed for the motor. He, then, created a dynamic pace controller to match the profile [4]. Dr. Matthew Holgate thought that using the dynamic pace controller was not the best solution and, instead, created a tibia based controller [2]. He used the tibia angular velocity and the tibia angle to create the controller [2]. He utilized average gait data and interpolated the points in between [2].

The interpolation of those points led to a minor problem. The controller only made the prosthetic ankle feel comfortable at one speed. Therefore, this thesis focuses on giving the amputee more control over the

prosthetic ankle. The first phase of this goal was to get the foot to drop to the floor when the patient was sitting. The second phase of this goal was to get able-bodied data and change the control algorithm to make the prosthetic feel comfortable at every speed.

LITERATURE REVIEW

EMG Systems

Electromyography (EMG) has been used to control prosthetic arms and legs for many years. Electromyography is a technique in which muscle activation potentials are gathered by electrodes placed on the subject [5]. These potentials can be used to track which muscles the subject is activating and with what force [5]. Correlating EMG activity with force is still controversial. There are two different types of interfaces for connecting the electrodes: invasive and non-invasive. The invasive methods gather control signals directly from the amputee's nervous system either using brain or muscle implants [6]. This allows the researcher to get a high quality signal from the exact location they want. However, a surgical procedure is rather involved, and the subject has to keep the implant wires sterile. There are also many psychological issues [6]. The non-invasive method does not have these problems. Thus, non-invasive electrodes are easier to handle and maintain. The problem with the non-invasive method is that it requires better signal conditioning. Since the signal conditioning is easy to implement and can be done with software, the non-invasive method is generally used by most researchers.

The concept of myoelectric control has been rather simple. The electrical signal from a muscle was used to control the movement of the prosthetic device. In a myoelectric controlled prosthesis, the electric

signal comes from a muscle remnant in the amputee's residual limb. This limb still had normal innervations so it was subject to voluntary control.

The control systems for prosthetics have come a long way in the past twenty years. The early control systems were cable controllers and switch controllers which use other parts of the body to control the prosthetic. More recently, researchers have used electromyography or EMG control systems. These systems used electrodes to gather electrical signals produced by muscles [5-11, 34]. The most cutting edge control systems are those that used neural or brain-machine interface, which connect the brain and prosthetics directly [25].

EMG Control of Upper Limb Prosthetics

Control is the biggest problem with most prosthetic arms. Fine control and the use of single joints is the issue which electromyographic controllers are trying to resolve. The grasping of door handles and car keys is almost impossible for most of the current models of prosthetic hands. The amount of force used is another cause for concern. The last, and probably most important concern, is that the user has no feedback. There are many well-known researchers that are working on these issues.

Several researchers in both the Netherlands and Spain developed an upper limb prosthesis called the MANUS-Hand [7]. They used the action potentials in the arm to command grasp modes and the hand controller implements stable grasps based on the stiffness control strategy. Unlike most upper-limb prosthetics that use flexor and extensor

muscles to control every axis of the hand, the MANUS-Hand uses pattern recognition to determine what the user wants the prosthetic hand to do. The researchers created a three bit control language. Each byte corresponds to a muscle contraction, which leads to twenty-seven possible combinations. However, since the controller cannot tell the difference between “AB1” and “1AB”, there are only eighteen unique combinations. In particular, those patterns in which a high EMG amplitude bit is followed by a lower EMG amplitude bit resulted in added difficulty. The master controller acquires EMG signals during 10 seconds at a sampling rate of 400 Hz to determine the noise level. The user is then asked to repeat the command to determine EMG thresholds. After calibration, EMG is acquired at 5 kHz and the master controller keeps recognizing valid commands [7].

Researchers in Italy at the ARTS Laboratory have created a controller that uses an electromyography discrimination algorithm to control their soft hand [8]. The algorithm is very simple and robust which allows the patient to open and close the hand. The researchers extracted electromyographic signals using commercially available electrodes placed on two antagonist muscles. These signals are the input into a simple finite state machine which is a mathematical model used to design computer programs and digital logic circuits. The finite state machine stays in the open state until the electromyography signal produced by flexor contraction of the forearm overcomes a relative threshold [8].

Also, in Italy, a prosthetic hand was produced at the CENTRO INAIL. Using the "all or nothing" type of EMG control algorithm caused by the contraction of two remaining muscles in the residual limb, it produced the opening or closing of the prosthetic hand at the fastest speed possible for the whole contraction time. The EMG signals from the residual limb were picked up by means of surface electrodes and processed by means of the following steps. The signal was first amplified by an amplifier with a gain of 80 dB and sensitivity 10 μ V. Then, the signal is passed through a high-pass filter and a band-eliminating filter to eliminate the power line noise. Following the filtering, the signal was double half-wave rectified. The negative half is eliminated so only the positive part is kept. The next step was to integrate the signal which corresponds to a low-pass type filtering with a cut off frequency of about 3 Hz. Next, the signal was sampled at a frequency of 100 Hz. The last step was sending the signal through a moving average filter. The output signal is used to move the arm but a proportional controller needed to be developed which uses an electromyography impulse to obtain a grip on an item. This impulse starts the closing of the hand. The force sensing resistors on the fingers produce a signal that is greater or equal to a "contact threshold" value. When the fingers stop, the object has been grasped [9].

In Japan, another group of scientists have designed a three degree of freedom exoskeleton for human upper-limb motion assistance. The researchers used the electromyography signals from the shoulder and the

elbow of the patient as an input to their controller. The exoskeleton is controlled by the EMG signal when the muscle activity is high and by a force based controller when the muscle activity is low. During times when there is intermediate muscle activity, both controllers work simultaneously. The EMG controller generates the desired torque for each joint of the exoskeleton based on the EMG signals of the related muscles. However, it is important to take into account the hand trajectory for the practical application. The experimenters used the root mean square method to process the raw EMG signals. The root mean square value is a measure of power of the signal and is widely used in most applications.

$$RMS = \sqrt{\frac{1}{N} \sum_{i=1}^N V_i^2} \quad (1)$$

The controller was built using nine shoulder controls and three elbow controls for a total of twenty seven different combinations. A combination was chosen based on twenty one rules that the researchers created [10].

EMG Control of Lower Limb Prosthetics

Electrodes may be connected with many muscles in the leg that facilitate pattern recognition. Researchers at the University of Michigan connected electrodes to the tibialis anterior, soleus, medial gastrocnemius, lateral gastrocnemius, vastus medialis, vastus lateralis, rectus femoris and medial hamstring [11]. They used these muscles to power their ankle-foot orthosis. The soleus electromyography (EMG)

activated the artificial plantarflexor muscle and inhibited the artificial dorsiflexor muscle in the orthosis. Tibialis anterior EMG activated the artificial dorsiflexor muscle. These signals were used as an input to a proportional myoelectric controller. This controller reduced the peak torque for plantar flexion and was able to supply more than enough torque for dorsiflexion. Researchers in the field of biomechanics can use this technique to study human muscle activation patterns in response to increased strength of the musculoskeletal system [11].

At the University of Rhode Island, Dr. He Huang, Dr. Todd Kuiken and Dr. Robert Lipschutz developed a new electromyographic pattern recognition strategy [12]. They used the gluteal and thigh muscles of eight able bodied and two amputee subjects to gather the data. The seven movement modes investigated were: level-ground walking, stepping over an obstacle, ascending and descending stairs, ipsilateral turning, contralateral turning and standing still. Each trial was repeated so that at least 10 complete stride cycles were recorded. The data was then used to create and test a phase-dependent classifier, which is a strategy for identifying locomotion modes. The classifier was able to recognize seven of the ten trials but had trouble with the time varying signal. The researchers therefore determined that the system although very accurate was not good enough to be used as the independent controller of a lower leg prosthetic [12].

In Israel, at Tel Aviv University, in collaboration with the Kaye College of Education, researchers looked at the gait of trans-tibial amputee mainly on the time-distance parameters and EMG activity. Fourteen male subjects were asked to walk at a comfortable pace for a distance of 3.6 meters. A portable wireless EMG system was used to record muscle activity signals during ambulation. The electrodes were placed on the vastus medialis and the biceps femoris of both legs. The researchers found that the gait of both legs was symmetric. The research showed that the major factors which lead to gait symmetry were the residual limb being pain-free, the optimal fit of the socket, the proper alignment of the socket and the physical condition of the amputee [13].

At Catholic University of Leuven in Belgium, a group of researchers used hip and residual limb muscles to gather electromyographic signals to control the trans-femoral prosthetic. The purpose of their research was to find the minimum amount of data required to recognize the locomotive mode and detect the intent to change to a different mode. The researchers connected electrodes to the rectus femoris, gluteus maximus, gluteus medius, adductor longus, tensor fasciae latae and the hamstring muscles. Three experiments were conducted: walking on level ground, walking up a slope and walking down a slope. Twenty-four able-bodied subjects were utilized in the first set of experiments, which tested only mode recognition. The data collected was used as a reference for the second group, which included a single amputee. Both the mode

recognition and intent of the subject were tested by the amputee. The results indicated that muscular activity of three muscles: gluteus maximus, gluteus medius, and tensor fasciae latae combined with foot-ground contact signals, supply sufficient information for mode recognition in the able-bodied population. The data from the amputee seems to agree with the able-bodied population except for the muscle activity in the tensor fasciae latae which was higher for the amputee. The intent recognition data allowed the researchers to see the transitions between the different modes. Although the data was repeatable, the researchers suspect that they will see different patterns for each individual [14].

At Drexel University, Dr. Gordon Moskowitz and his graduate students have been using EMG signals for pattern recognition for use in a trans-femoral prosthetic. They have been trying to determine if spatial patterns of EMG from muscle sites about the thigh and hip can be separated into: patterns resulting from knee flexion, patterns resulting from knee extension and patterns resulting from hip activity involving no knee activity. They ran two sets of experiments. The first was with the knee and hip moments held fixed while the joint angles were varied. In each configuration both flexion and extension torques were applied at the hip and knee, with each joint acting first in isolation from the other, and then with each joint acting in all possible combinations with the other. In the second experimental set, leg position was fixed and knee torque was varied. No attempt was made to keep hip torques constant throughout the

set. Twenty-five data points were then selected and used as training data for the classifiers. Linear discriminant analysis (LDA) was employed to classify the data. LDA is a method used in statistics and pattern recognition to find a linear combination of features which separate the three patterns [15].

Dr. Samuel K. Au, Dr. Paolo Bonato, and Dr. Hugh Herr at the Massachusetts Institute of Technology developed an EMG-position controlled system for an active ankle-foot prosthetic. They developed a biomimetic EMG-controller which predicts the amputee's motion by simulating the dynamics of the missing limb's neuromuscular system. Given EMG signals and the corresponding desired ankle-foot trajectory, they estimated the parameter values using the Matlab Optimization Toolbox. During the experiment, the amputee controlled the residual muscles, which previously actuated his ankle, to mimic pre-programmed motion trajectories on the graphical display. Since the goal of this investigation was to develop an EMG-controlled, ankle-foot prosthesis that mimics natural human ankle movements, it was desirable to measure EMG signals from those residual limb muscles that previously actuated the biological ankle before amputation. Thus, using fine wire electrodes, they recorded from the gastrocnemius and soleus muscles for prosthetic ankle plantar flexion control, and from the tibialis anterior for prosthetic ankle dorsiflexion control. The biomimetic EMG controller appeared to

generate a smoother, more natural ankle movement pattern than the neural network EMG-controller [16].

At the University of Waterloo in Canada in collaboration with Southern Illinois University, researchers performed a biomechanical analysis using force plate data along with a six channel EMG telemetry system on below the knee amputees. The five muscles that were monitored using surface electrodes were: gluteus maximus, bicep femoris, semitendinosus, rectus femoris, and vastus lateralis. Out of the five amputees, researchers were only able to create complete EMG profiles for three of them. These three amputees were asked to walk back and forth on a 10 meter walkway for 32 seconds. Fifteen strides were taken from this trial and were plotted along with data from an able-bodied subject. The amputees have a modified motor pattern derived from the knee and hip muscles. This can be caused by the amputees' neural system recognizing the asymmetric motor pattern and trying to compensate for it [17].

An electromyographic system would be an excellent choice to control SPARKy because it is reliable and simple. From the EMG literature, it is apparent which leg muscles most researchers find important for gait. This literature as well as the literature from the King's College School of Medicine and Dentistry presented later allowed us to determine where the force sensing resistor should be placed. Our FSR system was designed to be simple and reliable using a standard voltage divider circuit.

Peripheral nerve interface

At Arizona State University, Dr. Jiping He and his graduate students developed neural interfaces for rehabilitation of lower limb function in spinal cord injuries. Their research focused mainly on recovery after spinal cord injuries and restoration of limb mobility. The rehabilitation technique that they use is known as functional electrical stimulation (FES) and epidural spinal cord stimulation (ESCS). FES has a long dated history and can automatically adjust stimulation parameters to achieve a specified function. ESCS is a relatively new technique being discovered by Cook and Weinstein in 1973. These stimulations decrease the effort and limb heaviness, which in turn leads to improved locomotion performance [18].

Researchers at Case Western University have developed a hybrid orthosis for standing, walking and stair climbing for patients with spinal cord injuries. Their device was a hybrid Neuroprosthetic that combines mechanical braces and functional electrical stimulation. The mechanical braces consisted of two ankle foot orthosis that have a metal brace that runs from the knee to the hip. At the hip, there was a rotary actuator and a brace that wraps around the human. The device can be seen in figure 2.

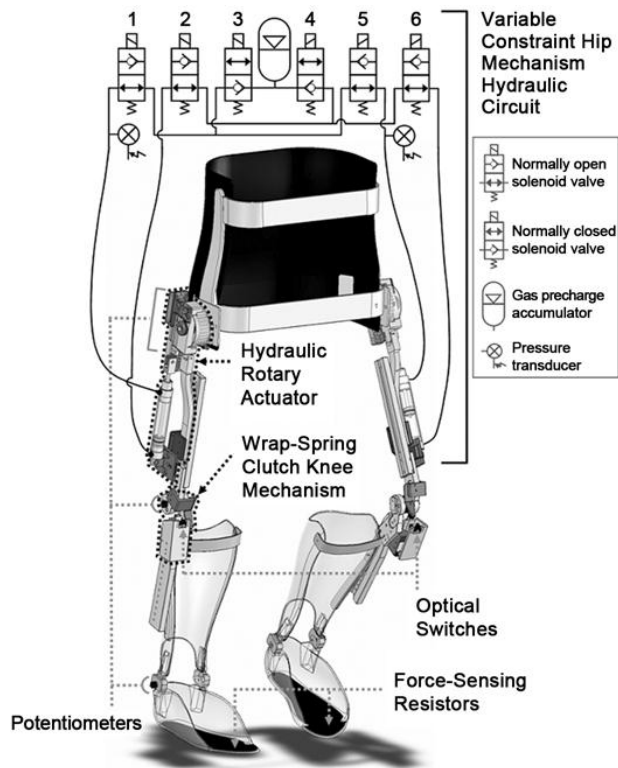


Figure 2: Exoskeleton. This exoskeleton helps patients stand, walk and ascend stairs.

The functional electrical stimulation is used in combination with the controller to achieve the necessary range of hip and knee motion to accomplish a task. The entire device was then bench tested, and tested using an able bodied subject. The last set of experiments was performed with a patient with paraplegia, an impairment in motor or sensory function of the lower extremities. During this set of experiments, the investigators specified a set of muscle stimulation for walking. The researchers deem the device feasible to help a patient learn to walk, ascend and descend stairs and stand [19].

At Lehigh University in Pennsylvania, Dr. Kenneth Horch and Dr. Gurpreet Dhillon used electrodes to control and give sensory feedback to

patients operating the Utah prosthetic arm. Electrodes were implanted inside median nerve fascicles in four male patients. The researchers then determined which electrodes they could use for touch/pressure sensation by sending a short pulse train to the electrodes. Next, the investigators had to determine which nerve fascicles could be used for motor control. This was done by connecting individual electrodes to motor neuron fibers and using a reference electrode as input to a differential amplifier and the neural firing rate was recorded. The actuators were controlled in force and torque mode by leaky integration of the neural firing rate with a linear decay rate. The results of the study show that it is feasible to give the patient sensory feedback about grip strength and limb position using implanted electrodes. In addition, the results show that it is possible to create a volitional controller for a prosthetic arm using the peripheral nerve [20].

Innervated nerve system from Northwestern

In the United States, a major breakthrough at the Rehabilitation Institute of Chicago by Dr. Todd Kuiken has revolutionized electromyographic controllers. He performed a surgery known as targeted muscle reinnervation, which takes the nerves from the amputated arm and moves them into the chest cavity. This procedure has been done four times: three times with a male subject and once with a female subject. Electrical burns, an automobile accident and an industrial accident injured

the male subjects. The female was injured in a motorcycle accident. Grids of monopolar surface EMG electrodes were placed over the chest cavity where the nerves were innervated. The chest contractions send out electric signals that can be interpreted by the grid of electrodes and have the possibility to be interpreted by a microprocessor. The signals can then be given as an input to a prosthetic device that will move according to the patients' thoughts. This procedure also allowed sensory feedback to be given back to the patient [21].

There were sixteen trials conducted in which the patient was asked to imagine and actuate movements of the missing arm. Eleven repetitions were run on each of the movements. After the signal was processed using a 5th order Butterworth filter, it was used as the input to a pattern recognition algorithm that would move the prosthetic arm [22].

The study shows that the targeted muscle reinnervation procedure increased the accuracy of the pattern recognition algorithm. This in turn caused the prosthetic arm to move in a more realistic way and made the patient feel like the prosthetic device is their real arm. The target muscle reinnervation procedure also gave more information about the control data as related to the physiology of the missing limb [22].

More recently, the researchers at the Rehabilitation Institute of Chicago Bionics lab have tried to apply the pattern recognition algorithm to lower limb prosthetics. They connected electrodes to the following nine muscles: semitendinosus, sartorius, tensor fasciae latae, adductor

magnus, gracilis, vastus medialis, rectus femoris, vastus lateralis, and the long head of the biceps femoris. The location of the electrodes was determined using researcher knowledge of the anatomy of the leg and test contractions. Twelve seconds of data was collected for each of a series of movements including plantar flexion and dorsiflexion. The data was then fed into the pattern recognition algorithm. Next, the researchers gathered another set of twelve second data clips to improve the accuracy of the algorithm. Then, two virtual limbs were created and the patients were asked to do a series of trials making the virtual limb replicate a specific motion. Future work will involve the use of a physical prosthetic and more subjects [23].

EMG in the socket

In another attempt at creating a control algorithm and to keep the electrodes protected, researchers have tried to use electromyography inside the socket. A set of researchers at University of Toronto have created a self-contained mechanomyography-driven externally powered prosthetic. They define mechanomyography as the measurement of the mechanical activity produced by contracting muscles. The researchers created a silicone liner for the socket in which they implanted accelerometer sensors. Then, they recruited two trans-radial amputees and built custom prosthetics using the Otto Bock hand that is controlled by surface electrodes. The researchers, next, recorded both the

electromyographic and mechanomyographic signals when the amputee was asked to contract his muscle in order to open and close the hand. The amputees had many triumphs controlling the prosthetic hand with success rates above seventy percent. Success was achieved when the amputee was able to open and close the hand. The only issue the amputees were having was that the new prosthetic hand was difficult to control at first but with practice the amputees were able to control it reasonably well [24].

Another set of researchers at Illinois Institute of Technology, in collaboration with the Rehabilitation Institute of Chicago, created implanted myoelectric sensors for upper-extremity prosthesis control. Contrary to popular belief that there were only three or four possible independent surface EMG sites for control, these researchers thought that there could be many more independent EMG sites with the use of implantable electrodes. The objective was to implant myoelectric sensors to control and drive an eight degree of freedom arm. The myoelectric sensors were implanted subcutaneously and the output signal was fed to a telemetry reader outside the skin. The implants were able to send signals with the use of a small coil that also powered the electrodes and was able to send and receive telemetry signals. The output signal of the reader was then read by the controller, and using fuzzy logic, the controller determined what the patient intended to do. The controller was first tested using surface electrodes to see if the controller was able to

decode the signal. Since the subject was able to control the prosthetic and the controller was able to decode the signal, the researchers then tried using their implanted myoelectric sensors. This test was very successful because the subject was able to control the prosthetic and the controller was able to decode the signal. The researchers next explored the intra-muscular signal independence. Another single patient experiment was run using seven muscles in the forearm. The researchers were trying to determine if the EMG signals were capable of controlling a two degree of freedom wrist and a three degree of freedom hand simultaneously. Since the amputee was able to control the hand and wrist, the researchers were able to prove their hypothesis that multiple independent EMG sites can be found with implantable electrodes [25].

The problem that researchers had with EMG inside the socket is keeping the electrodes sterile. The inside of the socket gets very warm and very little air gets in, so sweat builds up. Wet and dark places are a great place for bacteria to grow and thus can cause problems with the electrodes. Another problem was the electrode wires rubbing against the socket and creating noise.

Uses of Force Sensing Resistors in lower limb prosthetics

Researchers at the University of Waterloo, in collaboration with the Rehabilitation Center in Ottawa, have developed a knee ankle foot orthosis. They placed a series of force sensing resistors on the bottom of

the footplate, in order to establish when the subject is weight bearing on the orthosis. The output signal from the force sensing resistors is sent to the controller on a computer. The controller then compares the signals to a pre-determined threshold to see if the subject is weight bearing. If the threshold is overcome, the controller deactivates the solenoid thus bracing the subject's leg [26].

Researchers at Case Western University have developed a hybrid orthosis for standing, walking and stair climbing for patients with spinal cord injuries. They placed potentiometers at the knee and ankle joints to measure angles and force-sensing resistors were placed in the insoles to measure foot-to-floor contact. The force sensing resistors were used to determine what part of the gait cycle the subject was in. They were also used to make sure that the knee was locked against flexion allowing the leg to swing freely. When the contralateral force sensing resistor signals are high, it indicates that the contralateral limb is in stance phase and the orthosis needs to support the user. When the ipsilateral heel force sensing resistor signal is low, it indicates the transition from double- to single-limb support [19].

METHODS

The hypothesis of this thesis was to give the subject more control over the prosthetic ankle in two phases. The first phase was to allow the amputee to drop the foot to the floor when he was sitting. The second phase was to make the prosthetic ankle feel more comfortable when the amputee was walking at any speed. A force-sensing resistor was used because researchers in the Human Machine Integration Laboratory have used them before and had great success.

Background on Force Sensing Resistors

Franklin Eventoff invented the force-sensing resistor, or FSR, in 1977. FSRs are variable resistors which change their resistivity in a predictable manner when a force, pressure or weight is applied [27]. They are made of the following layers: a layer of electrically insulating plastic, an active area, a plastic spacer and a flexible substrate coated with a thick polymer conductive film [27]. This can be seen in the figure below.

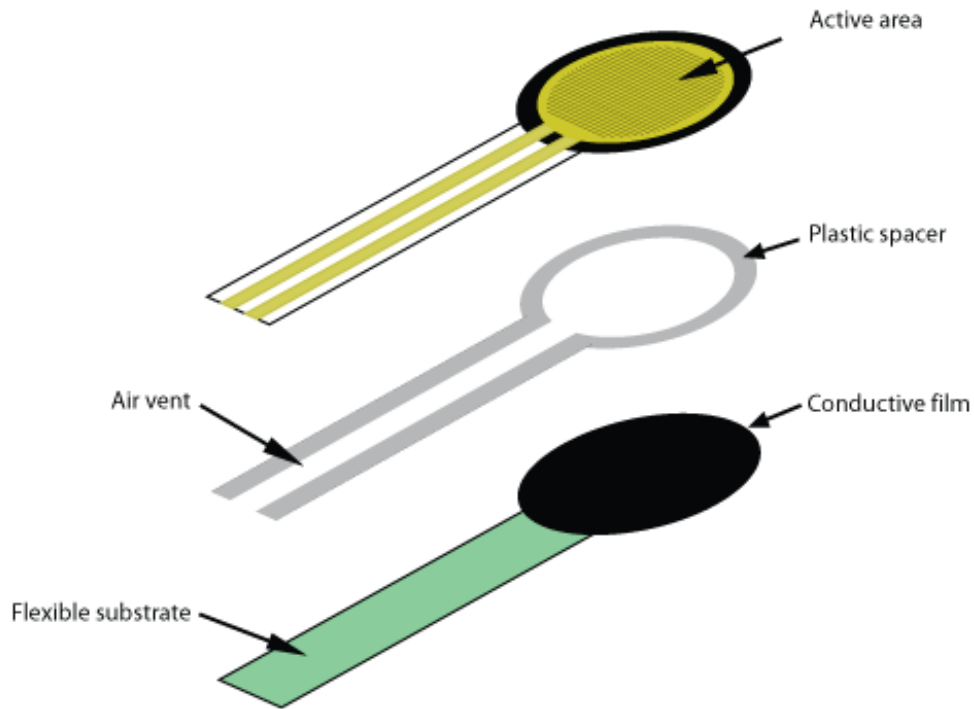


Figure 3: Force Sensing Resistor. The diagram above shows how a force sensing resistor is put together [27].

The active area is connected to the leads on the tail to be charged with an electrical voltage. The conductive film consisted of both electrically conducting and non-conducting particles suspended in matrix. When a force is applied to the sensor, the plastic spacer allowed all the air to leave through a vent and caused the particles to touch the conducting electrodes. This, in turn, changed the resistance of the film with the force being inversely proportional to resistance. For example, with no force on the sensor the resistance value could be close to 1 mega ohm. This led to one disadvantage; the force had to be able to lower the resistance to around 100 kilo-ohm that is the sensor's turn-on threshold [27].

The output signal was a logarithmic monotonic function that allowed for a wider range response with more accuracy. When enough force was

applied, this function changed slope quickly due to sensor saturation.

After this point, the output will not be significantly affected by an increase in applied pressure [28].

Background on Voltage Divider Circuit

One of the simplest circuits to build using a force sensing resistor was a voltage divider. This circuit allowed for a simple force to voltage conversion. To build this circuit, a power supply, a breadboard, a force sensing resistor, a fixed value resistor and some wire were needed. The power supply was set to +5 volts and a wire was used to connect the voltage to the breadboard. Then, one of the leads from the force sensing resistor was connected to the voltage side and the other to a location on the breadboard. The fixed value resistor was, then, placed with one end next to the second force sensing resistor wire and the other end to ground. In this way the FSR was able to measure the voltage drop across a resistor. The resistance value of the second resistor determined the output range of the sensor. The output voltage was collected at the node where the two resistors met. A figure depicting the circuit is shown in Figure 4.

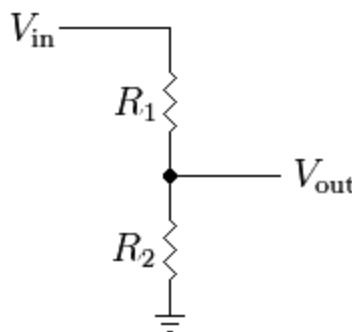


Figure 4: Voltage Divider Diagram. The figure above depicts the circuit schematic for a voltage divider circuit. R_1 varies and R_2 is fixed.

To determine the output voltage Ohm's law was used and the following formula was created:

$$V_{out} = \frac{R_1 V_{in}}{(R_2 + R_1)} \quad (2)$$

where R_1 was the force sensing resistor and R_2 was the fixed valued resistor. The value of the fixed value resistor is determined to maximize the force sensitivity range and to limit the current. For my case, this resistor was 2.2 kilo-ohms.

The output voltage of a voltage divider is not fixed but varies according to the load. To obtain a reasonably stable output voltage, the output current should be a small fraction of the input current. For a finite load, R_1 , an output voltage was reduced by voltage division by the factor $R_1 / (R_1 + R_2)$, where R_2 was the fixed value resistance. Likewise, as the term short-circuit implies, the output current delivered to a load, R_1 , was reduced by current division by the factor $R_2 / (R_1 + R_2)$. The overall gain was reduced below the gain estimated using an ideal load by the same current division factor.

Voltage dividers are often used to produce stable reference voltages. The term reference voltage implied that little or no current was drawn from the divider output node by an attached load. Thus, use of the divider as a reference, required a load device with a high input impedance to avoid loading the divider disturbing its output voltage.

Background on calibration

Due to the lack of accuracy of force sensing resistors, calibration of the sensors was essential. This lack of accuracy was due to hysteresis, defined as sensitivity to shear force and alterations in response properties with prolonged use. There has been some research into the techniques that dramatically improve both the reliability and accuracy of measuring force with FSRs. Since the resistance of the conductive polymer sensor dropped in an exponential fashion as the applied force was increased, the output voltage was a nonlinear function of applied force [29].

Researchers at EAFIT University in Colombia, ran two sets of tests. The first was a static test in which a device placed calibration weights on the force sensing resistors and the output voltage was obtained using a data acquisition device [28]. When a plot of force vs. time was created, the researchers noticed a large amount of sensor creep which was the increase in output voltage with constant weight being applied [28]. To compensate for this, they programmed an if/else algorithm when a threshold was reached, the derivative was taken [28]. This allowed the researchers to determine whether or not the increase in the output signal was from an increase of the applied load or creep [28]. Then, a dynamic test was run. The researchers used their same test device but this time the calibration weights were loaded and unloaded [28]. This created a new

unwanted behavior called hysteresis which is defined as the dependence of a system not only on its current environment but also on its past environment. The researchers saw one pattern when the weight was being loaded on the sensor and a second pattern when the weight was being unloaded [28].

Researchers at Simon Frazier University had also seen this hysteresis and found a solution. To correct the hysteresis, a fourth order polynomial function was obtained by multiple regression as well as a moving integral algorithm [29].

Since Interlinked Electronics Incorporated provided calibration curves, as seen below, and the researchers in the Human Machine Integration Laboratory were able to replicate it, a test device was not built. Instead, the force sensing resistor was taped to a table and a voltage divider circuit was built. A power supply provided the +5 volts and a 2.2 kilo-ohm secondary resistor was used. Then, using a small metal cylinder to spread the force over the entire area of the sensor, weights of ranges from one to ten pounds were loaded. The calibration curves for this set of experiments are shown in Figure 6. The theoretical curve was created using equation (2) with R_1 at 2.2 kilo-ohms, V_{in} of +5 Volts and various R_2 values.

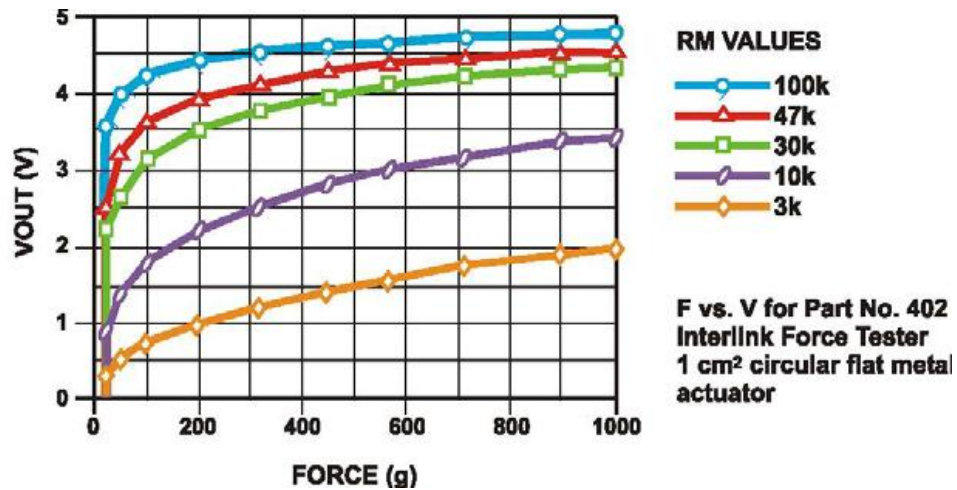


Figure 5: Calibration Curves. The chart above shows the calibration curve for a 1cm² circular force sensing resistor. The purple curve for 10 kilo-ohm R_m value was the target [27].

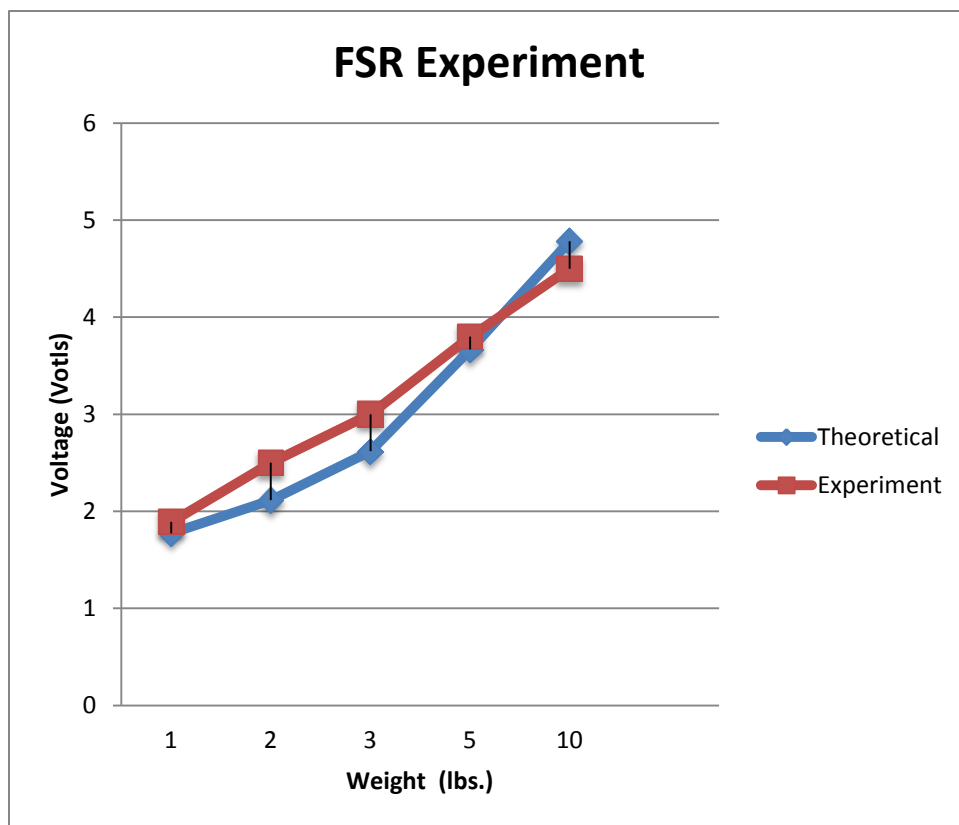


Figure 6: Experimental Calibration Curves. The chart above shows the calibration curve for a 1cm² circular force sensing resistor using a voltage divider circuit. Notice the similarity between the theoretical and the experimental values.

Background on Wheatstone bridge

If two voltage divider circuits are connected together, a special circuit called a Wheatstone bridge is created. Even though Hunter Christie invented the circuit, it was named after Sir Charles Wheatstone who was a very talented and versatile scientist [30]. Below is a schematic of the Wheatstone bridge circuit.

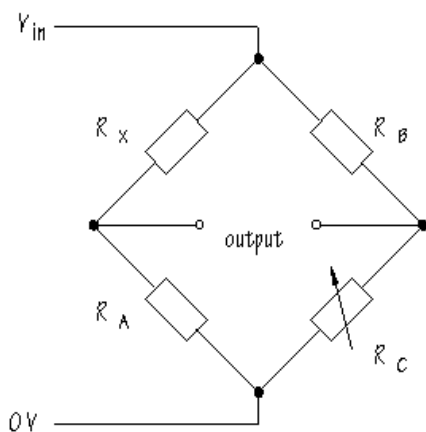


Figure 7: Wheatstone Bridge Schematic. The diagram above shows a typical Wheatstone bridge with R_C being a force sensing resistor [30].

If the value of R_X is unknown, the value of the R_C is adjusted until the values V_{out} of the two voltage divider circuits are equal. When the output voltages are equal, the bridge is said to be balanced. This can be checked by placing either a voltmeter or ammeter across the output terminals. When the circuit was balanced the ratio of R_X / R_A is equal to R_B / R_C . To solve for R_X the following equation can be used.

$$R_X = \frac{R_A \cdot R_B}{R_C} \quad (3)$$

Thus, if the values of R_A , R_B , and R_C are known, it is easy to calculate R_X . In Wheatstone bridge instruments, R_A and R_B are fixed and R_C is adjusted on a sliding scale in such a way that the value of R_X can be read off directly. A Wheatstone bridge circuit can be used to determine the weight of an object. The change in R_C required to balance the bridge represents the change in resistance. When this value is multiplied by the modulus of elasticity, the result yields the change in stress. The change in stress multiplied by the cross-sectional area produces the change in load, which is used to determine the weight [31].

A Wheatstone bridge circuit could be used by to remove the weight of the user from the signal for walking and standing during the experiments conducted in this research. The bridge also eliminates sensor saturation.

Calculation of socket forces when sitting and standing

When an amputee was sitting in a chair, the keel was usually at a forty five degree angle. Since this was not very comfortable, some amputees preferred to remove their prosthetic when they are sitting. Thus, the first goal of my research project was to give the amputee control over the prosthetic foot when sitting. This was done by placing a force sensing resistor inside the socket wherever the amputee thought was best. Then, the amputee was asked to put the prosthetic on and stay seated. This was so that the researchers could get a baseline for all sets of experiments.

Next, the amputee was asked to hold the prosthetic in the air and contract his calf muscles. This test was conducted in twenty second intervals because the prosthetic was heavy. Subsequently, the amputee was told to stay seated, keep the prosthetic on the ground and contract his calf muscles. The data gathered from this set of experiments is summarized in data table 1 in the result section below. This data was then used to modify the resistance of the fixed value resistor of a voltage divider circuit. This allowed the amputee to move the prosthetic foot to the ground when the threshold voltage was obtained.

The next goal of the project was to have the amputee standing with equal weight on both legs. The idea was to find a baseline for the walking experiments as well as to find a good location for the force sensing resistor. The forces inside the socket in the tangential direction were the shear and stress forces where the socket comes into contact with the residual limb. These locations were ideal spots to put the force sensing resistors since they would not allow the sensors to move when pressed. Researchers at the King's College School of Medicine and Dentistry have shown that there were three posterior locations where the socket and amputee were connected. The researchers called these three locations the popliteal depression, the medial gastrocnemius and the lateral gastrocnemius. The popliteal depression was located behind the knee and gave the most stress force which would make it the most ideal spot to place a force sensing resistor. The gastrocnemius was located in the back

part of the lower leg [32]. These three locations were the basis of the experiments that were conducted.

Researchers at Vanderbilt, led by Dr. Michael Goldfarb created a volitional controller for a prosthetic knee using surface electromyography. Goldfarb and some of his students divided volitional control into nonweight-bearing and weight-bearing [33]. Nonweight-bearing happened when the amputee was sitting, flexing his knee and when putting on or taking off shoes as well as tying shoes. Weight-bearing occurred when the amputee was standing or moving. They focused on nonweight-bearing volitional control. The researchers created an impedance controller that utilized the surface EMG signals from the quadriceps and hamstring muscles. The signal from the electrodes was then differentially amplified and filtered using a band pass filter. The filtered signals were next read into Matlab through a data acquisition card. The signal was high pass filtered, rectified and finally low pass filtered. The resultant signal was put into a database of a classifier as training data. The classifier was then given a threshold value for each subject. For checking, a signal was given to the classifier and it determined whether the amputee was attempting to flex or extend their knee. The controller was very effective in predicting whether the amputee was trying to flex or extend their knee [33].

Calculation of socket forces when walking

The forces inside the socket changed substantially when the amputee was walking because the sum of the tangential forces was no longer zero. The forces in the normal direction were half of the body weight of the subject and the weight of the prosthetic which was opposed by the normal force from the socket. In the tangential direction, the forces were friction, pressure of the socket on the amputee and the pressure from the amputee on the socket. In order for the amputee to be able to walk, he had to overcome the weight of the prosthetic in the normal direction and friction in the tangential direction.

Dr. Joan Sanders and Dr. Colin Daly at the University of Washington in Seattle have been working on the modeling of the normal and shear stresses on a residual limb in a prosthetic socket during ambulation [34]. They were using a technique known as finite element analysis and comparing the results with experimental measures. There were three input requirements in doing finite element analysis: geometries, boundary conditions and material properties. Dr. Sanders and Dr. Daly used magnetic resonance imaging to determine the geometries of the residual limb. For the boundary conditions, force and moments were measured using 20 strain gauges inside the pylon, which was the piece that connected the socket to the prosthetic. The material properties of skin and muscle were taken from literature. However, to simplify the problem, Dr. Sanders and Dr. Daly made the assumption that the skin and muscle were linear and uniform materials. The material properties were also

found for the socket using uniaxial compression, which was the compressive strength of a material when it failed completely. Once all the input data was collected, the finite element analysis was run in the six residual limb force and moment directions. For the human subject experiments, Dr. Daly and Dr. Sanders built four transducers. They used them to find the normal and shear stresses as well as the forces and moments of the residual limb during walking. The researchers had to add two waveforms in order to compare the forces and moments of the finite element analysis to the experimental data because they only had four transducers. Dr. Daly and Dr. Sanders, therefore, averaged the waveforms collected in the same alignment during the experiments. The experimental and analytical data seemed to match pretty well except for the peak magnitudes during plantarflexion alignment, during stance phase and toe off. These are due to the assumptions made by Dr. Sanders and Dr. Daly. Therefore, finite element analysis seems to be a good tool in modeling the forces and moments during walking [34].

Researchers at Tsinghua University used a musculoskeletal modeling technique to model and simulate the muscle forces of trans-tibial amputees to study the effect of prosthetic alignment [35]. They claimed that the musculoskeletal models were the most widely used in biomechanics and were the best at predicting muscle forces, ligaments and articular loading, which were the forces of the joints. The researchers started developing a model with two degrees of freedom in the sagittal

plane which represents the thigh, the prosthesis, the residual limb and the pelvis. The 2 degrees of freedom are the angle of the residual limb and the thigh relative to the horizontal. In the model, the prosthetic was supposed to be connected to the patellar tendon and the prosthetic foot was modeled as a particle mass for simplification. The model thus represented seven different functional muscle groups in the amputated leg. To allow for the model to be dynamic, the Lagrange laws were used. Lagrangian mechanics take Newton's second and third law and allow them to be used without being constrained to the Cartesian coordinate system. To determine the validity of the model, a male trans-tibial amputee was asked to walk. The kinematic data of the amputee was gathered using a motion capture system and force plates in the floor were used to gather reaction forces. The musculoskeletal model seems to match the experimental data fairly well. However, the model is limited because it only takes into account the sagittal plane. The researchers plan on making improvements to this model to better understand the biomechanics of trans-tibial amputees [35].

Tibia Based Controller

The current controller that SPARKy uses is called the tibia controller. Dr. Matthew Holgate developed the tibia based controller as part of his dissertation. He was trying to find a measurable variable to determine a mathematical relationship between the tibia angle and ankle

angle. He decided to use the tibia global angular position over all alternatives because it creates a simple shape when plotted and the relative ease of gathering data. The plot of gait percent versus tibia angle, at different stride lengths, depicts almost identical curves with the only difference being a stride length factor (see figure 8) [2].

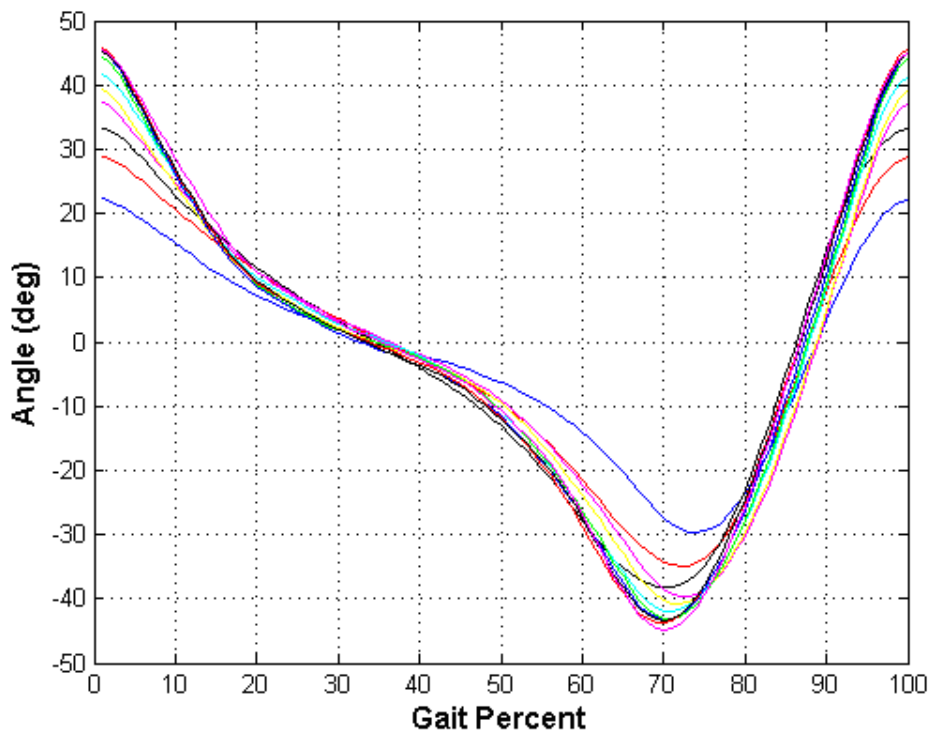


Figure 8: Gait percent vs. Angle [2].

To gather data for the tibia angle, a sensor can be attached to the prosthetic so there is no need for extra sensors. Both of these points make the global tibia angle the best choice for prosthetic controller input. The next step that Dr. Holgate took in creating the tibia controller was to decide whether it needed to be continuous or discrete. After some trials with if-then logic and seeing that the controller would get fooled

sometimes, the decision was made to go with a continuous controller. This meant that a one-to-one invertible relationship between the tibia angle and desired ankle angle needed to be established. Looking back at figure 8, the tibia angle versus gait percent plot is not invertible as a whole, so a new relationship needed to be formed. Dr. Holgate figured out that if he plotted tibia angular velocity versus tibia angle, a pattern emerged. The curves shown in figure 9 get larger with increased stride length. To make the pattern more evident the coordinate system was changed from Cartesian to polar. When Dr. Holgate did this, he came to the conclusion that there has to be a relationship between the polar angle and gait percent for each of the stride lengths. This relationship can be seen in figure 10. Something of significance is the fact that for each of the stride length curves, the plot of the polar angle versus gait percent appears very similar. However, the plot of the polar angle versus gait percent is invertible while the stride length curves plot is not [2].

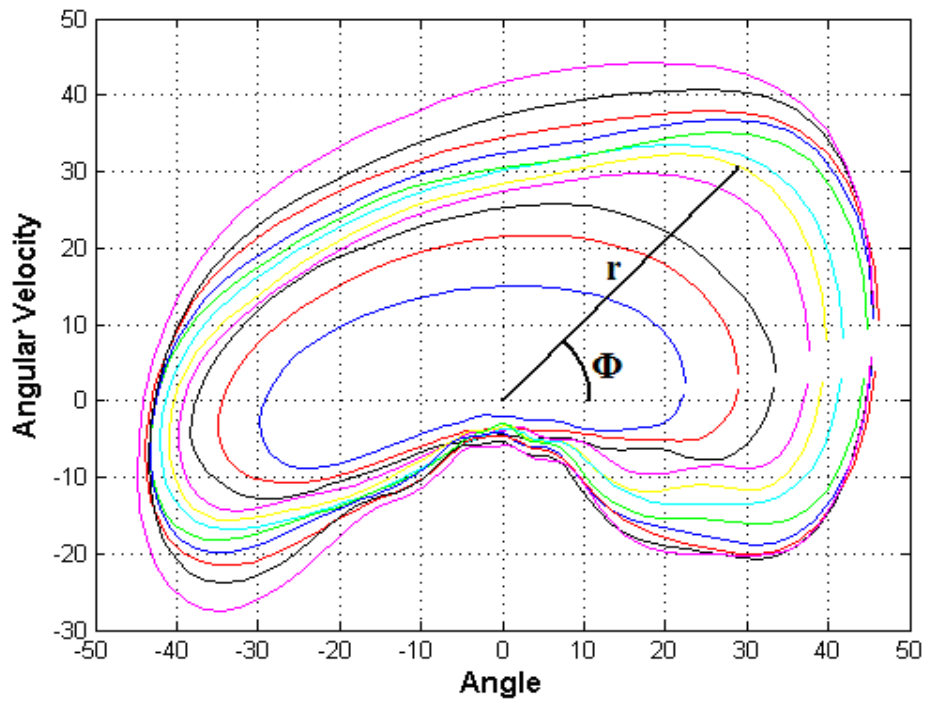


Figure 9: “Bean Graph.” This graph depicts tibia angle vs. angular velocity and received its name because the graph looks like a bean [2].

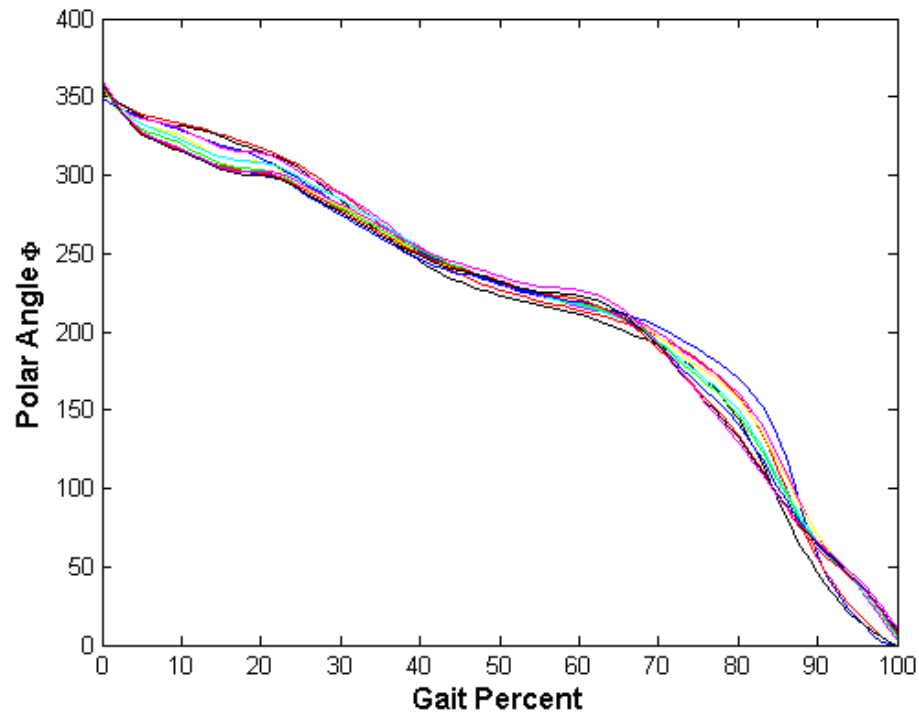


Figure 10: Gait percent vs. polar angle [2].

Then, Dr. Holgate found a fitted function in which if the tibia global angle and angular velocity are measured and the polar angle is calculated, the gait percent could be found. The fitted equation seems to not directly rely on stride length, but the stride length must be known to calculate the ankle angle. Thus, Dr. Holgate went back and took another look at figure 9 and saw that the longer the stride length, the longer the polar radius. So he tried to relate stride length to polar radius and came to the conclusion that the polar angle had to be part of the equation. Dr. Holgate then plotted the polar radius versus stride length versus gait percent and the result is figure 11. The figure makes it seem as if a relationship could be found but the inverse relationship of stride length as a function of gait percent and polar radius needs to be found. There is a problem though. Numerous times on the plot, the figure seems almost vertical and has multiple values for a single point. This problem is easier to see in figure 9, where all the stride length lines get all bunched up. A simple fix to this problem is to use a simple first order filter on the polar radius which causes the curves to separate (see figure 12). Remarkably, the plot seems to flatten out and for every combination of radius and gait percent, there is one value of stride length. The filter causes a small problem in that it causes some attenuation and phase lag. To minimize the lag, the measured and filtered polar radius needs to be compared to the expected filtered radius plot. The ankle angle can be found in motion capture data and used to find the stride length and gait percent. The resulting plot can

then be fit with a function or a look up table. Depending on what robot is being controlled, the controller will generate a desired position [2].

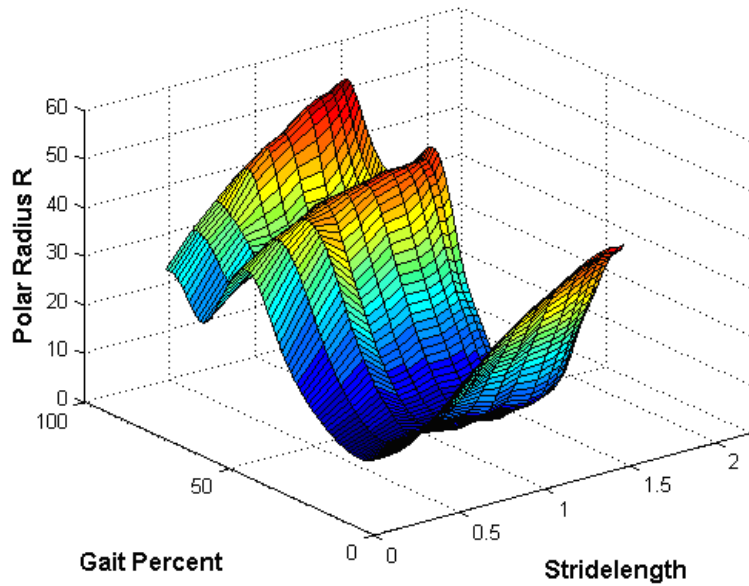


Figure 11: Gait percent vs. stride length vs. polar radius [2].

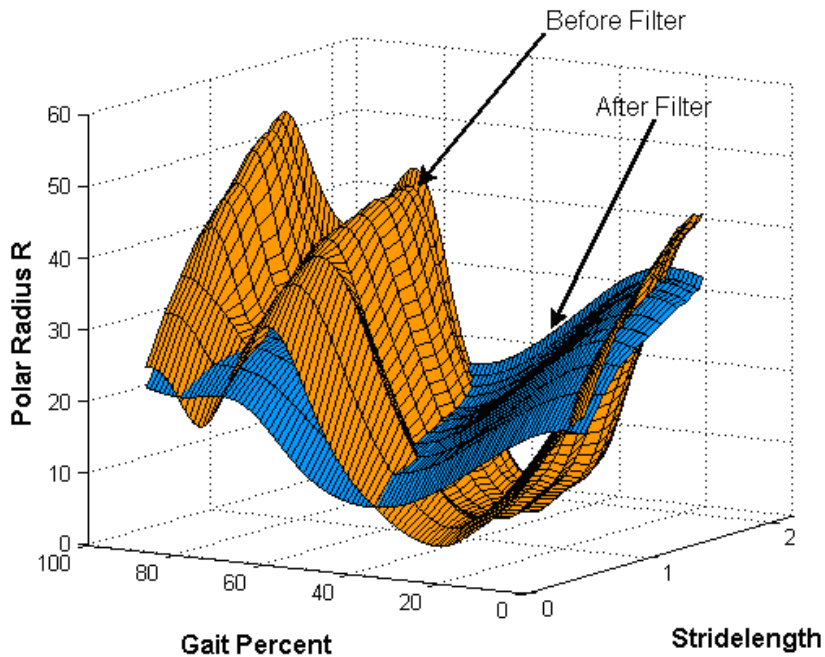


Figure 12: Gait percent vs. stride length vs. polar radius. The orange curve is the same as Figure 11 and the blue curve shows the filtered orange curve [2].

The biggest problem when Dr. Holgate was trying to get the prosthetic to execute the tibia controller was how to accurately measure the global tibial angle and angular velocity. During the experiments, he was able to use an angular rate sensor, which outputs a voltage proportional to the rate at which it turns. He then integrated the signal to find the angle. There were three major issues with the experimental data: the sampling is discrete, the output sensor was noisy and the integration is numerical. Therefore, the measured angle had lots of error that in turn causes the reference ankle angle to be inexact. Two ways to correct this would be strap down integration and a Kalman filter. However, they would have involved the use of extra sensors and would have increased system complexities. When Dr. Holgate was examining the data for multiple subjects, he noted that all the tibia angles were distinctive. He also noticed that the curves were different shape and were shifted. However, Dr. Holgate had an explanation for each of these. The shapes were different because everyone has their own unique gait pattern. The shifts were most likely caused by the discrete integration. Dr. Holgate figured out a way around this by using a transfer function which pseudo integrates instead of simple integration. [2].

$$\frac{\tau^2 s}{(\tau^2 s + 1)} \quad (4)$$

Dr. Holgate created a bode plot and noticed for high frequencies that the transfer function approximates the integration magnitude and phase really well. [2].

The initial testing of the controller was done on able body subjects and then was transferred to SPARKy. The first amputee tests yielded great results when they were able to walk at different stride length and then at different speeds. There are multiple advantages of the tibia controller. The first advantage is the controller's ability to update the ankle position as fast as the sampling time of the sensors. In addition, the controller is never committed to one state of operation. Third, the tibia controller can be adapted to work with any kind of users. The last and most unique feature is that the controller gives the users the ability to walk backwards, up and down slopes and up and down stairs [2]

RESULTS

FSR system for sitting

The results below come from one 65 kg able-bodied subject and one 70 kg trans-tibial amputee. Below is a data table that shows the minimum, maximum, mean and standard deviation for three tests: the patient sitting with prosthetic in air, the patient sitting with prosthetic on the ground, and the patient standing with both feet on the ground. There is data for contractions and non-contractions. The data is the measured V_{out} from the voltage divider circuit

Data Name		Min	Max	Mean	Standard deviation
Sitting air					
	No contraction	0.4246	1.364	0.827305	0.397144
	Contraction	0.5115	0.5427	0.528205	0.013038
Sitting ground					
	No contraction	0.4536	1.374	0.841732	0.395187
	Contraction	0.4885	0.5169	0.505741	0.010057
Standing ground					
	Contraction	1.21	1.603	1.415455	0.166098

Table 1: Force sensing resistor voltages for the amputee

The three tests were also conducted with an able-bodied subject and the results are shown in the data table below.

Data name		Min	Max	Mean	Standard deviation
Sitting air					
	No contraction	0.0364	0.0777	0.06029	0.002055
	Contraction	0.0464	0.0751	0.06027	0.002019
Sitting ground					
	No contraction	0.0477	0.0708	0.06068	0.001998
	Contraction	0.0469	0.0743	0.0601	0.002052
Standing ground					
	Contraction	0.9101	2.1041	1.60909	0.324996

Table 2: Force sensing resistor for the able-bodied subject voltages

The data for both the able-bodied subject and the amputee seem to agree fairly well except for the magnitude. This may be due to the fact that the amputee is in phenomenal shape and has been using the residual limb to power their passive prosthetic for the past ten years. In Figure 13, the amputee is sitting with his feet on the ground. He is contracting his muscle on and off.

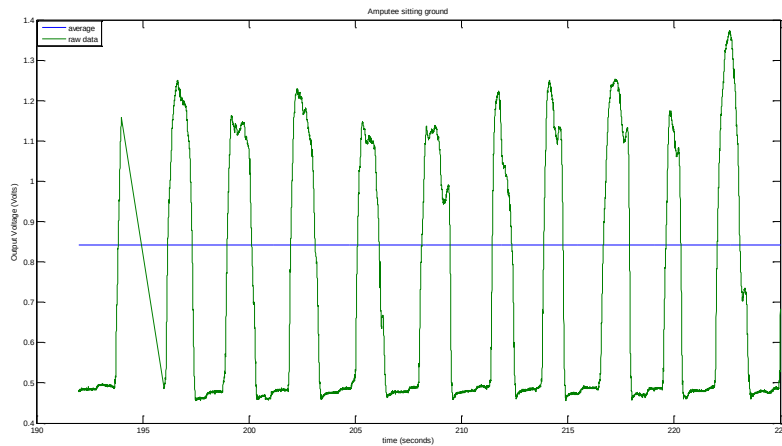


Figure 13: Amputee sitting. Raw data and average are shown.

The next step was to glue a force sensing resistor inside the socket which would allow the subject to drop the foot when sitting. The wires from the sensor were then connected to the motor control board on SPARKy. Subsequently, a virtual voltage divider circuit was created in Matlab that used the resistance values of the force sensing resistors. The block then computed the output voltage and compared the output voltage to a threshold voltage. The threshold voltage is equal to the mean value shown in Table 1 and is depicted in figure 13 as the blue line. The amputee was asked to contract their muscle to control the ankle motion and drop the prosthetic foot to the ground. After a few minor adjustments, the amputee was able to drop the prosthetic foot every time that they contracted their gastrocnemius muscle. The amputee was extremely happy and exclaimed that the prosthetic finally felt like his real foot. Below are a series of photographs of the amputee dropping the foot to the ground.



Figure 14: Amputee moving foot to ground.

The picture to the left in Figure 14 is the prosthetic foot at rest. Notice how the foot is at a forty-five degree angle which is very uncomfortable for the subject. The picture to the right in Figure 14 is the amputee moving the prosthetic to the floor.

FSR system for standing

To obtain a baseline for the walking test, the amputee and the able-bodied subject were asked to stand with both feet on the ground and weight evenly distributed. The minimum, maximum, mean and standard deviation for this data can be seen in Tables 1 and 2. The data collected was used to determine how the signal changed when the able bodied subject and the amputee were putting weight on the sensor.

FSR system for walking

Subsequently, the able-bodied subject was asked to walk on the treadmill while contracting his gastrocnemius muscle. There was one set of tests done walking forward. The subject walked at 2.5 and 3.2 mph. Each experiment was recorded for approximately three minutes at each speed. The graphs will be coupled, the one on the left is the raw data split into individual steps and the on the right is the average of all the steps together. The data depicted is from the gyroscope that was on the front of the tibia and a force sensing resistor on the medial gastrocnemius.

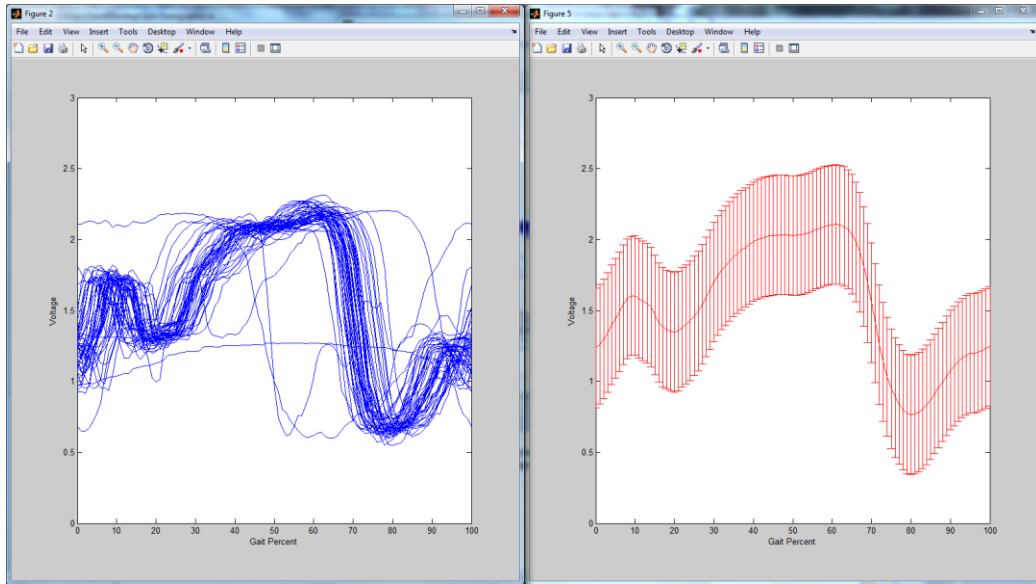


Figure 15: 2.5 FSR Walking of able-bodied subject. The graph on the left shows steps at 2.5 mph and the one on the right is an average of the steps.

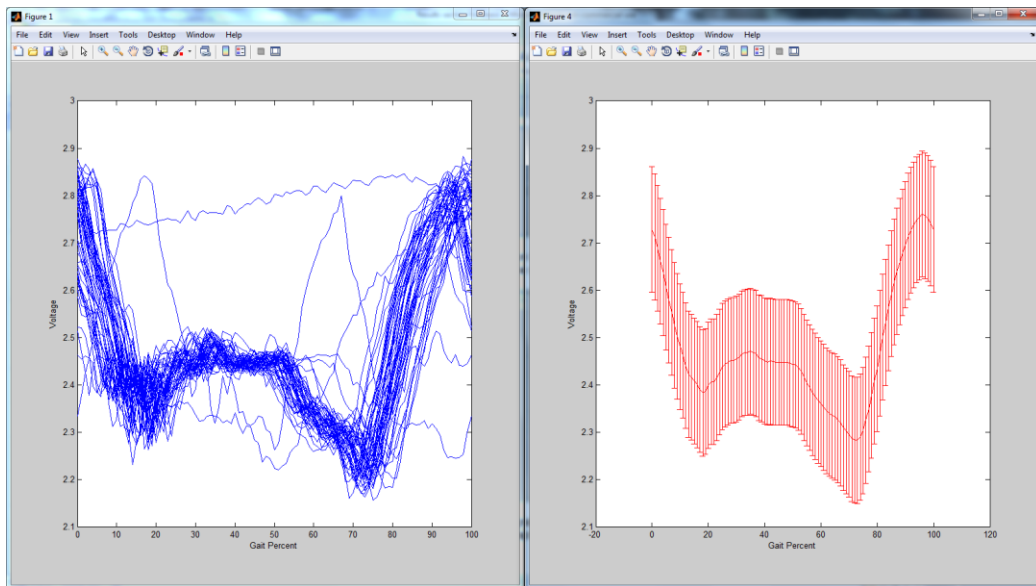


Figure 16: Gyroscope Walking of able-bodied subject.. The graph on the left shows steps at 2.5 mph and the one on the right is an average of the steps.

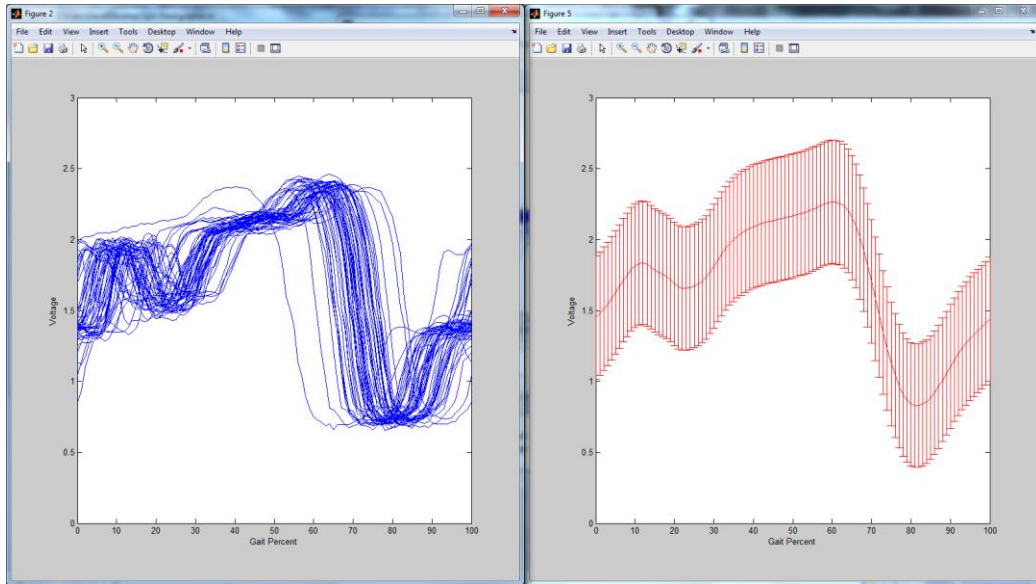


Figure 17: FSR Walking of able-bodied subject.. The graph on the left shows steps at 3.2 mph and the one on the right is an average of the steps.

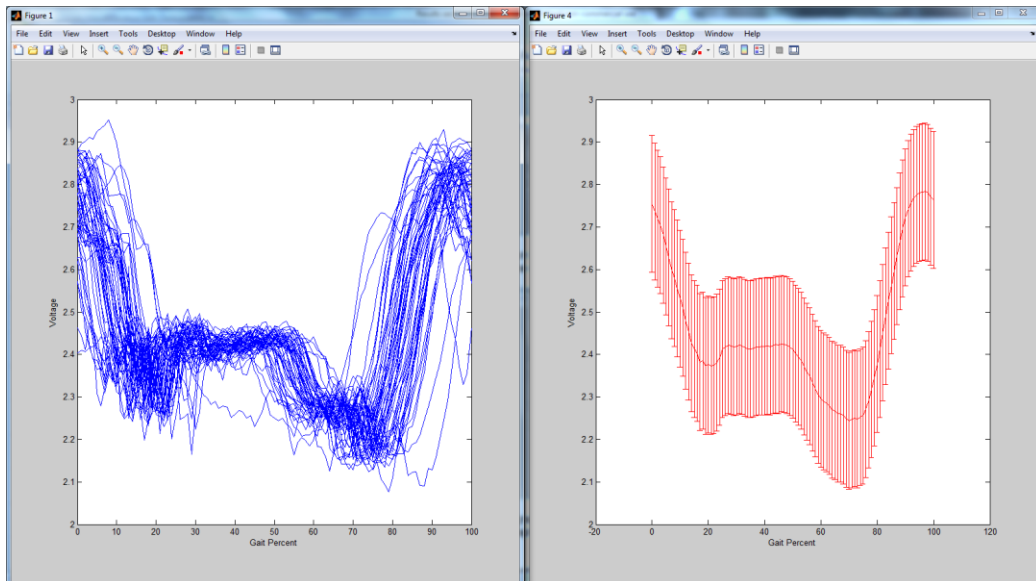


Figure 18: Gyroscope Walking of able-bodied subject.. The graph on the left shows steps at 3.2mph and the one on the right is an average of the steps.

The key thing to note is the maximum part of the gait cycle occurs earlier in the 3.2 mph graph as compared to the 2.5 mph graph. This

makes sense since the subject needs “push off” to occur sooner to get the foot through the swing phase of the gait cycle. Another key thing to note is how the second peak in the gyroscope data becomes larger as the speed increases. This is due to the fact that the leg is swinging faster as the speed increases.

These experiments were also conducted with the amputee shown in Figures 19 and 20. There is no gyroscope data because we have been unable to get a reliable connection from the control board to the data acquisition device.

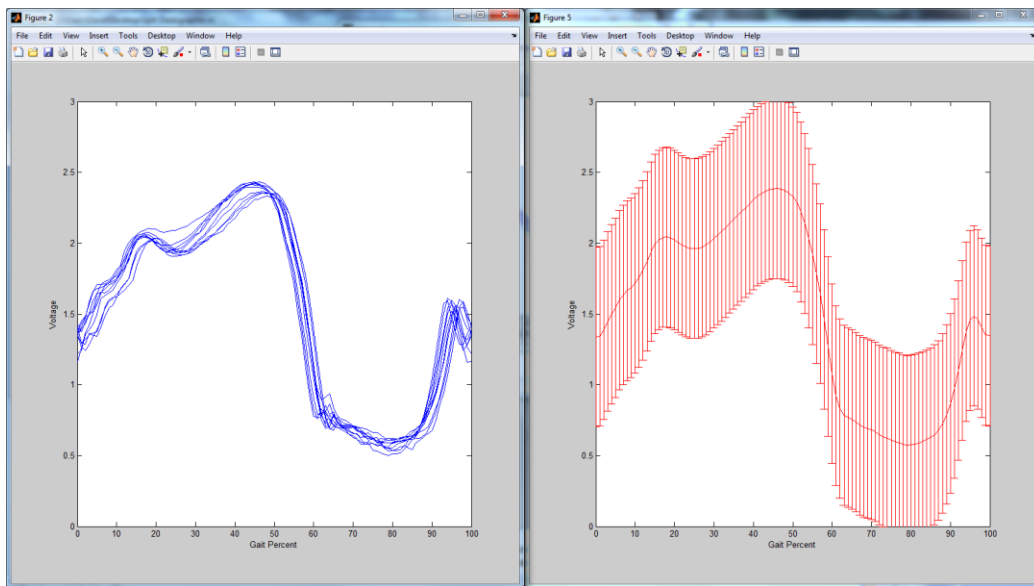


Figure 19: FSR Walking of amputee. The graph on the left shows steps at 2.5 mph and the one on the right is an average of the steps.

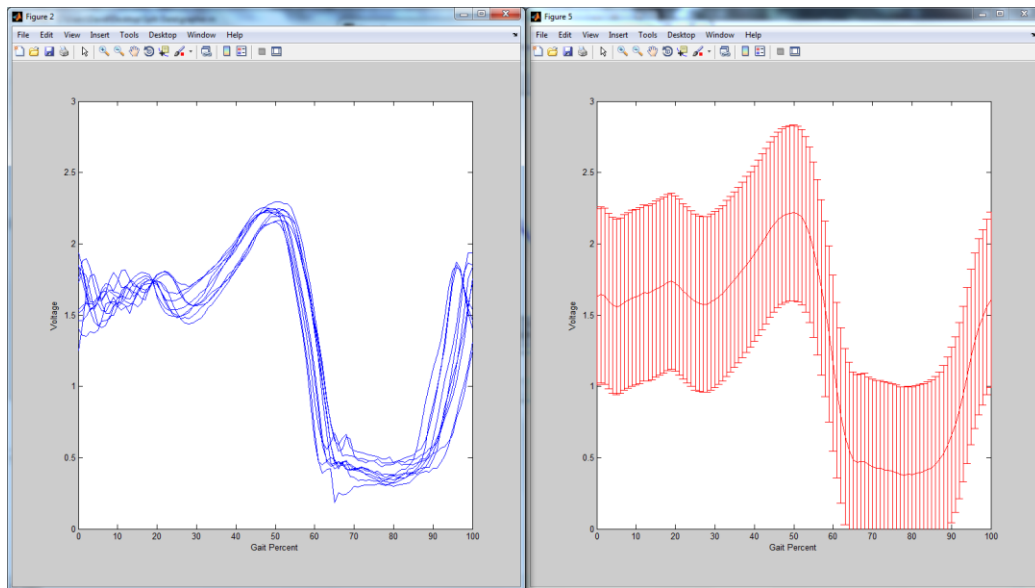


Figure 20: FSR Walking of amputee. The graph on the left shows steps at 3.2 mph and the one on the right is an average of the steps.

The data from the able-bodied subject and the amputee seem very similar with the exception of where push off occurs. This is due to the fact that the amputee is taller which leads to a different gait pattern between the subjects.

CONCLUSIONS

The overall goal of these experiments was to give the amputee more control over the prosthetic ankle. The first phase of this goal was to get the foot to drop to the floor when the patient was sitting. As seen in the results section and from feedback from the patient, this phase was very successful. The amputee was able to proportionally control the ankle movement with the force sensing resistor. The second phase of this goal was to research able-bodied data and change the control algorithm to make the prosthetic feel comfortable at every speed. This phase was not very successful. There are a multitude of reasons that can be given for this. For example, every person has a different gait. This is due to the fact that people have different heights, weights and have different body mass. In addition, even two people who are the same height can have two completely different gait patterns depending on how fast they walk. Thus, using an able-bodied subject data to change the control algorithm for an amputee is very difficult due to the fact that the able-bodied subject and the amputee have two very different gait cycles. Therefore, in order to create a controller that makes the prosthetic feel comfortable at any speed, individualized controllers have to be created using data collected on that subject.

FUTURE WORK

It was shown that the FSR sensor could easily be used to volitionally control the ankle movement when sitting. Future work will need to improve the algorithm to enhance the control of walking gait. Two areas will need to be researched: (1) improve the timing of push-off, and (2) increase push-off power when needed such as walking up an incline or stairs. The voltage divider circuit would help future researchers to be able to adjust the tibia based controller for when push off occurs at different speeds. Also, researchers will need to use the Wheatstone bridge circuit to remove the weight from the signal when the person is standing and walking.

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