Development and Evaluation of Biarticular Transtibial Prostheses for Level-Ground Amputee Walking

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Submitted to the Department of Mechanical Engineering in partial fulfillment of the requirements for the degree of

Doctor of Philosophy

at the

MASSACHUSETTS INSTITUTE OF TECHNOLOGY

February 2017

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Abstract

In the last several years, great strides have been made toward the development of transtibial prostheses that match the functionality of their corresponding biological structures. However, these devices are fundamentally limited because they only emulate the biological ankle-foot complex. In contrast, the gastrocnemius calf muscle spans both the ankle and knee joint, and thus provides biarticular function. Therefore, it may be prudent for transtibial prostheses to include actuation at both the ankle and knee joints of an affected limb.

This thesis presents the development of two such biarticular prosthesis systems. The first, tested on two participants, employed a quasi-passive clutched-spring knee orthosis, approximating the largely isometric behavior of the biological gastrocnemius. The second device, tested on six participants, provided positive net mechanical work with a tethered knee orthosis, controlled using a neuromuscular model. Both devices utilized a commercial powered ankle-foot prosthesis. Participants with unilateral transtibial amputation walked with these biarticular prostheses in two separate studies on an instrumented treadmill while motion, force, electromyographic, and metabolic data were collected. Data were analyzed to determine differences resulting from the activation of each knee orthosis, compared to the orthosis behaving as a free-joint. We hypothesized that the active conditions would reduce joint kinetic demands, and consequently metabolic cost, compared to the control conditions.

The quasi-passive system was capable of reducing both affected-side knee and hip moment impulse and positive mechanical work in both participants during the late stance knee flexion phase of walking, compared to the control condition. The metabolic cost of walking was also reduced for both participants. The powered artificial gastrocnemius reduced affected-side biological knee flexion moment impulse by 0.022 +/- 0.018 Nm/kg (p = 0.03), and affected-side hip positive work by 0.074 +/- 0.025 J/kg (p = 0.004) during late stance knee flexion, compared to the control condition. However, the data did not support our hypothesis that metabolism would decrease, as two of the participants did not display a metabolic reduction. These results highlight the kinetic benefits of an artificial gastrocnemius for transtibial amputee gait. However, further study is warranted to determine the requirements for achieving a consistent metabolic improvement.

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Acknowledgments

There are so many to whom I'd like to thank that I feel like I am giving an Oscar speech. First, I would like to thank my advisor, Professor Hugh Herr, for supporting me throughout my entire graduate career while providing an environment where I could so freely explore, experiment, and learn. Most importantly, thank you for convincing me to pursue the PhD in the first place. I look forward to doing “creative work for the rest of [my] life”, thanks to your very persuasive argument for me to stay on after my master's work.

To my committee, Professor David Trumper and Professor Sangbae Kim, your words of wisdom, encouragement, and gentle pushes in the right direction over the past several years have helped get me to this day. Thank you for your patience and understanding, especially when my research updates were less-than-perfectly clear.

I thank NASA, for funding the larger project of which I became a part these last few years. Without your funding, the entire tethered system would not have existed.

A hearty thank you to my good friends and colleagues here in Biomech over the years. You’ve made the lab more of a family than a workplace, and I’ve learned more from you than you know: Madalyn Berns, Bruce Deffenbaugh, Grant Elliott, Todd Farrell, Matt Furtney David Hill, Bevin Lin, Luke Mooney, Arthur Petron, David Sengeh, and Jing Wang. Special thanks to Jiun-Yih Kuan for letting me on the bandwagon with the simulator project and being so gracious in showing me the ropes (or should I say cables). Ken Pasch, thank you for your insight with this simulator project; you always had just the right questions to ask. Jared Markowitz, thank you for taking so much of your post-graduation time to guide me through your very complex modeling and optimization code. Ken Endo, your groundbreaking work with the first artificial gastrocnemius enabled me to get a running start. Jean-François Duval, your strain gauge amplifiers were, quite literally, instrumental. Cameron Taylor, your inexplicably enthusiastic support with clinical trials very likely saved my over-tired mind from forgetting to hit record on multiple occasions. Pavitra Krishnaswamy, Elliott Rouse, and Olli Kannape, you helped steady me during crises, and I am so grateful to have your support. Ernesto Martinez-Villalpando and Sam Au, you were two of the first people I met at the lab, and I’ve cherished both your friendship and mentorship over all these years.

To all of my study participants (you know who you are), I literally could not have done this without you. Thank you for bearing with me, through the forms, travel, boring down-time, hardware failures, grueling long sessions, and repeat visits. You are so admirable and truly inspiring, and it has been my honor to work with you.

To the Biomech admins throughout the ages, especially Tesha Myers, Sarah (Hunter) Ralston, and Lindsey Reynolds· you made me feel at home from the start, and I always felt I had a sympathetic ear in your presence.
Thank you to everyone who has worked as a UROP with me: Deema, Priya, Raul, Rachel, Laura, Dabin, and Arden. I appreciate your eagerness to help in whatever was needed.

To my friends and family everywhere (and double-counting some of Biomech here), you've helped me maintain a life outside of work, and part of the reason I'm so happy is because I have all of you in my life.

Ozzie, I know you can't read this, but I'll give you a bunch of treats after turning this in as a thank you for the countless hours you spent keeping me company while I put this dissertation together.

Finally, to my family: Steven, Heléna, Sammi, and Jenna, I never felt I could fall very far with all of you there to catch me. Thank you for always being there with your boundless love.
Contents

Chapter 1 .......................................................................................................................... 15
  1.1 Specific aims ........................................................................................................... 15
  1.2 Hypotheses .......................................................................................................... 15
  1.3 Thesis contributions ........................................................................................ 16

Chapter 2 .......................................................................................................................... 17
  2.1 Motivation ........................................................................................................... 17
  2.2 Sign conventions .............................................................................................. 20
  2.3 Normal ankle and knee biological function ..................................................... 21

Chapter 3 .......................................................................................................................... 23
  3.1 Introduction ......................................................................................................... 23
  3.2 Methods .............................................................................................................. 24
    3.2.1 Hardware ........................................................................................................ 24
      3.2.1.1 Powered Ankle-Foot Prosthesis ............................................................... 24
      3.2.1.2 Clutch-Spring Joint ............................................................................... 26
    3.2.2 Modeling and Spring Stiffness Selection .................................................... 27
      3.2.2.1 Target Biological Behavior .................................................................... 27
      3.2.2.2 Target Musculoskeletal Model .............................................................. 28
      3.2.2.3 Optimization .......................................................................................... 30
    3.2.3 Control ........................................................................................................... 31
      3.2.3.1 Control Electronics ............................................................................... 31
      3.2.3.1 Control Algorithm ............................................................................... 32
    3.2.4 Experimental Protocol .............................................................................. 36
    3.2.5 Data Processing ........................................................................................... 37
  3.3 Results .................................................................................................................... 39
    3.3.1 Modeling ....................................................................................................... 39
    3.3.2 Ankle Prosthesis Net Work ............................................................................ 39
    3.3.3 Affected Knee Kinematics .......................................................................... 40
    3.3.4 Affected Knee Kinetics ............................................................................... 41
    3.3.5 Affected Hip Kinetics ................................................................................... 43
    3.3.6 Affected Knee Electromyography ............................................................... 45
    3.3.7 Metabolism .................................................................................................. 47
  3.4 Discussion .............................................................................................................. 48
Figures

Figure 1: The BiOM powered ankle-foot prosthesis. .......................................................... 18
Figure 2: Calf muscle anatomy highlighting the gastrocnemius muscle ................. 20
Figure 3: Leg joint sign convention in the sagittal plane ........................................ 20
Figure 4: Knee function in the human gait cycle for normal level-ground walking . 22
Figure 5: Joint mechanism of the quasi-passive artificial gastrocnemius ............... 25
Figure 6: Quasi-passive artificial gastrocnemius ....................................................... 26
Figure 7: The Endo-Herr leg model using quasi-passive elements ....................... 29
Figure 8: Quasi-passive leg model with monoarticular gastrocnemius spring ...... 30
Figure 9: Example action of the clutch engagement .................................................. 31
Figure 10: Finite state machine for the quasi-passive artificial gastrocnemius ..... 33
Figure 11: Affected-side knee kinematics with the QPAG ........................................ 40
Figure 12: Affected-side knee flexion moment components with the QPAG ....... 41
Figure 13: Affected-side knee power components with the QPAG ......................... 42
Figure 14: Affected-side hip flexion moment components with the QPAG .......... 43
Figure 15: Affected hip power with the quasi-passive artificial gastrocnemius .... 44
Figure 16: Affected-side knee muscle EMG for Subject 1 with the QPAG ......... 46
Figure 17: Affected-side knee muscle EMG for Subject 2 with the QPAG .......... 47
Figure 18: Metabolic power changes with the QPAG .............................................. 48
Figure 19: Motor drive unit and tensioner ............................................................... 53
Figure 20: Conduit linkage for the powered artificial gastrocnemius ................. 53
Figure 21: Conduit element .................................................................................... 54
Figure 22: Powered artificial gastrocnemius joint mechanism ............................ 56
Figure 23: Sectional view of the powered artificial gastrocnemius joint .......... 57
Figure 24: Knee brace and socket attachment ....................................................... 58
Figure 25: Custom knee braces ............................................................................. 59
Figure 26: Custom strain gauge-based torque sensor .......................................... 60
Figure 27: Sensor calibration mounting platform ................................................... 61
Figure 28: Swept sine input for system identification .............................................. 62
Figure 29: Torque control schematic ..................................................................... 65
Figure 30: Trapezoidal torque response ................................................................. 68
Figure 31: Open-loop system response .................................................................. 70
Figure 32: Closed-loop system response ............................................................... 71
Figure 33: Metabolic power of non-amputees with the tethered knee orthosis .... 72
Figure 34: Representative closed-loop system response during walking trials .. 72
Figure 35: Hill-type muscle model ......................................................................... 81
Figure 36: Empirically-determined muscle metabolic power ............................... 90
Figure 37: Neuromuscular controller including spinal reflex loop..................... 94
Figure 38: Final cost values for the morphological parameter optimization ...... 96
Figure 39: Experimental setup with the powered Artificial Gastrocnemius ....... 98
Figure 40: Individual joint angles during walking for amputee participants .... 106
Figure 41: Components of affected-side knee moment with the powered AG .... 107
Figure 42: Net joint moments with the powered AG ................................................ 108
Figure 43: Affected-side knee extension moment impulse in late stance flexion ....110
Figure 44: Joint powers during walking with the powered AG ............................. 111
Figure 45: AG effect on affected-side hip positive power ................................. 113
Figure 46: Changes in joint positive work in late stance knee flexion .............. 114
Figure 47: EMG changes when walking with the powered AG ......................... 115
Figure 48: Percent change in hamstrings EMG with the powered AG .............. 116
Figure 49: Average changes in quadriceps EMG with the powered AG .......... 117
Figure 50: EMG changes compared to baseline activity ................................. 117
Figure 51: Affected-side ankle dorsiflexion angle with the Powered AG ........... 127
Figure 52: Affected-side knee flexion angle with the Powered AG .................. 128
Figure 53: Affected-side hip flexion angle with the Powered AG .................... 129
Figure 54: Affected-side ankle dorsiflexion moment with the Powered AG ....... 130
Figure 55: Affected-side knee flexion moment with the Powered AG ............. 131
Figure 56: Affected-side hip flexion moment with the Powered AG ............... 132
Figure 57: Affected-side ankle power with the Powered AG ............................ 133
Figure 58: Affected-side knee power with the Powered AG ............................ 134
Figure 59: Affected-side hip power with the Powered AG .............................. 135
Figure 60: Affected-side biceps femoris activity with the Powered AG .......... 136
Figure 61: Affected-side semimembranosus activity with the Powered AG ...... 137
Figure 62: Affected-side vastus lateralis activity with the Powered AG .......... 138
Figure 63: Affected-side vastus medialis activity with the Powered AG .......... 139
Tables

Table 1: Parameter values for the QPAG controller..........................................................34
Table 2: Amputee participant body parameters ....................................................................36
Table 3: Net work per step by the powered ankle-foot prosthesis with the QPAG ...39
Table 4: Affected-side peak stance knee extension with the QPAG........................................40
Table 5: Late stance affected-side knee flexion moment impulse with the QPAG .......42
Table 6: Affected-side knee positive work in late stance flexion with the QPAG...........43
Table 7: Hip flexion moment impulse in late stance flexion with the QPAG ..........44
Table 8: Affected-side hip positive work in late stance flexion with the QPAG ........45
Table 9: Affected-side semimembranosus activity with the QPAG ..............................................45
Table 10: Affected-side vastus lateralis activity with the QPAG ..............................................46
Table 11: Metabolic power of the amputee participants with the QPAG ................47
Table 12: Design parameters of the powered artificial gastrocnemius .........................69
Table 13: Non-amputee body parameters ..............................................................................77
Table 14: Muscle-specific model parameters in the musculoskeletal model.................84
Table 15: Optimized parameter bounds for the morphological optimization ..............91
Table 16: Optimization settings for the morphological optimization ...............................91
Table 17: Optimization settings for reflex optimization .......................................................93
Table 18: Amputee Matching to Non-Amputee Participants ..............................................95
Table 19: Metabolic cost of transport for non-amputee participants ..............................95
Table 20: Optimized morphological parameters .............................................................96
Table 21: Optimized neuromuscular model reflex parameters .........................................96
Table 22: Walking conditions for minimal metabolic power ........................................103
Table 23: Metabolic cost of transport using the Artificial Gastrocnemius ....................104
Table 24: Kinematic differences with the powered artificial gastrocnemius ...............105
Table 25: Joint moment impulse in late stance flexion with the powered AG ..........109
Table 26: Joint work in late stance knee flexion with the powered AG .......................112
Table 27: Mean EMG for the knee muscles for walking with the powered AG ......116
Chapter 1

Thesis Summary

1.1 Specific aims

- Develop electronics, and control system for a clutched-spring autonomous artificial gastrocnemius knee orthosis.
- Measure the biomechanical and metabolic effects of the quasi-passive artificial gastrocnemius when worn by transtibial amputees
- Develop a powered artificial gastrocnemius knee orthosis with off-board power
- Verify the metabolic transparency and kinetic efficacy of the powered gastrocnemius orthosis with non-amputees
- Collect walking data of non-amputee participants to inform neuromuscular modeling schemes
- Develop a controller, based on a neuromuscular model, to produce torque commands to the powered artificial gastrocnemius knee joint
- Evaluate the biomechanical and metabolic effects of the powered artificial gastrocnemius when worn by transtibial amputees

1.2 Hypotheses

- A biarticular transtibial prosthesis, consisting of a powered ankle-foot prosthesis and actuated knee orthosis, will reduce the joint moment impulse and positive mechanical work in the affected-side knee and hip during the late stance knee flexion phase of level-ground walking, as compared to walking with a monoarticular powered ankle-foot prosthesis with identical weight distribution.
• These kinetic changes will correspond to a reduction in metabolic cost of walking.

1.3 Thesis contributions

• The electronics and controller for a quasi-passive autonomous artificial gastrocnemius was developed

• Biological affected-side joint moment impulse and positive mechanical work in late stance knee flexion was shown to decrease for the knee and hip joints of two amputee participants wearing the quasi-passive artificial gastrocnemius, compared to the control condition with just an ankle-foot prosthesis

• Metabolic cost of walking was shown to decrease in two amputee participants wearing the quasi-passive artificial gastrocnemius compared to the control condition with just an ankle-foot prosthesis

• A robotic joint, tether, and attachment for a tethered, powered artificial gastrocnemius was developed

• A low-level torque control scheme was developed for enforcing torque commands at the tethered artificial gastrocnemius joint

• A biophysically-based controller, using a neuromuscular model, was developed, using previously-described techniques

• Biological affected-side knee moment impulse and hip positive work was shown to decrease during late stance knee flexion for six amputee participants wearing the powered artificial gastrocnemius compared to the control condition with just an ankle-foot prosthesis

• Individual transtibial amputees using the powered artificial gastrocnemius and powered ankle-foot prosthesis displayed a reduced metabolic cost of walking
Chapter 2
Background

2.1 Motivation
The loss of a leg below the knee can have a formidable impact on one’s quality of life. For those living with a leg amputation, everyday tasks such as walking, running, stair navigation, and even leaning against a wall can pose challenges. Prosthetic technology has yet to advance to a sufficient level as to fully restore the functionality of the missing biological structures. Consequently, pathologies exist in the aforementioned tasks, and quality of life is compromised. A great need exists to improve the prosthetic technology in order to enhance the quality of life for those living with amputation. In this thesis, I describe the development of two biarticular prostheses for transtibial (below-the-knee) amputees that each comprise both a powered ankle-foot prosthesis and an actuated orthosis at the affected-side knee.

The widespread, passive ankle-foot prostheses on the market today provide only a rudimentary approximation to the function of a human ankle joint. Instead of providing net mechanical work to the wearer during walking, these passive devices act at best in a spring-like manner: they can only provide as much mechanical energy return as is provided to them by the wearer, and they do not provide the articulation normally seen in the biological ankle-foot complex during walking. As evidence of this technological limitation, transtibial amputees display a variety of pathological features of their walking gaits. Specifically, transtibial amputees naturally select a 30-40% slower walking speed than those without amputation, and when walking at the same pace as a non-amputee, these amputees require 20-30% more metabolic power than their non-amputee counterparts [1]-[4]. Higher than normal levels of hip positive power at the end of stance phase has been observed as
well, which is thought to be a compensatory response to lack of calf muscle function, and may contribute to the aforementioned increase in metabolism while walking.

In the last several years, robotic advances in prosthetic technology have led to the introduction of powered ankle-foot prostheses (BiOM, BionX Medical Technologies, Inc., Bedford, MA), shown in Figure 1, which, unlike the passive conventional devices, provide levels of mechanical work comparable to those provided by the human ankle-foot complex. As a result of this functional improvement, many of the aforementioned gait pathologies have been drastically reduced; amputees using the powered prostheses have preferred walking speed, metabolic cost at a given speed, and contralateral limb impacts that are not significantly different from those of non-amputees [5], [6].

![The BiOM powered ankle-foot prosthesis.](image)

**Figure 1: The BiOM powered ankle-foot prosthesis.**
This prosthesis was the first such device to produce a significant reduction in transtibial amputee metabolic cost during walking.

Photo credit: Chris Conti Photography
These new prosthetic devices are, however, limited to acting at the ankle-foot complex alone, and consequently cannot restore the full function of the powerful gastrocnemius muscle, shown in Figure 2. The gastrocnemius provides not only a plantar flexion moment at the ankle, but also a flexion moment at the knee. Without this knee-flexing function, compensatory mechanisms are required. Indeed, transtibial amputees exhibit higher hamstring muscle activity during level-ground walking than non-amputees, possibly as an attempt to stabilize or flex the knee in place of the non-functional gastrocnemius muscle [7]. This higher muscle activity is still apparent when amputees walk with the powered ankle-foot prostheses, indicating that a monoarticular intervention at only the ankle-foot complex may not be sufficient to restore biological function. It is possible that this pathological muscle activity has detrimental effects on amputee gait. However, little work has been done to develop devices for restoring this missing gastrocnemius functionality. Such a device is an artificial gastrocnemius (AG), which is a device that provides the function of the missing biological ankle-foot complex, and also assists the affected-side knee joint by providing joint moments and powers resembling those from the biological gastrocnemius muscle.

This thesis explores two AG mechanisms. The first was a continuation of the work by [8] which used a clutched-spring at the knee joint, whereas the second was able to provide net mechanical work at the knee joint, in an attempt to more accurately reproduce biological gastrocnemius function. We hypothesized that, the applied knee torque by the AG devices would allow for a reduction in affected-side biological knee flexion moment by the amputees, during the latter half of the stance period, when the knee is flexing. Further, we hypothesized that the affected-side hip positive work would decrease in this same part of the stance phase, as a result of the combination of ankle push-off and knee flexion assistance.
2.2 Sign conventions

In this thesis, angles are defined for each of the three major joints of the human leg as shown in Figure 3. The ankle joint angles were defined as positive in dorsiflexion, where the toe pointed up toward the sky. The zero-angle for the ankle
was that at which the foot was perpendicular to the shank. The knee angles were
defined as positive in flexion, with a straight-leg corresponding to the zero-angle.
The hip angles were defined as positive in flexion, where the thigh was swung
forward with respect to the torso. The hip zero-angle was defined as that when the
leg was pointing straight down, in line with the torso. Joint moments were defined
in the same direction as the joint angles; positive moments indicated an internal
moment tending to rotate the joint in a positive direction.

2.3 Normal ankle and knee biological function

For the purposes of this thesis, we restrict our analysis to the sagittal plane. This
plane is one that divides the body into right and left halves. Thus, motions in the
sagittal plane are those most grossly associated with forward motion, and rotation
of the lower extremities to propel the body along its path.

Human walking may be analyzed in a periodic sense, as it is cyclic by nature.
Walking gait is typically analyzed by using a representative period, or gait cycle
(GC). A gait cycle for level-ground walking is most often represented as the period
from heelstrike to the following heelstrike of a given limb. The gait cycle can be
divided into two phases: stance phase (~60% of the GC) is defined as when the foot
is on the ground, from heelstrike to toeoff, and swing phase (~40% of the GC) is
defined as when the foot is off the ground, from toeoff to the subsequent heelstrike.

The stance phase may be further divided into sub-phases. For the purposes of this
thesis, the knee behavior is described. After heelstrike, during loading response, the
knee flexes in early stance flexion as it helps to absorb the initial impact, while
providing an extensor moment for support. The knee then extends during mid-
stance, in knee extension as the torso progresses up and over the stance limb, and
the knee begins to produce a flexion moment to slow this progression. Finally, in
late stance flexion, defined from mid-stance maximum knee extension to toeoff, the
knee flexes in preparation for lifting the leg into the air for the swing phase.
This last phase of stance, *late stance flexion*, is particularly important in the present work, since transtibial amputees walking with conventional prostheses exhibit increases in hip power during this phase of gait, possibly in response to the lack of power from the calf muscles, including the gastrocnemius [4].

Figure 4: Knee function in the human gait cycle for normal level-ground walking
Chapter 3
Quasi-passive Artificial Gastrocnemius

3.1 Introduction
In order to develop appropriate assistive devices, it is important to first understand how the gastrocnemius muscle functions. In-vivo ultrasonography shows that the gastrocnemius muscle fascicle length is largely isometric during the early and mid-stance phases of level-ground walking [9], indicating that the passive tissues such as the Achilles tendon are responsible for much of the power delivery from the gastrocnemius. Hence, the gastrocnemius may be approximated by a clutched-spring, with the spring representing the compliant tendinous structures, and the clutch representing the isometrically-acting muscle fibers. Indeed, a model of human gait by Endo and Herr with only spring-clutch structures at the knee joint was able to achieve human-like metabolic economy while capturing the dominant kinetic features of human walking [10]–[12]. It is therefore possible that a physical clutch in series with a spring can provide much of the missing knee function of the gastrocnemius in transtibial amputees.

Researchers have begun to study the effects of a robotic device at the affected-side knee to provide the knee flexion moment of the missing gastrocnemius. In the first such study [8], a knee orthosis, dubbed an artificial gastrocnemius (AG) was built, based on the spring-clutch representation of the gastrocnemius in the Endo-Herr model. This device was quasi-passive, in that it did not provide net positive mechanical power to the wearer, but still required an active controller and electrical power to operate. The AG was physically realized as clutchable rotary spring at the orthosis knee joint. Two copies of this AG, worn along with a powered-ankle-foot prosthesis (PAFP), was tested on both legs of one bi-lateral transtibial amputee, with promising results: the amputee’s metabolic energy expenditure was reduced
with the AG-PAFP condition, compared to the amputee walking at the same speed with only conventional leaf-spring-type ankle-foot prostheses. However, since the PAFP was not tested independently of the AG, it is not clear what incremental effects the AG had over the PAFP alone. A need, therefore, exists, to evaluate the incremental biomechanical and metabolic effects of the AG knee component.

In this study, we hypothesized that applying a clutched-spring at the affected-side knee joint of transtibial amputees, in conjunction with a powered ankle-foot prosthesis, would reduce biological knee flexion moment of the affected-side knee joint, as compared to the same conditions with only the ankle-foot prosthesis with identical mass distribution. We expected that this reduction in knee flexion moment would stem from lower knee and hip moments and powers and associated reductions in hamstring muscle activity, and we further hypothesized that these changes would provide a corresponding reduction in metabolic cost of walking.

3.2 Methods

3.2.1 Hardware

The quasi-passive artificial gastrocnemius (QPAG) comprised a powered ankle-foot prosthesis to provide ankle function in the sagittal plane, and a clutched-spring joint, mounted on a knee orthosis on the affected-side knee. Together, these two devices represented a bi-articular prosthesis, which could act at both joints independently.

3.2.1.1 Powered Ankle-Foot Prosthesis

A BiOM powered, ankle-foot prosthesis (BionX, Bedford, MA), shown in Figure 1, was used as the prosthesis for all clinical trials. This prosthesis had the capability of providing positive net power at levels comparable to the human ankle-foot complex [5]. At the core of the prosthesis was a series elastic actuator, comprising a brushless DC motor, ballscrew, and carbon fiber leaf spring. For dorsiflexion angles,
a rigid hard-stop engaged, both to reduce the torque required by the motor in dorsiflexion, and to act as a safety feature in the event of a power failure. The total mass of the prosthesis and battery was 1.8 kg.

The prosthesis controller employed a positive force feedback strategy, which served to approximate the biological muscle dynamics and neural reflexes. A wireless communication link enabled real-time tuning of the control parameters.

Figure 5: Joint mechanism of the quasi-passive artificial gastrocnemius
The exploded view of the joint design (bottom) is shown along with a rendering of the assembled joint without the cover (top).
3.2.1.2 Clutch-Spring Joint

The QPAG, shown in Figure 6, was a knee orthosis, comprising a pair of polycentric hinges that connected a thigh cuff to an aluminum bracket. The bracket connected to the base of a participant's socket, between the prosthetic pylon attachment and the socket. A custom, dog-tooth rotary clutch-spring with series compliance was attached lateral to the knee, in parallel with the hinges.

This clutch-spring joint was based on the design by [13], but had the addition of series compliance. It included two rings of 90 dog teeth each, brought together by the action of a solenoid. As in Figure 5, the clutch controlled the relative rotation of the Housing to the Rotating Clutch Ring. A solenoid was built into the middle of the joint, and translated in the medial-lateral direction under action of its actuation power and a return spring. When the solenoid was energized, the Translating Clutch Ring was engaged with the Rotating Clutch Ring. The translating clutch plate constrained the rotation of the Rotating Plate, which in turn was connected to the Housing through two tangential linear compression springs. Thus, when the solenoid was energized, the Housing was coupled to the Rotating Clutch Ring.
through the springs, and when the solenoid was inactive, these two parts were free to rotate with respect to each other.

The Housing of the clutch was bolted to the thigh cuff of the knee orthosis. The Rotating Clutch Ring connected to an output link with a linear ball bearing and radial ball bearing, acting in series. These bearings accommodated for the kinematic difference between the polycentric hinges of the orthosis and the single-axis rotation of the clutch joint (Figure 6). The total weight of the orthosis, including battery and electronics, was 1.9 kg.

The QPAG had onboard sensing for use in the control algorithm. Knee angle of the QPAG was measured using a 10 kOhm rotary potentiometer. Torque provided by the spring-clutch element was estimated by measuring the deflection of one of the two identical tangential springs with an 8 kOhm linear potentiometer. Prosthesis-side ground contact was detected using a resistive pressure sole footswitch (model: FSW, B&L Engineering, Santa Ana, CA), inserted into the shoe between the prosthetic foot and the insole.

3.2.2 Modeling and Spring Stiffness Selection
The selection of the spring constants for the tangential springs allowed for the control of rotary stiffness of the clutch-spring element. It was desirable for the stiffness of the clutch-spring joint to be such that the spring-clutch behavior of the artificial gastrocnemius would most closely reproduce the gastrocnemius muscle behavior of a healthy gastrocnemius. To this end, walking data from healthy non-amputees was used to select the spring stiffness values for the spring-clutch joint.

3.2.2.1 Target Biological Behavior
Kinetic and kinematic walking data were collected at the Gait Laboratory of Spaulding Rehabilitation Hospital, Harvard Medical School, in a study approved by the Spaulding committee on the Use of Humans as Experimental Subjects. A
healthy adult male (81.9 kg weight, 1.89 m height) was asked to walk at self-selected walking speeds across a 10 m walkway in the motion capture laboratory after informed consent was given. The motion-capture was performed using a VICON 512 motion-capture system, comprising eight infrared cameras. Reflective markers were placed at 33 locations on the subjects' bodies in order to allow the infrared cameras to track said locations during the trials. The cameras were operated at 120 Hz and were able to track a given marker to within approximately 1 mm. The markers were placed at the following bony landmarks for tracking the lower body: bilateral anterior superior iliac spines, posterior superior iliac spines, lateral femoral condyles, lateral malleoli, forefeet and heels. Wands were placed over the tibia and femur, and markers were attached to the wands over the mid-shaft of the tibia and the mid-femur. Markers were also placed on the upper body at the following sites: sternum, clavicle, C7 and T10 vertebrae, head, and bilaterally on the shoulder, elbow, and wrist joints. Ground reaction forces were measured using two staggered force plates (model no. 2222 or OR6·5·1, by Advanced Mechanical Technology Inc. Watertown, MA, USA), which were incorporated into the walkway. The precision of these force plates in measuring ground reaction force and center of pressure was approximately 0.1 N and 2 mm respectively. The force plate data was collected at 1080 Hz and synchronized with the VICON motion capture data.

3.2.2.2 Target Musculoskeletal Model
A simulation of human walking was used to select the desired knee spring stiffness of the QPAG. The parameters were optimized for the Endo-Herr sagittal-plane human leg model [11], [12], [14] using the method described in [8]. In this model, all muscle-tendon units, with the exception of the monoarticular hip flexors and extensors, and the monoarticular ankle plantar flexor were modeled as unidirectional spring-clutch elements (Figure 7), based on the hypothesis that these muscles largely behave isometrically during walking at a self-selected speed. The
remaining muscle-tendon units were modeled as unidirectional force sources in series with springs. When a given clutch was disengaged, it exerted no force, and the joint was allowed to rotate freely without any resistance from the corresponding spring element. Conversely, when a given clutch was engaged, its series spring would produce a force proportional to the spring deflection if the spring-length was greater than the spring’s rest-length, and zero force otherwise. The spring forces were applied to the respective joints via moment arm lengths from literature [15]. For the purposes of the AG spring selection, this model was modified to simulate a transtibial amputation. The ankle joint was removed, and the biarticular gastrocnemius spring-clutch element was modified to be a monoarticular knee flexor, as a representation of the QPAG knee joint (Figure 8).

Figure 7: The Endo-Herr leg model using quasi-passive elements
Figure credit: Ken Endo PhD thesis [14].
Here, the biarticular gastrocnemius element was made into a monoarticular element for the purposes of modeling the knee component of the quasi-passive artificial gastrocnemius device. Figure credit: Ken Endo PhD thesis [14].

3.2.2.3 Optimization

The parameters of the modified musculoskeletal model were selected with an optimization aimed at producing simulated joint moments at the knee and hip that best matched those moments from non-amputee target walking data. The optimized parameters of the model included seven spring constants and five spring rest lengths. Joint kinematics for the knee and hip were used as input to the model, while the parameters were optimized to minimize the following cost function, which was minimized subject to the condition that the net hip moment from the simulation matched the biological moment data throughout the gait cycle.

\[
f(x) = a \sum_{j=1}^{2} \sum_{i=1}^{100} \frac{\tau_{bio}^{i,j} - \tau_{sim}^{i,j}}{\tau_{bio}^{i,j}} + W^{+}_{act}
\]

Here, \(x\) is a vector of the 12 stiffness and engagement parameters, \(\tau_{bio}^{i,j}\) and \(\tau_{sim}^{i,j}\) are the joint moments applied about joint \(j\) at the \(i\)th percentage of the gait cycle from the biological data and the model, respectively. The parameter \(W^{+}_{act}\) is the total
positive mechanical work performed by the hip force sources during the gait cycle, and \(a\) is a weighing coefficient. The value of \(a\) was made as low as possible (0.5) while not greatly reducing the \(R^2\) values.

### 3.2.3 Control

The quasi-passive artificial gastrocnemius produced no positive net mechanical work to the wearer, but still required a control system to determine the appropriate times to engage and disengage the clutch. This controller took knee joint angle and stance information as input, and engaged the clutch appropriately through each gait cycle. The clutch was controlled to engage at maximum stance knee flexion angle, enabling the spring to store energy during the subsequent knee extension and flexion into swing phase, as shown in Figure 9.

![Figure 9: Example action of the clutch engagement.](image)

A typical knee flexion angle profile is shown for level-ground walking. The dashed red section indicates the region in the gait cycle when the clutch in the QPAG would be engaged.

### 3.2.3.1 Control Electronics

The computer platform for controlling the QPAG was a commercial single-board computer (model: Raspberry Pi Version B, Raspberry Pi Foundation,
Cambridgeshire, UK). The computer was equipped with an 800 MHz ARM11 processor, 512 MB SDRAM, with Linux Debian.

3.2.3.1 Control Algorithm

Finite State Machine

High-level control was implemented using a finite state machine, implemented in Python. The gait cycle was divided into four states: (1) SWING; (2) EARLY STANCE; (3) CLUTCHED; (4) LATE STANCE (Figure 10).

The SWING state was triggered from any other state when the affected-side foot left the ground, as detected by a drop in footswitch signal, $FSW$, to less than a fraction, $foot_{SW}$ of the maximum possible signal. During SWING, the clutch was disabled, allowing the knee to swing freely.

The EARLY STANCE state was triggered from the SWING state at foot contact of the affected-side with the ground, defined as FSW increasing to the stance threshold, $foot_{ST}$, provided that the time elapsed in the current state, $t_{state}$, was at least the minimum time required for the SWING state, $t_{swing}$. During the EARLY STANCE state, the clutch was disabled, and the controller monitored the knee angle of the brace for maximum stance knee flexion, at which point the CLUTCHED state would be engaged. A least-squares algorithm, similar to the one used previously [16], continually predicted the time remaining before maximum knee flexion angle. This prediction provided time to initiate the engagement of the clutch, so that the clutch would be fully engaged as close as possible to the time of maximum knee flexion. In addition to this knee flexion detection algorithm, two safety features were in place to ensure that the motions detected were resulting from a walking gait. The CLUTCHED state could only be enabled if the following two conditions were also met: 1) the amount knee was flexed a minimum angle of $\theta_{ES}$ from the angle at heelstrike, $\theta_{HS}$, 2) the maximum knee flexion angular velocity, $\dot{\theta}_{max}$ measured during the EARLY STANCE state was at least $\dot{\theta}_{ES}$. The values $\theta_{ES}$
and $\theta_{ES}$ were experimentally determined during early testing as the lowest values that successfully prevented false-triggering of the \textit{CLUTCH} state outside of a steady, level-ground walking gait.

![Finite state machine for the quasi-passive artificial gastrocnemius](image)

**Figure 10:** Finite state machine for the quasi-passive artificial gastrocnemius

The \textit{CLUTCHED} state served to activate the clutch near the maximum knee flexion during stance phase of walking. This state was activated when the following criteria were met: 1) the maximum stance flexion angle was predicted to occur within the time $t_{delay}$ of the current time, 2) the knee angle was flexed more than a threshold, $\theta_{ES}$, and 3) the maximum knee flexion angular velocity, $\dot{\theta}$ during the \textit{EARLY STANCE} state exceeded a threshold $\dot{\theta}_{ES}$. Conditions 2 and 3 were used to differentiate a walking gait with slow, non-gait motions, the latter of which did not warrant activation of the clutch.
The *LATE STANCE* state served to turn off the clutch after it engaged. Once the clutch spring began to develop force, the clutch teeth would bind, preventing the clutch from disengaging until the spring force dropped sufficiently. Therefore, to save electrical power and to allow a smoother clutch disengagement, the clutch solenoid was deactivated by entering the *LATE STANCE* state when the time elapsed in the current, *CLUTCH*, state, \( t_{state} \), exceeded the clutch timeout threshold, \( t_{cl} \).

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Description</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>( f_{ootST} )</td>
<td>Footswitch threshold for entering the <em>EARLY STANCE</em> state; fraction of the maximum possible value</td>
<td>0.2</td>
</tr>
<tr>
<td>( f_{ootSW} )</td>
<td>Footswitch threshold for entering the <em>SWING</em> state</td>
<td>0.15</td>
</tr>
<tr>
<td>( t_{swing} )</td>
<td>Minimum time in the <em>SWING</em> state before it is possible to exit <em>SWING</em></td>
<td>200 ms</td>
</tr>
<tr>
<td>( \theta_{ES} )</td>
<td>Minimum knee flexion angle (rad) during the <em>EARLY STANCE</em> state before the clutch can be engaged</td>
<td>0.087 rad</td>
</tr>
<tr>
<td>( \dot{\theta}_{ES} )</td>
<td>Minimum knee flexion angular velocity that must be observed during the <em>EARLY STANCE</em> state before the clutch can be engaged</td>
<td>1.2 rad/s</td>
</tr>
<tr>
<td>( t_{delay} )</td>
<td>Delay time required between activation of the solenoid and the full engagement of the clutch</td>
<td>50 ms</td>
</tr>
<tr>
<td>( t_{cl} )</td>
<td>Maximum time to energize the solenoid</td>
<td>400 ms</td>
</tr>
</tbody>
</table>

**Table 1: Parameter values for the QPAG controller**

**Prediction Algorithm**

In order to maximize spring energy storage and return, it was desirable to engage the clutch as near as possible to the moment of peak knee flexion in the *EARLY STANCE* state. In fact, the clutch solenoid needed to be engaged slightly before the desired clutching time, as to account for the time required to close the gap between the two sets of clutch teeth. To achieve the necessary prediction of peak stance flexion, a look-ahead algorithm was used.
First, the knee angle during the *EARLY STANCE* state was approximated as parabolic, given this assumption, the location of the vertex of the parabola may be found by performing a linear fit to the knee angular velocity data, via running sums, and solving for the zero-crossing. The algorithm used here was similar to one described by [16]. However, the earlier algorithm assumed a fixed time-step, which was not applicable with the system in this study, as the computer platform was not a truly real-time system. Hence, the algorithm proposed here did not presume a fixed time step.

The parameter $\beta = [\beta_0 \beta_1]^T$ that minimizes the error in the least squares sense of linear model $\dot{\gamma}(t) = \beta_0 + \beta_1 x(t)$ to time-series data $(x(t), y(t))$ is

$$\beta = (X^T X)^{-1} X^T y$$

(2)

where the matrix $X$ has elements $X_{i,j} = \frac{\partial y}{\partial \beta_j} = \begin{cases} 1, & j = 0 \\ x_i, & j = 1 \end{cases}$

Expanding Equation 2 yields

$$\beta = \frac{[\Sigma x^2 \Sigma y + \Sigma x \Sigma x y]}{W \Sigma x^2 - (\Sigma x)^2}$$

(3)

For the task of estimating the angular velocity over time, the values of $x$ were timestamps, and the values of $y$ were knee angular velocity values. For the window of size $W$, the most recent $W$ values of $x$ and $y$ were maintained in a queue. At each new timestep, the oldest values of $x$ and $y$ were popped from the queue while the current $x$ and $y$ values were added to the queue. This method allowed the sums to be updated each timestep without needing to store all values of the computed sums. Finally, the estimated time of the angular velocity zero-crossing was $\dot{t}_{\text{flex}} = -\beta_1/\beta_0$. It was found experimentally that the minimum error in clutching-time between the initiation of the development of clutch torque and the peak knee flexion angle occurred when the clutch was engaged 50 ms prior to the predicted maximum knee angle. As a result, the time delay parameter, $t_{\text{delay}}$, was set to 50 ms.
3.2.4 Experimental Protocol

Two participants with below-knee amputation were involved in this study (Table 2). Both participants had right-side unilateral amputation and were of generally good health. The clinical evaluation was conducted at MIT (Cambridge, MA) and was approved by MIT’s Committee on the Use of Humans as Experimental Subjects (COUHES). Each participant provided written, informed consent was obtained from before data collection was initiated.

<table>
<thead>
<tr>
<th>Height (cm)</th>
<th>Weight (kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1 (S1)</td>
<td>180</td>
</tr>
<tr>
<td>Subject 2 (S2)</td>
<td>193</td>
</tr>
</tbody>
</table>

Table 2: Amputee participant body parameters

An infrared camera system (model T40s, Vicon Motion Systems Ltc, Oxford, UK) was used to track the three dimensional motion, recorded at 100 Hz, of reflective markers, placed at 47 anatomical locations on the participants’ bodies, based on the Helen Hayes marker model. Ground reactions forces and center of pressure locations were measured using a dual-belt instrumented treadmill (Bertec Corporation, Columbus, OH) with a sampling rate of 1 kHz. Electromyographic (EMG) signals were collected at 2 kHz using a wireless surface system (Trigno, Delsys Inc, Natick MA) for the semimembranosis, and vastus lateralis muscles. The net metabolic cost of walking during each condition was estimated using standard open-circuit gas exchange techniques (model: K4b2, COSMED, Rome, Italy).

At the beginning of each session, the participant was asked to don the powered prosthesis in place of their conventional prosthesis. The knee orthosis of the QPAG was then affixed to the prosthesis and donned by the participant. A short time period was given to ensure that both the prosthesis and the artificial gastrocnemius orthosis were functioning properly.

For each participant, the prosthesis controller’s power setting was adjusted using the commercial tuning app as the participant walked over a treadmill at 1.25 m/s so
as to achieve net work from the prosthesis per step that was within one standard deviation of the mean for non-amputees walking at the same walking speed [17]. It was verified that the level of prosthesis net work remained within the desired range for both walking conditions for a given participant (0.045 to 0.16 J/kg) and, more importantly, that this level of net work stayed reasonably consistent across walking conditions for each participant.

Both the powered prosthesis and artificial gastrocnemius were worn for all trials. The participants were asked to perform one standing trial to measure standing metabolism. Walking trials were performed on the treadmill at a speed of 1.25 m/s.

Two walking conditions were tested: 1) a baseline condition (BASELINE) in which the QPAG acted as a free-joint at the knee with the spring-clutch disabled, and 2) an active condition (ACTIVE), in which the QPAG was controlled as a clutched spring at the knee with the described control algorithm. For both conditions, the powered ankle-foot prosthesis was active. The BASELINE condition represented a monoarticular transtibial prosthesis, as the knee joint was a free-joint when the clutch was inactive. Yet the mass distribution of the device was identical to the ACTIVE condition. Thus, a direct comparison could be made to determine the incremental effects of the clutched-spring knee joint.

### 3.2.5 Data Processing

Fourth-order Butterworth filters were used to filter the marker position and ground reaction force data with 6 Hz and 25 Hz cutoff frequencies, respectively. The marker and force data were post-processed through the SIMM (Musculographics Inc., Evanston, IL) inverse dynamics module to produce joint moments and angles in three dimensions. Only sagittal plane dynamics were considered. Affected-side biological knee moment contribution was computed by subtracting the measured QPAG knee moment from the total knee moment estimated from the inverse dynamics.
Gait events were determined using vertical ground reaction force data from the embedded forceplates. Approximate event timing was found by determining the times when the force increased beyond a 40 N threshold. Exact heelstrike and toeoff times were found by progressing backward, and forward in time, respectively, until the force value dropped to zero. Data were then cut to gait cycles based on heelstrike times, and resampled to 101 points. Gait cycles were discarded for the beginning and end of each trial, during the speed transients of the treadmill. Gait cycles in which the stride times were below 0.7 seconds or above 1.3 seconds, or in which a foot crossed the mid-line of the treadmill were also discarded.

Joint powers were computed as the product of joint moments and joint velocities from SIMM-derived joint kinematics, where positive power was defined as that produced by the joint on the environment. Positive joint work during late stance knee flexion was computed as the positive contribution of the time-integral of joint power from maximum stance knee flexion to toeoff. This region of the gait cycle was chosen for analysis because, as the knee flexes from the maximum extension angle, it provides an opportunity for the AG to provide positive power to the wearer. Joint flexion moment impulse was computed using the same integral for joint flexion moment. Net prosthesis work was computed by integrating the SIMM-derived joint torque over joint angle over the gait cycle.

Amputee EMG signals were first processed internally by the EMG system by applying a 4th-order bandpass filter with a pass band from 20 Hz to 450 Hz. Any DC offset was removed, and the signals were then rectified, and smoothed using a 3 Hz, 3rd-order Butterworth filter. The EMG data were then cut to gait cycles and averaged in the same manner as the motion and force data. The resulting EMG data were then normalized so that the average values of the activation signals matched those from [18] during the range of activation for the literature data. For each subject and walking condition, the EMG signals were averaged over the period, when the spring-clutch was, on average, engaged and generating torque.
Metabolic cost for each walking speed was computed by taking average oxygen and carbon dioxide data over a two-minute window at the end of each six-minute trial. The metabolic power was computed using the equation

\[ P = K_{O_2} \dot{V}_{O_2} + K_{CO_2} \dot{V}_{CO_2} \]  

(4)

where \( P \) is the metabolic power in Watts, \( \dot{V}_{O_2} \) is the volume flow rate of Oxygen inhaled, \( \dot{V}_{CO_2} \) is the volume flow rate of carbon dioxide exhaled, and \( K_{O_2} \) and \( K_{CO_2} \) are constants with values from literature [19], given as \( K_{O_2} = 16,580 \) W/L and \( K_{CO_2} = 4,510 \) W/L. The above equation is only valid for conditions when the metabolism is primarily aerobic. As a verification of this condition, the respiratory exchange ratio (RER), defined as \( \dot{V}_{CO_2}/\dot{V}_{O_2} \) was monitored, and only metabolic results with RER values less than 1.1 were considered.

3.3 Results

3.3.1 Modeling

The spring stiffness of the clutched-spring gastrocnemius element, derived from the modeling, was 101 Nm/rad [8]. To produce this rotary stiffness about the knee joint, two linear compression coil springs were selected, each with 51 N/mm stiffness. These springs acted on the joint with a moment arm of 32 mm, causing the equivalent joint rotary stiffness to be 105 Nm/rad.

3.3.2 Ankle Prosthesis Net Work

As shown in Table 3, the net work produced by the powered ankle-foot prosthesis was within the desired range from literature.

<table>
<thead>
<tr>
<th>Subject</th>
<th>BASELINE (J/kg)</th>
<th>ACTIVE (J/kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.113 +/- 0.028</td>
<td>0.130 +/- 0.024</td>
</tr>
<tr>
<td>Subject 2</td>
<td>0.077 +/- 0.016</td>
<td>0.082 +/- 0.019</td>
</tr>
</tbody>
</table>

Table 3: Net work per step by the powered ankle-foot prosthesis with the QPAG

The net work values were kept within the literature range of 0.045 to 0.16 J/kg.
3.3.3 Affected Knee Kinematics

Plots of the affected-side knee angle are shown in Figure 11. A reduction in peak knee extension angle was observed during mid-stance phase for both subjects (average angle reduction of 0.047 radians), with the ACTIVE condition producing an increased flexion angle, compared to the BASELINE condition.

<table>
<thead>
<tr>
<th>Subject</th>
<th>BASELINE (rad)</th>
<th>ACTIVE (rad)</th>
<th>Difference (rad)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.184 +/- 0.030</td>
<td>0.240 +/- 0.027</td>
<td>+0.056</td>
</tr>
<tr>
<td>Subject 2</td>
<td>0.019 +/- 0.015</td>
<td>0.057 +/- 0.024</td>
<td>+0.038</td>
</tr>
</tbody>
</table>

Table 4: Affected-side peak stance knee extension with the QPAG.

Peak angle values for both walking conditions are shown, as well as a difference value, which was computed by subtracting the BASELINE value from the ACTIVE value.

Figure 11: Affected-side knee kinematics with the QPAG shown for the BASELINE condition where the clutch was disabled (thick solid blue line), the active condition (thin solid red line). For reference, biological knee moment data are also shown for the non-amputee from which the clutch spring was tuned (thick black dashed line) +/- 1 standard deviation (black dotted lines). The vertical lines indicate the typical engagement and disengagement times for the clutch.
3.3.4 Affected Knee Kinetics

Affected-side knee moments for both participants are shown in Figure 12. As summarized in Table 5. The biological component of knee flexion moment impulse during late stance knee flexion phase was reduced when subjects walked with the ACTIVE condition of the QPAG, compared to the BASELINE condition. The average reduction in biological knee moment impulse in this phase was 0.0105 Nm*s/kg.

Affected-side knee power is shown in Figure 13, and knee positive mechanical work in late stance knee flexion is summarized in Table 6. The biological positive mechanical work was reduced with both subjects when in the ACTIVE condition, as compared to the BASELINE condition.

![Figure 12: Affected-side knee flexion moment components with the QPAG](image)

Figure 12: Affected-side knee flexion moment components with the QPAG shown for the BASELINE condition where the clutch was disabled (thick solid blue line), the ACTIVE condition (thin solid red line), and the biological knee moment contribution during the ACTIVE condition (thin green dashed line). Shaded regions for these curves indicate +/- 1 standard deviation. For reference, biological knee moment data are also shown for the non-amputee from which the clutch spring was tuned (thick black dashed line) +/- 1 standard deviation (black dotted lines). The vertical lines indicate the typical engagement and disengagement times for the clutch.
Table 5: Late stance affected-side knee flexion moment impulse with the QPAG compared between the two walking conditions with the quasi-passive artificial gastrocnemius: the BASELINE condition, where the artificial gastrocnemius acted as a free-joint at the knee, and the ACTIVE condition, where the AG was appropriately engaged as per the controller. The values were averaged over the late stance knee flexion phase of the gait cycle.

<table>
<thead>
<tr>
<th>Subject</th>
<th>BASELINE moment impulse (Nm*s/kg)</th>
<th>ACTIVE moment (Nm*s/kg)</th>
<th>ACTIVE Biological moment (Nm*s/kg)</th>
<th>Total % Change from BASELINE</th>
<th>Biological % Change from BASELINE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.006 +/- 0.004</td>
<td>0.009 +/- 0.004</td>
<td>0.001 +/- 0.002</td>
<td>+32</td>
<td>-.78</td>
</tr>
<tr>
<td>Subject 2</td>
<td>0.055 +/- 0.018</td>
<td>0.056 +/- 0.013</td>
<td>0.039 +/- 0.011</td>
<td>+2</td>
<td>-.29</td>
</tr>
</tbody>
</table>

Figure 13: Affected-side knee power components with the QPAG for the BASELINE condition where the clutch was disabled (thick solid blue line), the ACTIVE condition (thin solid red line), and the biological knee power contribution during the ACTIVE condition (thin green dashed line). Shaded regions for these curves indicate +/- 1 standard deviation. For reference, biological knee power data are also shown for the non-amputee from which the clutch spring was tuned (thick black dashed line) +/- 1 standard deviation (black dotted lines). The vertical lines indicate the typical engagement and disengagement times for the clutch.
### Table 6: Affected-side knee positive work in late stance flexion with the QPAG

<table>
<thead>
<tr>
<th>Subject</th>
<th>Total power (Nm*s/kg)</th>
<th>Biological power (Nm*s/kg)</th>
<th>Total % Change from BASELINE</th>
<th>Biological % Change from BASELINE</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.006 +/- 0.004</td>
<td>0.002 +/- 0.002</td>
<td>-6</td>
<td>-60</td>
</tr>
<tr>
<td>2</td>
<td>0.084 +/- 0.064</td>
<td>0.049 +/- 0.030</td>
<td>-26</td>
<td>-41</td>
</tr>
</tbody>
</table>

Compared between the two walking conditions with the quasi-passive artificial gastrocnemius: the BASELINE condition, where the artificial gastrocnemius acted as a free-joint at the knee, and the ACTIVE condition, where the AG was appropriately engaged as per the controller. The values were averaged over the late stance knee flexion phase of the gait cycle.

#### 3.3.5 Affected Hip Kinetics

Hip flexion moment is shown in Figure 14, and hip flexion moment impulse values are summarized in Table 7. Hip power is shown in Figure 15, and hip work values for both participants are shown in Table 8. As with the knee joint, both hip flexion moment impulse and positive mechanical work during late stance flexion were decreased for both participants.

---

**Figure 14:** Affected-side hip flexion moment components with the QPAG shown for the BASELINE condition where the clutch was disabled (thick solid blue line), the ACTIVE condition (thin solid red line), and the biological hip moment contribution during the ACTIVE condition (thin green dashed line). Shaded regions for these curves indicate +/- 1 standard deviation. For reference, biological hip moment data are also shown for the non-amputee from which the clutch spring was tuned (thick black dashed line) +/- 1 standard deviation (black dotted lines). The
vertical lines indicate the typical engagement and disengagement times for the clutch.

<table>
<thead>
<tr>
<th></th>
<th>BASELINE positive work (J/kg)</th>
<th>ACTIVE positive work (J/kg)</th>
<th>% Change from BASELINE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.0768 +/- 0.006</td>
<td>0.0567 +/- 0.006</td>
<td>-26</td>
</tr>
<tr>
<td>Subject 2</td>
<td>0.134 +/- 0.0134</td>
<td>0.115 +/- 0.012</td>
<td>-14</td>
</tr>
</tbody>
</table>

Table 7: Hip flexion moment impulse in late stance flexion with the QPAG compared between the two walking conditions with the quasi-passive artificial gastrocnemius: the BASELINE condition, where the artificial gastrocnemius acted as a free-joint at the knee, and the ACTIVE condition, where the AG was appropriately engaged as per the controller. The values were averaged over the late stance knee flexion phase of the gait cycle.

Figure 15: Affected hip power with the quasi-passive artificial gastrocnemius is shown for the BASELINE condition where the clutch was disabled (thick solid blue line), the ACTIVE condition (thin solid red line), and the biological knee moment contribution during the ACTIVE condition (thin green dashed line). Shaded regions for these curves indicate +/- 1 standard deviation. For reference, biological knee moment data are also shown for the non-amputee from which the clutch spring was tuned (thick black dashed line) +/- 1 standard deviation (black dotted lines). The vertical lines indicate the typical engagement and disengagement times for the clutch.
Table 8: Affected-side hip positive work in late stance flexion with the QPAG compared between the two walking conditions with the quasi-passive artificial gastrocnemius: the BASELINE condition, where the artificial gastrocnemius acted as a free-joint at the knee, and the ACTIVE condition, where the AG was appropriately engaged as per the controller. The values were averaged over the late stance knee flexion phase of the gait cycle.

<table>
<thead>
<tr>
<th>Subject</th>
<th>BASELINE positive work (J/kg)</th>
<th>ACTIVE positive work (J/kg)</th>
<th>% Change from BASELINE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.087 +/- 0.011</td>
<td>0.063 +/- 0.009</td>
<td>-27</td>
</tr>
<tr>
<td>Subject 2</td>
<td>0.117 +/- 0.016</td>
<td>0.095 +/- 0.014</td>
<td>-19</td>
</tr>
</tbody>
</table>

Table 9: Affected-side semimembranosus activity with the QPAG. Values are of normalized EMG, where 0 indicates no activation and 1 indicates maximal activation. Averages were taken over the period when the clutch was active, on average, during the ACTIVE condition. The values were averaged over the late stance knee flexion phase of the gait cycle.

<table>
<thead>
<tr>
<th>Subject</th>
<th>BASELINE (Normalized)</th>
<th>ACTIVE (Normalized)</th>
<th>% Change from BASELINE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.150 +/- 0.033</td>
<td>0.161 +/- 0.039</td>
<td>+7</td>
</tr>
<tr>
<td>Subject 2</td>
<td>0.216 +/- 0.051</td>
<td>0.133 +/- 0.042</td>
<td>-39</td>
</tr>
</tbody>
</table>
Table 10: Affected-side vastus lateralis activity with the QPAG
Values are of normalized EMG, where 0 indicates no activation and 1 indicates maximal activation. Averages were taken over the period when the clutch was active, on average, during the ACTIVE condition. The values were averaged over the late stance knee flexion phase of the gait cycle.

<table>
<thead>
<tr>
<th>Subject</th>
<th>BASELINE (Normalized EMG)</th>
<th>ACTIVE (Normalized EMG)</th>
<th>% Change from BASELINE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.092 +/- 0.039</td>
<td>0.104 +/- 0.037</td>
<td>+13</td>
</tr>
<tr>
<td>Subject 2</td>
<td>0.078 +/- 0.0061</td>
<td>0.089 +/- 0.0014</td>
<td>+13</td>
</tr>
</tbody>
</table>

Figure 16: Affected-side knee muscle EMG for Subject 1 with the QPAG shown for the BASELINE condition where the clutch acted as a free-joint (thick solid blue line), and the ACTIVE condition (thin solid red line). Shaded regions for these curves indicate +/- 1 standard deviation. For reference, biological knee EMG data are also shown from literature [18]. The vertical lines indicate the typical engagement and disengagement times for the clutch.
Figure 17: Affected-side knee muscle EMG for Subject 2 with the QPAG shown for the BASELINE condition where the clutch acted as a free-joint (thick solid blue line), and the ACTIVE condition (thin solid red line). Shaded regions for these curves indicate +/- 1 standard deviation. For reference, biological knee EMG data are also shown from literature [18]. The vertical lines indicate the typical engagement and disengagement times for the clutch.

3.3.7 Metabolism

The net metabolic power (with the power required for standing subtracted out) is summarized for the two participants in Figure 18 and tabulated in Table 11. Both participants displayed small percentage reductions in metabolic cost when walking with the ACTIVE condition, as compared to the BASELINE condition.

<table>
<thead>
<tr>
<th></th>
<th>Subject 1</th>
<th>Subject 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>BASELINE Power (W/kg)</td>
<td>3.80 +/- 0.19</td>
<td>3.04 +/- 0.11</td>
</tr>
<tr>
<td>ACTIVE Power (W/kg)</td>
<td>3.61 +/- 0.19</td>
<td>2.94 +/- 0.08</td>
</tr>
<tr>
<td>Percent Change</td>
<td>-5%</td>
<td>-3%</td>
</tr>
</tbody>
</table>

Table 11: Metabolic power of the amputee participants with the QPAG
Figure 18: Metabolic power changes with the QPAG
The BASELINE condition (blue) is compared to the ACTIVE condition (red) for the two amputee participants. Error bars indicate standard error of the mean within a trial.

3.4 Discussion
The findings from this study support the hypothesis that an artificial gastrocnemius would reduce biological knee flexion moment and positive work of the affected-side knee, compared to a similarly-weighted monoarticular prosthesis, during late stance knee flexion. Despite the short amount of time to get acclimated to the knee orthosis, the participants reduced their biological knee moment profiles as to maintain a generally invariant total knee moment profile. This behavior is consistent with other studies involving exoskeletal interventions at ankle [20] and hip joints [21]. The reduction of biological knee moment is beneficial from the standpoint of wearable device design, since it means that wearers of the QPAG were quickly able to replace biological function with that from the device.

As hypothesized, the reduction in affected-side biological knee moment impulse was expected to have a corresponding reduction in hamstring muscle activity. Indeed, the semimembranosis activity was reduced for Subject 2. However, no such decrease was found for Subject 1. The general lack of a change of muscle activity for this
subject indicates that he may have been utilizing other hamstring muscles for providing the knee flexion moment, and thus, a reduction in those muscles' EMG may have been observed with the QPAG if those data had been collected.

Also as hypothesized, the affected-side hip flexion moment impulse and hip positive work of both participants was reduced during late stance knee flexion. These reductions are likely a result of the energy return of the spring, which serves to help flex the hip.

As can be seen in Figure 17, the slight increase in activation of the vastus lateralis for Subject 2 occurred when the knee joint was extending (20-30% gait cycle). It is therefore likely this small increase in muscle activity was intended to assist in the initial extension of the knee and clutched-spring joint for storing energy in the device.

In contrast with the affected knee kinetics, the kinematics of the affected knee do change between the BASELINE and ACTIVE test conditions. The reduced extension of the knee in mid-stance could mean that the kinetic invariance comes with a kinematic cost. It may be possible that additional training with the device could allow participants to both maintain both kinetics and kinematics while still achieving a reduction in biological joint moments.

The QPAG, despite lacking an ability to generate positive net mechanical work, did demonstrate an ability to reduce amputee metabolic cost of transport for the two amputees tested. However, given the lack of statistical power, more work is needed to verify this effect on more participants. In addition, the slight increase seen in the net mechanical work from the ankle-foot prosthesis during the ACTIVE conditions could account for a non-negligible portion of this small metabolic benefit. If, however, the metabolic improvement is confirmed for other amputees when ensuring no increase in positive work from the prosthetic ankle, it would show that metabolic improvements are possible for transtibial amputees without the need for the injection of positive net mechanical work. These metabolic benefits may stem
from a combination of a reduction in knee and hip moments and powers in the affected-side leg.

Seeing as the biological gastrocnemius generates several joules of net work per step [22], [23], even greater benefits could possibly be achieved by providing this net work to amputees. By definition, this net mechanical work cannot be produced with a quasi-passive device, and, therefore, a different mechanism would be needed to test this hypothesis.
Chapter 4

Powered Artificial Gastrocnemius

4.1 Overview

The quasi-passive artificial gastrocnemius represents the approximate function of the biological gastrocnemius, but this quasi-passive representation is limited in describing the full function of the gastrocnemius muscle-tendon complex. Specifically, the gastrocnemius is responsible for 3.5 Joules of positive net mechanical work per step [22], [23], which is impossible to reproduce with a quasi-passive device. An active element, such as a motor must be present to inject energy into the amputee’s knee joint.

In this chapter, I describe a motorized artificial gastrocnemius for providing a more complete representation of the gastrocnemius muscle function, including the ability to provide positive net mechanical work to the affected-side knee joint of an amputee walking on a treadmill. This artificial gastrocnemius includes the mechanical design of the actuated joint and attachment to an off-board power source, a cable transmission, and torque-control strategies for enforcing desired behavior at the aforementioned joint.

4.1.1 Motivation

It has been shown that adding mass to the lower extremities can have a detrimental effect on walking metabolism [24]. Hence, the goal in this project was to provide interventions to a participant walking on a treadmill by remote actuation, so as to minimize mass affixed to the wearer. This remote actuation would then minimize the negative consequences of donned device mass. A cable-driven tethered exoskeleton system with an off-board actuation source was developed in order to minimize the weight being carried by a participant. This system was composed of
three modules for each joint of the human leg: the motor drive, the cable transmission, and the robotic joint donned by the wearer. For the purposes of the current work, the knee joint system is described.

4.2 Methods

4.2.1 Motor Drive Module

The core of the motor module was a powerful off-board actuation source, which included a 3kW AC servo motor (SGMSV-30A3A61, YASKAWA), a 3:1 gear reducer (042PLX0030-LB-04027, CGI INC), and a custom cable tensioner. The sensing and control system, designed for real-time control and system safety, included a host personal computer, an EtherCAT® master controller (SPiiPlusNTM, ACS MOTION CONTROL), local servo controllers (V200AE1A002000200, YASKAWA), and an EtherCAT® slave I/O system (750-534 EtherCAT Coupler, WAGO). This system had multiple analog and digital inputs, which were used for gathering feedback signals in real-time from the device for the purposes of data collection and feedback control.

A 90-mm diameter pulley was mounted to the transmission output shaft so to actuate the drive cable. A custom-built tensioner allowed the cable preload to be adjusted before use. The tensioner was a slider that moved in and out along rods that connected to the motor housing. A leadscrew was employed for providing the tension, and a series compression spring served to provide an approximate measure of cable tension. When the desired tension was reached, clamping screws were tightened to lock clamps on the tensioner in place. These clamps locked the sliders to the rods, thus and eliminating the series compliance of the tensioning spring.
4.2.2 Cable Drive

A cable linkage was used to transmit torque from the motor to the knee joint pulley in a similar way as a Bowden cable drive. However, instead of a flexible conduit throughout the entire tether, sections of rigid tubing were used (Figure 20). The tubing was made of 0/90 carbon fiber, with 22.2 mm outer-diameter and 19.1 mm inner diameter. This system had a distinct advantage over a traditional, flexible Bowden cable drive: the losses due to friction between the cable and conduit were near-zero through the tubing sections, as the cable did not contact the sides of the conduit.

Figure 19: Motor drive unit and tensioner

Figure 20: Conduit linkage for the powered artificial gastrocnemius
Shown are the conduit linkage (a) and close-up of the conduit linkage pulleys (b)
Some articulation was needed between the straight sections of tubing to allow translation of the joint through the walking volume. Short, flexible sections of conduit were used to link the straight conduit members together in order to provide this flexibility. Typical Bowden cable conduits have the tendency to straighten when tension is applied to the cable. Rather than using such a conduit, short, interlocking aluminum conduit elements were used (Figure 21). These elements were designed so that the length of the inner bore of the conduit did not change upon bending. Each element had a male and female end, which, when mated, produced a ball-and-socket joint, through which the cable could pass. These elements were strung together in series to produce flexible conduit sections of the desired lengths. Polytetrafluoroethylene liners (3 mm inner diameter, 6 mm outer diameter) provided low-friction interface within these sections of conduit elements.

The conduit sections were arranged behind and above the treadmill, so as to provide minimal interference to a wearer of the knee joint (Figure 20). The conduit linkage allowed two degrees of freedom in the sagittal plane, via the vertical and horizontal hinging of the conduit sections (Figure 20). Pulleys were used to allow for this large bend in the cable conduit assembly without introducing significant friction. The remaining flexibility needed near the waist of a human subject was provided by a section of several conduit elements linked in series, forming a flexible section of conduit.

![Figure 21: Conduit element](image-url)
4.2.2.1 Cable Selection
Steel cable was chosen to transmit power through the conduit system from the motor to the wearable joint. Steel cable was chosen for the desirable properties of high axial stiffness, and low frictional coefficient with the PTFE conduit liners.

4.2.2.2 Cable Tensioner
A spring-based tensioner system was used to preload the cable. This tensioner was a bracket with an internal sliding joint, driven by a lead screw (Figure 19). This joint enabled the bracket to expand to take up cable slack. A spring was placed in series between the sliding element of the bracket and the stationary element. The spring served as an indicator of cable preload. As the lead screw was rotated, the bracket expanded and provided tension on the cable. The reaction force from the cable conduit was transmitted through the bracket, and loaded the spring. Therefore, by measuring the spring deflection during the cable preloading, the cable tension could be approximated. Once the cable was pre-tensioned, screws were tightened to lock the adjustment mechanism and make the tensioner a rigid structure. Therefore, the tensioning springs did not introduce series compliance to the system.

4.2.3 Joint Mechanism
The actuated knee joint mechanism consisted of a pulley, driven by the tether cable, which sat lateral to the wearer's affected-side biological knee joint (Figure 22). This pulley was free to rotate on ball bearings in the sagittal plane about the knee center with respect to the joint housing. The pulley itself was connected to the output link through a torsional spring, which acted as a series compliance between the cable drive and the output link.
Figure 22: Powered artificial gastrocnemius joint mechanism
The assembled joint, without proximal and distal attachments, is shown in the top section. An exploded view of the same mechanism is shown in the bottom section.
Figure 23: Sectional view of the powered artificial gastrocnemius joint

Parts that connect as a rigid body are shaded with the same color. The housing (blue) attached to the proximal mount, which was affixed to the thigh cuff. The pulley (red) rotated with respect to the housing via the main bearing (teal). The output link (green) was affixed to the distal mount which connected to the socket. This link rotated with respect to the pulley via the inner bearing, with the torsional spring connecting the output link to the pulley. The potentiometer measured the relative rotation of the housing and the output link. Hard stops (not shown) mounted to the pulley assembly and engage with the housing at the limits of joint rotation.

4.2.4 Custom Knee Brace

A custom knee brace was built individually for each amputee participant to use the AG (Figure 25). Each amputee participant was brought to a certified prosthetist to get a cast made of their residual limb while their normal prosthetic socket was being worn. The prosthetist then created carbon fiber orthoses that were form-fitted for each individual. An orthosis could be strapped to the affected-side leg while the prosthetic socket was being worn. The orthoses each comprised a carbon fiber thigh cuff, and a carbon fiber shank cuff that fitted over the socket. These cuffs were joined by a pair of polycentric knee brace hinges (Figure 24)
4.2.5 Socket Attachment Bearing

A difference in kinematics existed between the robotic joint, which acted as a pin-joint, and the knee brace joints, which had polycentric motion. As a result of this difference, and to account for other errors in alignment with the knee brace, a series of bearings were used to provide a mechanical interface between the output link of the robotic joint and the shank cuff of the knee brace. This set of bearings were designed to transmit only tangential, sagittal plane forces from the robotic joint to the shank cuff of the custom brace. This bearing set consisted of 1) a linear bearing to allow for distal-proximal motion (1.7 kN dynamic loading capacity), 2) a ball joint to provide 3-axis rotational freedom, and 3) A linear sleeve bearing to allow medial-lateral motion.

Figure 24: Knee brace and socket attachment
The left image shows a rendering of the powered AG knee brace with actuated joint and powered ankle-foot prosthesis. The center image shows a side-view of the same, with arrows indicating the intended applied torque from the knee actuator and the resulting tangential force on the socket from the actuator. The right figure shows the bearing assembly which was mounted on the socket attachment. This assembly transmitted the tangential force from the actuator while allowing for motion from the polycentric knee joints as well as misalignment in the brace.
Figure 25: Custom knee braces
Shown is a custom knee brace for non-amputees including attachment mounts for the robotic joint mechanism (a), and custom knee brace for an amputee, attached to a socket (b)

4.2.6 Safety Features
Multiple hardware safety features were implemented, including emergency stop buttons available to both the researchers and the study participants, and a trigger switch, held by the participants: if at any point any of the stop buttons were pushed, or if the trigger was released, the motor drive was immediately disabled. Hard stops were mounted on both the motor and at the knee joint. Further, limits switches at the motor would disable the motor if the motor stops were approached.
4.2.7 Sensors

A rotary potentiometer (8 kOhm), mounted between the joint housing and the output link, was used to measure joint angle. This potentiometer was mounted on the lateral side of the joint, with the potentiometer shaft going through the center of the joint and attaching to the output link on the medial side (Figure 22). Torque sensing was accomplished via the use of a full-bridge strain gauge (SGT-2/350-FB13, Omega Engineering, Inc), mounted on the output link. The output link itself was designed in an attempt to maximize strain on the strain gauge surface, while minimizing strain variation across the surface, all while keeping maximum stresses within the material limits of the 7075-T6 Aluminum structure. See Figure 26 for the stress distribution resulting from finite element analysis of the final design.

![Figure 26: Custom strain gauge-based torque sensor](image)

The torque sensor was built from the output link of the joint mechanism. The dashed line represents the axis of the knee joint pulley. The rectangle indicates the location of the strain gauge.

The strain gauge was connected to a custom preamplifier, which included a common-mode input filter to reject electromagnetic interference, and a second-order low-pass filter with a cutoff frequency of 585 Hz. A footswitch (model: FSW, B&L
Engineering, Santa Ana, CA), worn as an insole in the shoe on the leg wearing the knee brace, was used to detect stance and swing phases.

4.2.7.1 Sensor Calibration
The strain gauge was calibrated by affixing the output link to a load cell, while rigidly connecting the housing of the pulley mechanism to the same mounting platform as the load cell (Figure 27). The motor was commanded a series of trapezoidal curves, in which the motor torque slowly ramped up to a maximum value, and then slowly ramped back down, bi-directionally. This method was used in order to facilitate comparison of strain gauge output to the load cell readings.

![Sensor calibration mounting platform](image)

Figure 27: Sensor calibration mounting platform

4.2.8 Open-Loop System Characterization
A linear, 2nd-order lumped-parameter model was used to represent the actuator and human system. The system model had the form:

\[
\frac{\tau_{\text{joint}}}{\tau_{\text{command}}} = \frac{K_e}{J_e s^2 + B_e s + K_e}
\]  

(5)

where \(K_e\) is the effective system compliance, \(J_e\) is the effective system inertia, and \(B_e\) is the effective system damping.
The effective inertia, $J_e$, was assumed to be dominated by the motor inertia ($7 \times 10^{-4}$ kg·m²), reflected through the 3:1 planetary transmission, and the transmission output inertia ($1.16 \times 10^{-4}$ kg·m²). The reflected motor inertia was calculated using the relation $J_e = J_m R$, where $J_m$ is the motor inertia and $R$ is the transmission ratio of 3.

![Motor torque command (Nm) vs. Time (seconds)](image)

**Figure 28: Swept sine input for system identification**

System identification tests for this AG model were performed using a knee brace built for non-amputee participants. The compliance of the human tissues has a significant effect on the overall dynamics of the human-machine system, and therefore, it was important to perform these system identification tests with a human in the loop. The brace was worn by a healthy adult male participant. The participant stood still, keeping his knee bent slightly, but resisting motion from the device. A swept-sine torque signal was commanded to the motor, with frequencies ranging from 0.1 to 200 rad/s in increments of 10 rad/s (Figure 28). Additional data was then collected with the flexible conduit section in the cable conduit linkage bent approximately 15 degrees, so as to capture the range of friction in this section. The magnitude of the command signal was adjusted for each frequency to maximize
torque while preventing excessive discomfort to the wearer. Joint moment data were collected from the AG strain gauge.

The above data collection assumed a dominantly linear response between motor torque command and torque output at the knee joint. However, it was also necessary to characterize the non-linear effect of coulomb friction in the transmission and cable drive. Hence, an additional test was performed to evaluate this frictional contribution to the system response. For these tests, instead of a swept sinusoid, a slowly-changing trapezoidal torque profile was commanded at the motor. This minimized the contribution of dynamical effects to the system response, while enabling the estimation of frictional effects.

System identification was performed using the time-domain signal of motor torque command as the input signal, and the strain-gauge measured joint torque as the output signal. These signals were provided to the system identification app in MATLAB to estimate parameters of the 2nd-order system model of the form indicated in Equation 5. Signal content was restricted to 0-200 rad/s for this analysis. The open-loop bandwidth and phase margin were then estimated from the resulting linear model.

4.2.9 Joint Torque Control
For the purposes of torque control, the gait cycle was divided into two phases: stance and swing. Stance phase began at heel-contact and ended at toeoff. Swing phase began at toeoff and ended at subsequent heel-contact. A running counter was used to estimate the elapsed fraction of the current gait cycle by comparing the time spent in the current gait cycle to the total time of the previous gait cycle: \( gc = \frac{\text{counter}}{\text{lastsamples}} \) where \( gc \) is the estimated elapsed fraction of the current gait cycle, \( \text{counter} \) is the number of control cycles elapsed in the current gait cycle, and \( \text{lastsamples} \) is the total number of control cycles of the previous gait cycle.
4.2.10 Stance Phase Torque Control

The high bandwidth and stability of the motor torque-based control enabled an open-loop control methodology for the stance phase of the gait cycle. A feed-forward friction compensation term, $\tau_{ff}$ with iterative learning was used to reduce the torque errors $\tau_{err}$ resulting from friction in the cable drive. This friction compensation was implemented as a torque profile as a function of time within a gait cycle. The function was updated each gait step by smoothly incorporating the observed torque error in two steps:

$$\tau_{learn}(gc,n + 1) = \tau_{err}(gc,n) \cdot \alpha + \tau_{ff}(gc,n) \cdot (1 - \alpha)$$  \hspace{1cm} (6)

and

$$\tau_{ff}(gc,n + 1) = \tau_{learn}(gc,n)$$  \hspace{1cm} (7)

where $gc$ is the elapsed fraction of the gait cycle, $n$ is the gait cycle number, and $\alpha$ is the learning rate. Once a new learned function $\tau_{learn}$ was computed, it was subsequently used in the feed-forward term $\tau_{ff}$ on the subsequent gait cycle. Thus, the error recorded during a gait cycle would be incorporated in the controller two gait cycles later. A value of $\alpha$ of 0.5 was found to provide a reasonable learning rate, for which the learned friction compensation profile was updated such that a new learned behavior was fully incorporated in approximately 10 seconds. Similar iterative learning methods have been successful in wearable robotics in the presence of cable frictional losses [25].

4.2.11 Swing Phase Torque Control

In an effort to further reduce resistance of the AG, a feedback torque controller was implemented to enforce the joint torque during the swing phase. The following control law was used:

$$\tau_c = K_i \int (\tau_{des} - \tau_{meas})dt$$  \hspace{1cm} (8)
which, in the Laplace domain, takes the form

\[ C(s) = \frac{K_i}{s} \]  

(9)

where, \( \tau_{\text{des}} \) is the desired joint torque, \( \tau_{\text{meas}} \) is the measured joint torque, and \( \tau_c \) is the motor torque command. Standard loop-shaping techniques were used, along with simulations using MATLAB to determine this control law and the parameter \( K_i \). For these simulations and loop-shaping, the open-loop model Equation 5 was used as the plant. In switching from stance phase to swing phase, the internal states of the closed-loop compensator were reset to zero, so to prevent any unwanted integral windup accumulated during the stance phase.

![Torque control schematic](image)

**Figure 29: Torque control schematic**

The control system for the AG was composed of an inner torque-controller loop, which enforced desired torque at the joint, and an outer loop involving a neuromuscular model as a controller, which took joint kinematics as input. The inner control loop could be switched between open-loop control (Stance) and closed-loop torque control (Swing), to be used in the appropriate phase of the gait cycle.

### 4.2.11.1 Software Limits

Several software safety features were implemented. These included:

1) A maximum motor torque was limited to 90 Nm
2) Maximum contribution of the feed-forward friction compensation was limited:
   a. $1.4 \times$ desired torque for the generation of flexion moments
   b. 10 Nm for the generation of extension moments
3) Integral anti-windup, preventing the commanded torque from integrator from exceeding 30 Nm

4.2.11.2 Closed-Loop System Characterization
System characterization tests were performed for the closed-loop system, with desired joint torque as input and measured strain gauge joint torque as output. The frequency characteristics of this system were determined in the same manner as described for the open-loop system. A closed-loop system model was developed by simulating the closed-loop torque controller applied to the open-loop linear model in MATLAB, including a 2 ms delay in the forward loop path to simulate latencies in the analog-to-digital converter. The closed-loop bandwidth (-3 dB) and the phase margin were estimated from this resulting closed-loop linear model.

4.2.12 Metabolic Validation
Clinical metabolic tests were performed to test the transparency of the robotic knee joint; it was required that the described torque-control techniques could provide sufficiently low resistance to wearers as not to adversely affect their metabolism. Success was defined as a non-significant difference in metabolism between two conditions during treadmill walking: 1) torque-controlled AG knee orthosis worn on one leg with zero torque commanded, and 2) AG knee orthosis worn, acting as a free-swinging joint (with the cable drive disconnected). For the purposes of this metabolic validation trials with non-amputees, a custom-built, carbon fiber brace, molded by a prosthetist, was used to mount the robotic joint on the leg of non-amputee participants (Figure 25).
Six healthy participants were asked to walk on a treadmill while metabolic data were collected. The clinical evaluation was conducted at MIT (Cambridge, MA) and was approved by MIT's Committee on the Use of Humans as Experimental Subjects (COUHES). Each participant provided written, informed consent was obtained from before data collection was initiated. Each participant was asked to walk at a typical walking speed of 1.25 m/s under three conditions, in random order: 1) walking normally without the AG knee orthosis, 2) wearing the orthosis with the cable drive disconnected, and 3) wearing the orthosis with torque control set to zero torque. For all conditions, participants wore a portable oxygen consumption mask attached to a standard open-circuit gas analysis system (model K4B², Cosmed, Rome, Italy). This system estimated metabolic energy consumption based on measurements of oxygen inspired and expired carbon dioxide.

Metabolic cost for each walking speed was computed by taking average oxygen and carbon dioxide data over a two-minute window at the end of each six-minute trial. The metabolic power was computed using the equation

\[ P = K_{O_2} \dot{V}_{O_2} + K_{CO_2} \dot{V}_{CO_2} \]  

where \( P \) is the metabolic power in Watts, \( \dot{V}_{O_2} \) is the volume flow rate of Oxygen inhaled, \( \dot{V}_{CO_2} \) is the volume flow rate of carbon dioxide exhaled, and \( K_{O_2} \) and \( K_{CO_2} \) are constants with values from literature [19], given as \( K_{O_2} = 16,580 \) W/L and \( K_{CO_2} = 4,510 \) W/L.

The above equation is only valid for conditions when the metabolism is primarily aerobic. As a verification of this condition, the respiratory exchange ratio (RER), defined as \( \dot{V}_{CO_2}/\dot{V}_{O_2} \) was monitored, and only metabolic results with RER values less than 1.1 were considered.
4.3 Results

4.3.1 Mechatronic Design Specifications
The designed robotic joint had physical parameters within the desired values (see Table 12). Although the robotic joint itself had a mass of only 0.6 kg, the brace for attaching it to the body required the addition of 1.3 kg for the non-amputee brace, 0.8 kg for each amputee brace, and 1.2 kg for the suspended conduit system.

![Graph showing torque response](image)

**Figure 30: Trapezoidal torque response**
Command and output torque with the cable linkage. If no friction was present, a slope of unity (thick red line) would indicate the relationship between motor command and output torque. However, frictional losses in the drivetrain cause a hysteresis curve to form (thin blue lines).
<table>
<thead>
<tr>
<th>Parameter</th>
<th>Desired Value</th>
<th>Achieved Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Joint range of motion (Radians)</td>
<td>1.3</td>
<td>2.1</td>
</tr>
<tr>
<td>Size</td>
<td>(Minimal)</td>
<td>50 mm radius, 53 mm wide</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>(Minimal)</td>
<td>0.6</td>
</tr>
<tr>
<td>COR deviation allowance (mm)</td>
<td>20</td>
<td>20</td>
</tr>
<tr>
<td>Maximum output torque (Nm)</td>
<td>60</td>
<td>90</td>
</tr>
<tr>
<td>Torque sensing resolution (Nm)</td>
<td>0.01</td>
<td>0.01</td>
</tr>
</tbody>
</table>

Table 12: Design parameters of the powered artificial gastrocnemius

4.3.2 Friction Characterization

Output joint torque was compared to the commanded motor torque as shown in Figure 30. If no friction was present, a perfectly straight line of slope = 1 would be expected. However, in the presence of drivetrain friction, a hysteresis band was present. Frictional effects manifested as a discrepancy, which scales approximately with torque command, between the motor torque and output joint torque. These frictional losses amounted to approximately 20% of the motor torque.

4.3.3 Torque-Control Response Characteristics

The effective inertia of the transmission elements at the output joint was calculated as 0.64 kg*m^2. The effective damping and stiffness of the second-order model in Equation 5 were estimated as 48.3 Nm*s/rad and 5912 Nm/rad, respectively.

When attached the human leg, as in the system identification experiments, the robotic joint had a -3 dB open-loop torque bandwidth of 16 Hz (100 rad/s) (Figure 31). When applying the swing-phase closed-loop torque controller, The closed-loop system bandwidth was 14 Hz (87 rad/s) (Figure 32), with a gain margin of 1.6 and a phase margin of 69 degrees.
Figure 31: Open-loop system response

Estimated frequency response from motor torque command to the joint strain gauge-measured torque. Shown are the frequency response data (red) and the linear, second-order model fit to the data (blue).
Figure 32: Closed-loop system response
Frequency response of the closed-loop system using the integral feedback compensator. The input to the system was desired joint torque, and the system output was the strain gauge-measured joint torque. The frequency response from the data is shown in red. Also shown for comparison is a simulated closed-loop controller applied to the linear open-loop model (blue).

4.3.4 Metabolic Validation
In the metabolic tests with non-amputees, it was found that the metabolic consumption of the participants walking with the zero-torque control condition was not significantly different from that of the disconnected condition, although significantly lower metabolism was found for the normal-walking condition without the AG (Figure 33).
Figure 33: **Metabolic power of non-amputees with the tethered knee orthosis.** Horizontal brackets indicate pairs of conditions in which the metabolism was significantly different (p < 0.05).

Figure 34: **Representative closed-loop system response during walking trials** Data were collected with an amputee participant. The torque error (blue) represents the difference in joint moment between the desired AG device torque (red) and the measured device torque (green).
4.3.5 Torque Errors During Walking

In addition to the transparency requirement, it was necessary to ensure that the non-zero desired torque values could be tracked by the AG within a reasonable range. The root mean squared stance-phase torque error, between the desired torque from the NMC and the measured joint torque, was 1.45 ± 0.72 Nm on average across the amputee participants (6 percent of the average maximum torque of 23 Nm). This error was well within the average step-to-step standard deviation in knee flexion moment during mid-stance phase for the non-amputee participants (4.8 Nm).

4.4 Discussion

Although it might seem that providing off-board power would resolve most of the mechatronic design challenges in a wearable device, the tethered system still required the tackling of several challenges in the hardware and control in order to be successful. First, the selection of motor transmission was a key factor. Any increase in transmission ratio drastically increased the motor reflected inertia at the output joint, making control very difficult. The transmission ratio was therefore carefully selected in order to minimize this effect, while still providing sufficient torque for the purposes of the experiments. Second, the cable transmission had to be developed. Next, even the best linear control strategies fail when in the presence of significant nonlinear effects, so the conduit linkage was developed in order to minimize sources of friction. The control system was designed for the purposes of replicating muscle function during the cyclic tasks of walking, so it was tailored to the specific requirements in the stance and swing phases. Finally, care was taken to provide a secure, and robust end-effector that securely attached to the legs of amputees, without becoming too cumbersome.

Overall, the efforts to produce transparency with the powered AG knee orthosis were successful. The difference in metabolism for the non-amputee participants walking with, compared to without the AG, may be attributed, at least in part, to
the weight of the device, since mass added to the limbs causes an increase in metabolic cost of walking [24].

The control strategies used in this study provided low errors between desired and measured joint torque across a wide dynamic range. Good torque tracking on the low end allowed for metabolic transparency, while the torque tracking at higher torque levels was comparable to the variations in normal human gait. This system was therefore considered well suited for simulating a gastrocnemius muscle at the knee for amputee participants.
Chapter 5

Neuromuscular Controller

5.1 Overview
Musculoskeletal morphology clearly plays an important role in human walking, in that the physical characteristics are finely adapted to produce gaits economically. Yet, humans are also able to react to disturbances, and seamlessly adapt to variations in terrain. It is desirable, in the development of prostheses, to encode these same characteristics into the control schemes, so as to achieve the same level of adaptability in the prostheses as seen in the human form. Although human motor control is thought to utilize both feed forward descending commands and reflexive feedback [26], simulation work has shown that purely reflexive control of leg muscles is able to produce robust walking in simulation [27]. Prosthesis controllers based on these reflex-driven muscle models have been developed to make use of these behaviors in physical form. These controllers have indeed shown biological-like gait characteristics, including terrain and speed adaptation [28], [29]. For the development of the powered artificial gastrocnemius, a neuromuscular model, based on the same model from [30], was used to produce a controller for the knee orthosis robotic joint.

This model used joint kinematics and neural stimulation signals as input, and produced joint moments as output. For control purposes, a simulated spinal reflex, as used in [29] produced neural stimulation signals using the internal state of the musculoskeletal model. Together, the musculoskeletal model and neural reflex is hereby referred to as the neuromuscular model. The neuromuscular model took only joint kinematics as input, and produced joint moments as output. This section describes the development of this model, and the application of the model to be a
neuromuscular controller (NMC), for the use in controlling the artificial gastrocnemius knee orthosis.

5.2 Methods

5.2.1 Non-Amputee Data Collection

The gait data of four healthy, non-amputees were used to inform the parameters of the neuromuscular model. These non-amputee participants were chosen so to match the amputees as closely as possible (see Table 13). The clinical evaluation was conducted at MIT (Cambridge, MA) and was approved by MIT's Committee on the Use of Humans as Experimental Subjects (COUHES). Each participant provided written, informed consent was obtained from before data collection was initiated. These data were collected in two phases.

In the first phase, kinematic, kinetic, and electromyographic data were collected for 90 seconds at each of six speeds (0.75 m/s, 1.00 m/s, 1.25 m/s, 1.50 m/s, 1.75 m/s, and 2.00 m/s). An infrared camera system (12 cameras, model: T40s, Vicon Motion Systems Ltd, Oxford, UK) was used to track the motion of subjects as they walked in the capture volume. Reflective markers were placed at 47 locations on each participant's body, based on the Helen Hayes marker model, and their three dimensional trajectories were recorded at 100 Hz. The marker locations were chosen specifically to track joint motion, as prescribed by the Helen Hayes marker set. The ground reaction forces and contact centers of pressure were measured using an instrumented force plate treadmill (Bertec Corporation, Columbus, OH) with a sampling rate of 1 kHz. Electromyographic signals were collected at 2 kHz using a wireless surface system (Trigno, Delsys Inc, Natick MA). Twelve muscles (tibialis anterior, soleus, medial gastrocnemius, lateral gastrocnemius, rectus femoris, vastus lateralis, vastus medialis, semimembranosus, biceps femoris long head, adductor magnus, illiacus, gluteus maximus) on the right leg of each subject were recorded, with symmetry being assumed for the other leg. Prior to the walking
trials, the muscle maximum voluntary contraction (MVC) tests were performed for each of the muscles of interest, in which the participants were instructed to push as hard as possible, three times in succession for 3-5 seconds each with the given muscle group, while electromyographic data were collected.

<table>
<thead>
<tr>
<th>Participant number</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NA3</td>
<td>175</td>
<td>73.4</td>
</tr>
<tr>
<td>NA4</td>
<td>180</td>
<td>90.3</td>
</tr>
<tr>
<td>NA6</td>
<td>188</td>
<td>95.4</td>
</tr>
<tr>
<td>NA7</td>
<td>180</td>
<td>103</td>
</tr>
</tbody>
</table>

Table 13: Non-amputee body parameters

Once the first phase was completed, the markers and electrodes were removed, and the participants were equipped with an open-circuit gas exchange measurement system for estimating metabolic cost (model: K4b2, COSMED, Rome, Italy). The participants were asked to perform one standing trial to measure standing metabolism, and then walking trials were performed on the treadmill at a speed of 1.25 m/s. The same six walking conditions were repeated, with each walking trial lasting six minutes.

5.2.2 Data Processing

Fourth-order Butterworth filters were used to filter the marker position and ground reaction force data with 6 Hz and 25 Hz cutoff frequencies, respectively. The marker and force data were post-processed through the inverse dynamics module of SIMM (Musculographics Inc., Evanston, IL), to produce joint moments and angles in three dimensions. Only sagittal plane dynamics were considered.

Gait events were determined using vertical ground reaction force data from the embedded forceplates. Approximate event timing was found by determining the times when the force increased beyond a 40 N threshold. Exact heelstrike and toeoff times were found by progressing backward, and forward in time, respectively,
until the force value dropped to zero. Data were then cut to gait cycles based on heelstrike times, and resampled to 101 points. Gait cycles were discarded for the beginning and end of each trial, during the speed transients of the treadmill. Gait cycles in which the stride times were below 0.7 seconds or above 1.3 seconds, or in which a foot crossed the mid-line of the treadmill were also discarded.

Metabolic cost for each walking speed was computed by taking average oxygen and carbon dioxide data over a two-minute window at the end of each six-minute trial. The metabolic power was computed using the equation

\[ P = K_{O_2} \dot{V}_{O_2} + K_{CO_2} \dot{V}_{CO_2} \]  

(11)

where \( P \) is the metabolic power in Watts, \( \dot{V}_{O_2} \) is the volume flow rate of Oxygen inhaled, \( \dot{V}_{CO_2} \) is the volume flow rate of carbon dioxide exhaled, and \( K_{O_2} \) and \( K_{CO_2} \) are constants with values from literature [19], given as \( K_{O_2} = 16,580 \) W/L and \( K_{CO_2} = 4,510 \) W/L. The above equation is only valid for conditions when the metabolism is primarily aerobic. As a verification of this condition, the respiratory exchange ratio (RER), defined as \( \dot{V}_{CO_2}/\dot{V}_{O_2} \) was monitored, and only metabolic results with RER values less than 1.1 were considered.

5.2.3 Estimating Muscle Activations

Several processing steps were taken to prepare the EMG signal to be processed for the estimation of muscle activations. EMG signals were first processed internally by the EMG system by applying a 4th-order bandpass filter with a pass band from 20 Hz to 450 Hz. A 60-Hz notch filter was also applied to remove mains hum. Any DC offset was removed, and the remaining signal was saturated at 5 standard deviations from zero. The signal was then normalized to the new maximum value of 5 standard deviations and rectified.

The processed EMG data were further processed with the method described in [30] to achieve estimates of muscle activation from the measured EMG signals. Two
steps were taken to achieve these activation estimates. First, EMG signals were processed through the algorithm proposed by Sanger [31]. This method estimated the driving neural signal, \( x(t) \), that produced the measured EMG, using the equation

\[
\frac{dx}{dt} = \alpha dW + (U - x)dN_\beta \tag{12}
\]

where \( dW \) is the differential of Brownian motion with rate \( \alpha \), \( dN_\beta \) is the differential of a counting process with \( \beta \) events occurring per unit time, and \( U \) is a uniformly distributed random variable in \([0,1]\). The driving signal was estimated by using

\[
P(\text{emg}|x) = \frac{e^{-\frac{\text{EMG}}{x}}}{x} \tag{13}
\]

where \( P(\text{emg}|x) \) is the conditional probability of observing an EMG signal, given the driving signal, \( x(t) \). The driving signal \( x(t) \) was then solved recursively, using Bayes’ rule.

In the second step, the neural signals \( x(t) \) were used to estimate the muscle activations, which are essentially the neural signals passed through a low-pass filter. A filter of the form

\[
\frac{d\alpha(t)}{dt} = \begin{cases} 
(x(t) - \alpha(t))[x(t)/\tau_{\text{act}} + (1 - x(t))/\tau_{\text{deact}}], & x \geq \alpha \\
(x(t) - \alpha(t))/\tau_{\text{deact}}, & x < \alpha
\end{cases} \tag{14}
\]

was used [32], [33] to model the activation and deactivation dynamics of the muscles, where \( \tau_{\text{act}} \) and \( \tau_{\text{deact}} \) were the activation and deactivation time constants for a given muscle, and \( \alpha \) was the estimated muscle activation signal. Finally, the activation signals were thresholded to remove the noise floor, and the resulting signals were averaged across gait cycles to produce a single activation curve for each muscle. The activations were initially scaled by those resulting from the MVC trials, but it was found that the resulting scaled EMG signals were unreasonably scaled, when compared to literature values [18]. Therefore, the final magnitudes of the estimated activation signals were scaled so that the average values of the
activation signals matched those from [18] during the range of activation for the literature data.

As was found in a previous study [30], many of the muscle spanning the hip to have signals too weak to be reliably measured (RF, ILL, BFSH, HAB, ADDL, and ADDM). For these muscles, neural excitation profiles from wire electrode experiments [18] were delayed 40 ms to simulate the delay between the EMG signals and the development of force [34], and then passed through the same activation dynamics used for the collected electromyographic data. The resulting activation signals were then used as input to the musculoskeletal model along with those derived from the electromyography. A similar method was used successfully in a similar modeling study [30].

5.2.4 Musculoskeletal Model

The kinematic, kinetic, and muscle activation data were used to inform the parameter selection of a musculoskeletal model, for the purposes of deriving a biophysically-based controller. The model described in [30] was used. In this model, all three joints (ankle, knee and hip) of a human leg were modeled in the sagittal plane, with lumped muscle groups representing the dominant muscle functions. These muscle groups were the GAS group (medial and lateral gastrocnemius), the VAS group (vastus lateralis, medialis, and intermedius), the HAM group (semimembranosus, semitendinosus, and biceps femoris long head), and the ILL group (iliacus and psoas). For each muscle group, a series compliant element was modeled in series to represent the tendinous structures connecting the muscles to the skeleton. The iliofemoral ligament at the hip was modeled as a unidirectional spring that served to flex the hip at large hip extension angles: \( \tau_{HFL} = -K_{HFL}(\theta_{\text{hip}} - \theta_{0,HFL}) \), where \( \tau_{HFL} \) and \( \theta_{\text{hip}} \) were the hip flexion moment and hip flexion angle, respectively, and \( \theta_{0,HFL} \) was the hip engagement flexion angle of the ligament model.
5.2.4.1 Hill Model

Each muscle group of the musculoskeletal model was modeled as an effective muscle, with Hill-type contraction dynamics based on [27]. This Hill-type model comprised a contractile element (CE), parallel elasticity (PE) and series elasticity (SE), arranged as shown in Figure 35, and all modeling was performed as described in previous work [30].

![Figure 35: Hill-type muscle model](image)

The CE force was a function of activation, $\alpha$, the contractile element length, $l_{CE}$, and the contractile element velocity, $v_{CE}$. The CE force had the form

$$F_{CE} = \alpha F_{max} f_l(l_{CE}) f_v(v_{CE})$$

(15)

where $\alpha$ is the muscle activation, $l_{CE}$ is the contractile element length, $v_{CE}$ is the contractile element velocity, and $F_{max}$ is the maximum isometric force. The individual terms of this equation are described by the following equations,

$$f_l(l_{CE}) = \frac{-1}{w^2} \left( \frac{l_{CE}}{l_{opt}} \right)^2 + \frac{2}{w^2} \left( \frac{l_{CE}}{l_{opt}} \right) - \frac{1}{w^2} + 1$$

(16)

and

$$f_v(v_{CE}) = \begin{cases} 
\frac{v_{max} + v_{CE}}{v_{max} - K v_{CE}} , & v_{CE} < 0 \\
\frac{(N - 1)(v_{max} - v_{CE})}{7.56 K v_{CE} + v_{max}} , & v_{CE} \geq 0
\end{cases}$$

(17)
Equation 16 describes a bell-like curve, where the optimum fiber length, \( l_{opt} \) was the value of \( l_{CE} \) at which the muscle could provide maximum force under isometric conditions, and \( w \) describes is the width of the bell, in units of \( l_{opt} \), as the amount of fiber shortening or lengthening from \( l_{opt} \) before the force-length relation dropped to zero.

In Equation 17, \( v_{max} \) was the maximum lengthening velocity of the muscle, \( K \) was a curvature constant, and \( N=1.5 \) was the muscle force (in units of \( F_{max} \)) when \( v_{CE} = v_{max} \).

The parallel elasticity was a non-linear spring, engaging for CE fiber lengths greater than \( l_{opt} \). The elasticity developed force as

\[
F_{PE}(l_{CE}) = \begin{cases} 
F_{max} \left( \frac{l_{CE} - l_{opt}}{l_{opt}w} \right)^2, & l_{CE} \geq l_{opt} \\
0, & l_{CE} < l_{opt}
\end{cases}
\]

In addition, a numerical tool was implemented to prevent the contractile element from collapsing. This buffer elasticity developed force as

\[
F_{BE}(l_{CE}) = \begin{cases} 
F_{max} \frac{2}{w} \left( \frac{l_{CE} - l_{opt}(1-w)}{l_{opt}} \right)^2, & l_{CE} \leq l_{opt}(1-w) \\
x, & l_{CE} > l_{opt}(1-w)
\end{cases}
\]

Combining the contractile element and elastic elements, the total force from the Hill-type muscle was

\[
F(\alpha, l_{CE}, v_{CE}) = F_{CE}(\alpha, l_{CE}, v_{CE}) + F_{PE}(l_{CE}) - F_{BE}(l_{CE})
\]

Fast-twitch vs slow-twitch fiber composition of individual muscles was modeled by incorporating the fraction of fast-twitch fibers in a given muscle, FFT, into the parameters \( K \), and \( v_{max} \):

\[
K = 8(1 - FFT)
\]  

(18)
\[ v_{\text{max}} = 4.8(1 + 1.5FFT) \]  

The muscle-specific parameters for Equations 15 through 20 were taken from [30] and can be found in Table 14.

The series elasticity of the Hill-type model represented the tendon, and was modeled as a non-linear spring, with the force-strain relation defined by

\[ F_{SE}(\lambda) = \begin{cases} 
\frac{\exp \left( \frac{K_{sh}}{\lambda_{ref}} \lambda \right) - 1}{\exp(K_{sh}) - 1}, & \lambda > 0 \\
0, & \lambda \leq 0
\end{cases} \]  

where \( \lambda \) was the tendon strain, referenced from its slack length \( \lambda = \frac{l_{SE} - l_{sl}}{l_{sl}} \), \( K_{sh} \) was the shape factor, and \( \lambda_{ref} \) was the strain where the tendon force equaled the maximum muscle isometric force, \( F_{\text{max}} \).

5.2.4.2 Muscle-Tendon Dynamics

The muscle and tendon acted in series, forming a structure called the muscle-tendon complex (MTC), with the muscle fibers at a pennation angle \( \theta \) to the line of action of the MTC. The overall dynamics of the muscle-tendon complex were represented by

\[ F_{MTC}(t) = F_{SE}(t) = F(\alpha(t), l_{CE}(t), v_{CE}(t)) \cos(\theta(t)). \]  

The length of the MTC was

\[ l_{MTC}(t) = l_{SE}(t) + l_{CE}(t) \cos(\theta(t)). \]  

The pennation angle changed as a function of contractile element length [35] as

\[ \theta(l_{CE}) = \sin^{-1} \left( \frac{l_{opt} \sin(\theta_0)}{l_{CE}} \right) \]
Where $\theta_0$ was the pennation angle where $l_{CE} = l_{opt}$ [36].

Since multiple muscles act at each joint, the total muscle-driven joint moment was the sum of all MTC contributions at the joint:

$$\tau_{mod} = \sum_{i} F_{MT,i}(t) r_i(t)$$

(24)

where the time-varying moment arm of a given muscle was $r_i(t)$.

5.2.5 Morphological Optimization

5.2.5.1 Parameters

As in [30], there were two types of inputs to the musculoskeletal model. The first type were the time-varying inputs, which were estimated from the collected biological data: $a(t)$, $l_{mtc}(t)$, and $r_i(t)$. The second type of input was the set of free parameters for the model, which were determined using an optimization scheme: $F_{max}$, $K_{sh}$, $\lambda_{ref}$, $l_{opt}$ and $l_{slack}$, where the last two parameters were scaled together. The remaining parameters of the model were fixed to the values given in [30] (Table 14).

<table>
<thead>
<tr>
<th>Muscle Group</th>
<th>FFT</th>
<th>$\tau_{act}$ (ms)</th>
<th>$\tau_{deact}$ (ms)</th>
<th>$w$</th>
<th>$K$</th>
<th>$v_{max}$ [$l_{opt}/s$]</th>
<th>$\theta_0$ (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>TA</td>
<td>0.25</td>
<td>68</td>
<td>76</td>
<td>0.49</td>
<td>6.60</td>
<td>6.0</td>
<td>5</td>
</tr>
<tr>
<td>SOL</td>
<td>0.20</td>
<td>71</td>
<td>79</td>
<td>0.80</td>
<td>6.24</td>
<td>6.4</td>
<td>25</td>
</tr>
<tr>
<td>GAS</td>
<td>0.50</td>
<td>57</td>
<td>62</td>
<td>0.61</td>
<td>8.40</td>
<td>4.0</td>
<td>17</td>
</tr>
<tr>
<td>VAS</td>
<td>0.50</td>
<td>57</td>
<td>62</td>
<td>0.55</td>
<td>8.40</td>
<td>4.0</td>
<td>5</td>
</tr>
<tr>
<td>BFSH</td>
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<td>64</td>
<td>70</td>
<td>0.75</td>
<td>8.40</td>
<td>5.2</td>
<td>23</td>
</tr>
<tr>
<td>RF</td>
<td>0.65</td>
<td>49</td>
<td>65</td>
<td>0.76</td>
<td>9.48</td>
<td>2.8</td>
<td>5</td>
</tr>
<tr>
<td>HAM</td>
<td>0.35</td>
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<td>70</td>
<td>0.75</td>
<td>7.32</td>
<td>5.2</td>
<td>15</td>
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<td>ILL</td>
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<td>57</td>
<td>62</td>
<td>0.74</td>
<td>8.40</td>
<td>4.0</td>
<td>7</td>
</tr>
<tr>
<td>GMAX</td>
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<td>59</td>
<td>65</td>
<td>0.77</td>
<td>8.04</td>
<td>4.4</td>
<td>0</td>
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<tr>
<td>GMED</td>
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<td>0.77</td>
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<td>4.0</td>
<td>19</td>
</tr>
<tr>
<td>ADDL</td>
<td>0.35</td>
<td>64</td>
<td>70</td>
<td>0.74</td>
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<td>5.2</td>
<td>6</td>
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<tr>
<td>ADDM</td>
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<td>59</td>
<td>65</td>
<td>0.75</td>
<td>8.04</td>
<td>4.4</td>
<td>3</td>
</tr>
</tbody>
</table>

Table 14: Muscle-specific model parameters in the musculoskeletal model
5.2.5.2 Preferred Walking Speed

The morphological parameters of the musculoskeletal model were optimized with the assumptions that, 1) humans prefer to walk at the speed at which metabolic cost is minimal, and 2) human morphology has evolved so to minimize metabolic cost while walking at this self-selected speed.

The preferred walking speed was estimated for each of the non-amputee participants by computing cost of transport (COT) at each walking speed, defined as the non-dimensional metabolic energy consumed per unit distance:

\[ COT = \frac{P}{mgv} \]  

(25)

Where \( P \) is the average measured metabolic power, and \( m, g, \) and \( v \) are the subject mass, the gravitational acceleration, and the walking speed, respectively. The preferred speed was chosen as the one with minimum COT. For each subject's data set, the subsequent optimizations were performed for the walking data collected at this preferred speed for the given subject.

5.2.5.3 Optimization Strategy

The free parameters of the musculoskeletal model were chosen via optimization. The goal was to have the output moments from the optimization closely resemble those from the biological data. However, the presence of redundancy in the musculoskeletal system made the solution of the above problem non-unique. Previous studies have included metabolic cost of transport as an attempt to narrow down the search space to parameter sets that approximate both human kinetics and metabolism [14], [30], [37]. The present study made use of this technique, and used of both kinetic and metabolic information in the development of musculoskeletal models. Optimization methods were used to produce parameterized musculoskeletal models that, when provided with biological walking kinematics and electromyographic data, could replicate not only the joint moments, but also the COT of the humans from whom the data were taken.
An optimization scheme was used to select parameters for the musculoskeletal models, based on the collected non-amputee walking data. For each amputee participant, a corresponding non-amputee was chosen whose height and weight most closely matched that of the given amputee (Table 18). The parameter values of the musculoskeletal model were then optimized based on the selected walking data. This method produced a model that was individually tuned for each amputee.

The musculoskeletal model was simulated using MATLAB and Simulink, using the ode15s (stiff/NDF) solver. Each simulation iteration was run for two gait cycles, where only the second gait cycle was used for computation of the cost functions. This ensured that only steady-state behavior of the model was evaluated.

Since the goal of the optimization was to best match both kinetics and metabolism to the biological data, two optimization costs were used. The use of two costs meant that, instead of a single set of optimal parameters, a set of Pareto optimal parameter sets existed known as the pareto front, for which neither of the two costs could be decreased further without increasing the other. A multiobjective genetic algorithm, implemented with the MATLAB function gamultiobj, was used to find the parameter sets that were pareto optimal in this sense. Table 16 lists the optimization settings for this optimization function.

The first cost involved the error in joint moments between the model and collected walking data, henceforth referred to as the kinetic fit, using the following equation:

\[
Cost_{fit} = 1 - \left( \frac{R_{ank}^2 + R_{knee}^2 + R_{hip}^2}{3} \right)
\]

The second cost related to the simulated COT. Lower COT from the model resulted in lower optimization cost values.

Several variations of these cost functions have been utilized for determining the parameters of similar models [30], [37]. In the current work, two methods of evaluating the optimizations were evaluated.
Cost Function 1:

The optimization cost function was a metric for obtaining appropriate parameter values that reflected biological morphology. Two cost functions were explored in order to obtain the most biologically-consistent results possible for individual amputee participants. For the first cost function (used for Subjects 3 and 4), the musculoskeletal model parameters were optimized using the method described by [30]. Here, the metabolic cost was estimated using a well-known model of energy expenditure [38]. This model estimates metabolic power per unit mass as the combination of several terms:

$$\dot{E} = \dot{h}_A + \dot{h}_M + \dot{h}_{SL} + \dot{w}_{CE}$$

(27)

Here, $\dot{h}_A$ is the activation heat rate, $\dot{h}_M$ is the maintenance heat rate, and $\dot{h}_{SL}$ is the shortening/lengthening heat rate. $\dot{w}_{CE}$ is the mechanical work rate of the contractile element, normalized by muscle mass. For a full description of these terms, see [38] and the supplementary materials of [39].

The simulated metabolic cost of the model was found by integrating $\dot{E}$ for each muscle in the leg over the gait cycle, normalize each muscle's term by the corresponding muscle mass, and adding up all terms to determine the full-leg metabolism. Bilateral symmetry was assumed, so the full-leg metabolic power was doubled, and then added to the measured basal energy consumption to determine the full-body metabolic energy cost. Individual muscle masses were estimated using the following:

$$M_i = \frac{\rho F_{max,i} l_{opt,i}}{\sigma}$$

(28)

where the average density and specific tension of skeletal muscle were taken as $\rho = 1059.7 \text{ kgm}^{-3}$ and $\sigma = 0.25 \text{ MPa}$, respectively.

Using this method, it was found that the measured electromyographic data failed to produce satisfactory values for both kinetic fits and metabolic cost estimates. It was
found that the activation profiles derived from wire electrode experiments [18] provided both kinetic and metabolic results closer to that of biology. Therefore, for this optimization method, all muscle activation profiles were derived from the wire electrode literature [18].

Selection of Parameter Set

In the set of solutions on the pareto front, all can be considered optimal in the pareto sense. However, a single solution of a model’s parameter set in order to use for control purposes. To select this single solution, the same method described in [30] was used. This method depended on the fact that often, a single simulated muscle would show a disproportionate increase in metabolic cost for solutions with very good kinetic fit. This increase was considered an attempt by the optimizer to improve kinetic fit by sacrificing the metabolism of the given muscle, and was not indicative of biological morphology. Hence, the optimal solution was chosen as the one where the contribution of metabolic cost from the selected muscle was at a preset threshold, given by the equation from [30]:

\[
\Delta_{\text{muscle}} = \frac{F_{\text{muscle}}(R^2) - F_{\text{muscle}}(\min(R^2))}{F_{\text{muscle}}(\max(R^2)) - F_{\text{muscle}}(\min(R^2))}
\]  

(29)

where \(F_{\text{muscle}}\) is the fraction of overall metabolic cost attributed to the selected muscle, as a function of the mean coefficient of determination for the kinetic fit \((R^2)\). The threshold value of 0.63 for \(\Delta_{\text{muscle}}\) was also taken from [30]. For Subject 3, the HAB muscle group was used for this selection. For Subject 4, the VAS muscle group was used. These muscles were selected based on their dominance in the simulated fractional metabolic cost.
Cost Function 2:

Some biological walking data did not produce acceptable cost values with the first optimization method (those for Subjects 2, 6, and 7), or the resulting model produced insufficient average mechanical power when tested in the resulting controller (Subject 5). For these data sets, a different metabolic cost function was used.

This metabolic cost estimation function has been used in the development of a powered ankle-foot prosthesis controlled with a neuromuscular model [29]. For this method, the function for simulating metabolic cost was based on empirical measures of muscle metabolic energy consumption as a function of fiber contractile velocity and activation [40], [41], according to the following:

$$\dot{E}(t) = p(v_{CE}(t))(\alpha(t)F_{max}v_{max})$$

Here, $\dot{E}(t)$ is the instantaneous metabolic power for a given muscle, $p(v_{CE}(t))$ is the empirically-determined normalized power curve (see Figure 36), and $\alpha(t)$, $F_{max}$ and $v_{max}$ are activation, maximum isometric force, and maximum shortening velocity of the given muscle, respectively. The simulated metabolic cost was found by integrating $\dot{E}$ for each muscle in the leg over the gait cycle, and summing up all muscle for the leg, assuming bi-lateral symmetry. The measured basal energy consumption was added to the resulting value to determine the full-body metabolic energy cost.

As with the first cost function, both the simulated metabolic cost, and the average kinetic fit from Equation 26 were considered for the evaluation of parameter solutions.
Figure 36: Empirically-determined muscle metabolic power
This relationship was determined from a combination of studies on biological muscle. Figure credit: Markowitz PhD Thesis [42].

Selection of Parameter Set

It was found that the kinetic fit from Equation 26 was largely invariant in the range of solutions where the simulated COT was near that of the measured values. Therefore, in selecting a single solution, the parameter set with the best kinetic fit was chosen, provided that the simulated COT was within the standard error of the measured COT value. A similar parameter selection method had been used to select optimal parameter sets in previous work [10].

Parameter Ranges for Optimized Parameters

Upper and lower bounds for the model parameters can be found in Table 15, and were taken from [30]. These ranges were found to provide sufficient range for most
parameters, while restricting the parameter values to those within reasonable physiological ranges.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Lower Bound</th>
<th>Upper Bound</th>
</tr>
</thead>
<tbody>
<tr>
<td>$F_{\text{max}}$</td>
<td>$0.5 \times F_{\text{max SIMM}}$</td>
<td>$3.0 \times F_{\text{max SIMM}}$</td>
</tr>
<tr>
<td>$l_{sl}, l_{\text{opt mult.}}$</td>
<td>Ensure $l_m &lt; l_{\text{opt}}(1 + w)$</td>
<td>Ensure $l_m \geq l_{\text{opt}}(1 - w)$</td>
</tr>
<tr>
<td>$K_{sh}$</td>
<td>2</td>
<td>5</td>
</tr>
<tr>
<td>$\lambda_{\text{ref}}$</td>
<td>0.02</td>
<td>0.09</td>
</tr>
<tr>
<td>$\theta_{HFL}$</td>
<td>$-\frac{\pi}{18}$</td>
<td>$</td>
</tr>
<tr>
<td>$K_{HFL}$</td>
<td>$2 \times \max(\tau_{\text{hip}})/\min(\theta_{\text{hip}})$</td>
<td></td>
</tr>
</tbody>
</table>

Table 15: Optimized parameter bounds for the morphological optimization

<table>
<thead>
<tr>
<th>Optimizer Setting</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>PopulationSize</td>
<td>1000</td>
</tr>
<tr>
<td>EliteCount</td>
<td>25</td>
</tr>
<tr>
<td>Generations</td>
<td>50</td>
</tr>
<tr>
<td>MutationFcn</td>
<td>mutationadaptfeasible</td>
</tr>
<tr>
<td>CrossoverFraction</td>
<td>0.8</td>
</tr>
<tr>
<td>Vectorized</td>
<td>On</td>
</tr>
<tr>
<td>PopInitRange</td>
<td>Full Parameter Space</td>
</tr>
</tbody>
</table>

Table 16: Optimization settings for the morphological optimization

5.2.6 Spinal Reflex Model

For the purposes of device control, the muscle model one described above required a neural driving signal to produce the activation $a(t)$. Simulation work has found robust walking patterns can emerge with a gastrocnemius reflex based on positive force feedback [27], but other reflex arcs may also be useful for initiating the buildup of force [29]. For the purposes of the artificial gastrocnemius device, all three reflex types were included, where the individual gains were adjusted based on an optimization scheme.

The simulated reflex included feedback of muscle state, based on the internal sensors of biological muscle, namely force (from Golgi tendon organs) and length and velocity (from muscle spindles). These three quantities were suggested as candidates for input to a reflexive feedback loop [30]. We used the same form of
linear reflex as used in previous work in using a similar model to control an ankle-foot prosthesis [29]:

$$x(t) = G_F \cdot (F(t - \Delta t_F) - F_0) \cdot u(F - F_0)$$

$$+ G_l \cdot (l(t - \Delta t_l) - l_0) \cdot u(l - l_0)$$

$$+ G_v \cdot (v(t - \Delta t_v) - v_0) \cdot u(|v| - v_0) \quad (31)$$

Here, $G_F, G_l,$ and $G_v$ are the gains, $\Delta t_F, \Delta t_l,$ and $\Delta t_v$ are the time delays for the force, muscle length, and muscle velocity values, respectively. The function $u(\ldots)$ is a step function, indicating that each reflexes only activates for values greater than the corresponding threshold value ($F_0, l_0,$ and $v_0$ for force, length, and velocity, respectively). The time delays model the neural delay for a signal to traverse the reflex arc from muscle to spinal cord and back to the muscle. All three delays were fixed at 20 ms [43]–[45]. The reflex signal $x(t)$ represents the neural stimulation. The stimulation-activation dynamics in Equation 14 were used to obtain the muscle activation, $\alpha$.

5.2.6.1 Reflex Optimization

The reflex gains and threshold values were optimized for the gastrocnemius muscle by simulating the effects of the reflex loop with the musculoskeletal model, in the same manner as performed for a previous study for ankle-foot prosthesis control [29]. These parameters were optimized to minimize the mean squared error between the reflex-based activation signals and the muscle EMG patterns from EMG data. The optimization was performed using a genetic algorithm followed by a gradient descent method. The optimizer settings are listed in Table 17. For each data set, the gastrocnemius morphological parameters of the musculoskeletal model were taken from the results of the morphological optimization. The optimization bounds on the threshold values were determined by finding the maxima and minima of the muscle model's force, length, and velocity curves, when the model
was driven by the biological kinematic and electromyography data. The bounds for the reflex gains were set from zero to the gains that would allow any of the reflexes to provide the bulk of the EMG-based activation signals.

<table>
<thead>
<tr>
<th>Optimizer Setting</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>PopulationSize</td>
<td>100</td>
</tr>
<tr>
<td>PopInitRange</td>
<td>Full parameter space</td>
</tr>
<tr>
<td>Vectorized</td>
<td>On</td>
</tr>
<tr>
<td>EliteCount</td>
<td>10</td>
</tr>
<tr>
<td>Generations</td>
<td>100</td>
</tr>
<tr>
<td>TolFun</td>
<td>1e-8</td>
</tr>
</tbody>
</table>

Table 17: Optimization settings for reflex optimization

5.2.7 Neuromuscular Controller

For the purpose of prosthesis control, the musculoskeletal model took as input walking state information and produced desired torque commands to the robotic joint. From Figure 37, it can be seen that the existence of this reflex loop allowed the resulting neuromuscular model to require only time profiles of joint kinematics to uniquely define the internal state of the model. Hence, the combination of the musculoskeletal model with the neural reflexes could be considered an impedance-like controller, where kinematic sensor data was the input, and torque command was the output. This type of neuromuscular controller (NMC) has been used to control similar transtibial prostheses, with robustness to slopes [28] and walking speeds [29].

Unlike in the modeling steps, it was not possible to use the real-time application of the NMC precluded the use of the SIMM modeling software to obtain MTC lengths and moment arms. Instead, the mapping from joint kinematics to muscle tendon unit lengths and moment arms was obtained by performing polynomial fits to the non-amputee data, with ankle and knee angle profiles as input, and SIMM-
generated muscle-tendon lengths and moment arms as output. The muscle-tendon length and ankle joint moment arm were both fit to the ankle and knee angles using a two-dimensional second-order polynomial fit, with the two angles as input. Two-dimensional lookup tables were then generated based on these polynomial fits. The gastrocnemius knee moment arm did not depend on the ankle joint, so a 40-point lookup table, based only on knee angle, was generated that minimized error between the lookup table and the data in the least-squares sense. The NMC was disabled during the swing phase by setting the neural stimulation to zero using the stance-phase switch, and ignoring any residual commanded torque.

![Neuromuscular controller including spinal reflex loop](image)

Figure 37: Neuromuscular controller including spinal reflex loop

For additional tuning, the force gain of the reflex controller was made adjustable in real-time. This gain was chosen, since the controller’s output power in simulation, based on non-amputee data, was found to be the most sensitive to changes in this parameter.

The NMC was implemented in MATLAB Simulink, and compiled to a C executable. The SPiiPlus C library was then used to provide an interface between the executable and the EtherCAT® master controller that controlled the motor. The executable continuously polled the master controller for new sensory data.
new data was read, the executable ran one iteration of the NMC code, and updated
the master controller with a new desired joint torque value. Using this method, the
NMC produced updates to the desired torque command at 1kHz.

5.3 Results

5.3.1 Non-amputee matching

For each amputee participant, a corresponding model was used, which was based on
a height-weight matched non-amputee participant’s walking data (Table 18).

<table>
<thead>
<tr>
<th>Amputee Number</th>
<th>Years Since Amputation</th>
<th>Amputee Height (cm)</th>
<th>Matched Height (cm)</th>
<th>Amputee Weight (kg)</th>
<th>Matched Weight (kg)</th>
<th>Walking Speed (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>S2</td>
<td>9</td>
<td>180</td>
<td>180</td>
<td>92</td>
<td>90</td>
<td>1.25</td>
</tr>
<tr>
<td>S3</td>
<td>31</td>
<td>175</td>
<td>175</td>
<td>74</td>
<td>73</td>
<td>1.25</td>
</tr>
<tr>
<td>S4</td>
<td>22</td>
<td>193</td>
<td>188</td>
<td>89</td>
<td>95</td>
<td>1.5</td>
</tr>
<tr>
<td>S5</td>
<td>2</td>
<td>188</td>
<td>188</td>
<td>98</td>
<td>95</td>
<td>1.5</td>
</tr>
<tr>
<td>S6</td>
<td>9</td>
<td>175</td>
<td>180</td>
<td>110</td>
<td>103</td>
<td>1.25</td>
</tr>
<tr>
<td>S7</td>
<td>35</td>
<td>180</td>
<td>180</td>
<td>104</td>
<td>103</td>
<td>1.25</td>
</tr>
</tbody>
</table>

Table 18: Amputee Matching to Non-Amputee Participants

5.3.2 Non-amputee Data Collection

The COT values for all non-amputee participants across walking speed is shown in
Table 19. The participants all had minimal COT at either 1.25 or 1.5 m/s.

<table>
<thead>
<tr>
<th>Participant</th>
<th>0.75 m/s</th>
<th>1.0 m/s</th>
<th>1.25 m/s</th>
<th>1.5 m/s</th>
<th>1.75 m/s</th>
<th>2.0 m/s</th>
<th>Optimal Speed (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NA3</td>
<td>0.419</td>
<td>0.364</td>
<td>0.326</td>
<td>0.339</td>
<td>0.362</td>
<td>0.380</td>
<td>1.25</td>
</tr>
<tr>
<td>NA4</td>
<td>0.462</td>
<td>0.386</td>
<td>0.371</td>
<td>0.375</td>
<td>0.404</td>
<td>0.467</td>
<td>1.25</td>
</tr>
<tr>
<td>NA6</td>
<td>0.481</td>
<td>0.389</td>
<td>0.347</td>
<td>0.341</td>
<td>0.368</td>
<td>0.400</td>
<td>1.5</td>
</tr>
<tr>
<td>NA7</td>
<td>0.395</td>
<td>0.339</td>
<td>0.305</td>
<td>0.315</td>
<td>0.348</td>
<td>0.408</td>
<td>1.25</td>
</tr>
</tbody>
</table>

Table 19: Metabolic cost of transport for non-amputee participants

95
5.3.3 Morphological Parameter Optimization:
The optimized morphological parameters for the gastrocnemius muscle are shown in Table 20.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>S2</th>
<th>S3</th>
<th>S4</th>
<th>S5</th>
<th>S6</th>
<th>S7</th>
</tr>
</thead>
<tbody>
<tr>
<td>$F_{max}$</td>
<td>1094</td>
<td>1052</td>
<td>1094</td>
<td>1936</td>
<td>3049</td>
<td>3049</td>
</tr>
<tr>
<td>$l_{sl}$</td>
<td>0.4565</td>
<td>0.4115</td>
<td>0.4565</td>
<td>0.4568</td>
<td>0.4217</td>
<td>0.4217</td>
</tr>
<tr>
<td>$l_{opt}$</td>
<td>0.0504</td>
<td>0.0454</td>
<td>0.0504</td>
<td>0.0504</td>
<td>0.0465</td>
<td>0.0465</td>
</tr>
<tr>
<td>$K_{sh}$</td>
<td>2.58</td>
<td>2.78</td>
<td>2.58</td>
<td>2.722</td>
<td>2.84</td>
<td>2.84</td>
</tr>
<tr>
<td>$\lambda_{ref}$</td>
<td>0.0593</td>
<td>0.0684</td>
<td>0.0593</td>
<td>0.0599</td>
<td>0.0846</td>
<td>0.0846</td>
</tr>
</tbody>
</table>

Table 20: Optimized morphological parameters

<table>
<thead>
<tr>
<th>Cost Value</th>
<th>S2</th>
<th>S3</th>
<th>S4</th>
<th>S5</th>
<th>S6</th>
<th>S7</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R^2$</td>
<td>0.71</td>
<td>0.76</td>
<td>0.80</td>
<td>0.85</td>
<td>0.80</td>
<td>0.80</td>
</tr>
<tr>
<td>COT</td>
<td>0.360</td>
<td>0.324</td>
<td>0.246</td>
<td>0.330</td>
<td>0.301</td>
<td>0.301</td>
</tr>
</tbody>
</table>

Figure 38: Final cost values for the morphological parameter optimization
The $R^2$ values were computed using Equation 26. The COT values were the simulated metabolic cost of transport values, as computed by the musculoskeletal model, when given the final optimized parameter set and biological non-amputee data as input.

5.3.4 Reflex Optimization
The optimized parameter values for the reflex loop are shown in Table 21.

<table>
<thead>
<tr>
<th>S2 (NA4)</th>
<th>S3 (NA3)</th>
<th>S4 (NA6)</th>
<th>S5 (NA6)</th>
<th>S6 (NA7)</th>
<th>S7 (NA7)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$F_{gain}$</td>
<td>3.687e-4</td>
<td>6.238e-4</td>
<td>3.994e-4</td>
<td>6.185e-4</td>
<td>5.11e-4</td>
</tr>
<tr>
<td>$F_{thresh}$</td>
<td>0.1786</td>
<td>0.3935</td>
<td>0.3624</td>
<td>0.291</td>
<td>0.532</td>
</tr>
<tr>
<td>$L_{gain}$</td>
<td>6.88</td>
<td>5.22</td>
<td>1.3465</td>
<td>6.32</td>
<td>8.19</td>
</tr>
<tr>
<td>$L_{thresh}$</td>
<td>0.0507</td>
<td>0.0490</td>
<td>0.0528</td>
<td>0.0448</td>
<td>0.0382</td>
</tr>
<tr>
<td>$V_{gain}$</td>
<td>0.9606</td>
<td>1.34</td>
<td>2.00</td>
<td>0.0193</td>
<td>2.00</td>
</tr>
<tr>
<td>$V_{thresh}$</td>
<td>0.1230</td>
<td>-0.1175</td>
<td>-0.1168</td>
<td>-0.0897</td>
<td>0.0340</td>
</tr>
</tbody>
</table>

Table 21: Optimized neuromuscular model reflex parameters
Chapter 6

Powered Device Clinical Evaluation

6.1 Overview

The powered artificial gastrocnemius (AG) had a distinct advantage over the quasi-passive counterpart: the powered AG had the capability of producing positive net mechanical power to the wearer. This advantage allowed the powered device to make up for losses in the attachment. The neuromuscular controller was designed to provide this positive power in a biologically-consistent manner.

It was hypothesized that this powered AG would reduce the metabolic cost of walking for transtibial amputees, compared to the control condition where the AG behaved as a free-joint at the affected-side knee through the use of zero-torque control. It was also hypothesized that this metabolic advantage would result from large reductions in affected knee biological joint moments and powers, since the powerful motor would be capable of far higher torques and powers than the quasi-passive device.

6.2 Methods

6.2.1 Data Collection Protocol

The walking speed for a given amputee was set as the preferred speed determined from the matched non-amputee walking data (Table 19). This speed was set for all walking conditions.

The clinical evaluation was conducted at MIT (Cambridge, MA) and was approved by MIT’s Committee on the Use of Humans as Experimental Subjects (COUHES). Each participant provided written, informed consent before data collection was initiated.
Each amputee was first asked to don only the powered prosthesis. They were then instructed to walk on the treadmill while the prosthesis power was adjusted so to achieve average prosthesis power within two standard deviations from the average biological ankle power at the given walking speed [17].

Figure 39: Experimental setup with the powered Artificial Gastrocnemius
An amputee participant walks in an experimental trial with the powered artificial gastrocnemius system, including full motion capture, electromyography, and metabolic cost gas exchange measurement system.

Next the amputee donned the artificial gastrocnemius in along with the powered prosthesis. A short time period was given to ensure that both the prosthesis and the artificial gastrocnemius orthosis were functioning properly. Then the amputee was asked to walk on an instrumented treadmill at the given walking speed with the same data collection methods used for matched the non-amputee participants in Chapter 5, including the instrumented treadmill, reflective markers, and EMG
system. The EMG electrodes were selected on the affected leg to monitor muscle activity in response to the device. The following locations were chosen: hamstrings (semimembranosis, biceps femoris long head), and quadriceps (vastus medialis, vastus lateralis). Each trial consisted of 90 seconds of data collection.

In addition to using the optimized parameters of the neuromuscular controller (NMC), additional conditions were tested. The force feedback reflex gain of the NMC was chosen to be adjusted for these additional conditions, as it was found in simulation to have the greatest effect on average power output from the simulated controller. While a participant walked on the treadmill with the AG and powered prosthesis, the force gain of the controller was incrementally increased from its default value (the Default gain setting) until either further increases in gain were said to be undesirable by the participant, or that the average power from the device was no longer increasing with higher controller gain (the Max gain setting). The gain was then decreased until the average power coming from the device was in the range of the biological gastrocnemius, of 3-5 Joules (the Bio gain setting) [22], [23]. If the Default setting provided less than or equal to 3-5 Joules of net mechanical work per step, an additional gain was tested, which was set to mid-way between the Default gain and the Max gain. This additional gain setting was called Half.

Once the aforementioned conditions were recorded with the motion capture system, the same conditions were collected, in random order, while participants wore the same Cosmed gas analysis system used for the non-amputee participants in Chapter Chapter 5 for estimating metabolic energy consumption. The participants were asked to perform one standing trial to measure standing metabolism. Each metabolic trial lasted 6 minutes in order to achieve steady state metabolism.

6.2.2 Data Processing

Fourth-order Butterworth filters were used to filter the marker position and ground reaction force data with 6 Hz and 25 Hz cutoff frequencies, respectively. The marker
and force data were post-processed through the SIMM (Musculographics Inc., Evanston, IL) inverse dynamics module to produce joint moments and angles. A point-mass of 2 kg was added at the affected-side knee joint of the SIMM model to represent the dynamical effects of the robotic joint and conduit mass. Affected-side biological knee moment contribution was computed by subtracting the measured AG knee moment from the total knee moment estimated from the inverse dynamics.

Amputee EMG signals were first processed internally by the EMG system by applying a 4th-order bandpass filter with a pass band from 20 Hz to 450 Hz. Any DC offset was removed, and the signals were then rectified, and smoothed using a 3 Hz, 3rd-order Butterworth filter. The EMG data were then cut to gait cycles and averaged in the same manner as the motion and force data. The resulting EMG data were then normalized so that the average values of the activation signals matched those from [18] during the range of activation for the literature data. For each subject and walking condition, the EMG signals were averaged over the period, on average, when the spring-clutch was engaged and generating torque.

Metabolic cost for each walking speed was computed by taking average oxygen and carbon dioxide data over a two-minute window at the end of each six-minute trial. The metabolic power was computed using the equation

\[ P = K_{O_2} \dot{V}_{O_2} + K_{CO_2} \dot{V}_{CO_2} \]  

where \( P \) is the metabolic power in Watts, \( \dot{V}_{O_2} \) is the volume flow rate of Oxygen inhaled, \( \dot{V}_{CO_2} \) is the volume flow rate of carbon dioxide exhaled, and \( K_{O_2} \) and \( K_{CO_2} \) are constants with values from literature [19], given as \( K_{O_2} = 16,580 \) W/L and \( K_{CO_2} = 4,510 \) W/L. The above equation is only valid for conditions when the metabolism is primarily aerobic. As a verification of this condition, the respiratory exchange ratio (RER), defined as \( \dot{V}_{CO_2}/\dot{V}_{O_2} \) was monitored, and only metabolic results with RER values less than 1.1 were considered.
Gait events were determined using vertical ground reaction force data from the embedded forceplates. Approximate event timing was found by determining the times when the force increased beyond a 40 N threshold. Exact heelstrike and toeoff times were found by progressing backward, and forward in time, respectively, until the force value dropped to zero. Data were then cut to gait cycles based on heelstrike times, and resampled to 101 points. Gait cycles were discarded for the beginning and end of each trial, during the speed transients of the treadmill. Gait cycles in which the stride times were below 0.7 seconds or above 1.3 seconds, or in which a foot crossed the mid-line of the treadmill were also discarded.

The walking metabolism was compared for each amputee participant across trials, where each trial involved a different value for the force feedback reflex gain of the NMC. The optimal gain was found as the one that resulted in the lowest metabolic power for a given amputee. The walking condition (called the OPT condition) corresponding to this optimal gain was then compared to the walking condition where the AG was controlled to behave as a free-joint (called the ZERO condition).

Joint powers were computed as the product of joint velocity from SIMM-derived joint kinematics from motion-capture, and joint moments, where positive power was defined as that produced by the joint on the environment. Positive and negative joint work contributions were defined as the time integral of the positive and negative components of joint power, respectively, integrated over the late stance knee flexion phase and divided by the time in that phase. Flexion and extension moment impulse values were computed in the same way, but using joint moments instead of powers.

Where available, average mechanical power from the powered prosthesis was taken from prosthesis telemetry data, computed using the prosthesis onboard torque and angle sensors. For two subjects (S3 and S4), these data were not available, so mechanical power was computed in the same manner as the other joints, using SIMM-derived kinetics from the motion-capture data.
Average mechanical power from the knee orthosis was also analyzed by taking knee orthosis average power during late stance knee flexion, from mid-stance maximum knee extension to toeoff. This region of the gait cycle was chosen for analysis because, as the knee flexes from the maximum extension angle, it provides an opportunity for the AG to provide positive power to the wearer.

Theoretical reductions in metabolic power were computed by assuming the average mechanical power provided by the combined AG knee actuator and ankle prosthesis in the OPT condition, as compared to the ZERO condition (with zero net work assumed for the AG knee in the ZERO condition), could perfectly replace muscle fiber mechanical power. With the efficiency of skeletal muscle assumed to be 25% [46], the reduction in muscle metabolic power was computed as 4 times the observed mechanical power provided by the AG knee actuator and ankle-foot prosthesis in the OPT condition. The resulting change in metabolic power was then compared to the measured metabolic power from the ZERO condition to determine theoretical percent reduction.

Statistics were run on various gait features by performing a two-sided, paired t-test between the features during the OPT condition and ZERO condition. Statistically significant differences were defined as those with P-value less than or equal to 0.05.

Hamstring electromyography changes were summarized by first finding the region in the gait cycle where the AG was active in the OPT condition. For a given participant, this region was determined by finding the range where the AG applied knee moment was at least 10% of its average maximum value for the given participant. Each of the two hamstring muscles were then each averaged for each participant over the aforementioned ranges. This process was performed for each of the OPT and ZERO conditions, for each participant, resulting in averaged EMG values for the two walking conditions, across participants.
6.3 Results

6.3.1 AG overall device performance

From the metabolic analysis, the optimal (OPT) conditions were found for each participant. The force reflex gain settings and average output power from the AG joints are summarized in Table 22.

<table>
<thead>
<tr>
<th>Participant</th>
<th>S2</th>
<th>S3</th>
<th>S4</th>
<th>S5</th>
<th>S6</th>
<th>S7</th>
</tr>
</thead>
<tbody>
<tr>
<td>Condition Name</td>
<td>Bio</td>
<td>Max</td>
<td>Default</td>
<td>Default</td>
<td>Default</td>
<td>Half</td>
</tr>
<tr>
<td>Force Gain Scaling</td>
<td>0.1</td>
<td>8</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.5</td>
</tr>
<tr>
<td>Force Gain (1/N)</td>
<td>3.69e-5</td>
<td>4.99e-3</td>
<td>3.994e-4</td>
<td>6.185e-4</td>
<td>5.11e-4</td>
<td>7.66e-4</td>
</tr>
<tr>
<td>AG Mean Power (W/kg)</td>
<td>0.054</td>
<td>0.118</td>
<td>0.028</td>
<td>0.027</td>
<td>0.105</td>
<td>0.031</td>
</tr>
<tr>
<td>Change in BiOM average power from ZERO (W/kg)</td>
<td>-0.001</td>
<td>+0.020</td>
<td>+0.056</td>
<td>+0.077</td>
<td>+0.011</td>
<td>+0.009</td>
</tr>
</tbody>
</table>

Table 22: Walking conditions for minimal metabolic power

For the BiOM average power changes, positive values indicate that the given robotic joint produced greater net work in the OPT condition than in the ZERO condition.

6.3.2 Metabolic Power

Walking metabolism is shown in Table 23 for the walking conditions in which the metabolic cost was found to be lowest, compared to the theoretically-calculated values using device average power. Large percentage reductions in metabolism were found in two of the six participants. Two participants showed small metabolic reductions, and the two remaining participants had metabolism values higher with the AG intervention, compared to the zero-torque condition. Within the participants with a metabolic reduction in the OPT condition, these metabolic reductions were well-correlated with the theoretical values based on device mechanical power ($r = 0.9$), as shown in Table 23. However, Subject 2 and Subject 4 displayed increases in metabolism in the OPT condition, as compared to the ZERO condition, which was not predicted with the theoretical calculations.
Table 23: Metabolic cost of transport using the Artificial Gastrocnemius

Values given represent the net metabolic power, above that of basal metabolic rate. Positive values for percent change indicate that the participant required less metabolic power with the neuromuscular controller, compared to the control condition of zero-torque at the knee. Theoretical metabolic reduction, based on the change in knee orthosis and ankle-foot prosthesis applied average power from the ZERO condition to the OPT condition, and assuming 25% muscle efficiency, is also shown for comparison.

<table>
<thead>
<tr>
<th>Condition</th>
<th>S2</th>
<th>S3</th>
<th>S4</th>
<th>S5</th>
<th>S6</th>
<th>S7</th>
</tr>
</thead>
<tbody>
<tr>
<td>ZERO Condition Net Metabolic Power (W/kg)</td>
<td>2.95</td>
<td>3.51</td>
<td>3.60</td>
<td>5.06</td>
<td>3.34</td>
<td>3.52</td>
</tr>
<tr>
<td>OPT Condition Net Metabolic Power (W/kg)</td>
<td>3.13</td>
<td>3.08</td>
<td>3.87</td>
<td>4.95</td>
<td>2.97</td>
<td>3.38</td>
</tr>
<tr>
<td>Percent Change in Metabolic Power from ZERO condition to OPT condition</td>
<td>5.8</td>
<td>-12.1</td>
<td>7.4</td>
<td>-2.3</td>
<td>-11.1</td>
<td>4.0</td>
</tr>
<tr>
<td>Theoretical Percent Change in Metabolic Power</td>
<td>-7.2</td>
<td>-15.7</td>
<td>-9.3</td>
<td>-8.3</td>
<td>-14.1</td>
<td>4.6</td>
</tr>
</tbody>
</table>
6.3.3 Kinematics

Joint kinematics averaged across participants are shown in Figure 40, and scalar kinematic quantities are summarized in Table 24. The most prominent change in the kinematics was on the affected-side in the stance phase, near 50 percent gait cycle (terminal stance). On average, participants had greater ankle peak flexion angles, and greater knee, and hip peak extension angles. The remaining kinematic features remained largely consistent between walking conditions.

<table>
<thead>
<tr>
<th>Peak Joint Angle (rad)</th>
<th>Mean</th>
<th>Std. Dev.</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max Ankle Dorsiflexion Angle</td>
<td>0.0193</td>
<td>0.0099</td>
<td>0.005*</td>
</tr>
<tr>
<td>Max Ankle Plantar Flexion Angle</td>
<td>-0.0210</td>
<td>0.0321</td>
<td>0.17</td>
</tr>
<tr>
<td>Max Knee Stance Flexion Angle</td>
<td>0.0158</td>
<td>0.0144</td>
<td>0.04*</td>
</tr>
<tr>
<td>Max Knee Stance Extension Angle</td>
<td>0.0514</td>
<td>0.0323</td>
<td>0.01*</td>
</tr>
<tr>
<td>Max Knee Swing Flexion Angle</td>
<td>0.0019</td>
<td>0.0261</td>
<td>0.86</td>
</tr>
<tr>
<td>Max Hip Flexion Angle</td>
<td>-0.0015</td>
<td>0.0106</td>
<td>0.75</td>
</tr>
<tr>
<td>Max Hip Extension Angle</td>
<td>0.0617</td>
<td>0.0455</td>
<td>0.02*</td>
</tr>
</tbody>
</table>

Table 24: Kinematic differences with the powered artificial gastrocnemius

Differences in kinematics between the ZERO and OPT walking conditions with the Artificial Gastrocnemius. Positive values for the means indicate that the quantity was greater during the OPT condition, as compared to the ZERO condition. An asterisk next to a P-value indicate a significant difference between the two walking conditions, at the 5 percent level.
Figure 40: Individual joint angles during walking for amputee participants. Shown are kinematics over two walking conditions, averaged across participants. The zero-torque condition of the AG (thick solid blue line) is shown compared to the AG controlled with the neuromuscular model (thin solid red line). Shaded regions represent +/- 1 standard deviation for these curves. Non-amputee data, averaged across participants, is shown for reference, with the mean (dashed black line) +/- one standard deviation (dotted lines).

6.3.4 Joint Moments

Joint moments, averaged across participants is shown in Figure 42. The contribution of affected-side knee joint moment from the AG and biological knee is
shown in Figure 41. As shown in Table 25, the affected-side biological knee joint flexion moment impulse in late stance flexion was significantly reduced by an average of 0.0223 Nm*s/kg ($p = 0.03$), in response to the knee flexion moment applied by the AG. Meanwhile, the total affected-side knee moment impulse in late stance flexion did not significantly change between the two walking conditions, across participants ($p = 0.48$) (here, the difference in moment impulse was only 0.0083 Nm*s/kg.).

Figure 41: Components of affected-side knee moment with the powered AG

The plots are for during walking for amputee participants over two walking conditions, averaged across participants. The net value, which is the sum of biological knee and AG values, is shown for two walking conditions: zero-torque control of the AG (thick solid blue line) and with the AG controlled by the neuromuscular model (thin solid red line). The contribution from the biological knee joint is also shown for the neuromuscular model condition (dashed thin green line). Non-amputee data, averaged across participants, is shown for reference, with the mean (dashed black line) +/- one standard deviation (dotted lines).
Figure 42: Net joint moments with the powered AG

Plots are showing averaged data across amputee participants over two walking conditions. The zero-torque condition of the AG (thick solid blue line) is shown compared to the AG controlled with the neuromuscular model (thin solid red line). Shaded regions represent +/- 1 standard deviation for these curves. Non-amputee data, averaged across participants, is shown for reference, with the mean (dashed black line) +/- one standard deviation (dotted lines).
Table 25: Joint moment impulse in late stance flexion with the powered AG
Comparison of individual joint moment impulses during the period from peak knee extension angle in stance phase until toeoff. Shown are differences between the ZERO and OPT walking conditions with the Artificial Gastrocnemius. Positive values for the means indicate that the quantity was greater during the OPT condition, as compared to the ZERO condition. An asterisk next to a p-value indicate a significant difference between the two walking conditions, at the 5 percent level.

<table>
<thead>
<tr>
<th>Joint Moment Impulse (Nm*s/kg)</th>
<th>ZERO</th>
<th>OPT</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Affected Side</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee Flexion</td>
<td>0.0419 +/- 0.0263</td>
<td>0.0502 +/- 0.0448</td>
<td>0.48</td>
</tr>
<tr>
<td>Knee Extension</td>
<td>-0.0344 +/- 0.0351</td>
<td>-0.0315 +/- 0.0424</td>
<td>0.67</td>
</tr>
<tr>
<td>Knee Bio. Flexion</td>
<td>0.0413 +/- 0.0262</td>
<td>0.0190 +/- 0.0252</td>
<td>0.03*</td>
</tr>
<tr>
<td>Knee Bio. Extension</td>
<td>-0.0343 +/- 0.0352</td>
<td>-0.0508 +/- 0.0543</td>
<td>0.25</td>
</tr>
<tr>
<td>Hip Flexion</td>
<td>0.1320 +/- 0.0913</td>
<td>0.0959 +/- 0.1016</td>
<td>0.10</td>
</tr>
<tr>
<td>Hip Extension</td>
<td>-0.0438 +/- 0.0321</td>
<td>-0.0418 +/- 0.0277</td>
<td>0.82</td>
</tr>
<tr>
<td><strong>Unaffected Side</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle Dorsiflexion</td>
<td>0.0012 +/- 0.0018</td>
<td>0.0010 +/- 0.0013</td>
<td>0.82</td>
</tr>
<tr>
<td>Ankle Plantar Flexion</td>
<td>-0.3447 +/- 0.1076</td>
<td>-0.3482 +/- 0.1222</td>
<td>0.80</td>
</tr>
<tr>
<td>Knee Flexion</td>
<td>0.0812 +/- 0.0469</td>
<td>0.0892 +/- 0.0400</td>
<td>0.25</td>
</tr>
<tr>
<td>Knee Extension</td>
<td>-0.0295 +/- 0.0361</td>
<td>-0.0194 +/- 0.0259</td>
<td>0.09</td>
</tr>
<tr>
<td>Hip Flexion</td>
<td>0.1194 +/- 0.0679</td>
<td>0.1110 +/- 0.0582</td>
<td>0.23</td>
</tr>
<tr>
<td>Hip Extension</td>
<td>-0.0406 +/- 0.0254</td>
<td>-0.0317 +/- 0.0241</td>
<td>0.38</td>
</tr>
</tbody>
</table>

Looking more closely at this change in affected-side biological knee moment, it is clear that the observed changes do not only represent a decrease in biological knee flexion moment, but potentially an increase in knee extension moment as well. Plots of individual participants' knee moments Figure 55 indicates that, in the presence of AG flexion torque, the biological knee increases in extension moment against the device, for some participants. Figure 43 illustrates this effect. Note that the participants with the lowest increase in knee flexion moment impulse (S3, S6) are those with the greatest metabolic advantages with the AG device.
This plot shows the affected-side knee extension moment impulse during late stance knee flexion, taken from maximum stance knee extension angle to toeoff. The extension moment represents the extent to which the biological joint is actively fighting the flexion assist from the device.

6.3.5 Powers

Positive muscle work is far more metabolically costly than negative muscle work. Therefore, we review positive powers at individual joints in order to determine possible sources of the observed changes in metabolism.

Affected-side biological joint positive and negative work values during late stance knee flexion for both the ZERO and OPT walking conditions can be found in Table 26. The affected-side hip shows a decrease in positive work from the ZERO to the OPT condition by 0.074 W/kg (p = 0.004). As can be shown in Figure 44, this reduction primarily occurs at the end of stance phase in late stance knee flexion.

Further, average positive power of the affected-side hip was found to be well-correlated (r = 0.86) to the power from the AG device (Figure 45), where higher
powers from the device tended to reduce the positive power at the affected-side hip. This trend occurred for all participants, although to varying degrees.

Figure 44: Joint powers during walking with the powered AG
Net joint powers, averaged across participants, during walking for amputee participants over two walking conditions. The zero-torque condition of the AG (thick solid blue line) is shown compared to the AG controlled with the neuromuscular model (thin solid red line). Shaded regions represent +/- 1 standard deviation for these curves. Non-amputee data, averaged across participants, is shown for reference, with the mean (dashed black line) +/- one standard deviation (dotted lines).
Table 26: Joint work in late stance knee flexion with the powered AG

Comparison of individual joint positive and negative work during the period from peak knee flexion angle in stance phase until toeoff. Shown are differences between the ZERO and OPT walking conditions with the Artificial Gastrocnemius. Positive values for the means indicate that the quantity was greater during the OPT condition, as compared to the ZERO condition. An asterisk next to a p-value indicate a significant difference between the two walking conditions, at the 5 percent level.
Figure 45: AG effect on affected-side hip positive power

Relationship between AG average power per gait cycle and the average positive power produced by the affected-side hip joint. The vertical axis represents the difference in affected-side hip power between the given walking condition and the ZERO condition. (Positive values indicate an increase in affected-side hip power compared to the ZERO condition.) All walking conditions are included. Each point on the plot represents the average across all gait cycles within a walking trial. Each marker represents a different amputee participant.

Considering other joints (Figure 46), the participants with metabolic reductions had the greatest reductions in total positive work summed over all biological joints. Subject 6 displayed the largest positive power reduction in the affected-side knee, while Subject 3 displayed the largest reduction in affected side positive hip work. It is clear, however, that this very large change in positive hip work does not occur on average over the entire gait cycle for Subject 3. Looking closer at the hip power for Subject 3 in Figure 59, the affected-side hip power increases in the OPT condition early in the gait cycle before being drastically reduced later (45-60% gait cycle), compared to the ZERO condition.
Figure 46: Changes in joint positive work in late stance knee flexion.
The biological positive work contributions of the affected-side knee and hip are shown, in addition to the average total positive work of all three joints on the unaffected side, and the total positive work across all biological joints. The differences in power represent the change from the ZERO condition to the OPT condition. Each color represents a different study participant.

6.3.6 Electromyography
Average electromyographic time-series data can be seen in Figure 47. The hamstring EMG activity of the affected-side leg generally reflected the kinetic changes at the affected-side biological knee joint; the activity of the semimembranosus and biceps femoris muscles decreased in the mid to late stance part of the gait cycle, when the AG was producing at least 10% of its maximum torque. The biceps femoris activation change was statistically significant at the 5% level (p = 0.03) and the semimembranosus change showed a strong trend toward significance (p = 0.07).

Average hamstring muscle reduction values during this same range of the gait cycle are summarized in Figure 48. Four of the six participants displayed large
percentage reductions in hamstring muscle activity in the OPT condition, as compared to the ZERO condition, with these reductions ranging from 17 to 51 percent of the values in the ZERO condition.

Figure 47: EMG changes when walking with the powered AG. Changes in knee muscle activity of amputees walking with the Artificial Gastrocnemius. Positive values indicate a larger amount of activation in the OPT condition, compared to the ZERO condition.

There was a fairly high correlation ($r = 0.74$) between hamstring EMG reductions with the OPT condition and the amount of EMG activity in the ZERO condition. Figure 50 shows this relationship. The largest reductions in activation were in those with the greatest activation in the ZERO condition.

In contrast with the hamstrings, the quadriceps muscles generally revealed an increase in activation while the AG was active (Figure 49), where both the vastus lateralis and vastus medialis showed statistically significant ($p < 0.05$) increase in activation in the OPT condition as compared to the ZERO condition (Table 27).
<table>
<thead>
<tr>
<th>Muscle</th>
<th>ZERO</th>
<th>OPT</th>
<th>Difference</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Biceps Femoris</td>
<td>0.0919 +/- 0.0456</td>
<td>0.0718 +/- 0.0334</td>
<td>-0.0201 +/- 0.0161</td>
<td>0.03*</td>
</tr>
<tr>
<td>Semimembranosis</td>
<td>0.2005 +/- 0.0848</td>
<td>0.1474 +/- 0.0599</td>
<td>-0.0531 +/- 0.0561</td>
<td>0.07</td>
</tr>
<tr>
<td>Vastus Lateralis</td>
<td>0.0948 +/- 0.0346</td>
<td>0.1320 +/- 0.0676</td>
<td>0.0371 +/- 0.0354</td>
<td>0.05*</td>
</tr>
<tr>
<td>Vastus Medialis</td>
<td>0.1224 +/- 0.0861</td>
<td>0.1812 +/- 0.1275</td>
<td>0.0588 +/- 0.0471</td>
<td>0.03*</td>
</tr>
</tbody>
</table>

Table 27: Mean EMG for the knee muscles for walking with the powered AG

Data are averaged across participants, comparing between the ZERO and OPT walking conditions with the Artificial Gastrocnemius during the period where the AG was active (> 10% of maximum torque). An asterisk next to a P-value indicate a significant difference between the two walking conditions, at the 5 percent level. The difference column indicates the change in the quantity from the ZERO to the OPT condition, where positive values indicate that the quantity was higher in the OPT condition than in the ZERO condition.

![Figure 48: Percent change in hamstrings EMG with the powered AG](image)

Changes are shown in muscle activity from ZERO condition to the OPT condition. Values are averaged over the period in the gait cycle when the AG was providing at least 10% of the maximum torque for the given participant.
Figure 49: Average changes in quadriceps EMG with the powered AG. Changes are shown in muscle activity from the ZERO condition to the OPT condition. Values are averaged over the period in the gait cycle when the AG was providing at least 10% of the maximum torque for the given participant.

Figure 50: EMG changes compared to baseline activity. EMG reductions are correlated to EMG activity during the ZERO condition. The values here are averaged over the range of the gait cycle where the AG applied torque was at least 10% of its maximum.
6.4 Discussion

6.4.1 Kinetics
The kinetic changes support the hypothesis that the AG causes a reduction in affected biological knee flexion moment impulse. In light of the short trial times, this result indicates that participants were capable of rapid kinetic responses to the interventions, on the order of several minutes or less. The knee moment impulse reduction may stem from an attempt by the participants to prevent the total knee moment from changing in the presence of the external disturbance. This result is consistent with other studies, as humans have been observed to maintain invariant joint moments at both the ankle [20] and hip joints [21] when an external torque from an exoskeleton is applied. This finding has a potential consequence: if the applied joint moment from the AG exceeded the affected knee moment from the ZERO condition, the participant could attempt to maintain total joint kinetics by providing an active extension moment with their quadriceps, so as to prevent the total knee flexion moment from increasing. Indeed, the kinetic behavior of several participants supports this hypothesis, where four of the seven participants displayed increases in knee extension moment impulse during the late stance knee flexion phase (Figure 43). This behavior may put an upper-limit on the torque flexion torque that can be provided by the AG without incurring a cost from the quadriceps in the form of active knee extension efforts.

6.4.2 Kinematics
The kinematic changes with the active AG condition indicate that the participants may put a greater importance on the kinetic behavior, rather than the kinematic. Also, the greater leg flexion on average for the participants during the middle of stance phase may contribute to the increase in quadriceps activity: a more flexed knee could cause the ground reaction force vector to pass more posterior to the knee joint, thereby causing an increased external knee flexion moment that would need to be supported by an increased biological knee extension moment. However, it is
possible that these kinematic changes could be eliminated through longer training periods with the device.

6.4.3 EMG

In healthy non-amputees, the hamstring muscles are typically not active between midstance and toeoff, when the gastrocnemius is active [18], and thus, any hamstring activation during this period can be considered pathological. Thus, the reductions in hamstring activity with the powered AG represent a reduction in pathology. The hamstring activity reductions are also consistent with the observed changes in knee moment. Although some subjects did not show as much of a hamstring activation reduction, those typically had lower pathological hamstring activation initially.

The increase in quadriceps activation for most participants may partly be associated with co-contraction of the muscles in response to the forces on the body from the thigh cuff of the knee brace, or for general co-contraction for stabilizing the joint. However, in light of reductions in hamstring muscle activity, some participants likely use these muscles for the generation of the observed increases in biological knee extension moment (Figure 43), perhaps to maintain total joint moment in response to excessive applied device flexion torque.

6.4.4 Metabolic Results

Although a statistically-significant metabolic reduction across participants was not found in this study, the large percentage reductions in metabolism for some participants indicate that it is likely a metabolic advantage may be achieved with a powered artificial gastrocnemius, provided appropriate conditions are satisfied. The majority of participants displayed metabolic reductions comparable to the theoretical values. Therefore, if greater power could be provided by the device, it is likely that a larger, metabolic reduction would be observed across participants on
average. However, some of this mechanical power from the device may still be lost as a result of imperfect attachment to the body.

However, the over-activation of the quadriceps muscles may adversely affect metabolism, as these muscle are large and metabolically costly to activate. It is particularly important, therefore, that when additional power is provided, the flexion moments from the device are kept low enough as to avoid compensatory quadriceps activation.

6.4.5 Potential Mechanisms for Metabolic Reduction
First, it is possible that this reduction in knee moment itself may play a role in the metabolic reduction in some of the participants, as the production of muscle force exacts a metabolic cost [47]. However, the isometric generation of isometric muscle force alone is comparatively efficient, compared to the generation of mechanical work. Therefore, other factors are likely at play.

The similarity of the observed metabolic reductions to the theoretical values indicates that much of the power provided by the AG goes to replacing muscle fiber positive work, as modeled in the theoretical cost values. Therefore, we look to positive joint work to determine potential mechanisms for the observed metabolic results. The consistent trend in affected-side hip positive power vs AG applied average power indicates that energy may be transferred up to the hip from the knee. These hip power reductions coincide with the initiation of hip flexion during preswing, indicating that the AG probably helps to initiate leg swing. This result is consistent with what is known about the normal function of the human gastrocnemius muscle [48], [49].

Since the metabolic improvements varied greatly across participants, it is useful to consider likely mechanisms for these improvements individually. Subject 6 had the largest reduction in affected-side knee biological positive power, indicating that not all power from the device was transferred to other joints. Subject 3 increased his hip
power early in the stance period, but then drastically decreased it near the end of stance. This behavior could be an attempt to store energy in the AG with the hip early in the stance period, when the lower joint velocities could allow the hip musculature to be in a more optimal fiber state for efficiency. This increase in hip work could then be compensated by a comparatively large decrease in hip muscle metabolic cost near toeoff, when the large forces and velocities could be more metabolically costly. Subject 7, did not receive as much power from the AG, but nonetheless had a small metabolic improvement with the device. Here, his comparatively low increase in antagonistic quadriceps activity, paired with the reduction in hip positive power may have contributed to the metabolic reduction for this participant. Subject 2 and 4 who both did not show a metabolic reduction, had increased positive joint work in their unaffected side with the OPT condition, which likely contributed to their increase in metabolism. The slight metabolic reduction in Subject 5, despite the increase in knee extension moment, may have resulted from the relatively large increase in average mechanical power from the ankle-foot prosthesis, which occurred despite attempts to maintain consistent prosthesis power across conditions.

6.4.6 Conclusions
The findings from this study provide suggestive evidence that an artificial gastrocnemius can improve metabolic cost of walking in transtibial amputees, and that this benefit likely stems from an assistance in the initiation of leg swing through transfer of mechanical power to the affected-side hip.

However, one must be careful when designing the control system of the AG to ensure that the torque applied by the device does not approach or exceed the torque generated by the affected-side knee before the intervention.
Chapter 7
Conclusions and Future Work

7.1 Conclusions

This thesis presents both technological and scientific developments. Technologically, the two artificial gastrocnemius devices presented tackle the challenge of producing a functional, yet wearable device through two different approaches. The quasi-passive device made use of an approximation of the human gastrocnemius in order to keep the device simple, small, and lightweight. In contrast, the development of the powered system involved a more mechanistic approach, wherein a tether allowed for the full function of the gastrocnemius muscle to be emulated without incurring the cost of excessive donned mass.

The scientific findings in this thesis emphasize the difficulty of reducing metabolic demand in humans, even in the presence of an existing pathology. Even the smallest unwanted perturbations can severely hinder an intervention that would otherwise cause a successful metabolic improvement. Thus, in the development of assistive wearable devices, it is of utmost importance to consider the biomechanical consequences of applying torques and powers to the human body. The finding that humans tend to reduce their knee flexion moment in response to applied device torque means that great care must be taken in the selection of controllers for device torque generation. Otherwise, humans may compensate by using antagonistic muscle activity.

Despite the difficulty in reducing metabolic demand, however, it is encouraging to find that with sufficient power, and with the appropriate levels of applied torques, it may be possible to provide metabolic and kinetic assistance to transtibial amputee walking gaits beyond that provided by a powered ankle-foot prosthesis alone. In the ongoing pursuit of better assistive technologies for those with limb amputation, this
result helps stress the importance of multiarticular approaches to prosthetic joint design.

7.2 Future work

7.2.1 Mechatronic Design
Regarding the quasi-passive artificial gastrocnemius, the mechanical attachment to the human body could be improved. Specifically, the thigh cuff could be made wider, as to minimize contact pressure with the soft tissues on the upper leg. Additionally, the aluminum bracket that connected the QPAG to the socket could be made stiffer, so as to minimize series compliance with the designed joint compliance. Lightweight, strong materials such as carbon fiber composite is recommended for this application. Further, the clutch itself could be improved by increasing the clutching resolution. The 90-tooth design prevented the clutch from engaging in steps smaller than 4 degrees. A greater number of teeth, or a different clutching mechanism might be warranted; a greater resolution would enable the clutch to waste less stroke on inter-tooth spacing and potentially store slightly more energy in the springs.

The powered artificial gastrocnemius could also benefit from some design improvements. First, an improvement on joint torque sensing could be made by measuring the angular deflection of the joint’s series spring. Feedback from spring deflection provides a way to mechanically filter unwanted noise and high-frequency content from the torque measurement. This improved sensing would then allow for a more aggressive feedback control scheme, likely with the benefit of improved performance. Such a change might allow closed-loop enforcement of torque for stance phase, in addition to swing phase. Although the open loop torque-control performance was found to be adequate in this work, a closed-loop controller would reduce unwanted impedance to the wearers, and could potentially allow for greater improvements in clinical measures such as metabolism.
Further, the tether system used for the powered artificial gastrocnemius could be improved. The majority of the observed friction in the cable drive was most likely caused by the cable going through bends in the linkage where the conduit was built from the small interlocking elements. If pulleys or other devices could be used to minimize or eliminate these bends, friction could be reduced, allowing further improvements to the torque control systems.

7.2.2 Clinical Study Design

Regarding the quasi-passive artificial gastrocnemius, it would be beneficial to perform additional experiments which more broadly explore this device. Each individual has their own gait idiosyncrasies, and a matching to other human gait solely based on height and weight has its limitations. Therefore, it is possible that, although the joint stiffness of the device was chosen via optimization, that this stiffness value may not have corresponded to the optimal values for the given amputee participants. Additionally, inherent compliance in the brace itself may change the effective device stiffness from the designed specification. Future trials could more systematically vary the joint stiffness to find the metabolically-optimal value, and then compare it to that from simulation.

The findings in Chapter 6 indicate that, with sufficient average power from the artificial gastrocnemius knee joint, but without the application of too much knee torque, more consistent reductions in walking metabolism are likely possible. To test this hypothesis, a future study could provide varying levels of power to the affected-side knee joint while keeping peak sufficiently low, not unlike previous work with an ankle exoskeleton [50]. In fact, it may be wise to use prior knowledge of an amputee's joint kinetics to inform the controller before an intervention with the device is attempted. One potential control strategy would be to record the knee moment profile of an amputee prior to activating the AG knee joint. The controller may then be adjusted to limit device applied torque to some fraction of the recorded
torque profile. Power delivery could then be adjusted by varying torque or joint impedance within this upper bound.

Since muscle work is so connected with metabolism, any change in device joint power can have a measurable effect on the metabolic results. Although efforts were made to maintain the power delivery of the BiOM ankle-foot prosthesis across conditions, the prosthetic ankle power was not always completely consistent. In future studies, it may be beneficial to use a different control strategy in the prosthesis so to more accurately maintain desired mechanical power output levels.

Further work could also be pursued with the neuromuscular controller. Modeling of human physiology allows for the possibility of connecting multiple joints virtually by controlling the joints with a single neuromuscular controller. Although multiple joints were controlled in this thesis, the two joints were never controlled using a single, unified controller. A natural next step would be to use a controller similar to the one developed in Chapter 5 to provide joint actuator commands to both the knee and ankle joints of the powered artificial gastrocnemius. Such a control strategy would provide better synchronization between the knee and ankle actuators, and could lead to a more effective intervention.
Chapter 8
Appendix

8.1 Individual Joint Biomechanics

The following section summarizes the behavior at the affected-side ankle, knee, and hip joints, for all six amputee participants. Each plot compares the OPT condition to the ZERO condition for the given gait feature, for all six subjects. Where applicable, the biological component of affected-side knee moments and powers is also shown.

Figure 51: Affected-side ankle dorsiflexion angle with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 52: Affected-side knee flexion angle with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 53: Affected-side hip flexion angle with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 54: Affected-side ankle dorsiflexion moment with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 55: Affected side knee flexion moment with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line). The dashed green line is the biological moment contribution.
Figure 56: Affected-side hip flexion moment with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 57: Affected-side ankle power with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 58: Affected-side knee power with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line). The dashed green line is the biological power contribution.
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line).

8.2 Individual Electromyography

The following section summarizes the muscle activity for four muscle spanning the knee joint, for all six amputee participants with the powered AG. Each plot compares the OPT condition to the ZERO condition for the given gait feature, for all six participants.
Figure 60: Affected-side biceps femoris activity with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 61: Affected-side semimembranosus activity with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 62: Affected-side vastus lateralis activity with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Figure 63: Affected-side vastus medialis activity with the Powered AG
The ZERO condition (blue thick line) is compared to the OPT condition (thin red line)
Bibliography


