Replacement of an Active Metatarsophalangeal Joint with a Passive Spring-Damper System for Implementation in an Ankle Prosthesis

by

Eric S. Kline

Submitted to the Department of Mechanical Engineering in Partial Fulfillment of the Requirements for the Degree of Bachelor of Science in Mechanical Engineering at the Massachusetts Institute of Technology

June 2016

© 2016 Massachusetts Institute of Technology. All rights reserved.

Signature redacted

Signature redacted

Signature redacted

Signature redacted

Signature redacted

Signature redacted
DISCLAIMER NOTICE

Due to the condition of the original material, there are unavoidable flaws in this reproduction. We have made every effort possible to provide you with the best copy available.

Thank you.

The images contained in this document are of the best quality available.
Replacement of an Active Metatarsophalangeal Joint with a Passive Spring-Damper System for Implementation in an Ankle Prosthesis

by

Eric S. Kline

Submitted to the Department of Mechanical Engineering on May 17, 2016 in Partial Fulfillment of the Requirements for the Degree of

Bachelor of Science in Mechanical Engineering

ABSTRACT

Analytical modeling was used to determine the optimal configuration of a replacement metatarsophalangeal (MTP) joint to be used in a prosthetic ankle. A spring was added to the joint to store energy and release it during the part of the gait cycle where the highest torque is required, reducing the torque the motor must exert. A linear spring-damper system adapted for use on a rotational joint was found to exhibit similar behavior to the biological joint for the range of motion required. The optimal gear ratio for the ankle motor, spring constant, and damping constant for the MTP joint were found using a MATLAB program written for this purpose. A physical prototype was fabricated, and testing was performed on an Instron machine to validate the results.

Thesis Supervisor: Hugh Herr
Title: Associate Professor
Acknowledgements

Special thanks to Hugh Herr for being a fantastic advisor and his willingness to work with me, and to Tyler Clites for his determination and patience in helping out in the process of writing this thesis. The author would also like to thank the Pappalardo staff for all the assistance and use of their workspace for machining of the prototype, and especially Bill Cormier for his expertise and assistance. Additionally, thanks to Pierce Hayward for assisting in the setup and testing of the device and use of the 2.002 lab, and thanks to James Roggeveen for assistance in the manufacture of the prototype. Data and valuable assistance was provided by Allison Arnold-Rife from Concord Field Station during the preliminary and background work for the project.
# Table of Contents

Abstract ........................................................................................................... 3  
Acknowledgements ......................................................................................... 4  
Table of Contents .......................................................................................... 5  
List of Figures ................................................................................................. 6  
1. Introduction ............................................................................................... 7  
  1.1 Background and Motivation. ................................................................. 7  
2. Design Constraints ...................................................................................... 8  
  2.1 Low Peak Current. .............................................................................. 8  
  2.2 Minimizing Prosthesis Position Error. .................................................. 9  
  2.3 Minimizing Weight ............................................................................ 9  
  2.4 Actuation .......................................................................................... 10  
3. MATLAB Modeling. .................................................................................... 10  
  3.1 Background and Previous Work. ......................................................... 10  
  3.2 Optimization and Results ................................................................... 11  
4. Design of Physical System. ........................................................................... 12  
  4.1 Design Considerations. ..................................................................... 12  
5. Prototype Testing ....................................................................................... 14  
  5.1 Motivation. ...................................................................................... 14  
  5.2 Test Setup ......................................................................................... 14  
  5.3 Test Results ..................................................................................... 15  
  5.3.1 Spring Constant ........................................................................... 16  
  5.3.2 Damping Coefficient .................................................................... 18  
  5.4 Further Testing .................................................................................. 19  
6. Suggestions for Subsequent Iterations. ....................................................... 20  
7. Concluding Remarks. ................................................................................ 21  
8. Works Cited .............................................................................................. 22
List of Figures

Figure 1. ..................................................... 6
Figure 2. ..................................................... 11
Figure 3. ..................................................... 12
Figure 4. ..................................................... 13
Figure 5. ..................................................... 13
Figure 6. ..................................................... 14
Figure 7. ..................................................... 15
Figure 8. ..................................................... 15
Figure 9. ..................................................... 16
Figure 10. .................................................... 17
Figure 11. .................................................... 17
Figure 12. .................................................... 18
Figure 13. .................................................... 19
1. Introduction

1.1 Project Background and Motivation

The goal of the overall project is to create a prosthetic ankle for transtibial (below the knee) amputees that has bidirectional feedback, allowing for better control over varied terrains and situations. Many prostheses at this point in time are either passive or, if motorized, run through a pre-programmed gait trajectory [1]. This makes it difficult for amputees to participate in more complex activities such as running or hiking, especially over uneven terrain. The goal of the project is to produce an ankle-foot prosthesis that exhibits close biomimetic behavior and has bidirectional feedback. This will allow the user better control of the ankle and a higher level of adaptability.

Because the project has not progressed far enough to be in the human trials stage, the first phase will involve animal testing with pygmy goats as test subjects. This species works well because of the similarities in the musculoskeletal structure to humans, and because there is a large amount of data available. Biological gait data (position, joint torques and trajectories) for pygmy goats obtained using a motion tracking system was generously provided by Allison Arnold-Rife from Concord Field Station [2].

![Diagram of biological and prosthetic goat limb](image)

Figure 1: Schematic showing biological and proposed prosthetic goat limb, and designations of variables for modeling.

The ankle, foot, and hoof (shown in figure 1) will be replaced with a prosthetic limb. A number of complex systems will go into making this a reality. The author's portion of the project was to design a replacement for the metatarsophalangeal (MTP) joints, or the joints between the foot and the toes. The ankle will be powered electronically and motorized, but to keep the weight and power consumption low, the MTP joints will be replaced with a passive linear spring-damper system. The goal of the author's part of this research was to discover how to most closely follow actual gait data by adjusting the spring constant, damping.
coefficient, and transmission gear ratio. The physical device was modeled in MATLAB and extensive optimization was performed to determine the ideal configuration.

After optimization, a prototype was created in SolidWorks and then fabricated for the purpose of testing and validating the results. The foot-toe device was mounted on a special base and set up on an Instron machine. Subsequently, the device was run through a number of static and dynamic tests to determine the spring and damping ratios.

After confirming the parameters of the foot-toe joint, a new iteration of the first prototype was planned with the intent of lowering the weight without sacrificing strength or function.

2. Design Constraints

2.1 Low Peak Current

One of the primary design constraints for the prosthetic ankle being designed by Tyler Clites in the Biomechatronics Group of the MIT Media Lab is the peak current required by the motor. A small, frameless motor drives a ballscrew which pivots the ankle around a joint, providing the torque required to drive the foot. The amount of heating losses caused by Joule heating in the motor windings scales with the current and resistance of the motor:

\[
heating\ energy\ loss \propto I^2 \cdot R
\]

where \(I\) represents the current drawn by the motor and \(R\) stands for the resistance of the motor coils. As this relationship shows, as the current increases linearly, the amount of energy lost as resistive heating will increase exponentially. This leads to motor heating, which is undesirable in close contact to living tissue, can potentially cause melting of the wire insulation and subsequent short-circuiting, but also reduces motor efficiency. Thus, to produce higher efficiency and to avoid substantial problems with the motor it becomes necessary to minimize the current drawn by the motor. This reduces energy losses and leads to longer battery life. Since the internal resistance of the motor is inherent in the
motor, it cannot be adjusted without choosing a different motor. It is therefore crucial to reduce both the peak current the motor draws over a gait cycle and also to reduce the average current if possible.

2.2 Minimizing Prosthesis Position Error

Another critical parameter of the prosthesis, especially when it moves to human trials, is the ability to accurately control position trajectory. Optimally, the position difference between a point on a normal biological leg and the corresponding point in the prosthesis would be zero. Due to the difficulty of creating a system that successfully duplicates the exact trajectory of every point true to the biological norm, the design process for the prototype foot-ankle system was focused on minimizing the error in the knee joint. The fundamental assumption made to justify this was the following: if the knee joint followed the same position trajectory as the biological leg, what happened in the ankle, foot, and toes would not affect the user's experience significantly, as no physical perception below the knee occurs in the prosthesis. With a biomimetic knee trajectory, the user should be able to keep the center of gravity in the natural position during walking and avoid using techniques to overcompensate for the prosthesis, such as over utilizing other muscles [1]. The user experience should be very similar to his/her experience before the loss of the limb below the knee.

Parameters for a "suitable" amount of deviation were not established before analytical modeling, but it was determined that the position error should be minimized to give the user the least discomfort and to enable adaptive user behavior in compensating for an appropriately small offset.

2.3 Minimizing Weight

In order for the system to be most efficient, the weight should be low. Ideally, it would not exceed the weight of the amputated portion of the leg, but with a battery and motor, it is difficult to maintain the natural weight, at least in this trial. The aim was to create a foot and toe, joints included, that did not exceed 150 grams, although this could
vary a small amount depending on the mass of the subject. All calculations here are done for a goat with a mass of 28 kg, and normalized data from Concord Field Station was used with a correction factor for the weight. At this point in the process, the more important characteristics were the spring and damping constants, so minimizing weight was not highly prioritized in the first iteration. Subsequent prototypes and actual prostheses will need to have a much lower mass (see section 6).

2.4 Actuation

In the biological system, joints are actuated by a combination of muscle and tendons which collectively have energy storing and dissipating characteristics. Ideally, both the ankle and the MTP joint would be replaced with motors to accurately control the angular position and torque output of both joints. However, due to the weight constraints discussed in section 2.3, this is not feasible in this situation. It was determined to replace the MTP joint with a passive element that stores and dissipates energy in a similar manner to the biological joint, with the exception of not being actively powered.

Both the ankle joint and the MTP joint would be replaced with motors in an optimal situation, but in order to mimic biological behavior, the replacement prosthesis must be similar in geometry and mass to the original biological leg. Adding a motor would add some additional weight, but driving two motors would require more power and increase battery size. This would add too much weight to the prosthesis and interfere with operation, in addition to increasing the system complexity and necessity for a more intricate control system. To avoid these issues, a passive spring-damper system replaces the MTP joint. Storing the energy in the spring additionally reduces the load on the motor; energy is stored during the first part of stance in the gait cycle, and then released as the ankle requires its peak torque. This lessens the amount of current the motor must draw, helps with efficiency, and also aids in replicating biological leg trajectories.

The purpose of this simulation was to compare the analytical model to the normalized gait provided by the Concord Field Station. In order for the passive MTP joint replacement to be effective, it must approximate as closely as possible the action of the
biological goat joint. Both the biological ankle joint and MTP joint are comprised of elements that both store and dissipate mechanical energy. To replace the active biological joint with a passive mechanical one, a rotational spring damper system was considered. With the rotational model of a torsional spring damper system, the spring constant and damping constants were determined using the MATLAB code created by the author and Tyler Clites in January 2015 (see section 3.1).

The torsional spring dashpot system was determined to be unfeasible due to the large size and weight of most commercially available rotational dashpots. A linear spring dashpot was decided upon for this situation, as it was found to approximate the action of a rotational joint for the small angles it rotates through.

3. MATLAB Modeling
3.1 Background and Previous Work

Previously to this work, a MATLAB model was created to predict the behavior of a mechanical prosthesis in various configurations. The goal of this modeling was to determine characteristics of the system without needed to guess-and-check with physical prototypes, a system that would be neither time nor cost effective. In January 2016, the author (as an undergrad researcher) and Tyler Clites (a graduate student in the MIT Media Lab) created a MATLAB simulation based on an inverted triple pendulum model with a driving force at the top. (See figure 1 for the geometry and physical representation). The three bottom “links” in the pendulum represent the hoof, foot, and shank of a goat’s leg. The forces at the top of that diagram represent a combination of the forces acting on the goat’s knee. These take into account the acceleration of the animal’s center of gravity, movement of its muscles, etc. These data were taken from the paper published in Journal of Experimental Biology by Arnold, et al [2]. The kinematic equations for this system were found, and then the joint angles and torques were solved for, which allowed for comparison between the empirical biological model and the predictive model created in MATLAB. In the program, the parameters for the gear ratio in the motor, the spring constant, and the damping coefficient were all variable, and a range of these values was
run through the code in order to optimize. Later, in the fall of 2015 and beginning of 2016, another version of the code was created in order to represent a linear spring damper system to replace the MTP joint. This code was used extensively by the author to optimize these variables in the system.

3.2 Optimization and Results

The aforementioned MATLAB code was run to determine the optimal values for the spring and damper system. The code uses the kinematic model produced in January 2015 and runs this model iteratively, with a range of values for both the damping coefficient $b$ and the spring constant $k$ of the system for the linear model. The range for the damping coefficient was 100-220 N-s/m, in increments of 10 N-s/m. For the spring, a range of 8000-12000 N/m was chosen, with steps of 200 N/m. All possible combinations of these spring-damping constants were run, and then plotted against the maximum peak current (see figure 2) and the maximum offset (figure 3).

![Figure 2: Peak current drawn by ankle motor during gait cycle. Lower spring constants and damping coefficients perform better when it comes to lowering the peak current. Note that for all combinations, however, the maximum current predicted is still well below the specification of less than 9 A.](image-url)
Minimizing the amount that any point on the prosthesis deviates from the biological system (actual goat foot) is important, but the knee is chosen as a point of reference because it will still be part of the animal’s biological leg, and therefore needs to follow the same pre-prosthesis trajectory.

A linear spring constant of 8600 N/m and a linear damping constant of 160 N-s/m was chosen, which should have a peak current of 7.86 A and a predicted maximum offset from biological of 2.23 mm.

4. Design of Physical System
4.1 Design Considerations

The first prototype of the system was not required to be lightweight, but it was built to be robust and precise. Joints needed to rotate freely but have little to no lateral movement, and alignment of the spring and damper was important. For the overall system, the geometry was determined by an update to the MATLAB program created for the optimization of the spring and damping coefficients. The initial simulation was done with a rotational spring and rotational damper, but because of the difficulty in finding suitable ones, the model with a linear system was chosen. (See figures 4 and 5). Additionally,
commercially available springs and dampers with the exact constant required were not readily available, so the spring chosen has an actual constant of 8401 N/m and the damper has a coefficient of 128 N-s/m instead of 160, which puts the max offset at 2.54 mm and a max current of 7.79 A, which is still within specification.

Changing the rotational system to a linear one required additional calculation; the attachment points for the spring-damper system in the MTP joint change the behavior significantly. The damper being used has a travel of only 12 mm [4], so over the entire 22 degrees of travel, this was the maximum the length of that system could change. The introduction of an “offset angle” (figure 6) on the toe joint allowed for this, and the optimal angle was calculated using an updated version of the MATLAB code, in addition to the attachment points on the foot and toe.
Figure 6: The offset angle shown on the left aids in two respects; it decreases the amount of travel in the damper, which enables a commercially available mini-buffer to be used, and it also decreases the nonlinearity in the rotational spring constant. The blue arrows mark the attachment points for the spring-damper system, and these were carefully chosen based on the damper travel for the damper and spring used.

5. Prototype Testing

5.1 Motivation

Testing was performed in order to validate the model that was created, in addition to checking the spring rate and damping constant in the actual physical system. According to McMaster-Carr's website, from which the spring was purchased, there is a possible variation in the spring rate of $\pm 8\%$ [3]. This would mean that for the spring rate of 8406 N/m specified, the actual rate could vary from 7733 to 9078 N/m. Because of the high quality and consistency that would be required for a medically approved prosthesis, the spring constant would need to be verified for each prosthesis, or more practically, supplied by a more reliable source in terms of product specification quality and accuracy.

5.2 Test Setup

The entire foot-MTP joint mechanism was tested on an Instron test machine. The machined foot was placed in a custom designed mounting plate to hold it at an angle relative to the Instron force plate and actuator. See figure 7.
The foot was subjected to several cyclic tests in which the Instron was lowered by 6mm and to rotate the joint to the end of its range and then relaxed. Several tests for the leg joint were performed. A series of tests in which the joint was forced to a certain position, and then held there before returning to the initial position, was performed in order to clearly find the spring constant of this system. Since the system is moving a very small
amount (22 degrees in the model), the spring constant will fluctuate along the trajectory but for the most part will be linear. The differing rotational spring constant at different angles is slight but is included in the computer simulation.

5.3 Results

5.3.1 Spring Constant

The Instron was lowered to a point where it deflected the leg to its maximum point, 7.5 millimeters down, and then was held for 15 seconds to avoid any effects from the damper. Though the damper has its own spring constant, it is relatively small compared to that of the spring used and is negligible in these calculations. The position profile is highly accurate in the Instron, and the load was measured with a 100 N load cell for higher accuracy than the Instron load cell provided.

![Position Profile for Test](image1)

![Position Control Test on MTP Joint](image2)

**Figure 9:** Plots showing the deflection of the tip of the foot over the Instron test cycle (left) and the load required to deflect the toe (right)

The spring constants were calculated using the geometry of the leg in the position in which it is held in the test apparatus, and the results for the period the leg is held while the spring is compressed is shown in figures 10 and 11, for the linear spring constant and rotational spring equivalent, respectively.
Figure 10: Calculated average linear spring constant was found to be 8170 N/m, which falls within the spring's specification.

The spring rate is within 300 N/m of the one ordered (8406 N/m) and within the 8% possible variation according to McMaster-Carr.
5.3.2 Damping Coefficient

To test damping coefficient of the system, the spring was disengaged from the rest of the actuation, allowing the damper to be tested alone. The MTP joint with only the damper connected was subjected to a position displacement shown in figure 12:

![Position Profile for Damper Test](image)

**Figure 12:** Position control for damper-only test. Linear speed was 100mm/min.

The load-position response of this system demonstrates hysteresis, which is to be expected, and also exhibits some elastic characteristics due to a built in spring in the damper [4]. The damping coefficient $b$ was found by solving for the spring constant in the damper (derived from a test in which only the damper was moved to a position, held there, and the force output was measured), then calculating for $b$ using the equation of motion for a basic spring-damper system:

$$ F = kx + b\dot{x} $$

Solving for $b$ gives:

$$ b = \frac{F - kx}{\dot{x}} $$

Using the geometry of the device gives an average damping constant of 623.3 N-s/m in calculation, which is much higher than what the damper purchased was calculated
to be. This may have arisen from error in calculating, error in the tests, or simply means that more testing is needed to verify that this is correct. If it is not, a new damper must be found that exhibits the correct characteristics.

![Position vs Load output for MTP Joint Damper](image)

**Figure 13:** Position vs load required for a cyclic loading of the damper. The space between the curves represents the energy dissipation in the damper.

### 5.4 Further Testing

It is suggested that further tests are run that move the joint in the prosthesis trajectory profile created using the MATLAB simulation. This trajectory is similar to the biological trajectory, but will vary slightly as a result of the differences between the two. If the foot-toe combination were tested in this way, with the position and velocity profile required by the prosthesis model, how accurately the model approximates the actual prototype could easily be observed by comparing torque outputs of the tested joint to the model results.

Additionally, in further testing, the apparatus used to hold the foot-toe joint in place should be updated. The base for the plate was not stiff enough, and flexed a small amount
at the end of some of the loading cycles. Since the stiffness of the spring was far less than that of the base, this effect was neglected, but in future testing a different base should be fabricated to increase the accuracy of the tests.

6. Suggestions for Subsequent Iterations

Due to the high weight and suboptimal function of the first design for the foot-MTP joint assembly, a second design was produced. The goals of this new design were to create a new iteration of the first that held the geometry constant while lowering the weight and the bulkiness of the prosthesis. This design incorporates a more I-beam like geometry in the cross section of many of the components that doesn’t sacrifice much stiffness, yet produces a much more lightweight structure. This will reduce the amount of mass the ankle motor has to move, and help with system efficiency.

Additionally, the material should be evaluated for future iterations. The prototype is made from solid 6061 aluminum alloy, and is bulky and ill-suited for an actual prosthetic ankle. A suggested material for use is carbon fiber; the material has a good weight-to-strength ratio and its use would significantly decrease the weight of the device while maintaining strength and stiffness. A rubber surface with a high coefficient of friction should be added to the tip of the toe joint in order to prevent slipping and allow for pivoting without sliding along the point of contact.

The shoulder screws used in the initial design are very precisely made and function well as joints. A lighter solution, however, may be a press fit on one part of the joint, with a close fit to allow for rotation on the other. This would eliminate the weight of the screw head and nut. The possibility of becoming loose is a danger, and would need to be monitored.

In addition to this, a strain gauge will need to be added to the ballscrew attached to the motor in order to measure the torque output of the ankle, which is important to the control system.
7. Concluding Remarks

Replacing the MTP joint on an ankle prosthesis with a passive spring-damper system will allow for more accurate biomimetic behavior and will additionally reduce the amount of current the motor draws, increasing efficiency and allowing for a lighter battery. Assuming that the computer simulation is correct, and if the joint is built according to the parameters found in the optimization, the prosthesis should replicate closely the biological ankle-foot complex that it is meant to replace.

With further development and testing, the prosthesis will be able to be used to help those with below-the-knee amputations improve quality of life.
Works Cited


