A Novel Control Algorithm for Ankle-Foot Prosthesis

Arvind Parsan
Department of Mechanical and Materials Engineering
Florida International University
Miami, Florida 33174 USA
(954)-798-4883
apars002@fiu.edu

Sabri Tosunoglu
Department of Mechanical and Materials Engineering
Florida International University
Miami, Florida 33174 USA
(305)-348-1091
tosun@fiu.edu

ABSTRACT
Patients suffering from below knee amputation utilize ankle-foot prostheses to regain partial mobility. Research efforts have focused on improving device functionality to offer patients a higher standard of living. This study proposes a novel algorithm for device control, which utilizes input from two sensors to relay data into the control system. The system processes the data according to a set of rules, and outputs a respective foot angle value to mimic the normal motion of the ankle-foot complex in natural cadence. Matlab Simulink offers exceptional utility in developing a computational model for evaluating the efficacy of the rule system. Comparison of the true and theoretical foot angle values produced low levels of error underlining the overall effectiveness of the algorithm.

Keywords
Gait, Controller, Ankle-Foot Prosthesis, Micro-controller, Matlab, Simulink, Natural Cadence, Smart Prosthesis, Biomechanics, biomimetic, Ankle-Foot Emulator, Below-Knee Prosthesis

1. INTRODUCTION
Gait is the most basic of human actions, and has been the subject of considerable study for many decades. Many people learn to walk at a young age, and take for granted the 15 independent degrees of freedom which the body regulates to create the gait cycle. Gait studies can be divided into several segments including: temporal and stride measurements, kinematics, kinetics, electromyography, balance, posture, pathology, and age. Cadence is defined as the number of steps per unit of time and differs based upon someone's weight, height, age, gender, and pathology. Cadence can be further divided into three categories: slow, natural, and fast walking. Fast, natural, and slow walkers have an average cadence of 123.1, 105.3, and 86.8 steps per minute respectively. Our study focuses exclusively on natural cadence, and aims to replicate the foot angle of the smart prosthesis using input from sensors placed in strategic positions on the device [1].

Gait is cyclic in nature, and this fact has been utilized by researchers to simplify the gait cycle into smaller segments. The gait cycle can be divided into two distinct phases: the stance phase, and the swing phase [2]. The cycle begins with heel strike of one foot, proceeds to a foot-flat stance of the same foot, moves into heel strike of the opposite foot, continues with toe off of the original foot, proceeds with a forward swing of the original foot in air, and culminates with the heel strike of the original foot [3]. The stance phase consists of the first ~60% of the gait cycle beginning with the heel strike of the original foot and ending directly before toe off. The swing phase accounts for ~40% of the gait cycle and begins directly after toe off and continues until heel strike is once again initiated, beginning a new gait cycle [2].

The ankle-foot complex is of particular importance in human movement. During locomotion the complex plays a pivotal role in the support of body weight and the control of walking kinetics [4-7]. Three stages of the gait cycle are particularly significant in gait kinetics. At heel strike the heel absorbs the reaction forces that are created by contact between the heel and the ground. At this stage the total body weight of the individual is dually supported by the heel of one foot, and the toes of the opposite foot. As the opposite foot moves into its swing phase the ankle-foot complex solely supports the weight of the individual in the foot-flat stance. As opposite heel strike occurs the complex pushes off of the floor propelling the body forward into the subsequent swing phase. The ankle-foot complex supplies the body with the necessary kinetics required to drastically reduce the metabolic energy required for locomotion [7].

Currently there are 2 million Americans who suffer from an amputation and millions more worldwide. One in 190 Americans lives without a limb, and the number is expected to double by the year 2050 [8]. Every year 185,000 amputations occur in the United States alone [9]. Several factors contribute to the growing number of amputees such as diet, disease, and trauma. After surgery patients are presented with a limited number of options. They can use the traditional passive prosthesis that contains purely mechanical components, or they can opt for a more advanced micro-processor controlled prosthesis. Ankle-foot prostheses are designed specifically for patients suffering from below-knee amputations that require a device to replace the functions of the lower part of the leg [2]. Traditional prosthesis do not offer the patient any form of intelligent control or powered assistance for walking, and because of this amputated patients can expend up to 60% more metabolic energy in level walking than healthy subjects [10, 11].

Metabolic energy expenditure is a major concern with patients suffering from limb loss. Patients forced to exert an unusually high level of energy to complete basic tasks necessary for daily life are drastically discouraged from pursuing other activities, such as sports and hobbies, which would enrich their lives. Studies by the Kaufman group have confirmed that patients with microprocessor-controlled prosthesis have increased physical activity, and enhanced quality of life compared to those with traditional prosthesis. For this reason it is imperative that microprocessor-controlled prosthesis be a topic of continued research and development for the purpose of giving amputees a higher standard of life [12].

The objective of this study focuses exclusively on the control of an ankle-foot prosthesis. Ankle-foot prostheses have been under development for several decades and their progress is a process of continued improvement. Researchers have investigated ankle-foot prosthesis from a multitude of perspectives including: mechanical design and kinetics [13, 14], assistance in pathology [15], and overall gait control [16, 17]. Most ankle-foot prosthesis work by regulating the movement of a single linear actuator mounted where the tendon would be located in a healthy individual. During intended use the actuator moves up and down causing the foot to move through various angles.

Control of this device can be divided into four categories: biomechanical signals, electromyographic signals, peripheral nervous system signals, and central nervous system signals [4]. This investigation has focused on using biomechanical signals to control the prosthesis, because of the large amount of readily available data published in human gait studies. By using two sensors (heel contact, and knee angle) we propose a rule system for controlling the foot angle of the prosthesis. The goal is to mimic the motion of a healthy foot in natural cadence in order to produce enhanced device performance to ultimately minimize the metabolic energy required to walk. To evaluate the efficacy of the algorithm a simulation was developed in Matlab Simulink. Matlab Simulink was selected for computational modeling because of its ease of use, block programming language, and previous applications in the modeling of gait and robotic systems [18-21].

2. NOMENCLATURE

θ_h = Hip Angle

θ_{th} = Thigh Angle

θ_{tr} = Trunk Angle

θ_k = Knee Angle

θ_{lg} = Leg Angle

θ_a = Ankle Angle

θ_{ft} = Foot Angle

$\omega_k = \frac{d\theta_k}{dt}$ = Knee Angular Velocity

3. METHODS

3.1 Gait Modeling

The human leg can be modeled in three-dimensional space as having a total of seven degrees of freedom (DOF), where three DOF are in the hip, one is in the knee, and three are in the ankle [3]. Together both legs have fourteen DOF, but in actuality fifteen degrees of freedom are required to model the gait of the human body entirely. The additional DOF is the trunk angle which is depicted as θ_{tr} in Figure 1 [1]. Taking into consideration all 15 DOF can be a daunting task, and poses significant challenges in studying gait. For this study one of the two legs is healthy removing the need to model 7 DOF, and the trunk angle remains constant at 90 degrees in natural cadence which removes an additional DOF. The hip and the knee of the amputated leg are intact which removes an additional 4 DOF, leaving behind only the 3 DOF corresponding to the ankle-foot complex. In two dimensional motion studies of the ankle-foot prosthesis 2 DOF are unused reducing the study to examining 1 DOF. The sole DOF is represented in Figure 1 by the foot angle (θ_{ft}).

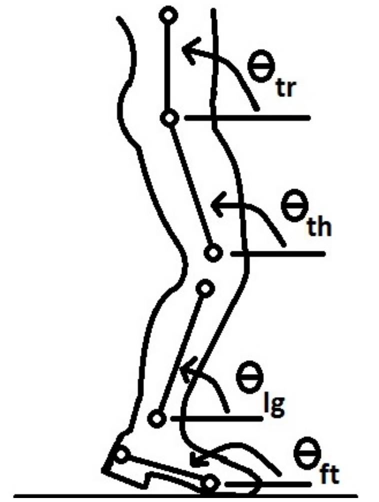

Figure 1. Outline of Major Angles in a Single Leg

Equations 1, 2, and 3 were acquired from [1], and represent three major angles that are reported extensively in literature. As one can see these angles are not directly measured in data collection, but are calculated from other angles.

As previously stated in natural cadence the trunk angle remains constant at 90 degrees [1]. Taking this into consideration equations 1-3 can be used to solve for the foot angle (θ_{ft}). The algorithm aims to match the foot angle, and its ability to do so is studied closely in the simulation. The derivation of equation 4 is included in the Appendix at the end.

Equation 1: $\theta_h = \theta_{th} - \theta_{tr}$

Equation 2: $\theta_k = \theta_{th} - \theta_{lg}$

Equation 3: $\theta_a = \theta_{ft} - \theta_{lg} - 90°$

Equation 4: $\theta_{ft} = \theta_a + \theta_h - \theta_k + 180°$

Equation 5: $\omega_k = \frac{d\theta_k}{dt}$ = Knee Angular Velocity

Table 3.32b of [1] details a collection of data for hip, knee, and ankle degrees recorded every 2% in stride for natural cadence. This data was input into equation 4 to compute the true foot angle value at various points of percent stride, which can be seen in Figure 8.

The values of true foot angle are input into the model as a standard for comparison against the theoretical foot angle values produced by the algorithm for the prosthesis.

3.2 Sensor Selection and Knee Analysis

3.2.1 Sensor Selection

Many ankle-foot prostheses use a multitude of sensors to feed information into the controller. These sensors include: accelerometers, rotary potentiometers, and force transducers [7]. In our algorithm the system receives input from only two sensors: a rotary potentiometer measuring knee angle, and a force transducer mounted at the bottom of the heel. Input from the rotary potentiometer undergoes a derivation with respect to time to produce the knee angular velocity, as seen in Equation 5. The heel sensor functions as a switch, and is only activated when the heel is in contact with the ground. Therefore, the rotary potentiometer delivers a quantitative measurement to the system, while the heel sensor delivers a purely qualitative measurement.

3.2.2 Knee Angle Examination

As one can see in Figures 2 and 3 both the knee angle and the foot angle exhibit non-linear behavior throughout the percent stride. This poses a significant challenge in developing the rule system. If the relationship was more linear fewer sensors would be needed to differentiate between different stages of the gait cycle. The knee angle exhibits two peaks: one in the stance phase, and the other in the swing phase. The foot angle plateaus near 175 degrees for a significant portion of the stance phase, and drops substantially prior to the initiation of the swing phase where the foot angle proceeds in a steep upward climb.

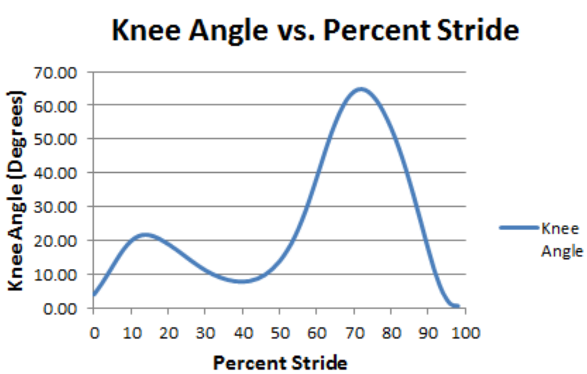

Figure 2. Knee Angle vs. Percent Stride

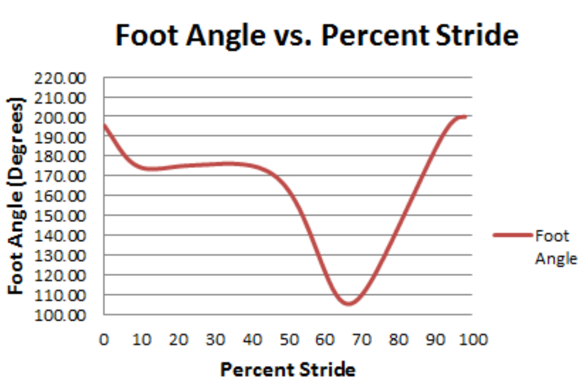

Figure 3. Foot Angle vs. Percent Stride

3.2.3 Knee Angular Velocity and Stride Segmentation

Figure 4 illustrates the knee angular velocity throughout a complete stride. As one can see the angular velocity is positive from 0-14 and from 40-72 percent of the stride. The angular velocity is negative from 14-40 and approximately 72-98 percent of the stride. This information is particularly important in developing the algorithm that will be discussed in greater detail. Taking into consideration that the angular velocity can be divided into two positive regions and two negative regions, and the heel sensor is on for one major region (0-40 percent of stride), and off for one major region (40-98 percent of stride) Figure 2 can be divided into 4 segments listed in Table 1.

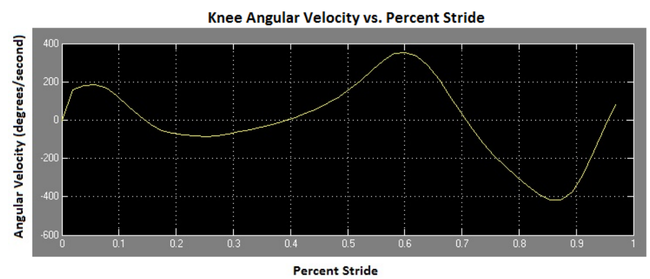

Figure 4. Knee Angular Velocity vs. Percent Stride

Table 1. Segmentation of Knee Angle vs. Stride Plot

Segment	Range of Percent Stride	Heel Sensor	Angular Velocity
1	0-14	On	Positive
2	14-40	On	Negative
3	40-72	Off	Positive
4	72-98	Off	Negative

3.3 Selection of Control Mechanism

Several system identification techniques were examined for potential in prosthesis control. Neural networks were considered, but ultimately dismissed due to the large amount of data required to properly develop an accurate control system. Fuzzy logic exhibited some promise for controlling the system, but modeling with the Fuzzy Logic Toolbox in Matlab produced large errors that could not be reduced. This was mostly likely because of the non-linear interaction between knee angle and foot angle in a complete stride. After these two techniques were discarded the model was developed using two switches placed in series to control the filtering of four functions. The objective is to use the correct function in the correct portion of the stride to accurately relate knee angle to foot angle throughout a complete stride.

3.4 Control Algorithm

Figure 5 displays the control algorithm for governing the foot angle of the prosthesis. The overall system aims to use the angle of the knee, the status of the heel sensor, and the value of the knee's angular velocity to determine the corresponding foot angle. Due to the non-linear behavior of the knee angle through a complete stride the graph in Figure 2 has been divided into four segments listed in Table 1. In the algorithm the first sensor reading is the knee angle measurement acquired from the rotary potentiometer. The derivative is taken to determine the angular velocity, and based upon the positive or negative value of the velocity the algorithm proceeds to evaluate either Segments 1 and 3, or Segments 2 and 4. Next, the heel sensor is read, and based upon the on/off status the algorithm determines which segment to use. In Table 2 it is seen that each segment has a corresponding function where the dependent variable is the knee angle, and the independent variable is the foot angle. In this manner the rule system is able to output a foot angle for each knee angle value across all four segments of the stride.

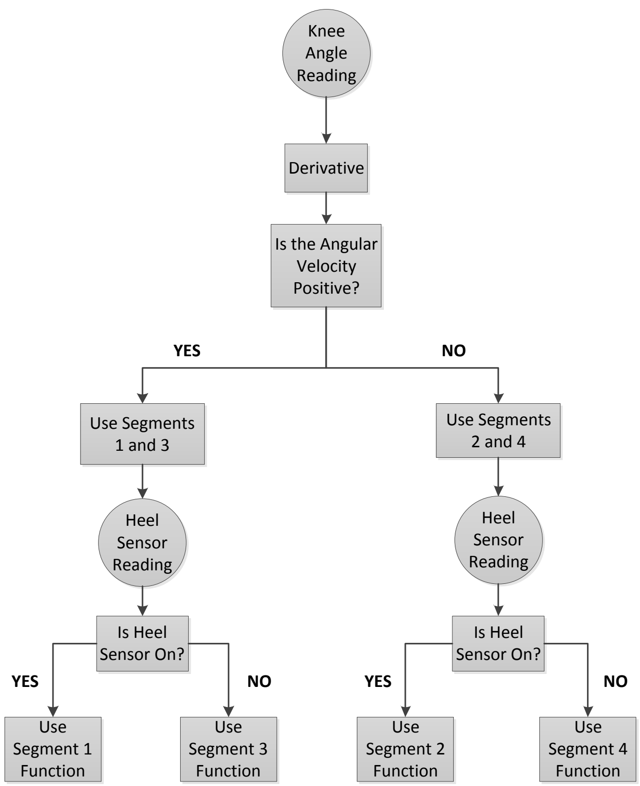

Figure 5. Algorithm governing use of functions

Table 2. Functions Governing Foot Angle

Segment	Segment Function
1	y = -1.2643x + 198.72
2	y = -0.1432x + 177.2
3	y = -1.2165x + 178.28
4	y = -1.1964x + 195.83

3.5 Matlab Simulink Model

Figure 6 exhibits the model that was built in Matlab Simulink. A magnification of this model is more clearly displayed in the Appendix. The model begins with the input of a series of physiologically relevant knee angles. A derivative is immediately taken to determine the knee angular velocity, and this value is the input for the velocity switch. As described in the algorithm segment functions 1 and 3 proceed through the model when the velocity is positive, and segment functions 2 and 4 move forward when the value is negative. The two signals which pass through the velocity switch are separated by the ''demux'' block, and input into the heel switch. The values for the heel sensor are input into the model as a source block, and provide the input for the heel switch. The switch behaves according to the rule system displayed in Figure 5. The output from the heel switch is the foot angle of the ankle-foot prosthesis. Other blocks function to calculate the error of the foot angle, and visually display the true value of the foot angle superimposed against the calculated value.

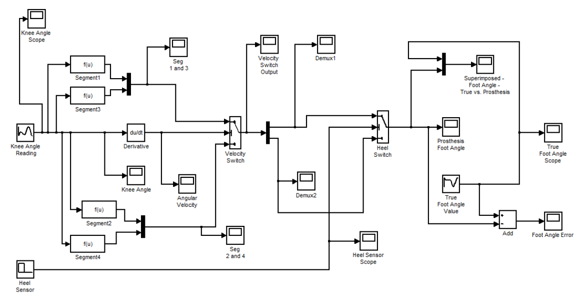

Figure 6. Simulink Model for Testing Algorithm

4. RESULTS AND DISCUSSION

Figures 2 and 4 display plots of the knee angle and the knee angular velocity utilized in the model. Figure 7 illustrates the spectra for the heel sensor seen over one complete stride. The heel sensor is activated in first 40% of the stride, noted by the value of one, and deactivated in the remaining 60% of the stride, noted by a value of zero. The heel switch contains a threshold of 0.9 allowing it to distinguish between the on and off statuses.

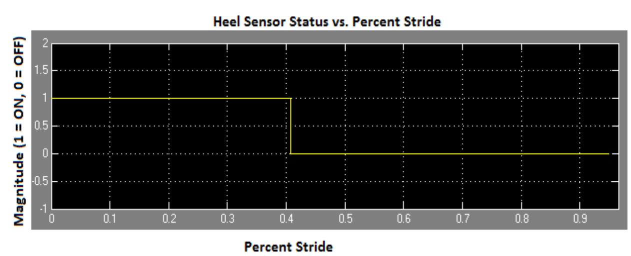

Figure 7. Heel Sensor vs. Percent Stride

Figure 8 displays the true foot angle utilized for evaluation of the theoretical results. Figure 9 illustrates the output of the control system referred to as the prosthesis foot angle. Figure 10 represents a superimposed plot of the true foot angle and the prosthesis foot angle versus percent stride. In this figure one can see that the algorithm functions well in mimicking the true value of the foot angle for the first two segments (0-40%) of the stride. In segment three (40-72%) there are regions of varying accuracy. From 41% to 52% there is a noticeable difference between the theoretical and true values. The largest deviation occurs at approximately 72% of the stride, and is subsequently followed by the beginning of the fourth segment. The function of the fourth segment maintains a fairly consistent error, and follows the overall path of the true foot angle plot.

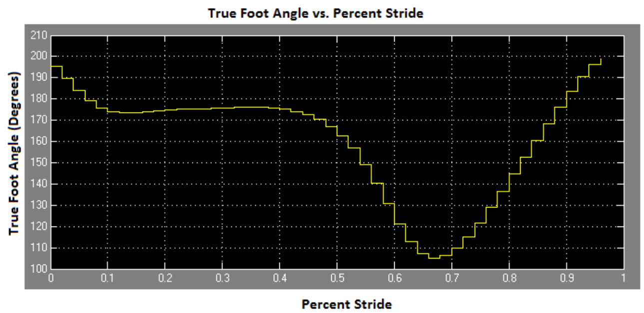

Figure 8. True Foot Angle vs. Percent Stride

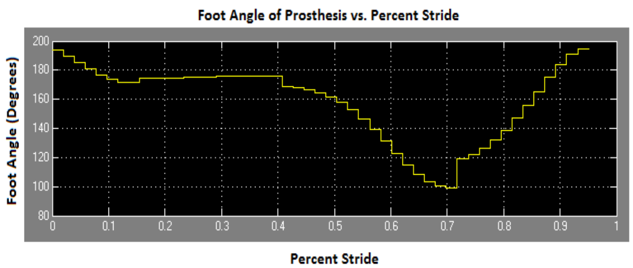

Figure 9. Prosthesis Foot Angle vs. Percent Stride

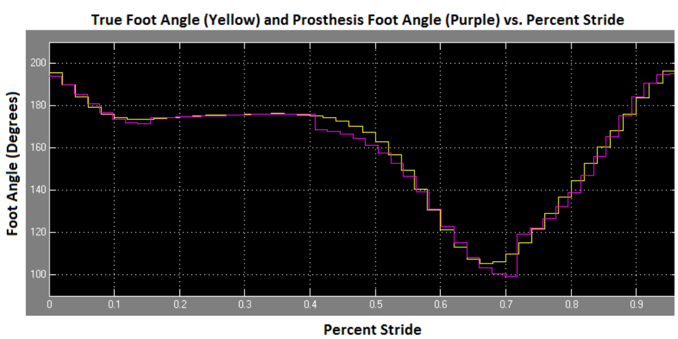

Figure 10. True and Prosthesis Foot Angles Superimposed

In Figure 11 one can see that each of the four segments have drastically different levels of error. In the first segment (0-14%) the error fluctuates from +5.5 to -1.7 degrees. Then, in the second segment (14-40%) the error ranges from +2.1 to -1 degrees. Subsequently, in the third segment (40-72%) the error varies from +10.4 to -9.9 degrees. Lastly, in the fourth segment the error ranges from +5.6 to -7.9 degrees. As one can observe different functions are more accurate at following the true path of the foot angle. At 67 percent stride the true foot angle makes a sharp change in slope that is difficult for the control methodology to follow. This change sets the precedent for the error observed at 70 percent stride.

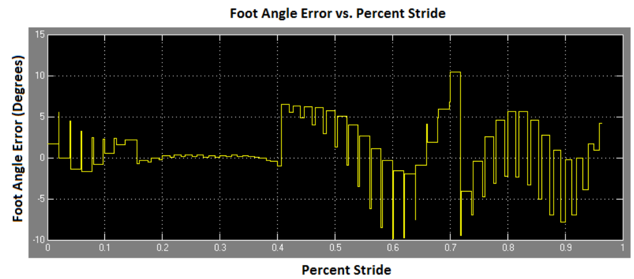

Figure 11. Foot Angle Error vs. Percent Stride

5. CONCLUSION AND FUTURE WORK

This study has successfully simulated a novel algorithm for controlling an ankle-foot prosthesis. The results indicate that this rule system has potential for direct application in micro-processor controlled prosthetics. Matlab Simulink has exhibited utility in modeling the complex nature of human gait, and provided a medium for evaluating the functionality of the algorithm. Matlab Simulink is an indispensible tool in computational modeling, and provides great utility in accelerated modeling and evaluation of control mechanisms.

The reported error is minimal, and can be further reduced through additional studies. Suggestions for future work include implementing advanced numerical method techniques to develop more sophisticated functions, and utilizing additional sensors to better mimic the natural motion of the ankle-foot complex. The study can be further expanded to develop additional control methods for governing foot angle positions in patients with slow and fast cadence. Furthermore, control of kinematics is but one of several important factors to consider in prosthesis design. Additional studies are recommended to investigate mimicking the natural kinetics of human gait to produce enhanced functionality of the prosthesis.

6. REFERENCES

[1] Winter, David. The Biomechanics and Motor Control of Human Gait: Normal, Elderly, and Pathological. Second Edition. Waterloo, Ontario: Waterloo Biomechanics, 1991.

[2] Au S., et al. “An Ankle-Foot Emulation System for the Study of Human Walking Biomechanics”. *2006 IEEE International Conference on Robotics and Automation* (2006): 2939-2945.

[3] Dollar M., et al. “Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art”. *IEEE Transactions on Robotics*. 24, NO. 1 (2008): 1-15.

[4] Zwipp H., et al. “Ankle Joint Biomechanics”. *European Journal of Food and Ankle Surgery*. 1 (1994): 21-27.

[5] Dul J., et al. “A Kinematic Model of the Human Ankle”. *Journal of Biomedical Engineering*. 7 (1985): 137-143.

[6] Surer E., et al. “A markerless estimation of the ankle-foot complex in 2D kinematics during stance”. *Gait and Posture*. 33 (2011): 532-537.

[7] Jimenez-Fabian R., et al. “Review of control algorithms for robotic ankle systems in lower-limb orthoses, prostheses, and

exoskeletons". *Medical Engineering & Physics*. 34 (2012): 397-408.
[8] Ziegler-Graham K., et al. "Estimating the Prevalence of Limb Loss in the United States: 2005 to 2050". *Archives of Physical Medicine and Rehabilitation2008*; 89 (3): 422-429.
[9] Owings M., et al. "Ambulatory and Inpatient Procedures in the United States, 1996". Hyattsville, Md.: U.S. Dept. of Health and Human Services, Centers for Disease Control and Prevention, National Center for Health Statistics; 1998.
[10] Varol H., et al. "Mutliclass Real-Time Intent Recognition of a Powered Lower Limb Prosthesis". *IEEE Transaction of Biomedical Engineering*. 53 (2010): 542-551.
[11] Mohanty R.K., et al. "Comparison of energy cost in transtibial amputees using "prosthesis" and "crutches without prosthesis" for walking activities". *Annals of Physical and Rehabilitation Medicine*. In Press Available Online April 9th, 2012.
[12] Kaufman K., et al. "Energy Expenditure and Activity of Transfemoral Amputees Using Mechanical and Microprocessor-Controlled Prosthetic Knees". *Archives of Physical Medicine and Rehabilitation.* 89 (2008): 1380-1385.
[13] Bellman R., et al. "SPARKy 3: Design of an Active Robotic Ankle Prosthesis with Two Actuated Degrees of Freedom Using Regenerative Kinetics". *Proceedings of the 2nd Biennial IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics*. (2008): 511-515.
[14] Holgate M., et al. "The SPARKy (Spring Ankle Regenerative Kinetics) Project: Choosing a DC Motor Based Actuation Method". *Proceedings of the 2nd Biennial IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics*. (2008): 163-168.
[15] Blaya J., et al. "Adaptive control of a variable-impedence ankle-foot orthosis to assist drop-foot gait". *IEEE Transactions on Neural Systems and Rehabilitation Engineering* (2004): 968-977.
[16] Eilenberg, MF., et al. "Control of a powered anklefoot prosthesis based on a neuromuscular model". *IEEE Transactions on Neural Systems and Rehabilitation Engineering*. (2010): 164-173.
[17] Holgate M., et al. "Control Algorithms for Ankle Robots: A Reflection on the State-of-the-Art and Presentation of Two Novel Algorithms". *Proceedings of the 2nd Biennial IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics*. (2008): 97-102.
[18] Lim C.M., et al. "Modeling of knee joint muscle during the swing phase of gait – a forward dynamics approach using *Matlab/Simulink*". *Simulation Modeling Practice and Theory*. 11 (2003): 91-107.
[19] Gill H.S., et al. "Heelstrike and the pathomechanics of osteoarthrosis: a simulation study". *Journal of Biomechanics*. Volume 36, Issue 11 (2003): 1617-1624.
[20] Mansouri M., et al. "A platform for dynamics simulation and control of movement based on OpenSim and MATLAB". *Journal of Biomechanics.* Volume 45, Issue 8 (2012): 1517-1521.
[21] Wang Z., et al. "Mobility analysis of the typical gait of a radial symmetrical six-legged robot". *Mechatronics*. Volume 21, Issue 7 (2011): 1133-1146.

7. APPENDIX

Derivation of Equations:

$$\theta_h = \theta_{th} - \theta_{tr}$$

$\theta_{tr} = 90°$ in natural cadence

$$\theta_h = \theta_{th} - 90°$$

$$\theta_{th} = \theta_h + 90°$$

$$\theta_k = \theta_{th} - \theta_{lg}$$

$$\theta_{lg} = \theta_{th} - \theta_k = \theta_h - \theta_k + 90°$$

$$\theta_a = \theta_{ft} - \theta_{lg} - 90°$$

$$\theta_{ft} = \theta_a + \theta_{lg} + 90°$$

$$\theta_{ft} = \theta_a + \theta_h - \theta_k + 90° + 90°$$

$$\theta_{ft} = \theta_a + \theta_h - \theta_k + 180°$$

$\omega_k = \frac{d\theta_k}{dt}$ = Knee Angular Velocity

Simulink Model: