
MASSACHUSETTS INSTITUTE OF TECHNOLOGY

**A Neuromuscular Model for Lower Leg
Prostheses Control**

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Abstract

The objective of this project was to create tangible progress toward the creation of a neuromuscular-model-based control system for a below-the-knee prosthesis and knee brace system. Ultimately, the prosthesis system — currently in hardware development in the MIT Media Lab’s Biomechatronics Group — will be used by transtibial amputees to actuate both the missing ankle joint and the incomplete knee joint. This project represents the first steps toward preparing an existing neuromuscular model for the first-ever use in a transtibial prosthesis with a fully-actuated artificial gastrocnemius. While assembling each piece of this optimization into a single pipeline proved beyond the scope of this project, this study constitutes a proof-of-concept for the proposed control system in the sense that all constituent pieces are explored and found to supply results consistent with previous modeling. Over the next few months, these pieces will be assembled into a final prosthesis control system whose efficacy will be evaluated empirically on transtibial amputee subjects.

1 Motivation

Below-knee amputees lack the powerful calf muscles that humans need to walk smoothly and efficiently. This fundamental lack of leg function leads to a variety of pathologies during basic walking tasks. Such amputees require more energy to walk at a given pace, walk at a slower pace than non-amputees, and have a higher chance of developing osteoarthritis and other problems throughout their bodies. The Biomechatronics Group at the MIT Media Lab has developed biomimetic ankle-foot prostheses in an attempt to mitigate these pathologies, but the existing ankle-foot prosthesis only restores the function of the calf muscles at the ankle joint. The lab has recently begun a project under PhD student Michael Eilenberg that aims to introduce an active knee brace to complete the function of the gastrocnemius muscle, which acts not only at the ankle joint, but at the knee joint as well. This project investigates a possible control infrastructure for the resulting system in simulation by adapting an existing full-leg neuromuscular model developed by former Biomechatronics PhD student Dr. Jared Markowitz to optimize only the lower-leg control parameters of interest. In the future, this control system will be implemented in hardware to control both the artificial gastrocnemius currently in development and a BiOM ankle-foot prosthesis to more fully simulate the lower leg of an intact subject.

2 Introduction and Context

2.1 Modeling Human Walking

Human walking is an extremely complex process that has been modeled in countless ways in scientific literature. At one extreme are models that treat the human body as a purely mechanical system, such as pendulum walkers, and at the other are large-scale dynamic optimizations that attempt to model as many muscles and their associated reflexes as possible, often at the expense of vastly overestimating the metabolic cost of transport. The system used in this project, developed by Dr. Jared Markowitz in the Biomechatronics Group, falls somewhere between these two extremes: the major muscle groups of the human leg are first modeled with a number of free muscle-tendon morphological parameters that are optimized for a given subject based on data from the subject’s walking trials, under the assumption that the human body has evolved to minimize the metabolic cost of walking at a self-selected speed. These morphological parameters are then fixed and the resulting model used to optimize more parameters associated with the neurological reflex response. This reflex response model can then be used to control the output torques of a lower-limb prosthesis based on input angles. In this sense, the system constitutes a forward dynamic, reflex-based neuromuscular model.

2.2 Limitations and Advantages of Neuromuscular Control

While controllers based on human biomechanics are often more computationally intensive (and more difficult to reason about) than more generic gait modeling approaches, attempting to more precisely mimic human physiology allows for additional gait features — like terrain and speed adaptation — that would otherwise be difficult to achieve without a complex

sensory infrastructure. These adaptive behaviors have been observed in past neuromuscular controllers, including the one adapted for this project, and it is theorized that these adaptive behaviors could prove robust to a wider variety of environments than those generated by simpler sensor-based models.

In non-amputees, walking gait — and motor control in general — is believed to be determined by both a feedforward neural drive (resulting from the subject’s conscious desire to move) and a reflexive (unconscious) feedback component. Most current lower-limb prosthesis control systems, including the one adapted for this project, consider only the latter, as strictly reflex-based controllers have been shown to produce qualitatively similar gaits to non-amputee human walking and effective control signals to model user intent are difficult to determine. This means that prosthesis users are inherently limited in their ability to externally influence the behavior of the device and are therefore reliant on the control system’s internal adaptation when encountering new environments.

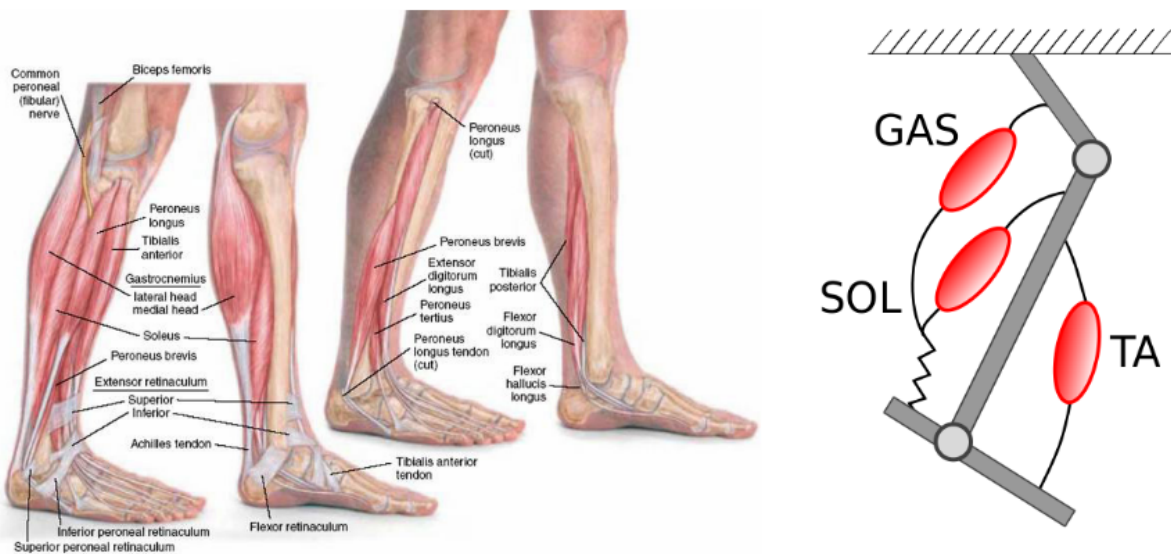


Figure 1: *Left:* The muscle groups of the lower leg. (source: <http://www.stepbystepfootcare.com/faqs/nakedfeet/>) *Right:* Leg abstraction used in neuromuscular model. The gastrocnemius (GAS), soleus (SOL), and tibialis anterior (TA) are modeled as Hill-type muscles, discussed in-depth below.

2.3 Lower Leg Musculature of Interest

The objective of this project is the control of a prosthesis for transtibial amputees. Thus, the major muscle groups of interest are those the amputees are missing — namely, the tibialis anterior, involved in ankle dorsiflexion, the soleus, involved in ankle plantarflexion, and the gastrocnemius, involved both in ankle plantarflexion and knee flexion. The model also considers the morphology of the Achilles tendon, which connects to both the soleus and the

gastrocnemius. For this system, the Hill muscle model is employed, in which each muscle-tendon unit’s contraction force is modeled in terms of the muscle’s length and velocity, as well as a neural “muscle activation” command.

To make analysis more tractable, most current models, including the one adapted for this project, consider movement only in the sagittal plane (i.e., the plane that divides the body into left and right halves), effectively reducing the problem to two dimensions. The system adapted for this project can be thought of as a “lumped parameter” model, in which the torque contributions of smaller muscles are abstracted into the three major muscle groups above.



Figure 2: The BiOM active ankle-foot prosthesis.

2.4 Previous Work in Hardware

Ultimately, the control architecture explored in this project will be used to control a BiOM — a commercial active ankle-foot prosthesis manufactured by iWalk, Inc. — and a powered knee brace that will serve as an artificial gastrocnemius that is currently in development in the Biomechatronics Group. The BiOM’s internal microcontroller and the gastrocnemius will communicate serially via UART using a Raspberry Pi computer. The code for the controller will either run on the Raspberry Pi or on a larger unit with more computational power. In the latter case, the computer running the controller will communicate wirelessly with the Raspberry Pi.

Prior to the start of this project, I completed code for the C++ UART interface between the BiOM’s microcontroller and the Raspberry Pi, proving that they can communicate reliably at 100 Hz, the limit imposed by the BiOM’s serial hardware.

3 System Development

The goal of this project is to make progress toward the creation of a standalone controller that takes as input each of the prosthesis’ joint angles and outputs joint torques. This will be accomplished by processing these angle inputs through a neuromuscular model that attempts to replicate the behavior of an intact biological system as closely as possible while minimizing computational complexity. Ahead of time, this model is itself run through an optimization procedure to set a series of morphological parameters based on walking data from a given subject. Each of these optimization steps is discussed in detail below.

My work for this project can essentially be considered an adaptation and extension of the work by Dr. Jared Markowitz on developing a neuromuscular model of human walking and applying it to prosthesis control. All of the code and Simulink models I employ in this project are drawn directly from his PhD thesis work in the Biomechatronics Group. Over the course of the project, I first got the existing full-leg optimization code up and running, modified it to optimize only the morphological parameters of interest in several ways, and compared the results with existing parameter sets.

3.1 Morphological Parameter Optimization

3.1.1 Existing Optimization Procedure

While the manner in which muscles and their associated reflexes generate a walking gait can in some sense be generalized across the human population, prosthesis design must account for variation across subjects. The first step in the creation of the control system explored in this project is therefore to use kinematic, kinetic, electromyographic (EMG), and metabolic data to estimate individual test subjects’ muscle-tendon states, joint moments, muscle activation, and energy expenditure, respectively. These estimates are then used to assign morphological parameters to an otherwise generic model of human muscles and tendons. This is accomplished through a multi-objective optimization function that attempts to minimize the metabolic cost of the walking gait associated with each possible parameter set while matching as closely as possible the observed gait. The optimization is “kinematically-clamped” in that it fixes subjects’ joint angle trajectories as estimated from kinematic data while allowing torques to vary across parameter sets.

The morphological parameters assigned in this optimization to each Hill-type muscle-tendon complex are muscle maximum isometric force F_{max} , maximal force length l_{opt} , tendon shape factor K_{sh} , and tendon reference strain λ_{ref} . At each iteration of the optimization, corresponding to one possible assignment of these parameters, the metabolic cost of transport and the “kinetic fit” of the data — i.e., how closely the modeled joint torques match those measured empirically — are both computed. Each is used as a cost function in a multi-objective genetic algorithm executed using MATLAB’s `gamultiobj` command.

3.1.2 Observations on MATLAB’s Multi-Objective Genetic Algorithm and Simulink Modeling

When performing the morphological optimization described above, the parameters of interest can be specified both in the MATLAB code running the genetic algorithm (by manually

specifying the parameters desired) and in the associated Simulink model (by including or omitting each muscle-tendon model). I explored several methods of optimizing only the lower leg parameters: namely, I compared the parameters associated with the gastrocnemius, soleus, and tibialis anterior as generated by

- the complete (50-parameter) leg model in both MATLAB and Simulink;
- the 12 parameters of interest in MATLAB calculated from the full Simulink model; and
- the 12 parameters of interest in MATLAB calculated from a Simulink model of only the muscle-tendon complexes of interest.

Each method above resulted in qualitatively reasonable parameter assignments. At the same time, each method seemed to explore different areas of the Pareto front more thoroughly than others and generated vastly different numbers of solutions. Using the same genetic algorithm settings, the first method produced only 179 possible solutions, while the second produced 237 and the third 343. This makes intuitive sense, as the first method is far more constrained by its additional parameters than those methods that consider only the parameters of interest. In the future, the precise reasons for these differences should be more thoroughly investigated in order to ensure that the genetic algorithm is exploring the entire possible solution space.

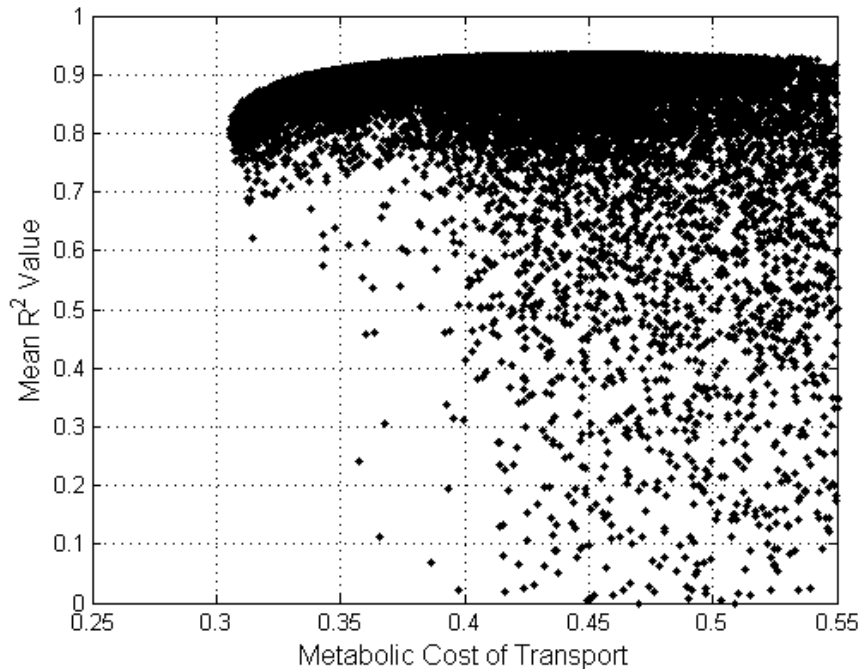


Figure 3: Example solutions plotted by cost function generated by the optimization pipeline on the full 50-parameter model for a single subject. The curved envelope at the top of the solution set constitutes the Pareto front.

3.2 Choosing an Optimal Morphological Solution

Because the parameter optimization is a multi-objective function, it generates a large set of arguably “optimal” solutions depending on how the two costs are weighted. Known as the Pareto front, these solutions comprise any parameter set in which one cost function cannot be decreased without increasing the other.

When modeling all muscle groups in the full leg, as in the work by Markowitz, a trend is evident when considering the distribution of energy expenditure among muscles in solutions across the Pareto front. While the metabolic cost of transport is relatively evenly distributed among muscle groups when the gait less precisely matches experimental data, there exists a point along the front at which one or more muscles — primarily the vastii — begin to work disproportionately hard relative to other muscles to achieve minor improvements in kinetic agreement. Thus, in Markowitz’s study, the point along the Pareto front with maximal kinetic agreement, but before this shift in relative metabolic cost among muscles occurs, was chosen as the final set of morphological parameters.

When modeling only the gastrocnemius, soleus, and tibialis anterior, on the other hand — the muscles relevant to this study — no such trend emerges. For the purposes of this project, I chose a set of parameters from near the center of the Pareto front for verification against previous models and found that it was indeed consistent with parameters generated for the lower leg by the full model in Markowitz’s study. When this optimization is eventually applied to prosthesis control, however, a more rigorous study of these curves’ characteristics should be undertaken to determine a reasonable selection paradigm.

3.3 Reflex Optimization

The last step in the creation of the neuromuscular control system is to determine the subject-specific parameters associated with each muscle’s reflexive neural activation — namely, appropriate gains, thresholds, and time delays for each muscle-tendon complex that result in predicted activations consistent with EMG data. Incorporating the morphological parameters calculated above, a second kinematically-clamped optimization is performed using MATLAB’s `ga` genetic algorithm function to determine these parameters, with the difference between modeled and measured muscle activations as the associated cost function.

While I was unable to explore this optimization step as fully as those above over the course of this project, as existing code was much less complete, I was able to implement a reflex optimization for the gastrocnemius muscle that generated qualitatively reasonable parameter values. Because each muscle can be reflex-optimized separately, this constitutes a proof-of-concept for the overall procedure.

4 Conclusions and Future Work

While each of the optimizations implemented in this project successfully generate control parameters consistent with existing neuromuscular models, the pieces of the pipeline are not yet fully integrated into a viable control system that could be deployed on a physical prosthesis. Thus, the next step in this project’s development is to streamline the process by

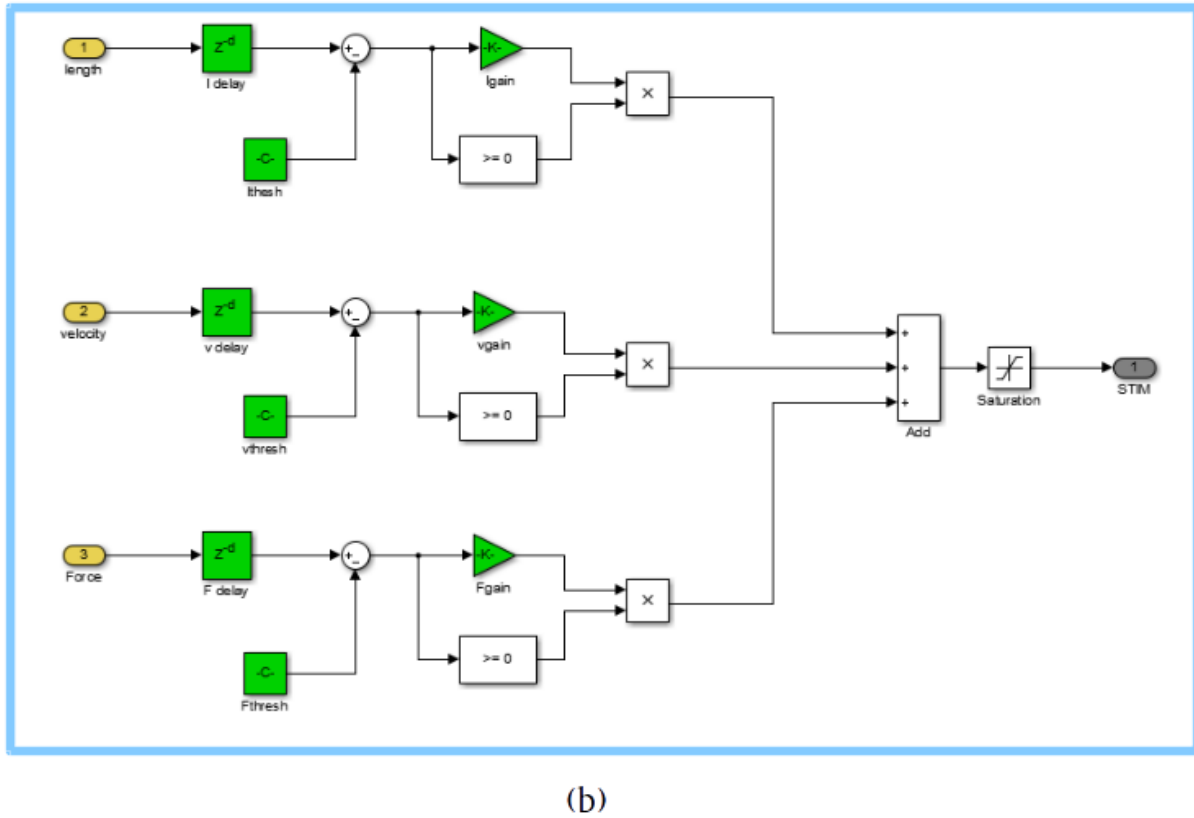
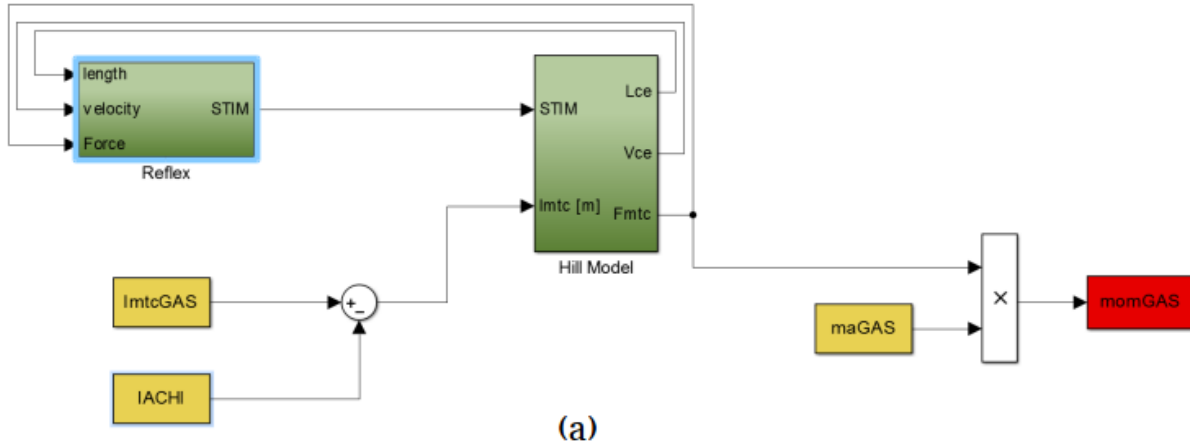


Figure 4: Simulink model used to calculate gastrocnemius reflex parameters: namely, gain, threshold, and time delay for the muscle length, velocity, and force associated with reflex stimulation. Insert (b) corresponds to the **Reflex** block in full control model (a). Once these gains are established via the optimization described above, the model can be used for prosthesis control, converting muscle-tendon length (**lmtcGAS**), muscle moment arm (**maGAS**), and Achilles tendon length (**IACHI**) to output torque (**momGAS**). Similar models can be used in reflex optimization and control of the soleus and tibialis anterior.

which data from test subjects is captured and processed to generate the subject’s morphological parameters, which can then be used in the reflex optimization to generate appropriate control parameters that can ultimately be used in a control system for the prosthesis of a biomechanically matched subject. Additional wrappers for the pipeline should also be generated to allow data of different formats to interface with the optimization procedure. This pipeline could prove useful for any number of new prosthesis and exoskeleton projects undertaken by the Biomechatronics Group, so packaging the code in an accessible manner is perhaps the most important contribution of this project.

Looking further ahead, there exist a number of more sophisticated avenues of exploration once the full pipeline is in place. The neuromuscular model, for example, is computationally intensive, and it would be worth exploring the efficacy of systems that replace one or more of the muscle models with a simpler spring-clutch system to reduce the computational burden on the device’s software infrastructure. Improvements could also be made to the hardware sensors themselves: currently, for example, the BiOM’s ankle torque is measured using a Hall effect sensor at the ankle joint but does not take into account the full angle of the carbon fiber “foot,” which flexes substantially during stance phase, similarly to the flexion in the forefoot structures of a non-amputee. Thus, measuring angle using an electrogoniometer attached to both the shank of the leg and the toe of the BiOM could generate training data more consistent with the proposed neuromuscular model.

5 Personal Reflection

This project was an opportunity to expand on my previous work in the Biomechatronics Group toward the creation of a real prosthesis. This inherent applicability — especially with the potential for philanthropic impact — is what drew me to electrical engineering and computer science in the first place. I’ve both furthered my knowledge in EECS, through developing hardware communication systems and coding in MATLAB and Simulink, and explored a variety of other fields, from the physics of modeling the human body to the mechanical engineering of designing a system to interface with it.

More than anything else, this project underscored the inherent complexity of biomechanical modeling, let alone building it into a control system. The project was a constant exercise in scope, with a steep learning curve, and required both a working knowledge of a wide variety of fields and the ability to mentally black-box components to gain intuition — skills that will serve me well as I enter graduate school in the fall.

6 Acknowledgements

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