Design of a Passive Ankle Prosthesis with Energy Return That Increases with Increasing Walking Velocity

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DESIGN OF A PASSIVE ANKLE PROSTHESIS WITH ENERGY RETURN THAT INCREASES WITH INCREASING WALKING VELOCITY

by

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A Thesis submitted to the Faculty of the Graduate School, Marquette University, in Partial Fulfillment of the Requirements for the Degree of Master of Science

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ABSTRACT
DESIGN OF A PASSIVE ANKLE PROSTHESIS WITH ENERGY RETURN THAT INCREASES WITH INCREASING WALKING VELOCITY

Alexander J. Folz, B.S.
Marquette University, 2017

Patients who undergo a transtibial (below the knee) amputation are often met with a difficult decision: selection of a prosthesis. Limitations of currently available prostheses motivate work on a new solution, the EaSY Walk, a passive device that mimics two key aspects of the natural ankle: non-linear rotational stiffness through implementation of a stiffening flexure mechanism and rotational work output that varies as a function of walking velocity to propel the user forward.

To achieve the latter, a strategy to convert the maximum available translational energy acquired from deflection along the leg into rotational energy about the ankle joint through coupling of these two degrees of freedom is used. This strategy utilizes maxima/minima of known ankle profiles to control timing of critical device functions as well as the quantity of energy input from leg deflection. In doing so, both consistent operation of the device and maximal energy output at a given walking velocity are theoretically obtained.

Optimizing for both aforementioned ankle criteria, 25.1% of the work of the average natural ankle was achieved for 15 mm of leg deflection, less deflection than is exhibited by many shock absorbing pylon prostheses. After fabricating and testing the optimized design using a repeatable robot trajectory, the device was found to convert 26.6% of input translational work as rotational work, accounting for 63.1% of modeled rotational work. Through human subject testing, the device was found to function inconsistently due to the large impact loadings associated with human gait. In order to achieve proper functionality with human gait, design modifications to the energy storage and release devices are recommended.
ACKNOWLEDGMENTS

Alexander J. Folz, B.S.

In no particular order, I would like to thank my family for their continued support of my academic pursuits, dating all the way back to preschool until now, as I sit on the cusp of finishing my Master’s Degree. I would like to thank my fiancee for her support of my endeavors in addition to giving me the motivation to keep pushing forward, through good times and bad. I would like to thank my advisor, Dr. Schimmels, as well as everyone else in our research group in addition to everyone from the engineering department who consistently supported me throughout my time as a student. Without your help, I would not be where I am today as a student or, perhaps more importantly, as a person. Thank you to you all, I can only hope that I am able to give back to others what you have given to me.
# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>ACKNOWLEDGMENTS</td>
<td>i</td>
</tr>
<tr>
<td>LIST OF TABLES</td>
<td>vi</td>
</tr>
<tr>
<td>LIST OF FIGURES</td>
<td>vii</td>
</tr>
<tr>
<td>1 INTRODUCTION</td>
<td>1</td>
</tr>
<tr>
<td>1.1 Background &amp; Problem</td>
<td>1</td>
</tr>
<tr>
<td>1.2 Natural Gait</td>
<td>3</td>
</tr>
<tr>
<td>1.2.1 The Gait Cycle</td>
<td>4</td>
</tr>
<tr>
<td>1.2.2 The Ankle - Quantitative Function</td>
<td>6</td>
</tr>
<tr>
<td>1.3 Ankle Prostheses: Current State of the Market</td>
<td>9</td>
</tr>
<tr>
<td>1.3.1 Passive Ankle Prostheses</td>
<td>10</td>
</tr>
<tr>
<td>1.3.2 Pseudo-Passive Ankle Prostheses</td>
<td>12</td>
</tr>
<tr>
<td>1.3.3 Active Ankle Prostheses</td>
<td>13</td>
</tr>
<tr>
<td>1.4 New Design Philosophy</td>
<td>14</td>
</tr>
<tr>
<td>1.5 1st Generation Design</td>
<td>16</td>
</tr>
<tr>
<td>1.6 2nd Generation Design</td>
<td>17</td>
</tr>
<tr>
<td>1.7 EaSY-Walk Design Requirements</td>
<td>19</td>
</tr>
<tr>
<td>1.8 Thesis Overview</td>
<td>19</td>
</tr>
<tr>
<td>2 EASY-WALK FUNCTIONAL DESCRIPTION</td>
<td>21</td>
</tr>
<tr>
<td>2.1 The EaSY-Walk</td>
<td>21</td>
</tr>
<tr>
<td>2.2 Functional Summary</td>
<td>24</td>
</tr>
</tbody>
</table>
IRB PROTOCOL
LIST OF TABLES

1.1 Leading Causes of Amputation ........................................ 2
3.1 Easy-Walk Modeled Function ........................................... 32
5.1 Robot Testing Data, Performance Metrics .......................... 47
C.1 EaSY-Walk Design Parameters ........................................ 64
C.2 EaSY-Walk Design Parameters - By Style ............................ 79
C.3 EaSY-Walk Design Parameters - By Sensitivity ..................... 79
C.4 EaSY-Walk Constant Design Parameters ............................ 80
C.5 EaSY-Walk Constant Design Parameters ............................ 81
# LIST OF FIGURES

1.1 Plantarflexion and Dorsiflexion ........................................... 4
1.2 Ankle Function ................................................................. 7
1.3 Typical Ankle Moment-Angle Relationship .............................. 8
1.4 Ankle Work and Ground Reaction Force ................................. 9
1.5 Example SACH Feet ............................................................. 10
1.6 The Vari-Flex XC Rotate ..................................................... 11
1.7 Example Passive Device Function ......................................... 11
1.8 The Proprio Foot ............................................................... 12
1.9 CESR Foot ........................................................................ 13
1.10 The BiOM ........................................................................ 14
1.11 Work Flow Diagram .......................................................... 15
1.12 Conceptual Design ............................................................. 15
1.13 Generation 1 Prototype ....................................................... 16
1.14 CamWalk Function ............................................................. 17
1.15 Robot Testing Moment-Angle Plots ....................................... 18
2.1 EaSY-Walk Function ........................................................... 22
3.1 EaSY-Walk Model Diagram .................................................. 26
3.2 Device Model Input/Output Diagram ..................................... 27
3.3 Modeled Moment Data ......................................................... 29
3.4 Average Walker Data ........................................................... 30
3.5 MU Subject 1 Data ................................................. 30
3.6 MU Subject 2 Data ................................................ 31
3.7 MU Subject 3 Data ................................................ 31
4.1 Physical Realization of the EaSY-Walk .......................... 34
4.2 EaSY-Walk, Exploded View .................................... 35
4.3 Conceptual Flexure ................................................. 36
4.4 Flexures, Drafted View .......................................... 37
4.5 Translational Jamming Mech., Drafted View ................. 38
4.6 Translational Reset Mechanism, Drafted View ............. 38
4.7 $K_1$ on Lower Ankle (B), Drafted View ..................... 39
4.8 Rotational Jamming Element .................................. 40
4.9 Toothed Block, Drafted View .................................. 40
5.1 EaSY-Walk Critical Positions ................................... 43
5.2 Robot Testing Path ................................................. 44
5.3 EaSY-Walk, Robot Test Data .................................... 45
5.4 EaSY-Walk Moment-Angle Curves ............................. 46
5.5 Moment Vs. Angle, Human Test Subject ..................... 49
B.1 Conceptual Friction Based Jamming Mechanisms ............ 60
B.2 Friction Based Jamming Mechanism, FBD .................. 61
B.3 Cantilever Beam with Point Load at End .................... 63
C.1 Device Model Input/Output Diagram .......................... 65
C.2 Coordinate Frames Attached to Device ....................... 66
C.3  Force Balance on Upper Ankle (C), Upper Leg Body ................. 68
C.4  Flexure Design ......................................................... 69
C.5  Upper Ankle (C) Force Balance, Active Jamming ................. 73
C.6  Flow Chart of EaSY-Walk Device Model .............................. 74
C.7  Flow Chart of EaSY-Walk Device GA Optimization Routine .... 80
C.8  Convergence Plot of the EaSY-Walk Device Optimization .... 81
E.1  Moment Vs. Angle Relationships for 10 Data Sets ............... 83
CHAPTER 1

INTRODUCTION

This document details the design, development, and testing of the EaSY (Early Stance Y Deflection) Walk, a novel two DOF ankle prosthesis that mimics natural ankle behavior over a wide spectrum of walking velocities. Successfully mimicking such behavior so would be potentially life changing for many amputees who have had a decreased quality of life following amputation due to restrictions of currently available prostheses. Within this chapter, Section 1.1 provides background on transtibial amputees and describes the difficulties they face selecting a proper prosthesis. Section 1.2 describes the gait cycle for able bodied walkers. Section 1.3 details the ankle prosthesis market as it stands today. Additionally, Sections 1.5 and 1.6 detail previous designs having similar functions. Finally, Section 1.7 provides design objective of the EaSY Walk and Section 1.8 details an overview of this document.

1.1 Background & Problem

An estimated 623,000 individuals are living with a major lower leg amputation (defined as any lower limb amputation greater than the amputation of toes) in the United States [1]. Reasons for needing to undergo this procedure are many, including cancer treatment, blunt force trauma, and lower leg infection. However, based on data taken for 135 patients of Sunnybrook Health Science Centre and Sunnybrook Centre for Independent Living in Toronto, the most prevalent cause for below the knee (BK) amputation (78%) is peripheral vascular disease (PVD, poor blood circulation in the extremities). With 58.1% of this population acquiring PVD as a side effect of type 1 or 2 diabetes, the percentage of transtibial amputations caused as a byproduct of diabetes is 45% [2]. (See Table 1.1).
Table 1.1: Leading Causes of BK Amputation (Adapted from [2])

<table>
<thead>
<tr>
<th>Cause of amputation</th>
<th>Number of Patients (%)</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unilateral Transtibial</td>
<td>Bilateral Amputees</td>
</tr>
<tr>
<td>PVD w/o diabetes</td>
<td>38 (35)</td>
<td>6 (22)</td>
</tr>
<tr>
<td>PVD w/ diabetes</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Type I</td>
<td>2 (2)</td>
<td>5 (19)</td>
</tr>
<tr>
<td>Type II</td>
<td>41 (38)</td>
<td>13 (48)</td>
</tr>
<tr>
<td>Trauma</td>
<td>23 (21)</td>
<td>3 (11)</td>
</tr>
<tr>
<td>Others</td>
<td>4 (4)</td>
<td>0</td>
</tr>
<tr>
<td>Total</td>
<td>108 (80)</td>
<td>27 (20)</td>
</tr>
</tbody>
</table>

With nearly half of all transtibial amputations being correlated to diabetes, a closer look at the disease is warranted. One striking observation is its correlation to low standing socio-economic status. From one study, it was found that being in a depressed socio-economic standing, defined as making less than $15,000 per year, doubles an individual’s risk of acquiring type 2 diabetes when compared to those living in secure economic standing, defined as making greater than $80,000 [3]. A second independent study found that individuals who had lived in an impoverished condition for any period of time in the past twelve years would have a 41% greater chance of being diagnosed with type 2 diabetes compared to their peers [4].

This data suggests there are a large number of ankle prosthesis recipients living in a poverty-stricken situation. Prostheses that are affordable to these individuals are inferior in matching natural ankle behavior. The inferiority of said prostheses may cause many quality of life detriments for the amputee including asymmetrical gait (for unilateral amputees), slower self selected walking speeds, higher metabolic cost per distance traveled and increased pain in the residual limb [5, 7]. That leaves the following challenge for research done in the field: design a prosthetic ankle that incorporates function generally reserved for high end products at a low cost. Design of such a product, allowing
a subject to feel more comfortable with their ankle prosthesis, could lead to large improvements in regards to quality of life for the recipient [8, 9].

1.2 Natural Gait

While walking may seem simplistic to those with two healthy lower limbs, this does not imply that the act of ambulation is without complexity. When walking, individuals subconsciously use many different control algorithms and compensatory mechanisms in order to walk efficiently. Even after decades of research, many of the control patterns applied during ambulation remain superficially understood. Further adding to the complexity of the problem, gait patterns vary significantly for each individual. However, one identical aspect is minimization of the metabolic cost of walking [10].

To gain an understanding of the importance of the novel ankle prosthesis presented here, one must have a basic understanding of the gait cycle and its associated nomenclature and terminology. Much of the terminology used in the gait analysis community, from [11, 12, 13] is as follows:

**Gait Cycle:** The gait cycle starts when one foot makes contact with the ground and ends when the same foot contacts the ground again. The cycle can be broken down into various phases and periods to determine normal and pathological gait.

**Sagittal Plane:** A sagittal plane is any plane which divides the body into right and left portions [11]. For the ankle in normal gait, the bulk of motion and loading occurs in this plane.

**Dorsiflexion:** Dorsiflexion is defined as the rotation of the top of the foot towards the shin and can also be defined by the orientation of the foot relative to the leg (See Fig. 1.1) where the angle between the two is less than 90°.
Figure 1.1: Diagram of plantarflexion and dorsiflexion; 0 is attached to the foot body and 0’ and 0” are attached to the leg body

**Plantarflexion**: Plantarflexion is rotation of the top of the foot away from the shin and can also be defined by the orientation of the foot relative to the leg (See Fig. 1.1) where the angle between the two is greater than 90°.

**Inversion/Eversion** Rotation about the ankle joint that occurs in the frontal plane (plane that divides the body into front and back), with inversion being an inward rotation, and eversion being an outward rotation.

**Stance/Swing Phase**: The stance phase is the portion of one gait cycle in which the foot is in contact with the ground. Conversely, the swing phase is the period in which the foot is not in contact with the ground.

**Single/Double Support**: Describes how many of an individual’s feet are in contact with the ground at an instance in time.

1.2.1 The Gait Cycle

While the gait cycle may at first appear dauntingly complex, subdividing it into sections based on various metrics/measures can make it far more digestible. Several of
Function of the ankle can be subdivided based upon ankle rotation. Doing so yields three distinct phases during the stance phase, as described below.

Controlled plantarflexion (CP), the first phase of ankle function, occurs between heel strike and foot flat ((1) and (2) of Figs. 1.2 & 1.3). The ankle begins at an angularly neutral position and plantarflexes to maximum plantarflexion at foot flat. This phase is also called weight acceptance, as the foot acclimates to the weight of the individual being applied to it and shock absorption occurs due to heel tissue and shoe material.

At foot flat, the controlled dorsiflexion (CD) phase begins. With the foot remaining flat on the ground past foot flat, the ankle now acts as a pivot point for the leg to rotate clockwise about. This phase continues until heel off ((4) of Figs. 1.2 & 1.3) which

<table>
<thead>
<tr>
<th>Table 1.2: The Stance Profile, Subdivided by Various Metrics</th>
</tr>
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<tbody>
<tr>
<td><strong>Phases</strong></td>
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<tr>
<td>-------------------</td>
</tr>
<tr>
<td>Stance Phase:</td>
</tr>
<tr>
<td>Leg in question is in contact with the ground</td>
</tr>
<tr>
<td>Double Support:</td>
</tr>
<tr>
<td>both feet in contact with the ground</td>
</tr>
<tr>
<td>Single Support:</td>
</tr>
<tr>
<td>one foot in contact with the ground</td>
</tr>
<tr>
<td>Single Support:</td>
</tr>
<tr>
<td>one foot in contact with the ground</td>
</tr>
<tr>
<td>Double Support:</td>
</tr>
<tr>
<td>see above</td>
</tr>
<tr>
<td>Swing Phase:</td>
</tr>
<tr>
<td>Leg in question is removed from the ground</td>
</tr>
<tr>
<td>Single Support:</td>
</tr>
<tr>
<td>see above</td>
</tr>
<tr>
<td>Feet Adjacent</td>
</tr>
<tr>
<td>Tibia Vertical</td>
</tr>
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</table>
typically occurs at approximately the same point as maximum dorsiflexion.

After maximum dorsiflexion occurs, the ankle enters the powered plantarflexion (PP) phase. This phase is named due to the musculature of the intact ankle providing a moment that propels the ankle into the plantarflexed position. Due to the increased moment, ankle power reaches a maximum and the ankle performs work on the body at average to fast walking speeds. The PP phase ends at toe off (5) of Figs. 1.2 & 1.3), where the swing phase begins.

In the table above, the timing of each of these subdivisions (stance/swing, foot contact, gait events, gait phases, gait phase at the ankle) is compared to one another.

1.2.2 The Ankle - Quantitative Function

Ankle function can also be described quantitatively through kinematic and kinetic profiles of joint function as shown in Fig. 1.3 from [14]. Here, profiles of force along the leg, angle of the ankle, and moment about the ankle joint are presented as a function of stance percent.

Specific events described in Section 1.2.1 are indicated in Figs. 1.3 and 1.4. These events include heel strike (1), foot flat (2), maximum leg force (2.1), maximum dorsiflexion/heel off (3), powered push-off (4), and toe off (5). Qualitative function described previously can be seen in these plots. The plots show ankle angle decreasing through the CP phase (reaching a minimum of $-7.8^\circ$), increasing through the CD phase (maximum of $6.9^\circ$) before decreasing once again during the PP phase (minimum of $-20.1^\circ$). The data also shows, for an average subject weighing 56.7 kg, force along the leg reaching a maximum of 596 N (10.51 N/kg, normalized per mass of subject) during CP and ankle moment reaches a peak of 89.6 Nm (1.58 Nm/kg, normalized per mass of subject) during the PP phase.
Another approach to graphically describe ankle function is plotting ankle moment against ankle angle, shown in Fig. 1.3. One functionality displayed is rotational work output. From the definition of rotational work:

$$W(\theta) = \int M(\theta) \cdot d\theta$$  \hspace{1cm} (1.1)

By the definition, integrating to find the area inside of the curve shown in Fig. 1.3 yields work, with a counterclockwise loop describing the ankle doing work on the body and actively assisting forward propulsion. Conversely, a clockwise loop describes the body doing work on the ankle, or the ankle dissipating work. From [14], the ankle performs 18.47 J of work on the body per stance cycle. Or, normalizing based upon the subject's mass, 0.326 J/kg of work.

Fig. 1.3 also displays ankle stiffness, where ankle stiffness is defined as:
Figure 1.3: Moment-Angle Relationship for Able Bodied Subjects in the Sagittal Plane from [14]

\[ k_{\text{ankle}} = \frac{dM}{d\theta} \]  

This manifests itself graphically as the slope of the curve. From the curve, ankle stiffness exhibits a large amount of nonlinearity in both plantarflexion and dorsiflexion and, in both cases, stiffens with increasing deflection. The figure also shows the different magnitudes of stiffness in the plantarflexion region (more compliant) compared to the dorsiflexion region (less compliant).

Modification of gait at varying walking velocities should be considered because designing for only a self selected walking velocity would lead to potentially sub-optimal results at other speeds. Experiments conducted by [15] as well as [16] both showed variation in ankle function based upon walking velocity.

One such variation manifested itself in the rotational work output of the ankle. Both articles showed that, as shown in Fig. 1.4, ankle work increases linearly with walking velocity. While each experiment resulted in a marginally different slope for the relationship \((0.27 \text{ vs. } 0.23 \frac{J/kg}{m/s})\) as well as crossover velocity at which the ankle acts elastically (0.84
Figure 1.4: Ankle Work and Max Ground Reaction Force as a Function of Gait Speed, from [16]

vs. 1.26 \( \text{m/s} \), [15] postulated that these differences could be attributed to differing methods of calculating ankle work for each experiment. Data from [16] also showed an approximately linear relationship between normalized maximum force along the leg and gait speed (shown in Fig. 1.4), with leg force increasing as a function of gait speed.

1.3 Ankle Prostheses: Current State of the Market

Current options for ankle prostheses include conventional mechanically passive devices and active (bionic) devices. Below is a description of various ankle prosthetics on the market today.
1.3.1 Passive Ankle Prostheses

While utilizing many different design strategies, passive devices share two common traits:

1. Exhibit elastic function; Passive devices lack an active element to provide rotational work about the ankle joint, but can provide non-linear stiffness.
2. Low cost; Lacking the expense of actuators, batteries, etc., these devices are typically relatively inexpensive.

One such passive design strategy is the SACH (Solid Ankle Cushioned Heel) foot. The bulk of compliance in the device is achieved through an adjustable, soft heel insert that deforms upon heel strike. A belting mechanism is used to achieve rotational compliance, but the belting is often designed with a high stiffness that is identical in plantarflexion and dorsiflexion. This high stiffness leads to a significantly reduced range of motion for the end user. One such example of a SACH foot is the Seattle SACH by Trulife, illustrated in Fig. 1.5.

A second design strategy, the ESAR (Energy Storage and Return) foot, features a more natural rotational compliance about the ankle joint when compared to a SACH foot [20]. Rotational compliance is generally achieved through the use of a flexure, oftentimes made of a carbon fiber composite, that is allowed to bend with increased loading. This is
the case in the Vari-Flex XC Rotate, pictured in Fig. 1.6. The Vari-Flex XC Rotate also utilizes a compliant pylon at the top of the device to provide shock absorption in the vertical direction as well as rotational compliance about the leg axis to provide comfort for the end user.

None of these devices provide any rotational work about the ankle joint. Due to internal damping in stiffness elements, the Moment-Angle curve for the average passive device looks similar to Fig. 1.7. The clockwise nature of the curve suggests passive devices dissipate rotational energy about the ankle joint, with the curve in Fig. 1.7 dissipating 7.89 J of work, or 0.124 J/kg when normalized by mass of the subject.
1.3.2 Pseudo-Passive Ankle Prostheses

A pseudo-passive ankle prosthesis does not use an actuator to generate work about the ankle, but does use some form of electromechanical system to achieve a more accurate representation of the natural ankle. Some devices in this category are already on the market while some are still being developed. One such ankle already on the market is the Proprio Foot by Össur, shown in Fig. 1.8.

The Proprio Foot uses a DC motor based actuator to control an individual’s ankle angle during the swing phase and is capable of dorsiflexing in this phase, thus increasing ground clearance of the foot [23]. Using an on-board accelerometer, it is also able to determine whether an individual is walking on flat ground, uphill, or downhill and adjust ankle angle accordingly. This adjustment is done with the goal of reducing the frequency of toe scrape related falls in transtibial amputees. However, the device does not provide any work about the ankle to propel the user forward.

An example of a pseudo-passive ankle prosthesis that is not on the market yet is the CESR (Controlled Energy Storage and Return) Foot (See Fig. 1.9). The CESR uses rotational work wasted during the CP phase of gait and converts it into useful work during powered push-off [24]. This is accomplished through the use of an actively controlled one
way clutch for determination of lock/unlock timing. Due to this activation, the device does not qualify as purely passive.

1.3.3 Active Ankle Prostheses

An active device uses electrical components (actuator, batteries, etc.) to provide positive rotational work about the ankle joint during powered push off. This behavior is ideal for the bulk of patients to regain near normal gait. However, the cost of such devices is typically very high. In addition, the active elements in these devices require a power source that must be replenished/recharged relatively frequently, a task that may be found as a nuisance to the end user.

To date, one of the more successful active prostheses on the market is the BiOM (shown in Fig. 1.10), produced by BionX Medical Technologies and originally researched DESigned at MIT. The BiOM uses a DC motor actuator in series with a spring element (SEA, series elastic actuator) placed about the ankle joint to achieve high ankle torques while commanding far smaller torques from the actuator. This style of design, now popular in active ankle prostheses, allows for far more efficient use of the actuator and also allows for a smaller actuator to be employed, decreasing device weight. Device cost is

Figure 1.9: The CESR (Controlled Energy Storage and Return) Foot [24]
Figure 1.10: Example of an Active Ankle Prosthesis, The BiOM [25]

still an issue for the BiOM. In 2013, two years after the product’s official launch, the
device cost approximately $50,000 and had only sold about 1,000 units [25].

Another active ankle prosthesis which has achieved success in the laboratory is the
Project SPARKy, developed at ASU [26]. The Project Sparky, functioning utilizing a DC
motor actuator in series with a spring element similar to the BiOM, differs from the BiOM
in its use of a lead-screw mechanism in conjunction with the DC motor actuator to
achieve rotational work about the ankle joint. The Project SPARKY was commercialized
by SpringActive as the Odyssey [6], again placing an actuator in series with a spring
element to achieve natural ankle behavior. Further, the Odyssey utilizes a microprocessor
which analyzes data from a gyro and motor encoder 1,000 times per second to ensure
robust function.

1.4 New Design Philosophy

The limitations of currently available ankle prostheses motivated work on a new
solution, a passive ankle device that mimics several key aspects of a natural ankle joint.
More specifically, the device will look to mimic nonlinear rotational stiffness about the
ankle joint and rotational work output (powered push-off about the ankle joint) that
increases with walking velocity, all while remaining relatively inexpensive. The overarching design philosophy is as follows: Utilizing translation along the leg and rotation about the ankle joint, compliantly coupled to one another, store translational work from along the leg and remap the work as rotational work about the ankle to achieve more natural ankle characteristics including active behavior from a passive device [27].

![Figure 1.11: Work Flow Diagram for the EaSY Walk Device](image)

An energy flow diagram representing this design philosophy is shown in Fig 1.11 and can be conceptually achieved through the device in Fig. 1.12. In this diagram, Body A represents the foot, Body B the lower ankle, and Body C the upper ankle. Between Body B and C exists the translational degree of freedom (DOF), denoted as $r$. Rotation is allowed between Body A and B and is denoted as $\theta$. Several springs also act between these three bodies. For example, $k_a$ provides a rotational stiffness about the ankle joint, $k_s$ a translational stiffness along the leg for shock absorption upon heel strike and $k_c$ is a

![Figure 1.12: Diagram of the Two Coupled Degrees of Freedom Conceptual Design](image)
stiffness of the energy storage element that couples the two DOF in the system.

1.5 1st Generation Design

The first generation design was meant as a proof of concept for the ideas put forth in Section 1.4. This includes the ability to harness and store translational work from along the leg and remap it as rotational work about the ankle joint in addition to achieving natural stiffness about the ankle joint using a network of translational compression springs.

This concept manifested itself as shown in Fig. 1.13. The energy stored in deflection along the leg was expected to be aiding in powered push-off. Testing of the mechanical realization of the device indicated that this did not occur. As spring $K_2$ always acted about the ankle joint (e.g., always acted between Body A/B and Body C), translational energy was released as rotational energy simultaneous to its storage [28].
1.6 2nd Generation Design

To ensure that translational energy captured is not released as rotational energy until powered push-off (e.g., after maximum dorsiflexion), the 2nd generation design (CamWalk) implemented a cam based timing mechanism to release stored energy as rotational energy used to propel the ankle into plantarflexion at a designer specified value of dorsiflexion angle [21].

Figure 1.14: CamWalk Function Throughout the Stance Cycle [21]
As shown in Fig. 1.14, the design utilized four springs to achieve the desired functionality. $K_1$, the shock absorption spring, is used to provide vertical compliance in the device at heel strike. This spring acts between Bodies C and F. $K_2$ compliantly couples the translational and rotational degrees of freedom in the device and acts between Bodies A and D. $K_3$, the nominal plantarflexion stiffness spring, provides a counter-clockwise moment about the ankle joint when the device is in plantarflexion. Last, $K_4$, the nominal dorsiflexion spring, provides a clockwise moment about the ankle joint when the device is in dorsiflexion. Both $K_3$ and $K_4$ act between Bodies A and B. See Figure 1.15 for conceptual function of the device through the stance cycle.

Through both robot testing (Fig. 1.15, (a)) and human subject testing (Fig. 1.15, (b)), it was found that the device was capable of yielding positive rotational work output about the ankle joint. Limitations of the device were also discovered during human subject testing. Translational energy storage occurred mid-stance and the resulting mid-stance deflection (occurring at approximately 60% of stance) was found to be somewhat unexpected and uncomfortable by some test subjects. Second, the CamWalk weighed a total of 4.9 lbs. and was found to be too heavy by some test subjects.

Finally, consistent device function was highly dependent upon the maximum dorsiflexion angle seen on a given stride as the cam device was not self selecting. A
subject walking with maximum dorsiflexion greater than the selected cam position resulted in decreased rotational work output from the device. Opposing that, a subject walking with maximum dorsiflexion less than the selected cam position would not experience the coupling of translation and rotation and therefore would see no rotational work benefit.

1.7 EaSY-Walk Design Requirements

During development, three design requirements were identified. The first of was the requirement for early stance leg deflection. Under normal operation of the EaSY-Walk prosthesis, deflection will occur near heel strike to make the leg deflection that is required of the device acceptable/comfortable to the user.

The second design requirement was that the EaSY-Walk will weight approximately the same as/less than the weight of a natural foot/ankle complex, and will fit within a design envelope small enough that the device fits within a US men’s size 10 shoe and beneath a standard pant leg.

The third design requirement was that energy return increases with increasing walking velocity. Research by [15] as well as [16] presented in Section 1.2.3 showed that the normal human ankle generates an increased amount of rotational work about the ankle joint at walking speeds greater than self selected. As such, to meet this design requirement, the EaSY-Walk will be required to output an increasing amount of rotational work about the ankle joint as walking speed increases. All design objectives and metrics considered are summarized in Appendix A.

1.8 Thesis Overview

This thesis presents the design and construction of the EaSY-Walk (Early Stance Y-Deflection), a two DOF passive ankle prostheses. Chapter 2 describes the conceptual design of the EaSY-Walk. Chapter 3 discusses the modeling and optimization process
used in the design process. Chapter 4 presents the mechanical realization of the optimized conceptual design. Finally, Chapters 5 and 6 describe experiments and the results of testing completed on the device as well as conclusions drawn based upon these results.
CHAPTER 2

EASY-WALK FUNCTIONAL DESCRIPTION

A conceptual design capable of meeting and/or exceeding the design requirements detailed previously in Chapter 1 was needed. This led to the conceptual development of the EaSY-Walk’s passive components. Within this chapter, a comprehensive description of the design, described in general in addition to throughout the stance cycle, is provided in Section 2.1 and a conclusion of the work completed is detailed in Section 2.2.

Two major design considerations were implemented with the EaSY-Walk. The first of these was the use of timing/sequencing methods for both translational energy storage and its conversion to rotational work designed to store translational energy at heel strike and convert it to rotational work at maximum dorsiflexion. These sequencing methods were also designed to allow for variable translational energy storage, with the amount of energy stored a function of maximum leg force and utilized mechanical diode mechanisms [29, 30]. Second, design of the nominal stiffness mechanisms was refined to minimize the design envelope of the device and provide greater stiffness non-linearity.

2.1 The EaSY-Walk

As shown in Fig. 2.1a, the device is made up of three bodies: foot (A), lower ankle (B), and upper ankle (C). Sagittal plane rotation is allowed between foot (A) and lower ankle (B) and translation along the leg axis is allowed between lower ankle (B) and upper ankle (C). Additional bodies are all related to the mechanical diode mechanisms, with pyramid adapter (D), translational jamming mechanism (E), rotational jamming mechanism (F), and reset mechanism (G) implemented into the design. See Appendix B for additional details regarding selection of the mechanical diode mechanisms.
Between these bodies exists four discrete springs, two of which are traditional translational springs and two of which are flexure based. See Chapter 4 for additional details regarding the flexure based stiffness mechanisms. The coupling spring $K_1$, is located between upper ankle (C) and rotational jamming mechanism (F), where rotational jamming mechanism (F) remains static with either the cantilevered hook on lower ankle (B) ($K_1$ acting along the translational DOF) or along the channel in foot (A) ($K_1$ acting...
along the rotational DOF). The translational jamming mechanism reset spring $K_2$ acts between the lower ankle (B) and reset mechanism (G) and resets the mechanism during the swing phase.

The plantarflexion and dorsiflexion stiffness flexures, $K_3$ and $K_4$ respectively, each act between foot (A) and lower ankle (B), with $K_3$ providing a clockwise moment about the ankle joint when the device is plantarflexed and $K_4$ providing a counter-clockwise moment about the ankle joint when the device is dorsiflexed.

Looking at the function of the device through the gait cycle, heel strike is shown in Fig. 2.1a. At heel strike, force along the leg begins to increase due to the walker’s gravitational load on the ankle. This increase in leg force causes a deflection in $K_1$ and $K_2$ and energy is stored from deflection along the leg. Leg force, and thereby deflection between lower ankle (B) and upper ankle (C) as well as in $K_1$, continues to increase past foot flat (approximately maximum plantarflexion and deflection of $K_3$ during stance) until maximum leg force is achieved (Fig. 2.1b).

At maximum leg force, shown in Fig. 2.1b, there is no instantaneous change in force. The sequencing of this point can be seen in Fig. 1.2. Past this point, force along the leg begins to decrease. Unimpeded, the deflection between lower ankle (B) and upper ankle (C) as well as in the coupling spring would begin to decrease, thereby decreasing the amount of stored energy in the spring. To avoid this outcome, a jamming element of the translational jamming mechanism (E) "wedges" between lower ankle (B) and upper ankle (C), thus constraining relative motion between the two. This locking ensures that the greatest amount of translational energy available is stored in $K_1$.

In addition to leg force decreasing after Fig. 2.1b, the ankle begins to dorsiflex which causes $K_4$ to provide a counter-clockwise moment about the ankle joint. This continues up to Fig. 2.1c and Fig. 2.1d, where the ankle reaches maximum dorsiflexion. Looking first at Fig. 2.1c, a roller of the rotational jamming mechanism (F) is still at rest on the lower ankle (B) and the coupling spring ($K_1$) is still acting along the translational
DOF. In Fig. 2.1d, the moment of decreasing dorsiflexion, a roller of the rotational jamming mechanism (F) rolls off of lower ankle (B) and comes into contact with the curved surface on the foot (A).

With a jamming element of the translational jamming mechanism (E) continuing to restrict motion between the lower ankle (B) and the upper ankle (C), $K_1$ acts between the foot (A) and the upper ankle (C), providing a moment about the ankle joint. Through the rest of the stance phase, part of which is shown in Fig. 2.1e as powered push-off, the coupling spring ($K_1$) continues to release the energy previously stored in deflection along the leg as a moment about the ankle.

As shown in Fig. 2.1f, the device enters the swing phase and leg force at the ankle reverses directions. The pyramid adapter (D) translates relative to the upper ankle (C) and, through a four bar linkage, releases the constraint previously applied to reset mechanism (G) and $K_2$. Now free to translate, $K_2$ releases energy previously stored, rotating a jamming element of the translational jamming mechanism (F) out of contact and resetting the translational DOF. With $K_1$ being constrained at its free length, the rotational jamming mechanism (F) is kinematically forced to traverse up the channel in the foot body (A) and back to the hook included in the lower ankle (B). The device is now fully reset and ready for the user's next stride.

2.2 Functional Summary

Based on the conceptual design chosen, the device is designed to satisfy the design requirements described in Section 1.7. Through the use of jamming mechanisms, the design of the device will capture translational work available at various gait speeds and will result in the bulk of deflection along the leg occurring early in stance. In addition, flexure based elastic components will lead to a non-linear stiffness similar to that of the natural ankle and allow the device to be fit in a compact envelope. Optimization of the
design is detailed in Chapter 3.
This chapter describes the modeling and optimization process completed for the EaSY-Walk. In Section 3.1, the model structure is described. Section 3.2 illustrates the optimization process used and the resulting optimal functionality of the device. Section 3.3 concludes the chapter.

3.1 Modeling Overview

A quasi-static model was employed for computer simulation of the device. In a quasi-static model, dynamic elements are assumed to be negligible relative to static elements. In the analysis of the EaSY-Walk, the potential energy stored in the springs and flexures is assumed to be significantly larger than kinetic energy stored in inertial elements (velocity of the foot, leg, etc.).

3.1.1 Model Inputs and Outputs

The EaSY-Walk has two (2) primary degrees of freedom, allowing deflection along the leg and rotation about the ankle joint. Thus, to calculate system behavior, two inputs (independent variables) are needed to find two outputs (dependent variables). Appendix C provides additional details regarding the modeling techniques used.

Figure 3.1: Diagram Displaying the Inputs and Outputs of the EaSY-Walk Model
Based on the functionality of the device and prior knowledge of gait, force along the leg, $F$, is the independent variable along the translational DOF and ankle angle, $\theta$, is the independent variable about the rotational DOF. These were selected as independent variables as data shows $F$ is independent of device function and that a walker will kinematically enforce $\theta$ based on their hip and knee angles. With $F$ and $\theta$ as independent variables, deflection along the leg, $r$, and moment about the ankle joint, $M$, are the two dependent variables in the simulation.

### 3.2 Optimization Process

To determine an optimal design (optimal spring end locations, optimal spring rates, etc.) for the EaSY-Walk device, an iterative optimization algorithm was utilized. The optimal design will be for a given set of design parameters which consists of the locations of spring ends as well as the rates for each of the springs in the system. These parameters can be further defined by their ability to be adjusted from subject to subject.
and the magnitude by which they effect the fitness function (further detailed in Table C.2 and C.3, respectively). The spring rate for each of the flexures will be based on the length, width and height of the respective flexure. Figure 3.2, shown above, graphically displays each of these parameters on the device (note: coordinate frame O is attached to the foot body and all parameters discussed are relative to this coordinate frame). A summary description of each of these parameters, $\vec{P}$, can be found in Appendix C.

Additionally, optimality is defined by a fitness for objectives function, which describes desirable aspects of the design numerically. For the EaSY Walk, the performance metrics considered in the objective function include work output, stress on the dorsiflexion flexure, stress on the platarflexion flexure, stiffness match in the early loading phase, stiffness match in the late loading phase, stiffness match in the unloading phase, and maximum ankle moment. Each of these values was normalized, multiplied by a weighting factor selected based on a previously selected importance of each design criteria, and summed in Eq. 3.1 shown below. More details regarding the fitness function can be found in Appendix C.

$$\text{Qual} = 0.45Q_{work} + 0.125Q_{\sigma,d} + 0.025Q_{\sigma,p} + 0.15Q_{EL} + 0.1Q_{LL} + 0.05Q_{UL} + 0.1Q_{M_{max}}$$  \hspace{1cm} (3.1)

### 3.2.1 Optimized Results - Average Walker

Design parameters, along with subject data including $F(t)$ and $\theta(t)$ from [14] were reinserted to the quasi-static device model to assess performance for the optimal parameter set, $\vec{P}_{opt}$. Using the model, the following plot comparing the ankle torque curves of the reference ankle from [14] to device performance was generated.

Moment vs. Angle curves, in addition to a Deflection vs. Stance Percent plot are shown in Fig. 3.4. Both the modeled and reference curves move counter-clockwise as a
function of time, showing positive work output for both the reference ankle as well as the modeled prosthesis. With the reference curve yielding 18.74 J of rotational work and the device yielding 4.71 J of work, the device generates 25.16% of the reference work. This represents a significant improvement over a typical passive prosthesis which would be expected to dissipate rotational work about the ankle joint. In addition, the Deflection Vs. Stance Percent plot shows the bulk of deflection (14.59 mm) occurring early in stance. Only 0.28 mm of deflection occurs in mid or late stance, proving that the device functions based upon an early stance deflection technique.

### 3.2.2 Optimized Results - Specific Walkers

As several device parameters can be varied from subject to subject (see Appendix C for additional details), the device was also optimized for three specific walkers, with 4 strides of data for Subject 1, 4 strides for Subject 2, and 3 strides for Subject 3. The utility of optimizing for additional walkers is two fold. First, it shows that the device can closely match natural ankle data for a variety of different walkers through the use of adjustable parameters. It also shows the robustness of the device over a large sample size of strides, now totaling 12 including [14]. Moment Vs. Angle data, divided into plots by subject, is
Figure 3.4: Comparison of Simulated Ankle Behavior to that for the Average Walker [14] shown in Figs. 3.4-3.6. Modeled ankle performance is in blue and reference data in red, with the line style varying for each stride.

Figure 3.5: Four Strides for MU Test Subject 1
The non-linear stiffness exhibited by the device and the rotational work output from the device can be seen in the Moment Vs. Angle plots. To get a more quantitative look at the device’s rotational work output, the work output for each trial and the corresponding percentage of reference work output are summarized in Table 3.1. Over the 11 trials, the device averaged 61.6% of the work output of the reference curves and, for Subject 3, Stride 1, exceeded the performance of the natural ankle, yielding 121.1% of the
work output of the reference curve.

Table 3.1: EaSY-Walk Modeled Functionality for 12 Data Sets

<table>
<thead>
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<th>Subject</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
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<th>2</th>
<th>3</th>
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<td></td>
<td></td>
<td></td>
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<td>Work Per. [%]</td>
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<td>60.3</td>
<td>92.3</td>
<td>46.7</td>
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<td>30.5</td>
<td>28.7</td>
<td>32.7</td>
<td>121.1</td>
<td>86.0</td>
<td>79.7</td>
</tr>
</tbody>
</table>

3.3 Design Optimization Summary

As modeled, the EaSY-Walk is capable of exhibiting non-linear rotational stiffness in addition to positive rotational work output about the ankle joint. Overall, the device averaged 58.5% of the reference work output of a natural ankle joint over the 12 data sets analyzed. Appendix C details the design parameters utilized to achieve such performance. The optimal set of parameters for the device are used in the physical realization of the design.
CHAPTER 4

DETAIL DESIGN DESCRIPTION

In this chapter, the ankle prosthesis design functions are realized physically. NX10.0 was used for 3D CAD modeling of device components and assembly. Section 4.1 describes the bulk overview of the device design. Additionally, specific device functions are described in Section 4.2 and fabrication of the device is described in Section 4.3

4.1 Bulk Device Overview

The physical prototype was designed to not only have the performance characteristics previously outlined but also to be usable, comfortable, and safe for a wide array of human test subjects. To accomplish these goals, the device had to be strong (Factor of Safety of 2 for a 250 lb. individual), compact (fit within a typical shoe), and lightweight. Manufacturing cost was also a consideration. The final device weighs 4.15 lbs., fits within a US Men’s Size 10 shoe, has a total height of 8.86 in., and is safe for use by a 250 lb. individual.

The bulk of the device was made out of Aluminum 7075 (aircraft grade aluminum), which was relatively low cost with a high strength to density ratio of 752,500 in. For comparison, 1045 Carbon Steel has a strength to density ratio of 320,400 in. In addition to Aluminum 7075, other materials used include a unidirectional carbon fiber/epoxy composite (for the flexure devices, which required superior strength and stiffness), ABS plastic (for the foot shell, where minimizing weight was essential) and Hardened Steel 4140 (for the jamming mechanisms, where robust surface properties were required). The EaSY-Walk prototype, shown in both modeled and physical form, is introduced in Fig. 4.1 and Fig. 4.2 shows an exploded view of the device.
These figures display many other design details not yet discussed. First, based on the slot geometry in the foot body, the device has a Range of Motion (ROM) of 20 degrees in plantarflexion and dorsiflexion. This ROM, much larger than what is seen in many typical passive prostheses, was found to be important to the comfort of several subjects who walked with the CamWalk device [21].

To ensure that only one DOF is allowed along the translational DOF, a ball spline shaft and nut were utilized. A spline shaft contains parallel grooves that are constrained to several tracks in the nut to ensure that no rotation is allowed about the axis of the shaft,
crucial to the functionality of the device. Between the foot body and the plastic foot insert (not shown in Fig. 4.2), 5 degrees of rotation are allowed in both inversion and eversion, implemented to mimic the roll allowed by the natural ankle about this axis. Two translational springs are placed 0.69 in. from this axis to provide inversion/eversion stiffness.
4.2 Detail Design

Several of the mechanisms employed by the EaSY-Walk, including the flexure and jamming mechanisms, are described in more detail below.

4.2.1 Flexure Mechanisms

Non-linear rotational stiffness is achieved using flexures (i.e., cantilever beams with a moving point load (as shown in Fig. 4.3)).

In the design, a plantarflexion and dorsiflexion flexure are fixed to the foot body. On the lower ankle, two sets of contactors (roller bearings) rotate with the body and deflect their respective flexure, creating a torque about the ankle. As the contactor is rotating about the ankle joint, it will translate both perpendicular to the beam (causing beam deflection) as well as along the axis of the beam (causing the effective length of the beam to shorten). The flexure support and its geometry were optimized to achieve a more natural stiffness non-linearity about the ankle joint in a small volume.

Due to the amount of deflection expected in each flexure mechanism and the loading required to match the stiffness of the natural ankle, a carbon fiber composite was chosen for the flexures due to its high strength (300,00 psi) and high elastic modulus (14 Msi). Additionally, ball bearings were selected as contactors for both $K_3$ and $K_4$ to reduce
wear. The final design of each flexure mechanism is broken out in Fig. 4.4.

4.2.2 Translational Jamming & Reset Mechanism

The translational jamming mechanism ensures that translational energy stored is released as rotational energy. To accomplish this within a small space, flats were machined onto the spline shaft and these flats become one of the two contact surfaces for the jamming element. The other surface was a small plate affixed to Body B. All bodies in this mechanism were made out of hardened steel to handle the large contact stresses present. These features can be seen in Fig. 4.5.

To reset this mechanism, a small amount of rotational energy must be applied to each jamming element (sprag) to remove it from contact. Desirably, reset of the jamming elements would occur at the start of the swing phase (leg force reverses directions). Reset is accomplished with the device in Fig. 4.6. This mechanism transforms vertical motion at the pyramid adapter into a horizontal motion that constrains the upper end of $K_2$ in stance but allows it to freely translate in the swing phase. With $K_2$ free, its energy is transferred
Figure 4.5: Translational Jamming Mechanism Acting Between the Lower Ankle (B) and Upper Ankle (C)

Figure 4.6: Translational Reset Mechanism. Stance Phase Operation is at Left and Swing Phase at Right

to the translational jamming element, resetting the device for the next stride.
4.2.3 Rotational Jamming & Reset Mechanism

The rotational jamming mechanisms transfers one end of the coupling spring from Body B to Body A when maximum dorsiflexion occurs. In the unlocked mode of operation, a bearing at the lower end of $K_1$ "rides" on a hook at the end of a cantilever arm on Body B. The bearing "riding" on the hook is displayed in Fig. 4.7. When on the hook, all loading in the coupling spring is constrained to act between Body B and C.

![Figure 4.7: Rotational Jamming Mechanisms Prior to Engagement (not providing a moment about the ankle)](image)

During increasing dorsiflexion, the jamming element sits between a curved surface on Body A and a cam surface attached to the lower end of $K_1$, as shown in Fig. 4.8, but does not apply a significant force on either. Once the ankle begins plantarflexing, the jamming roller wedges between these two bodies. Now, the lower end of $K_1$ is static relative to Body A as opposed to Body B.

With the lower end of $K_1$ static relative to Body A, the bearing rolls off of the hook and the toothed block, placed coaxially with the bearing, drops onto the toothed insert on Body A. This functionality is displayed in Fig. 4.9. The toothed nature of these two components ensures that the coupling spring does not translate to the end of the channel, which would allow a significant amount of energy stored in the coupling spring to be expended as non-useful work.
Reset of this mechanism occurs due to the coupling spring being constrained at its free length. After reset of the translational mechanism occurs, the toothed block is kinematically constrained to travel along the top of the slot shown in Figures 4.7-4.9 until the bearing re-seats itself in the hook on Body B.

4.3 Fabrication

Components machined for the EaSY-Walk Prototype were fabricated at ProtoLabs, Inc., Multi Tool LLC and Clearwater Composites (flexures only). Final cost of the device,
including purchased components, was approximately $9,406 (see Bill of Materials located in Appendix D). Device testing of the developed prototype is described in Chapter 5.
CHAPTER 5

DEVICE TESTING AND RESULTS

To confirm that the device functions as expected and also to determine if this functionality is subjectively satisfying to the end user, the prosthesis was tested with a robotic test setup and with several human test subjects. Section 5.1 describes the procedure used to complete robot testing and present details the quantitative results of tests. Section 5.2 describes human subject testing procedures and test results.

5.1 Robot Testing

To confirm validity of simulated results obtained from the quasi-static EaSY-Walk model, and to check for functional robustness of the device in a controlled setting, robot testing was completed.

5.1.1 Procedure

Robot testing was completed using a Staubli 6R robot arm platform. Spring stiffnesses and loading on the device were scaled to approximately 1/10 of what is expected of an average walker (56.7 kg.).

The pyramid adapter of the prosthesis was mounted to an ATI 6-axis force/torque transducer which was attached to the robot end effector. Force along the leg $F(t)$ was measured and moment about the ankle joint $M(t)$ was calculated using measured moment at the sensor and the distance between the sensor and the ankle joint.

Potentiometers were used to acquire kinematic data. A rotary potentiometer, placed between the foot (A) and the lower ankle (B), measured ankle angle, and a translational potentiometer, between lower ankle (B) and upper ankle (C), measured leg deflection. Both were wired into a National Instruments Elvis board and voltage data from
each was collected. Calibration curves were generated for each potentiometer to get deflection from voltage. To correct for the discrepancy in collection frequency, a Matlab script was used to efficiently synchronize each set of data.

Device kinematics were used to determine the path of the end effector for specified device function. For this analysis, the location of four points was needed: location of the pyramid adapter ($\vec{P}_{PA}$), the ankle joint ($\vec{P}_{AJ}$), the heel on the shoe body ($\vec{P}_H$) and the toe on the shoe body ($\vec{P}_T$). These are shown in Fig. 5.1.

During the stance phase of the test, ankle angle is manifested as rotation about $\vec{P}_{AJ}$ and leg deflection as translation between $\vec{P}_{PA}$ and $\vec{P}_{AJ}$. Ankle rotation and leg deflection are specified based upon desired device function.

Experimental force and motion data were acquired using a robot trajectory associated with a healthy walker and a deflection along the leg of 15 mm. Device function for the specified trajectory is shown in Fig. 5.2, with the reference frames shown matching those displayed in Fig. 2.1. This result was compared to simulated results from a quasi-static model of the device. Using $F(t)$ and $\theta(t)$ as inputs, the values of $r(t)$ and $M(t)$
5.1.2 Results

The EaSY-Walk functioned as expected during robot testing. The device displayed an expected amount of non-linearity in stiffness in addition to positive rotational work output about the ankle joint. One shortcoming that was noted during this testing was the delay between maximum dorsiflexion and the coupling spring transferring the moment.
Figure 5.3: EaSY-Walk Moment and Deflection Curves, Comparison of Experimental and Modeled

from the lower ankle (B) to the foot (A), with moment transfer occurring later than the point of maximum dorsiflexion by approximately 4.9 degrees. This delay can be attributed to slip in the rotational jamming mechanism.

Qualitatively, the two outputs of the quasi-static model (ankle moment, $M(t)$ and leg deflection, $r(t)$) are plotted with their experimental counterparts in Fig. 5.3. From the plot of ankle moment, the maximum value from the experimental data is 9.66 Nm, compared to 8.93 Nm from modeled data (a discrepancy of 8.2%). From the deflection
plot, both experimental and modeled results show the bulk of deflection occurring early in stance.

The moment-angle relationship for both the modeled case and experimental case are displayed in Fig. 5.4. Note that arrows represent the progression of the path over time and that a counter-clockwise loop represents positive rotational work done. In this plot, the delay between maximum dorsiflexion and rotational work conversion (4.9 deg.) can be observed. In addition, the experimental data indicates that the prosthesis achieved 0.187 J of rotational work output, whereas the model predicted 0.273 J of work output. This result shows that the experimental work output of the device yielded 68.5% of the simulated case, indicating the device as tested lost 32.5% of the simulated rotational work output. This loss of work can be traced back to the distance the bottom end of $K_1$ must travel to transfer from the lower ankle (B) to the foot (A), which allows energy from the coupling spring to be expended as non-useful work. Additional datasets were acquired and are described in Appendix E.
Similar to the single dataset previously discussed, the EaSY-Walk over the summation of 10 data sets averaged an input of 0.65 J of translational work and an output of 0.17 J of rotational work, yielding an average of 26.6% work conversion from translation to rotation. Additionally, the modeled device averaged 0.27 J of rotational work. Based on this, the device yielded 63.05% of the modeled work suggested by the quasi-static model.

Table 5.1: Robot Testing Data, Performance Metrics

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<th>Average Value</th>
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</tr>
<tr>
<td>Rotational Work, Modeled [J]</td>
<td>0.27</td>
</tr>
<tr>
<td>Rotational Work, Actual [J]</td>
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<td>Percent of Translational Work [%]</td>
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<tr>
<td>Percent of Modeled Rotational Work [%]</td>
<td>63.05</td>
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</table>

5.2 Human Subject Testing

The EaSY-Walk was tested with human test subjects to assess functionality as well as subjective performance in a typical use case of the EaSY-Walk device. Human Subject testing was completed with IRB approval, Protocol HR-2537. Please see Appendix F for a listing of the specific procedures to be followed listed in the IRB Protocol form.

5.2.1 Procedure

Human subject testing was comprised of two portions: kinematic/kinetic data acquisition testing and energetics of walking data acquisition. Regarding the former, kinetic data was acquired using two (2) 6-axis force plates upon which the subject walks and kinematic data was acquired using the Vicon imaging system. From this data, it is possible to generate the Moment Vs. Angle plots for the prosthesis during walking and
quantify the rotational work output of the device. For the latter, energetics of walking data was acquired using an oxygen uptake measurement system to measure the respiratory intake of a subject as they walk on a treadmill at a self selected walking velocity. From this data, it is possible to quantify the energy a subject expends using the EaSY-Walk compared to the energy they expend utilizing their prescribed prosthesis.

Human subject testing was completed on three unique test subjects over a total of four testing sessions (two of the subjects were tested once, one subject was tested twice). Out of the three test subjects, kinetic/kinematic data was acquired for each of the subjects and energetics of walking data was acquired for two of the three subjects.

5.2.2 Results

Between the four test trials, consistent functionality of the device was not achieved, as proper reset and coupling for both the rotational jamming mechanism and translational jamming mechanism were found to function inconsistently and data was not captured for a stride in which the device functioned properly.

While kinematic and kinetic data was not acquired for a stride in which the device functioned as designed (i.e., a stride in which a non-negligible amount of translational work is converted into rotational work), kinematic and kinetic data was still acquired for each of the test subjects. An example data set, acquired for Subject 3, is shown below. Two facets of the device functionality are visible in the data.

First, the plots show (in a fashion similar to plots detailing robot testing data displayed in Section 5.2) non-linear characteristics in regards to rotational stiffness about the ankle joint. Secondly, the data shows that the EaSY-Walk prosthesis dissipated rotational work about the ankle joint during the stride. While a positive quantity of rotational work about the ankle joint would not be expected due to the coupling feature not functioning as expected, the dissipation of rotational work would not necessarily be
expected. However, it can be attributed to friction inherent with the mechanical components of the design.

Additionally, oxygen uptake data was acquired during treadmill testing. In results acquired from oxygen uptake data, it was found that the test subject’s self selected gait velocity was slower with the EaSY-Walk than with the subject’s prescribed prosthesis. This can be attributed to the subject having more familiarity with their prescribed prosthesis in addition to the EaSY-Walk not showing consistent functionality.

The most significant issue facing the rotational jamming mechanism (F) was related to the reset of the mechanism. The reset mechanism was designed such that during the swing phase, the friction block would be kinematically constrained to translate along the channel on the foot (A) until the lower coupling bearing returned to its initial location on the hook of the lower ankle (B). However, due to machining tolerances, the bearing returned to a position located a small distance above the surface of the hook on the lower ankle (B) during reset, leaving a slight gap between the bearing and the hook.

At heel strike the prosthesis comes in contact with the ground and, as discussed previously, leg force reverses from acting in the upwards direction as a function of the weight of the prosthesis to acting in the downwards direction as a function of the weight of the subject. As designed, the translational energy acquired would be stored as potential
energy in the coupling spring.

However, due to the aforementioned gap between the coupling spring bearing and the surface of the hook on the lower ankle (B), a portion of translational energy is instead manifest as kinetic energy and the bearing rapidly accelerates, closing the gap between the bearing and the surface of the hook. As a large actuation force retaining the bearing in the hook is not present at heel strike, the bearing "ramped" (e.g., a rolling motion) off of the hook as opposed to seating on the hook at heel strike, causing coupling to occur at heel strike as opposed to maximum dorsiflexion.

As a result of early coupling, translational energy acquired from deflection along the leg was not stored and later converted into rotational energy. Instead, with the coupling spring acting about the ankle joint through the entire stance phase in instances where coupling occurred at heel strike, any deflection/loading applied to the coupling spring was manifest as increased stiffness about the ankle joint as opposed to rotational work about the ankle joint.
CHAPTER 6

CONCLUSION

Gaps in the ankle prosthesis market motivated work on a new solution: the EaSY-Walk, a device that mimics several key aspects of the natural ankle. Through testing, the novel device was found to function consistently in a controlled setting, but with significantly diminished performance when testing with human subjects.

6.1 Contributions

Development of the EaSY-Walk centered around several design objectives, including having the energy storage associated with deflection along the leg early in stance, increasing rotational energy return in response to increasing walking velocity, and a design envelope small enough that the device fits within a US men’s size 10 shoe and beneath a standard pant leg.

To meet the above objectives, jamming mechanisms were implemented to control both the timing of translational energy storage and rotational energy release. This approach allowed early stance leg deflection and allowed rotational work output to increase with increasing walking velocity. Also, flexure based nominal stiffness mechanisms decreased the overall design envelope of the device and increased the non-linearity in the rotational stiffness to more closely match the rotational stiffness of a natural ankle joint.

Additionally, a quasi-static model of the EaSY-Walk design was generated and focused on the modeling of each of the novel design concepts discussed above. This model includes the timing of energy storage and release utilizing jamming mechanisms in addition to modeling of the flexure based nominal stiffness. Using this model, a set of device parameters were optimized to best match the stiffness profile and work output of the natural ankle using Genetic Algorithms. From the optimized results, a prototype of the
device was constructed for testing.

6.2 Conclusions

From robot testing, the experimental data indicates that the prosthesis achieved 0.187 J of rotational work output, whereas the model predicted 0.273 J of work output. This shows that the experimental work output of the device yielded 68.5% of the simulated case. Additionally, two limitations of the device were identified in robot testing. First, the conversion of translational work to rotational work occurs later than the point of maximum dorsiflexion by approximately 4.9 degrees. This delay can be attributed to slip in the rotational jamming mechanism. Second, the device as tested lost 32.5% of the simulated rotational work output. This can be traced back to the distance the bottom end of $K_1$ must travel to transfer from the lower ankle (B) to the foot (A), which allows energy from the coupling spring to be expended as non-useful work.

Through human subject testing, the EaSY Walk did not show robust, consistent functionality similar to what was shown in robot testing. The majority of the problems can be attributed to an inconsistent reset of the coupling spring. This behavior significantly diminishes the amount of translational work converted to rotational work.

Combining the knowledge gained from robot testing and human subject testing, it can be concluded that, based on the friction based jamming mechanisms employed, the EaSY-Walk is only able to achieve desired functionality when the loading applied to the device is applied in a slow and controlled manner, thereby satisfying the quasi-static assumption made during the development phase. As such, additional revisions to the EaSY-Walk will be needed to achieve the functionality required of a commercially viable product.
6.3 Future Work

To ensure that future generations of the device are able to achieve robust functionality regardless of the loading applied to the device, it is recommended that significantly tighter tolerances be held on all parts/components related to the jamming mechanisms of the device. While doing so would incur additional costs, it would also greatly benefit the consistency of functionality of the device. Additionally, a mechanism to apply an initial actuation force to ensure the coupling spring remains on the "hook" of the lower ankle body at heel strike would be required.

Alternatively, future generations of the device could replace the kinetic/friction based locking mechanisms with a kinematic/ratcheting based solution. While employing such a solution would decrease the number of locking positions from a theoretically infinite amount to a finite amount based upon a discrete distance between locking positions, a ratcheting based solution would also be significantly more robust to the impact loadings and variability associated with human gait, ensuring both consistent coupling and consistent reset of the coupling mechanisms.

In addition to refining the device such that it functions robustly and consistently regardless of loading, it is also necessary to make performance refinements to the device for it to become a commercially viable product. Necessary performance refinements further include ensuring that translational energy lost during the coupling event is minimized, and that coupling of the translational and rotational degrees of freedom occurs in closer proximity to maximum dorsiflexion. Further, any design changes implemented to achieve the above goals must not increase the weight or decrease the robustness of the device.
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APPENDIX A

DESIGN OBJECTIVES AND METRICS

Shown below are a table of customer needs and a table of design metrics including the customer needs they can be mapped back to. Each of these tables includes an importance of each need and design metric, where a 1 signifies a low need metric and 5 signifies a high need metric.

<table>
<thead>
<tr>
<th>No.</th>
<th>Category</th>
<th>Need</th>
<th>Importance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Reduces Energy</td>
<td>Reduces energy expended while walking</td>
<td>5</td>
</tr>
<tr>
<td>2</td>
<td>Comforability</td>
<td>Lightweight</td>
<td>5</td>
</tr>
<tr>
<td>3</td>
<td></td>
<td>Operation is nearly silent</td>
<td>3</td>
</tr>
<tr>
<td>4</td>
<td></td>
<td>Usable on a staircase</td>
<td>2</td>
</tr>
<tr>
<td>5</td>
<td></td>
<td>Minimal shock felt by user</td>
<td>4</td>
</tr>
<tr>
<td>6</td>
<td></td>
<td>Usable on a sloped surface (ie, inversion/eversion)</td>
<td>4</td>
</tr>
<tr>
<td>7</td>
<td></td>
<td>Aesthetically pleasing</td>
<td>1</td>
</tr>
<tr>
<td>8</td>
<td></td>
<td>Mass is comfortably distributed</td>
<td>2</td>
</tr>
<tr>
<td>9</td>
<td>Compactness</td>
<td>Fits beneath a standard pant leg</td>
<td>3</td>
</tr>
<tr>
<td>10</td>
<td></td>
<td>Fits inside of a standard sized shoe</td>
<td>3</td>
</tr>
<tr>
<td>11</td>
<td></td>
<td>Height is small enough to fit most users</td>
<td>3</td>
</tr>
<tr>
<td>12</td>
<td>Safety</td>
<td>Can safely support a heavy user</td>
<td>5</td>
</tr>
<tr>
<td>13</td>
<td></td>
<td>Has minimal pinch points</td>
<td>1</td>
</tr>
<tr>
<td>14</td>
<td></td>
<td>Has a long usage life</td>
<td>2</td>
</tr>
<tr>
<td>15</td>
<td>Customizability</td>
<td>Allows for springs of varying stiffness</td>
<td>4</td>
</tr>
<tr>
<td>16</td>
<td></td>
<td>Allows for multiple deflection settings</td>
<td>4</td>
</tr>
<tr>
<td>17</td>
<td>Cost</td>
<td>Reasonable Purchase Cost</td>
<td>3</td>
</tr>
<tr>
<td>No.</td>
<td>Corr. Needs</td>
<td>Metric</td>
<td>Importance</td>
</tr>
<tr>
<td>-----</td>
<td>-------------</td>
<td>---------------------------------------------</td>
<td>------------</td>
</tr>
<tr>
<td>1</td>
<td>1</td>
<td>Converts Trans. Work to Rot. Work</td>
<td>5</td>
</tr>
<tr>
<td>2</td>
<td>1</td>
<td>Proper Energy Mapping</td>
<td>4</td>
</tr>
<tr>
<td>3</td>
<td>1,3</td>
<td>Minimal Energy lost to friction, etc.</td>
<td>4</td>
</tr>
<tr>
<td>4</td>
<td>1,2</td>
<td>Total Mass</td>
<td>5</td>
</tr>
<tr>
<td>5</td>
<td>3</td>
<td>Operational Noise Level</td>
<td>3</td>
</tr>
<tr>
<td>6</td>
<td>4</td>
<td>Proper Stair Operation</td>
<td>2</td>
</tr>
<tr>
<td>7</td>
<td>5</td>
<td>Maximum Jerk of Pyramid Joint</td>
<td>3</td>
</tr>
<tr>
<td>8</td>
<td>7</td>
<td>Max allowable Inversion/Eversion</td>
<td>4</td>
</tr>
<tr>
<td>9</td>
<td>9</td>
<td>Transverse size of upper ankle region of prosthesis</td>
<td>3</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
<td>Length and width of foot portion of prosthesis</td>
<td>3</td>
</tr>
<tr>
<td>11</td>
<td>11</td>
<td>Overall Height of Prosthetic</td>
<td>3</td>
</tr>
<tr>
<td>12</td>
<td>15</td>
<td>Spring Compatibility</td>
<td>4</td>
</tr>
<tr>
<td>13</td>
<td>16</td>
<td>Deflection Settings</td>
<td>4</td>
</tr>
<tr>
<td>14</td>
<td>17</td>
<td>Low Build Cost</td>
<td>3</td>
</tr>
</tbody>
</table>
APPENDIX B

CONCEPTUAL DETAIL DESIGN - JAMMING MECHANISMS AND FLEXURES

This appendix will further detail several of the mechanical subsystems utilized by the EaSY-Walk.

B.1 Friction Based Jamming Mechanisms

To achieve the energy storage and release behavior described above, mechanical diodes will be employed. A commonly used device used in electrical, hydraulic and even mechanical circuits, a diode is a device that restricts the flow variable to only flow in one direction regardless of the direction of the effort variable.

Take, for example, a hydraulic diode (e.g., a check valve). In a check valve, fluid is only able to flow from right to left but not vice versa. Thus, the mechanical diodes employed by the device will allow motion (either translational or rotational) in one direction but will constrain motion (regardless of the direction of force/torque) in the opposite direction, similar to the clutches in the CESR Foot [24] discussed in Section 1.3.2.

Mechanisms capable of this functionality include a ratchet and pawl mechanism, a multi-position latching mechanism, and a sprag/roller friction based jamming mechanism [30]. Each has many advantages and disadvantages that must be considered when selecting a mechanism to implement into the prosthesis.

A sprag/friction based device was selected for translational jamming/energy storage and a roller/friction based mechanism was selected for rotational jamming/energy conversion. Friction based mechanisms were selected primarily for their ability to have a theoretically infinite number of locking positions, whereas number of locking positions in both the ratchet and pawl mechanism as well as the multi-position latching mechanism are
Both sprag and roller based friction jamming mechanisms work on the same basic principle: using friction to jam (or wedge) an element between two surfaces, thereby locking the positions of the two surfaces relative to one another [29]. This friction is caused by an eccentricity, either located on the element itself (sprag mechanism) or on one of the two surfaces abutting the element (roller mechanism). A diagram of each of these mechanisms is shown in Fig. B.1.

In the roller mechanism shown in Fig. B.1, the eccentricity in the system can be seen in the outer cam surface. For the sprag mechanism, the eccentricity is manifested in the distance/offset between the center of the two half circles making up the jamming element. Based on this offset, the element boasts an effectively infinite number of radii (four of which are illustrated) as the element is rotated. Moving to the far right diagram, the jamming element is optimized for dynamic operation. While each of the three diagrams in Fig. B.1 show rotational systems, the same techniques can be applied to a translational system.

To ensure self actuation, two design variables must be considered: the coefficient of friction between the jamming element/contact surfaces and strut angle of the system. In a sprag clutch mechanism, the strut angle is a function of the eccentricity present in the jamming element. Conversely, the strut angle in a roller clutch mechanism is a function of
the angle between the inner and outer contact surfaces and can be observed visually in Fig. B.2.

In both cases, it will be the frictional load \( F_{f(0)} \) and \( F_{f(i)} \) that opposes motion in the system. Looking to the Sprag FBD, \( \alpha(o) \) and \( \alpha(i) \) (the outer and inner strut angles) are, in general, marginally different values as a function of the difference in radii of the inner and outer raceway surfaces. However, as the sprag device in the EaSY-Walk is between two flat surfaces, the inner and outer strut angles will be equal (deemed \( \alpha \)). Knowing that \( F_i = F_o \) for an equilibrium case, \( F \) can be defined as the total force vector at each contact point, \( F_n \) the force normal to each contact point, \( F_f \) the frictional load at each contact point, \( \alpha \) as the strut angle and \( \mu_s \) the frictional coefficient between the jamming element and each of the raceways. From this, the following equations are established for each contact point:

\[
F_f = F \sin(\alpha) \quad \text{(B.1)}
\]

\[
F_n = F \cos(\alpha) \quad \text{(B.2)}
\]
\[ F_f \leq F_n \mu_s \] \hspace{1cm} (B.3)

\[ \tan(\alpha) \leq \mu_s \] \hspace{1cm} (B.4)

This sets up the relationship between strut angle and coefficient of friction that must be considered. The tangent value of the strut angle must be less than the coefficient of friction between the jamming element and the raceways. However, decreasing the strut angle also increases the normal load required to achieve a desired frictional force. Increasing the normal load will increase system stresses and will necessitate added material and weight to accommodate for this. As such, material selection (influencing strength and coefficient of friction) in addition to strut angle selection will be crucial to the success of the jamming mechanisms.

**B.2 Flexure Based Nominal Stiffness**

In both the Generation 1 and 2 devices, a translational spring with one end fixed and the other allowed to pivot about the ankle joint was used to achieve a nominal rotational stiffness. This design was considered advantageous as the pivoting design would, ideally, generate a non-linear rotational stiffness about the ankle joint. However, through modeling and optimization it was determined that the non-linearity of the natural ankle could not be matched with translational springs. It was also found during the mechanical design phase of the EaSY-Walk that use of translational springs to achieve nominal stiffness about the ankle joint would decrease the compactness of the final design.

Based on the aforementioned shortcomings, alternatives were considered and a flexure based mechanism was selected. Flexures, for the purpose of this design, act as a cantilever beam with a point load applied at some position on the beam by a contactor, shown conceptually in Fig. B.3.
In the design, a plantarflexion and dorsiflexion flexure are constrained to remain static with the foot body. On the leg body, two sets of contactors (roller bearings) rotate with the body and deflect their respective flexure, creating a reaction force. This reaction force on the contactor imparts a moment about the ankle joint. As the contactor is rotating about the ankle joint, it will translate both perpendicular to the beam (causing beam deflection) as well as along the axis of the beam (causing the effective length of the beam to shorten). Through optimization, it was found that this design strategy could decrease the spacial envelope of the device and achieve a more natural stiffness non-linearity about the ankle joint.

Figure B.3: Cantilever Beam with Point Load at End
APPENDIX C
MODELING AND OPTIMIZATION DETAILS

C.1 Design Parameters

Table C.1: EaSY-Walk Design Parameters

<table>
<thead>
<tr>
<th>Component</th>
<th>Parameter</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coupling Spring</td>
<td>$K_1$</td>
<td>Coupling spring stiffness.</td>
</tr>
<tr>
<td></td>
<td>$X_{ab1},Y_{ab1}$</td>
<td>Loc. of $K_1$ on Body (A)/(B). Defined by the pos. of the &quot;hook&quot; on Body(B).</td>
</tr>
<tr>
<td></td>
<td>$X_{bc1},Y_{bc1}$</td>
<td>Loc. of $K_1$ on Body (B)/(C).</td>
</tr>
<tr>
<td>Dorsiflexion Flexure</td>
<td>$X_d,Y_d$</td>
<td>Location of the flexure contactor.</td>
</tr>
<tr>
<td></td>
<td>$X_{wd}$</td>
<td>Location in the X direction of the rigidly fixed end of the flexure. Note: Y-location of this point determined by $Y_d$.</td>
</tr>
<tr>
<td></td>
<td>$B_w$</td>
<td>Width of the flexure at the base/wall end of the flexure.</td>
</tr>
<tr>
<td></td>
<td>$B_e$</td>
<td>Width of the flexure at the tip/contactor end of the flexure.</td>
</tr>
<tr>
<td></td>
<td>$H$</td>
<td>Thickness of the flexure.</td>
</tr>
<tr>
<td>Plantarflexion Flexure</td>
<td>$X_p,Y_p$</td>
<td>Location of the flexure contactor.</td>
</tr>
<tr>
<td></td>
<td>$Y_{wp}$</td>
<td>Location in the Y direction of the rigidly fixed end of the flexure. Note: X-location of this point determined by $X_p$.</td>
</tr>
<tr>
<td></td>
<td>$B_p$</td>
<td>Width of the flexure.</td>
</tr>
<tr>
<td></td>
<td>$H_p$</td>
<td>Thickness of the flexure.</td>
</tr>
</tbody>
</table>

The model as described in Section 3.2 will require a set of parameters to fully define the design of the prosthesis. It is this set of parameters that will be optimized, with the optimization process described in greater depth below. These parameters consist of the locations of spring ends as well as the rates for each of the springs in the system. The spring rate for each of the flexures will be based on the length, width and height of the respective flexure. Figure C.1, shown below, graphically displays each of these parameters.
on the device (note: coordinate frame O is attached to the foot body and all parameters discussed are relative to this coordinate frame). A summary description of each of these parameters, $\vec{P}$, is shown in Table C.1.

C.2 Device Kinematics

From parameters $\vec{P}$, the initial location of each of the spring and flexure ends is known. However, locations of these points relative to any body they are not rigidly affixed to will not be constant. To find the deflection in the springs, the location of both ends of each spring relative to a single reference point or body are needed.

Begin by affixing a coordinate reference frame to Foot (A) (0), Lower Ankle (B) (0’) and Upper Ankle (C) (0’’), with rotation allowed between Foot (A) and Lower Ankle
(B) and translation allowed between Lower Ankle (B) and Upper Ankle (C). The application of the described coordinate frames is as shown in Fig. C.2.

To transform points from the 0 frame to 0’ frame and vice versa, the rotation matrix for the planar case (defined below in Eq. (C.1), \( \theta \) is defined as the ankle angle) will be utilized. This rotation is shown in Equation (C.2).

\[
R = \begin{bmatrix}
\cos(\theta) & -\sin(\theta) \\
\sin(\theta) & \cos(\theta)
\end{bmatrix}
\]

\(
\begin{bmatrix}
X'_{ab1} \\
Y'_{ab1}
\end{bmatrix} = R \begin{bmatrix}
X_{ab1} \\
Y_{ab1}
\end{bmatrix} \iff \begin{bmatrix}
X_{ab1} \\
Y_{ab1}
\end{bmatrix} = R^{-1} \begin{bmatrix}
X'_{ab1} \\
Y'_{ab1}
\end{bmatrix}
\)

Points on the upper leg body (Upper Ankle (C)) can also translate along the Y-axis in frame 0’ or relative to Lower Ankle (B). With \( r \) defined as deflection along this axis and \( C_{BC} \) defined as the initial distance between coordinate frame 0’ and 0”, the transformations between these two coordinates frames (using the top end of \( K_1 \) as an example) are as below:
Finally, combining these two sets of equations to go directly from the 0 frame to 0" frame and vice versa yields the following:

\[
\begin{bmatrix}
X''_{bc1} \\
Y''_{bc1}
\end{bmatrix}
= R
\begin{bmatrix}
X'_{bc1} \\
Y'_{bc1}
\end{bmatrix}
- \begin{bmatrix}
0 \\
C_{bc} - r
\end{bmatrix}
\iff
\begin{bmatrix}
X'_{bc1} \\
Y'_{bc1}
\end{bmatrix}
= R^{-1}
\begin{bmatrix}
X''_{bc1} \\
Y''_{bc1}
\end{bmatrix}
+ \begin{bmatrix}
0 \\
C_{bc} - r
\end{bmatrix}
\] (C.4)

Next, spring deflections will be considered. Start by finding the free length of the spring in question. Using Pythagorean Theorem and \(K_1\) in its initial state, free length \((L_{0,K1})\) is defined in Eq. (C.5) and subsequently, deflection \((d_1)\) is as detailed in Eq. (C.6).

\[
L_{0,K1} = \sqrt{(Y_{bc1,0} - Y_{ab1,0})^2 + (X_{bc1,0} - X_{ab1,0})^2}
\] (C.5)

\[
d_1 = L_{0,K1} - \sqrt{(Y_{bc1} - Y_{ab1})^2 + (X_{bc1} - X_{ab1})^2}
\] (C.6)

C.3 Force and Torque Equations

To properly model the EaSY-Walk, it will be required to determine the force balance on Upper Ankle (C). When the translational jamming mechanism is not locked, the force balance (neglecting \(K_2\) due to its low stiffness) is as shown in Fig. C.3. With \(F_{leg}\) being defined purely in the Y-axis of frame 0’ and \(F_{K1}\) placed into the 0’ frame, the resulting force balance is shown in Eq. (C.7).

\[
F_{leg} = d_1 K_1 \cos \left( \sin^{-1} \left( \frac{X'_{bc1} - X'_{ab1}}{L_{0,K1} - d_1} \right) \right)
\] (C.7)

As \(F_{leg}\) is a model input, \(X_{bc1}, X_{ab1}, K_1\) are design parameters, and \(L_{0,K1}\) is derived from design parameters, it will be required to solve for spring deflection \(d_1\) in Eq. (C.7).
Due to the angle of the spring changing as a function of $d_1$, the solution to the equation is non-linear and can’t be solved using conventional methods.

Instead, a mathematical method, the fzero function, will be used. Fzero is a function built into Matlab that finds the root of a non-linear function when given a function and a seed point. Rearranging Eq. (C.7) and using the deflection found at a previous frame as the starting point, fzero can be used to efficiently solve for $d_1$. Treating the spring as a linear element, force $F_{K1}$ can be found using $d_1$ as in Eq. (C.9).

$$0 = d_1 K_1 \cos \left( \sin^{-1} \left( \frac{X'_{bc1} - X'_{ab1}}{L_{0,K1} - d_1} \right) \right) - F_{leg}$$ \hspace{1cm} (C.8)

$$F_{K1} = K_1 d_1$$ \hspace{1cm} (C.9)

Additionally, splitting the spring force $F_{K1}$ into components yields:

$$\begin{bmatrix} F_{K1,x} \\ F_{K1,y} \end{bmatrix} = \frac{F_{K1}}{L_{0,K1} - d_1} \begin{bmatrix} X_{bc1} - X_{ab1} \\ Y_{bc1} - Y_{ab1} \end{bmatrix}$$ \hspace{1cm} (C.10)

To determine moments $M_{K1}$ about the ankle joint utilizing the forces exhibited by the coupling spring previously found, the cross product of position of the spring end and spring force will be used.
Since $M_x$ and $M_y$ from the cross product are equal to zero, spring moments about the Z-axis can be summed to find the overall moment about the ankle joint, shown below:

$$M_{ankle} = M_{K1,z} + M_{K3,z} + M_{K4,z}$$

(C.12)

C.4 Flexure Stiffness Modeling

Both $K_3$ and $K_4$ will act as cantilever beams in the system, with a contactor applying a point load at the end. $K_3$ will be of constant cross section but $K_4$ will have a linear taper along the width of the beam as shown in Fig. C.4. This taper will further increase the non-linearity observed in the stiffness of $K_4$, as the cross section of the flexure at the contact point will increase as the effective length of the beam decreases.

Analyzing $K_3$, the rotation matrix given by Eq. (C.2) can be utilized to find the location of contact between flexure and contactor ($X_{flex}$ and $Y_{flex}$, Eq. (C.13)), from which the deflection ($d_p$, Eq. (C.14)) and the effective length ($L_{eff,p}$, Eq. (C.15)) of the
flexure can be found. When finding effective length of the flexure, the curvature of the flexure is negligible and Pythagorean’s Theorem is valid.

\[
\begin{bmatrix}
X_{flex} \\
Y_{flex}
\end{bmatrix}
= R
\begin{bmatrix}
X_{p,0} \\
Y_{p,0}
\end{bmatrix}
\tag{C.13}
\]

\[d_p = X_{p,0} - X_{Flex}\]
\tag{C.14}

\[L_{eff,p} = \sqrt{(Y_{wp} - Y_{Flex})^2 + d_p^2}\]
\tag{C.15}

From elementary beam theory, the force imparted on the contactor is as defined in Eq. (C.17), where \(E\) is Young’s Modulus of the flexure and \(I_p\) is the second moment of area about the centroid of a rectangle as in Eq. (C.16).

\[I_p = \frac{B_p H_p^3}{12}\]
\tag{C.16}

\[F_{K3,x} = \frac{3EI_p d_p}{L_{eff,p}^3}\]
\tag{C.17}

However, this assumes that the loading on the beam remains orthogonal to the original beam axis. Due to the contactor design, the load will be applied normal to the surface of the flexure. To determine this load, the angular deflection of \(K_3\) at the point of contact must be determined. Using this contact angle, \(\gamma_p\), the force in spring \(K_3\) in the y-direction \(F_{K3,y}\) can be found.

\[\gamma_p = \frac{F_{K3,x} L_{eff,p}}{2EI_p}\]
\tag{C.18}

\[F_{K3,y} = F_{K3,x} tan(\gamma_p)\]
\tag{C.19}
Analyzing $K_4$, the dorsiflexion flexure, the overall process will be similar to above. However, due to the taper along the width of $K_4$, a more generalized form of beam theory will be used. Using the dummy-load approach, an extension of Castigliano’s Theorem [31], the deflection in the dorsiflexion flexure, $d_d$, is as below:

$$\delta_e = d_d = \int_0^{L_{eff,d}} \frac{Mm}{EI_d(x)} dx$$  \hspace{1cm} (C.20)

Further, in Eq. (C.20), $Mm$ can be defined as the following, wherein $F_{K_4,y}$ is the force normal to the dorsiflexion flexure and $x$ is the length of the flexure from the contact point of the contactor to the rigid end.

$$Mm = F_{K_4,y}x^2$$  \hspace{1cm} (C.21)

Due to the tapered nature of the beam, it is known that the second moment of area for $K_4$ is some function of $x$, based on the slope of the taper. The derivation of this slope (a modified version of the slope $z$ is determined in Eq. (C.23)) using know values such as $I_w$ (second moment of area for the flexure at the fixed support, constant) and $I_e$ (second moment of area for the flexure at the contactor location, function of $L_{eff,d}$) and its usage are shown below.

$$I_e = I_w \frac{L_{eff,d,0} - L_{eff,d}}{L_{eff,d,0}}$$  \hspace{1cm} (C.22)

$$z = \frac{I_w}{I_e} - 1$$  \hspace{1cm} (C.23)

$$I_d(x) = I_e + I_e \frac{z x}{L_{eff,d}} = I_e \left( \frac{L_{eff,d} + z x}{L_{eff,d}} \right)$$  \hspace{1cm} (C.24)

$$d_d = \frac{F_{K_4,y}L_{eff,d}}{EI_e} \int_0^{L_{eff,d}} \frac{x^2}{L_{eff,d} + z x} dx$$  \hspace{1cm} (C.25)
The substitution method can then be used to solve for the above integral and rearranged to find the force normal to the surface of the dorsiflexion flexure $F_{K4,y}$. Having solved for $F_{K4,y}$, it can then be used to solve for the angle of the flexure at the point of contact, $\gamma_d$, shown in Eq. (C.27).

\[
F_{K4,y} = \frac{EI_e d_d z^3}{L_{eff,d}^3 (0.5z^2 - z + \log(1 + z))}
\]  

\[
\gamma_d = \frac{F_{K4,y} L_{eff,d}^2 (z - \log(1 + z))}{z^2 EI_e}
\]  

Similar to $K_3$, contact angle can be used to find force along the X-axis. Combined with the location of the contactor found previously, $M_{K4,z}$ can be found.

C.5 Energy Storage and Return Mechanisms

One of the key design features of the EaSY-Walk is the sequencing of both translational energy storage and its conversion to rotational work output. Discussed in Chapter 2, the friction based jamming mechanisms employed function based upon the time based derivatives of known ankle function profiles.

Analyzing the function of the translational jamming mechanism, the force balance on Upper Ankle (C) was previously described when the translational jamming mechanism is unlocked. In this case, the only two forces acting on the body (again, negating $F_{K2}$) are $F_{leg}$ and $F_{K1}$. However, as described above, leg force will begin decreasing just past foot flat. When this occurs, $F_{K1}$ would decrease in order to retain a force balance. As this is an undesirable outcome, the translational jamming mechanism will lock at this point, applying a downwards force on Upper Ankle (C) (as shown in Fig. C.5) to retain the force balance. This jamming mechanism is capable of switching modes multiple times during the same stance cycle, dependent upon the magnitudes of $F_{leg}$ and $F_{K1}$. The new force balance on Upper Ankle (C) becomes the following:
Recalling Figure 2.1c, the rotational jamming mechanism is tasked with transferring the coupling spring from Lower Ankle (C) to Body A at maximum dorsiflexion. This transfer is accomplished by requiring two separate criteria for locking: the device must be in dorsiflexion (to avoid early locking during CP) and the device must be plantarflexing, signaling the end of CD and the start of the PP stage.

In the model, the lower end of $K_1$ in the lower leg frame ($X'_{ab1}$ and $Y'_{ab1}$) will remain constant when the device is unlocked. Once locking occurs and $K_1$ is transferred to Body A, the lower end of $K_1$ will now remain constant relative to the foot frame ($X'_{ab1}$ and $Y'_{ab1}$). Opposed to the translational jamming mechanism, the device cannot return to the unlocked state subsequent to locking during the stance phase.

Within the Matlab model of the device, both of these mechanisms can be modeled together using if statements in addition to for and while loops. Shown in Fig. C.6 is the flowchart of the code, which can describe device function at any point in the stance phase of gait.
C.6 Energy Methods Modeling

Energy method modeling serves two purposes: first, it will be used to determine the rotational work output and the translational work stored by the device and second, to verify the model. Starting with determining work input/output, a numerical integration technique will be employed to solve for both since the model inputs will be a discrete set of data points. Using $M(t)$ and $\theta(t)$ as inputs and defining $N$ as the number of discrete points in the data set yields the following equation for determining the rotational work output, $W_{rot,\,tot}$, for the device.

$$W_{rot,\,tot} = \sum_{i=2}^{N} M(t_i) [\theta(t_i) - \theta(t_{i-1})] \tag{C.29}$$

Additionally, the same technique for translational work $W_{trans,\,tot}$:

$$W_{trans,\,tot} = \sum_{i=2}^{N} F(t_i) [r(t_i) - r(t_{i-1})] \tag{C.30}$$
Energy methods can also be used to verify the model. To do this, energy stored in each spring element as well as the rotational and translational energy exhibited in the system will be summed and, in a valid system, will equal zero. This was completed using the following set of equations, where \( j \) is an arbitrary frame in the data set and \( E_{K2} \) is negated due to its low stiffness.

For the translational energy input to the system \( E_{rot}(j) \) and the rotational work output \( E_{trans}(j) \):

\[
E_{rot}(j) = \sum_{i=2}^{j} M(t_i)[\theta(t_i) - \theta(t_{i-1})] \tag{C.31}
\]

\[
E_{trans}(j) = \sum_{i=2}^{j} F(t_i)[r(t_i) - r(t_{i-1})] \tag{C.32}
\]

This can additionally be completed for the energy stored in the coupling spring, \( E_{K1}(j) \):

\[
E_{K1}(j) = \frac{1}{2} K_1 d_1(j)^2 \tag{C.33}
\]

For the flexures, where it will be required to use energy methods for solid bodies to find the bending, shear and axial energy stored in each of the flexure devices:

\[
E_{flexure}(j) = E_{axial} + E_{bend} + E_{shear} \tag{C.34}
\]

\[
E_{K3}(j) = \frac{1.356}{12} \left[ F_{K3,x}(j)^2 \left( \frac{L_{eff,p}(j)^3}{2EI_p} + \frac{L_{eff,p}(j)}{2GH_pB_p} \right) + \frac{F_{K3,y}(j)^2 L_{eff,p}(j)}{EH_pB_p} \right] \tag{C.35}
\]

\[
E_{K4}(j) = \frac{1.356}{12} \left[ F_{K4,x}(j)^2 \left( \frac{L_{eff,p}(j)^3(0.5z(j)^2 - z(j) + log(1 + z(j))}{2EI_dz(j)^3} \right) + \frac{L_{eff,p}(j)log(1 + z(j))}{2GH_{B_e}(j)z(j)} \right] + \frac{F_{K4,y}(j)^2 L_{eff,d}(j)log(1 + z(j))}{2EH_{B_e}(j)z(j)} \tag{C.36}
\]
Finally, summing each of the energy values previously found: Having found the energy stored in each storage element in addition to the energy input and output to the system, the values can be summed in order to determine the total system energy $E_{total}$.

Note that to verify the model, $E_{total}$ should always equal 0.

$$E_{total}(j) = E_{rot}(j) + E_{trans}(j) + E_{K1}(j) + E_{K3}(j) + E_{K4}(j) \quad (C.37)$$

C.7 Optimization Process

Using the model described, functionality of the device can be found for any parameter set $\vec{P}$. A formal optimization method will be employed to go from an arbitrary parameter set $\vec{P}$ to the optimal parameter set, $\vec{P}_{opt}$. To complete the optimization process, four things will be needed. First, a fitness function will be required to determine what defines the optimal result. Second, design constraints will be needed such that the final set of design parameters is feasible to construct. Third, data sets (containing $F(t)$, $\theta(t)$ and $M_{ref}(t)$) from the gait of able bodies individuals will be needed for both inputs to the model ($F(t)$, $\theta(t)$) as well as for comparison to device function ($M_{ref}(t)$). Finally, an optimization algorithm will be chosen. The selection process for each of these is described in the following sections.

C.8 Fitness Function

A fitness function is often used to fully quantify a set of design objectives. Design objectives stated previously define several of the fitness function terms, starting with matching rotational work output from the device ($W_{rot, tot}$, found in Eq. (C.29)) to work output of the natural ankle ($W_{rot, ref}$, using $M_{ref}$). After normalizing with respect to $W_{rot, ref}$, this gives us the following fitness term for work:
Next, the moment vs. angle profile of the device will be optimized. For this term, minimize the error between the experimental profile of the natural ankle and the modeled profile of the device. To allow for weighting various regions of the profile differently, error calculations were split into three regions (Early Loading (EL), Late Loading (LL) and Unloading (UL)). Reference error values were employed to normalize terms.

\[
Q_{work} = \frac{W_{rot,tot} - W_{rot,ref}}{W_{rot,ref}} \quad (C.38)
\]

\[
Q_{EL} = \sqrt{\sum_{i=1}^{EL} \frac{(M(t_i) - M_{ref}(t_i))^2}{(0 - M_{ref}(t_i))^2}} \quad (C.39)
\]

\[
Q_{LL} = \sqrt{\sum_{i=EL+1}^{LL} \frac{(M(t_i) - M_{ref}(t_i))^2}{(0 - M_{ref}(t_i))^2}} \quad (C.40)
\]

\[
Q_{UL} = \sqrt{\sum_{i=LL+1}^{UL} \frac{(M(t_i) - M_{ref}(t_i))^2}{(0 - M_{ref}(t_i))^2}} \quad (C.41)
\]

\[
Q_{M_{max}} = \frac{abs(max(M_{ref}) - max(M))}{max(M_{ref})} \quad (C.42)
\]

It will also be desired to minimize the stress felt by the flexure devices. However, as each flexure also has a tensile strength that failure will occur at, a combination fitness term/design constraint will be applied. This will be done using a piecewise function with a fitness value several orders of magnitude higher (as it is desired to minimize this term) in the failure region than other regions. Knowing the tensile strength of the flexures (to be made of a carbon fiber composite) is 300,000 psi and the approximate fatigue life curve for the material yields the following fitness equations:
\[ Q_{\sigma,d,\sigma,p} = \begin{cases} (8.5451E10)e^{(1.1302E-04)\sigma_{d,p}}, & \sigma_{d,p} < 193,357 \text{psi} \\ 10,000, & \sigma_{d,p} \geq 193,357 \text{psi} \end{cases} \quad (C.43) \]

Finally, summing these up and applying weighting factors based on the importance of each performance metric (defined in Appendix A) yields Eq. (C.44) shown below.

From the equation, it can be seen that workout output (0.45 for \( Q_{\text{work}} \)) weighs slightly larger than the torque profile match (0.4 when summing \( Q_{EL}, Q_{EL}, Q_{EL}, Q_{EL} \), and \( Q_{M_{max}} \)) and approximately 3 times as large as the weighting for stress in the flexures.

\[ \text{Qual} = 0.45Q_{\text{work}} + 0.125Q_{\sigma,d} + 0.025Q_{\sigma,p} + 0.15Q_{EL} + 0.1Q_{LL} + 0.05Q_{UL} + 0.1Q_{M_{max}} \quad (C.44) \]

C.9 Design Parameters and Constraints

As described by \( \overrightarrow{P} \), a total of 16 parameters (10 locations, 1 spring rate, 5 flexure dimensions) fully define the prosthesis in the model. Among these, 9 will be constant for the designed prototype and 7 will be adjustable from subject to subject.

Through parameter sensitivity testing, several of these variables were found to have a near negligible effect on the value fitness value. As every additional parameter to be optimized comes with a significant computational cost, the insensitive parameters were either hand picked or solved for with a rudimentary optimization.

To ensure that the device is optimal for the average walker, several sets of data will be Utilized. These data sets include [14] (considered as an "average" walker) as well as 11 total data sets acquired over 3 unique able bodied individuals (4 data sets for Subject 1, 4 for Subject 2, and 3 for Subject 3) tested at Marquette University. Over these sets of data, the 6 "constant" parameters to be optimized will remain static while the 5 "adjustable"
parameters will be optimized separately for [14] as well as each of the 3 unique test subjects from Marquette (i.e., the optimization process will result in 1 set of "constant" parameters and 4 sets of "adjustable" parameters, one for each subject). This will make for a 26 parameter final optimization.

Constraints on each of these parameters will be based on physical constraints of the design space. In addition, the device was constrained to allow for up to 15 mm of deflection along the leg. This constraint was set to ensure comfort for the end user.

C.10 Genetic Algorithms Routine

To find the optimal parameter set, a Genetic Algorithm (GA) method in the Matlab Optimization toolbox will be utilized. GA’s function by taking a large initial population of parameter sets (each with differing parameter values), each parameter falling in a predetermined range (i.e., constraints) based on the design requirements of the device, and retaining the parameter sets that resulted in the strongest fitness function values. These strongest parameter sets will then be mutated with a new population of random parameter
After allowing the GA optimization routine to run to its natural termination (see Fig. C.8 for a plot of the optimal fitness value at each optimization iteration, showing convergence to a singular value), an optimal parameter set of both constant and adjustable parameters for the device was found. Each parameter value is as tabulated below, with constant values in Table C.4 and adjustable values in Table C.5.

Table C.4: EaSY-Walk Constant Design Parameters

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<th>Parameter [in]</th>
<th>( X_{ab1} )</th>
<th>( Y_{ab1} )</th>
<th>( X_{bc1} )</th>
<th>( Y_{bc1} )</th>
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Figure C.8: Convergence Plot of the EaSY-Walk Device Optimization

Table C.5: EaSY-Walk Constant Design Parameters

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APPENDIX D

PURCHASED AND MACHINED COMPONENT DETAILS

Shown below is a Bill of Materials detailing the purchase price for the parts comprising the EaSY Walk, sorted by supplier:

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<th>MultiTool ($)</th>
<th>Misumi ($)</th>
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APPENDIX E

ROBOT TEST RESULTS - MULTIPLE TRIALS

While a single dataset representative of a typical dataset acquired from robot testing was described in Chapter 5, multiple sets of robot testing data were acquired. In the plot shown below, ankle moment is plotted against ankle angle for a total of 10 data sets acquired using an identical robot path as well as an identical level of translational deflection (approximately 15 mm).

Similar to the data set displayed in Fig. 5.4, each of the datasets plotted in Fig. E.1 show the delay between maximum dorsiflexion and rotational work conversion. Additionally, performance metrics for each of these 10 data sets were tabulated as shown below in Table 5.1.

Figure E.1: Moment Vs. Angle Relationships for 10 Data Sets
The data sets include several key performance metrics for the device including translational work in Joules, modeled rotational work in Joules, experimental (actual) rotational work in Joules, percentage of translational work converted to rotational work, and percentage of modeled work which was manifested experimentally.

Similar to the single dataset previously discussed, the EaSY-Walk over the summation of 10 data sets averaged an input of 0.65 J of translational work and an output of 0.17 J of rotational work, yielding an average of 26.6% work conversion from translation to rotation. Additionally, the modeled device averaged 0.27 J of rotational work. Based on this, the yielded 63.05% of the modeled work suggested by the quasi-static model.
APPENDIX F

IRB PROTOCOL

Specific Procedures to be Followed:

This study will involve one visit with the study team at the Marquette University Exercise Science Motion Analysis Lab. Following completion of the consent form, the participant will perform standard gait tests using their prescribed prosthesis followed by the same test using the new prosthesis.

Specifically, the procedure will be as follows:

1. The participant will be given a pre-test survey to ascertain their current health status and to gather general perception of the performance of their current prosthesis.
2. The participant will be seen by a certified prosthetist (CP) to ensure that their current prosthesis is functional and properly aligned.
3. A typical gait analysis session (approximately 2-3 hours) will follow.
   a. Reflective markers will be placed at bony landmarks on lower extremities with double sided adhesive tape. The landmarks will be cleaned with alcohol.
   b. EMG electrodes will be placed on the skin at locations on the outer thigh and below the knee, determined in accordance with the literature, and connected to a 16-channel wireless EMG system.
   c. The participant will stand in the middle of the gait analysis walkway in a comfortable position and the cameras will film the marker locations for calibration.
   d. The participant will walk the length of the walkway (about 10 meters) at a normal comfortable pace (as well as a rigorous and leisurely pace) while the motion analysis system records data from the video cameras, force plates, and electrodes.
e. The participant will complete multiple walking trials (approximately 15-20), with several trials being completed at each of the paces described above in step 3d. They may rest between walking trials if necessary. A chair will be provided for seated rest if desired.

f. The participant will be taken to a long hallway in the building and asked to walk for 2 minutes following the standard 2 Minute Walk Test (2MWT) procedure.

g. The participant will walk on a treadmill for 3 minutes where oxygen utilization will be performed using a breath-by-breath cardiopulmonary exercise testing device. The device consists of a mask which is connected to an oxygen and carbon dioxide analyzer. The mask is mounted on the head via straps.

4. The participant will be outfitted with the newly developed prosthesis at the default deflection setting.
   a. Alignment and fit will be performed again by the CP.
   b. The patient will walk with assistance to gain familiarity using the prosthesis.
   c. The prosthesis may be removed to change springs at the judgment of the designer. Steps 4a-4b will be repeated.
   d. Step 4c may be repeated until the prosthesis is in an acceptable configuration for the walker.

5. The gait analysis protocol described in steps 3a-3e will be repeated with the new prosthesis at the current setting.

6. The prosthesis will be removed and set at a second deflection setting. Steps 4 and 5 will be repeated.

7. The prosthesis will be removed and set at a third deflection setting. Steps 4 and 5 will be repeated.

8. The prosthesis will be outfitted at the subjects preferred setting. Step 4 will be repeated if necessary. The gait analysis protocol described in steps 3f-3g will be repeated at the preferred setting of the new prosthesis.
9. The participant will have his/her original prosthesis returned and adjusted by the CP.

10. The participant will be asked to fill out an exit survey similar to the original survey to ascertain the general perception of the performance of the new prosthesis.

11. A short interview by the team will be conducted on the performance of the prosthesis compared to the original prosthesis.