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Anthony J. Anderson

On the Development of a Robotic Biarticular Prosthesis Emulator

Anthony J. Anderson

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Reading Committee:

Patrick M. Aubin, Chair

Brittney C. Muir

Sawyer B. Fuller

Katherine M. Steele

Program Authorized to Offer Degree:

Mechanical Engineering

University of Washington

Abstract

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Anthony J. Anderson

Chair of the Supervisory Committee:

Dr. Patrick M. Aubin

Department of Mechanical Engineering

People with transtibial limb loss experience decreased walking ability and have long-term musculoskeletal issues, such as knee osteoarthritis. Currently available ankle-foot prostheses, which only actuate the ankle joint, do not replicate the biarticular nature of the gastrocnemius muscle, which may limit their clinical efficacy. It is possible that transtibial prosthesis designs that include knee flexion assistance through a knee exoskeleton or exosuit could improve gait and quality of life for people with transtibial limb loss. However, the design space of these devices is large, and it is difficult to determine how biarticular gait assistance from different device designs might help people with limb loss. Therefore, developing a method to rapidly test and compare a variety of biarticular prosthesis designs is crucial for improving the lives of people with transtibial limb loss.

The objectives of this work were to design, build, and evaluate a biarticular prostheses emulator – a device that combines a robotic ankle-foot prosthesis and knee exoskeleton that can behave like a variety of hypothetical device embodiments to facilitate scientific exploration in a laboratory environment. This work was separated into three studies. The first study detailed the design, control, and evaluation of a uniarticular robotic ankle-foot prosthesis with offboard actuation and control. The prosthesis design incorporated novel

design choices compared to other prosthesis emulators, including a commercial foot plate and parallel torsion spring to provide dorsiflexion moments during walking. The prosthesis was found to be able to emulate both idealized passive devices and powered devices in walking experiments. The second study explored the design and control of a robotic knee exoskeleton emulator specifically for people with transtibial amputation. The third and final study presented the design and evaluation of an integrated biarticular prosthesis emulator that combined the robotic ankle-foot prosthesis with a second-generation design of the robotic knee exoskeleton. The biarticular prosthesis emulator design and control were evaluated with benchtop experiments and a walking demonstration that showed the prosthesis emulator's ability to provide simultaneous knee and ankle assistance during walking.

The primary contributions of this work are the design and development of several novel assistive devices for people with transtibial limb loss. Due to our modular design approach, future research may use either the ankle-foot prosthesis or knee exoskeleton as stand-alone uniarticular assistive devices, or together as a biarticular prosthesis. The work in this dissertation lays the foundation for a future research program that uses these devices to explore biarticular gait assistance and compare different hardware embodiment for prostheses that improve health and quality of life for people with transtibial limb loss. The potential impact of this work is significant, as the development of more effective prostheses for people with transtibial limb loss can significantly improve their mobility and overall quality of life.

TABLE OF CONTENTS

	Page
List of Figures	iii
List of Tables	viii
Chapter 1: Introduction	1
1.1 Research Motivation and Focus	1
1.2 Significance and Contributions	3
1.3 Organization and Layout	5
Chapter 2: Background	7
2.1 Transtibial Limb Loss, Mobility, and Health	7
2.2 Modern Ankle-Foot Prosthesis Technology	9
2.3 Functional Roles of the Soleus and Gastrocnemius Muscles	12
2.4 Biarticular Prostheses for Restoring Gastrocnemius Function	15
Chapter 3: Design, Control, and Evaluation of an Ankle-Foot Prosthesis Emulator	17
3.1 Abstract	17
3.2 Introduction	18
3.3 Mechatronic System Design	21
3.4 Control System Design	29
3.5 Experimental Methods	35
3.6 Results	40
3.7 Discussion	41
3.8 Conclusion	47

Chapter 4:	Design and Evaluation of an Knee Flexion Assistance Exoskeleton for People with Transtibial Amputation	48
4.1	Abstract	48
4.2	Introduction	49
4.3	Mechatronic Design and Control	51
4.4	Experimental Methods	58
4.5	Results	61
4.6	Discussion	64
Chapter 5:	A Robotic Emulator for Exploring Biarticular Prosthesis Designs for People with Transtibial Limb Loss	66
5.1	Introduction	66
5.2	Mechanical Design	69
5.3	Control System	75
5.4	Methods	79
5.5	Results	84
5.6	Discussion	91
5.7	Conclusion	95
Chapter 6:	Conclusions	96
6.1	Summary	96
6.2	Future Work	98
Bibliography		104

LIST OF FIGURES

Figure Number		Page
3.1	(a) Schematic of the hardware layout for the COBRA system. Control and actuation hardware are placed on a stationary cart. Flexible Bowden cable tethers transmit forces to the prosthesis end-effector while a participant walks on an instrumented treadmill. (b) Photograph of the ankle-foot prosthesis end-effector attached to a research participant with transtibial limb loss. The end-effector consists of a custom ankle joint that attaches to standard clinical prosthesis hardware, including a pyramid adapter and various sizes of carbon fiber prosthetic footplates.	22
3.2	Primary COBRA actuator with components labeled. Motor torque and power are delivered to a Bowden cable through a transmission consisting of a flexible shaft coupler (purple/gold), a drive shaft supported by bearings (blue), and a pulley that is fixed to the shaft. A wave spring (red) is included to apply axial preload to the bearings and reduce shaft vibration. The bearing hub block is shown as a partial section view to expose the drive shaft.	23
3.3	Ankle prosthesis as assembled (left) and as an exploded view (right). Primary components are labeled. The custom prosthesis joint attaches to standard clinically-available prosthesis componentry, including the pyramid adapter and Ossur LP Variflex footplate. 3D printed components are shown in red. An additional cover for the encoder mount and 3D printed component that clamps the Bowden cable to the Bowden cable inlet are not shown for clarity.	25
3.4	Mechanical characterization of the prosthetic ankle. Experimental data is red and best-fit lines are black. Curve fit equations are run in real-time on the prosthetic ankle torque controller. Top: Distance from the Bowden cable line of action to the prosthetic ankle joint center as measured by the 3D solid model used to design the joint. Given that the pulley at the transmission has a 25.1 mm radius, the transmission ratio between the motor and the ankle joint ranges from 3 to 3.4. Bottom: Dorsiflexion torque provided by the parallel torsion springs.	28

3.5	Torque control system for the COBRA Ankle. Torque setpoints, τ_r , are generated using ankle angle, θ_a , and ground reaction forces, F_z . Torque setpoints are transformed into cable force setpoints, F_r , using models of the ankle's lever arm and spring torque. A cascaded control architecture is used to track cable force setpoints, where an outer force tracking loop generates motor velocity commands, θ_r , for an inner motor velocity loop. The force tracking loop uses feedback on measured cable force error with a damping term on measured motor velocity, $\dot{\theta}_m$. The learning controller generates a feedforward trajectory as a function of gait phase, ϕ	30
3.6	Finite state machine diagrams for the COBRA Ankle control system. (a) Finite state machine for ankle torque setpoint generation when the ankle is in passive mode. The passive torque setpoint generator switches between two linear stiffnesses depending on the plantarflexion angle of the prosthesis. (b) Finite state machine for ankle torque setpoint generation when the ankle is in active mode. The active torque setpoint generator is similar to the passive setpoint generator, provides push-off in late stance when the requested torque exceeds a tunable threshold. When the threshold is exceeded, an additional constant offset torque is added to the passive stiffness, which is equivalent to an instantaneous change in the rest angle of the virtual spring.	32
3.7	Benchmark test results of the prosthesis and control system. Left: Step responses collected with the foot mounted in a test stand. Right: Frequency response collected with the foot mounted in a test stand. Discrete sine waves of torque setpoint were sent to the controller with an offset of 100 Nm and amplitude of 150 Nm. A second-order transfer function model with a time delay was fit to the experimental gain and phase data.	41
3.8	One gait cycle on the COBRA Ankle. (a) Initial contact between the prosthesis and treadmill is detected in real-time. (b) Foot-flat occurs. Between (a-b), the prosthesis produces a net dorsiflexion torque due to the parallel torsion spring. Near (b), the ankle angle moves through the virtual neutral angle and the prosthesis switches to a net plantarflexion torque. (c) During mid-stance, the prosthesis bears the full bodyweight of the participant. (d) The peak dorsiflexion angle corresponds with the peak plantarflexion torque. When the prosthesis is in active push-off mode, push-off is triggered near this moment. (e) The prosthesis leaves the ground and returns to the virtual neutral angle during swing. (f) Heel-strike for the next stride occurs.	42

3.9	COBRA Ankle biomechanics across three stiffness conditions. The top row shows data from Participant I and the bottom from Participant II. Peak ankle torques remained relatively constant across conditions, while peak ankle dorsiflexion angles decreased as stiffness increased. Peak ankle power was lowest for the stiffest condition.	43
3.10	Ankle power during active push-off mode tests. As the push-off torque parameter w from (5) was increased, peak ankle power increased.	43
4.1	Knee flexion moments for healthy control subjects and people with limb loss walking on passive and powered prosthetic ankles, digitized from [1]. Flexion moments are notably lower on the residual limb as compared to those from the control subjects, regardless of whether the prosthesis is active or passive. Proposed knee flexion exoskeleton assistance strategies are overlaid in black.	52
4.2	Schematic of the experimental setup. The participant walks on a treadmill as a stationary motor delivers torque to the knee exoskeleton through a flexible Bowden cable tether. The cart also contains a high-speed computer for taking sensor measurements and sending motor commands. The experimenter interface is a desktop PC that streams data to and from the controller.	53
4.3	Rendering of the knee exoskeleton with labels for the primary components described in the text.	54
4.4	Photo of the participant walking with the robotic exoskeleton.	60
4.5	Plots of relevant real-time controller signals for a representative trial. Vertical ground reaction forces are measured as an analog input from the split-belt instrumented treadmill. Gait phase is estimated in real-time from ground reaction force data. The knee exoskeleton encoder measures rotation of the biological knee joint as well as additional rotation due to soft tissue compression during actuation. In the exoskeleton torque plot, the blacked dashed line is the setpoint and the red line is the measured exoskeleton torque.	63
4.6	Torque tracking performance for the knee exoskeleton. Signals show mean strides for each condition $\pm$ one standard deviation. The average biological knee moment from control subjects in [1] is overlaid in grey to give context to the timing and magnitude of the exoskeleton torque profiles.	64

5.1	The potential design space for biarticular prostheses, partitioned by the devices' ability to deliver net positive work at each joint over a stride. Examples of each device embodiment from the literature are shown for each combination. The robotic prosthesis introduced in this work was designed with the ability to emulate devices in all four quadrants.	70
5.2	A schematic of the biarticular prosthesis emulator with primary components labeled. A research participant with transtibial limb loss walks on an instrumented treadmill while wearing the robotic ankle-foot prosthesis and exoskeleton. The stationary cart holds two servomotor actuators, a high-voltage servo circuit, a high-speed industrial controller, and a desktop computer that acts as an experimenter interface. Sensors from the prosthesis, exoskeleton, and treadmill are read by the industrial controller.	71
5.3	A photograph of a participant walking in the biarticular prosthesis. Key components of the exoskeleton hardware are labeled.	74
5.4	A diagram of the controller and signal flow for the biarticular prosthesis. On the real-time controller, heel strikes times are used to estimate gait phase. Desired torque signals are generated for the knee and ankle using a phase-based and angle-based controller, respectively. These desired torque signals are transformed to desired cable forces and are tracked with feedback controllers that output motor velocity commands for each motor. Bowden cable transmissions apply torques to the devices, and wearable sensor data is fed back into the real-time controller.	76
5.5	Knee torque command parameterization for the knee exoskeleton and the average biological knee flexion moment from healthy subjects, digitized from [1]. Gait phase is estimated in real-time from historical heel-strike data and is used as the domain for the commanded exoskeleton torque signal.	78
5.6	Expanded schematic of the cable force tracking controller. A desired torque is sent to the cable tracking controller from an upstream torque controller and transformed into the Bowden cable force required to achieve the requested torque. For the knee exoskeleton and ankle-foot prosthesis, the encoder measurement is used to compute the instantaneous lever arm of the device. For the ankle-foot prosthesis, the encoder measurement is also used to estimate the torque provided by the parallel spring.	78

5.7	Biarticular prosthesis communication and connectivity diagram. The host PC sends control parameters to and receives telemetry from the real-time controller via a local area network (LAN) and ethernet router. Signals from the wearable sensors and instrumented treadmill are read by an analog-to-digital converter on the real-time controller. Control algorithms are implemented on the real-time controller and motor velocity setpoint commands are sent to the servo drives via an etherCAT protocol.	80
5.8	Photograph of the test stand setup for the ankle-foot prosthesis (left) and knee exoskeleton (right). The prosthetic footplate was removed from the prosthesis in order to efficiently lock the ankle joint. The lateral, actuated joint was removed from the knee exoskeleton for benchtop tests in order to simplify the design of the test stand.	81
5.9	Bode plot for exoskeleton torque and motor velocity tracking controllers. The exoskeleton torque controller shows a resonant peak near 4 Hz. Points show individually tested frequencies, and a spline connects the points for visualization.	85
5.10	Torque tracking performance with two of the sinusoidal torque commands at 0.5 Hz (top) and 3 Hz (bottom). Nonlinearities due to Bowden cable friction cause higher tracking errors when the loading rate changes directions.	86
5.11	Influence of the iterative learning controller as shown in the benchtop experiment. The torque command is shown in black, the proportional-damping controller (PD) in blue, and the proportional-damping controller with iterative learning control (ILC) in green. The significant phase lag in the PD controller is largely addressed by the feedforward nature of the ILC algorithm.	87
5.12	Coordinated step responses between the ankle-foot prosthesis and knee exoskeleton. Step commands were conducted in three batches, corresponding to low, medium, and high magnitudes for each device. For a given batch, torque steps were commanded to the knee and ankle simultaneously. All five step responses are shown for each batch at both joints.	88
5.13	Average measurements of the biarticular prosthesis signals during walking. The ground reaction force is shown on the top row, the prosthesis torques in the middle row, and the exoskeleton torque on the bottom row.	90
5.14	The participant walking with the robotic biarticular prosthesis. As the participant moves into swing phase, the knee exoskeleton cable is nearly slacked, allowing for low joint impedance during swing.	91

LIST OF TABLES

Table Number	Page
3.1 Ankle-foot Prosthesis Design Specifications	26
3.2 Participant Demographics	37
3.3 Prosthesis Settings for Passive Emulation Mode	38
3.4 Stance Phase Torque Tracking Errors in Passive Emulation Mode	42
4.1 Mass Distribution of Wearable Exoskeleton Components	56
4.2 Exoskeleton Performance Metrics Across Conditions	62
5.1 Step Response Metrics	89

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DEDICATION

to Liz, my wife, my love

PREFACE

At the time of writing, Chapters 3, 4, and 5 of this dissertation are stand-alone scientific articles at various stages of the publication process. Chapter 3 is currently under peer review, Chapter 4 was published in the 2022 International Conference on Rehabilitation Robotics, and Chapter 5 is currently in preparation for submission. Contributing authors are listed at the beginning of each chapter.

Chapter 1

INTRODUCTION

1.1 Research Motivation and Focus

People with transtibial limb loss experience reduced mobility, health, and quality of life [1–3]. Many of the health issues experienced by transtibial amputees (TTAs) are caused by disrupted ankle-foot function. Transection of the calf and ankle musculature during amputation leads to altered walking patterns that immediately disrupt mobility, and the resulting cumulative asymmetries in limb loading during walking lead to long-term musculoskeletal impairment [4]. Improving our ankle-foot prosthesis technologies is one way to improve the health and quality of life of the hundreds of thousands of people living with transtibial limb loss in the United States [5]. Robotic technologies have an important role to play in improving prosthesis technologies, both in supplementing our state-of-care in the form of bionic limbs or as tools that help us understand the fundamental issues in gait biomechanics and neuromuscular control [6–8].

Modern ankle-foot prostheses are generally mechanically passive devices that are designed to replicate human ankle function during locomotion [9, 10]. However, passive prostheses are fundamentally limited in their ability to replicate the function of the biological human ankle, which changes impedance throughout the gait cycle [11], changes its fundamental mechanical behavior across activities [12], and delivers positive net mechanical work across each stride [13]. Robotic prostheses with sensors, actuators, and controllers are necessary if we want to restore ankle function and mobility to people with limb loss. However, robotic ankle-foot

prostheses are in their infancy, and studies of their efficacy in helping people with limb loss have had mixed conclusions [6, 14–17]. Much research attention has been paid to the problem of designing naturalistic prosthesis controllers [18–21], but another potential issue is that the mechanical structure of our prostheses does not reflect the underlying muscular anatomy of the human calf. The gastrocnemius is a biarticular muscle that provides simultaneous knee flexion and ankle plantarflexion moments during walking, a feature that is missing from all modern ankle-foot prostheses for people with transtibial limb loss. It is currently unknown how this impairment affects TTAs walking on passive and powered ankle-foot prostheses, but new devices that incorporate knee flexion assistance similar to the gastrocnemius may improve walking outcomes and quality of life.

Future intelligent ankle-foot prostheses, whether uniarticular or biarticular, have the potential to improve mobility and quality of life for people with transtibial limb loss. One barrier to technological advancement is that it is challenging to predict what types of prosthesis design and control strategies will benefit users. Design and control strategies based on biomechanical models of walking, for example, have so far failed to reduce the metabolic cost of steady-state walking, even when using advanced human-in-the-loop optimization methods [8, 22, 23]. These challenges are rooted in the difficulty of predicting the human locomotor response to a new robotic assistive device, which involves learning, adaptation, preference, and individual variation – all of which make modeling the human-robot system difficult [24–28]. Therefore, our current scientific situation is such that we stand to learn the most about prosthetic design and control from rapid experiments, and we need to ensure that our research tools and programs are designed with this idea in mind.

Off-board robotic actuation systems or ‘emulators’ are tools used in biomechatronics research that allow wearable device designers to explore new device architecture and assistance strategies rapidly and without committing time and resources to the minutia of, e.g., actuator design, electrical design, or battery management [29–31]. These laboratory-based

systems use powerful off-board motors and computers with flexible transmission tethers and lightweight end-effectors to allow for rapid experiments with human subjects in a controlled environment. Findings from these types of experiments can then inform and inspire designs for untethered prostheses and exoskeletons [32].

This dissertation focuses on the design, control, and evaluation of a modular biarticular prosthesis emulator that combines a custom robotic ankle-foot prosthesis with a custom robotic knee exoskeleton, both of which are independently driven by an off-board actuation and control unit. We intended to develop and validate a research tool that will enable us to conduct future experiments that will 1) provide insights into the basics of amputee gait biomechanics with and without biarticular assistance, 2) facilitate comparisons between hypothetical BP designs in the laboratory, and 3) inspire designs for new, untethered biarticular prostheses that can improve mobility for end-users outside of the laboratory.

1.2 Significance and Contributions

This dissertation contributes to the bionic prosthesis and exoskeleton design communities by presenting and validating novel hardware, software, and control systems for wearable robotic devices for assisting people with transtibial amputation. We developed and validated a biarticular prosthesis emulator with novel design choices that contribute to a wider diversity of tools for wearable robotics research. By building these new research tools with an emphasis on modularity and reconfigurability, we have developed a platform that will streamline future research for a variety of wearable robotic devices with the potential to improve health and mobility in people with transtibial amputation and other lower limb impairments. The specific contributions of this dissertation are listed below.

- **The design and development of the Configurable Off-Board Robotic Actuator (COBRA):** COBRA is a custom off-board robotic actuation system and platform for wearable robotics research. COBRA consists of a mobile cart that holds two

high-power servomotors with custom Bowden cable transmissions, a high-voltage servo circuit, a high-performance industrial controller, and a computer that serves as an experimenter interface. Along with the hardware, COBRA has a suite of generalized software programs that serves as templates for controlling a wide variety of wearable robotic devices, i.e., prostheses and exoskeletons. The performance of COBRA was verified and validated across three research studies that included benchtop and walking experiments.

- **The design, control, and evaluation of the COBRA Ankle-Foot Prosthesis:**

The COBRA Ankle-Foot Prosthesis is a custom robotic end-effector that is actuated and controlled by COBRA. The prosthesis is capable of providing biologically relevant peak torque, power, and range of motion for people with transtibial limb loss walking at typical speeds. We made several unique design choices that allow for the COBRA Ankle-Foot Prosthesis to achieve behaviors unavailable to other state-of-the-art ankle-foot prosthesis emulators, including 1) a parallel spring architecture that allows for control of both plantarflexion and dorsiflexion moments with a single offboard actuator, 2) the ability to fit inside of a cosmetic foot shell and shoe, and 3) the ability to accommodate a variety of foot lengths and shoe sizes by attaching to clinically available prosthetic footplates. The device's robotic performance was characterized by benchtop tests, and a case study of the device's performance during walking was conducted with two participants with transtibial limb loss.

- **The design, control, and evaluation of the COBRA Knee Exoskeleton:** The COBRA Knee Exoskeleton is a custom robotic knee flexion exoskeleton with off-board actuation and control. The exoskeleton was designed to assist the residual knee joint in people with transtibial limb loss. Two generations of the COBRA Knee Exoskeleton were developed during the course of this dissertation, the second of which made several

design improvements focused on user comfort. The exoskeleton’s robotic performance was characterized by benchtop tests, and the was evaluated in a walking experiment with one participant with transtibial amputation. This experiment was the first to provide active robotic assistance to the residual knee of a transtibial amputee while walking with a passive ankle prosthesis.

- **The design, development, and validation of the COBRA Biarticular Prosthesis Emulator:** The COBRA Biarticular Prosthesis combines the COBRA Ankle-Foot Prosthesis and second-generation COBRA Knee Exoskeleton for walking experiments focused biarticular assistance strategies. Given the actuation and sensor layout, the device is capable of being programmed to emulate several types of future potential biarticular prostheses, including passive, active, and quasi-passive embodiments. The device behavior was validated for one embodiment (emulated passive ankle prosthesis, active robotic knee exoskeleton) in a case study with a single participant with limb loss.

Together, this work lays the technical foundation for future clinically-oriented experiments that investigate the physiological and biomechanical effects of biarticular prosthesis assistance for people with transtibial limb loss. Further, both the robotic ankle-foot prosthesis and knee exoskeleton can be used independently to explore other potential assistive devices or control strategies for uniarticular prostheses and exoskeletons. COBRA will serve as a flexible platform for research into other types of wearable assistive devices for other clinical populations.

1.3 Organization and Layout

Chapter 2 provides a literature review on the effects of transtibial limb loss, the functional roles of the soleus and gastrocnemius muscles, and the prior development of other biarticular

prostheses for restoring gastrocnemius function. Chapters 3, 4, and 5 are technical chapters that describe three chronological research studies that describe the robotic hardware design, development, and evaluation process for the biarticular prosthesis emulator. First, we designed, built, and evaluated a custom robotic ankle-foot prosthesis with offboard actuation and control (Chapter 3). We found our device to be capable of providing biological levels of ankle torque and power with low torque tracking error during walking experiments with transtibial amputee participants. Second, we designed and developed a knee flexion exoskeleton for users with transtibial amputation (Chapter 4). We found that the exoskeleton was able to provide repeatable assistive torques at multiple magnitudes to one amputee subject during walking. Finally, we redesigned our knee exoskeleton with a focus on user comfort and adjustability in the frontal plane. We paired the redesigned exoskeleton with our robotic ankle prosthesis and conducted a case study with one participant walking with the device in uniarticular and biarticular assistance modes (Chapter 5). We found that the coupled devices were able to operate together and provide both soleus-like and gastrocnemius-like torques to the user during walking.

Chapter 2

BACKGROUND

People with below-knee amputation might benefit from prostheses that emulate the mechanical layout of the biological calf muscles. This literature review discusses the consequences of below-knee amputation, modern prosthetic ankle technologies, the function of the ankle plantar flexors, and prostheses that aim to replicate the role of the gastrocnemius muscle.

2.1 Transtibial Limb Loss, Mobility, and Health

Limb loss is a growing problem in the United States. In 2005, an estimated 1.6 million people were living with the loss of a limb, a number projected to double by 2050 [5]. Rates of major lower limb amputation are increasing primarily due to increases in diseases associated with obesity, like diabetes and peripheral arterial disease [33]. Rates of worldwide obesity are increasing [34], and the worldwide rate of diabetes is expected to double between 2000 and 2030 [35]. As one indicator of the growing issue, the Department of Veterans Affairs reported a more than 40% increase in the number of lower limb amputations in veterans between 2008 and 2018, a vast majority of which were due to complications of vascular disease [36]. Treating this growing problem effectively will require clinical practice and assistive technology improvements.

Walking is an integral part of daily life disrupted by limb loss. People with transtibial (below-knee) amputation adopt compensatory walking patterns when using conventional prostheses. Spatiotemporal parameters like step length, step width, and step time are asymmetric between the intact and prosthetic-side limbs, leading to relatively more time spent

on the intact limb over each stride [37, 38]. Joint kinematics during walking are altered significantly, with asymmetries between limbs and decreased range of motion at the knee and prosthetic ankle compared to non-disabled people [1, 39]. There are also bilateral asymmetries in the ground reaction forces, joint moments at the hip, and joint forces [1, 40–42]. Overall, the walking patterns of people with transtibial amputation are markedly different from those of the general population.

Gait changes associated with transtibial amputation decrease people’s ability to navigate their daily environments. Transtibial amputees (TTAs) have slower self-selected walking speeds [43, 44] and reduced metabolic energy efficiency [45–48]. Clinical mobility tests, e.g., the Four-Square-Step-Test, also indicate reduced locomotor performance in TTAs [1, 49]. People with limb loss experience relatively high rates of falling and fear of falling, which can reduce physical activity and community mobility [50, 51]. Daily step counts, as measured with wearable sensors, are lower for TTAs than for the general population. Studies of community-based physical activity have found that American adults generally walk between 4,000 and 6,000 steps per day [52, 53]. In contrast, studies of lower-limb amputees report values between 2,377 and 4,249 steps per day [54–56]. These statistics are concerning, given our knowledge of the physical and cognitive benefits of walking [57, 58] and the detriments of sedentary lifestyles [59].

Altered walking patterns and reductions in physical activity often lead to further musculoskeletal disease in lower-limb amputees [4]. One study of traumatic lower limb amputees found knee and hip osteoarthritis prevalence to be 17 and 12 times greater than the general population [60]. Norvell et al. found that transtibial amputees report 30% more knee pain than non-amputees [2]. Similarly, people with transtibial amputation report high rates of chronic lower back pain [61]. One large study of people with limb loss found that 62.3% reported consistent back pain, one quarter of which rated their pain as “extremely bothersome” [62].

Overall, the loss of a limb can have a profoundly negative effect on a person’s quality of life. Several studies have found that people with lower limb loss report a lower quality of life than the general population due to issues including chronic residual limb pain, reductions in social connectedness, and loss of employment [3, 63–65]. Clinical solutions to improve mobility and health are direly needed for this population.

Many issues that negatively impact quality of life are directly or indirectly related to the TTA’s loss of functional mobility. There is an opportunity for engineers, clinicians, and end-users to collaborate on developing new assistive devices that improve mobility for people with transtibial amputation. The development of intelligent assistive devices is an active area of research that is beginning to improve mobility outcomes across clinical populations [66–70]. However, the state of care for ankle-foot prostheses requires improvement.

2.2 Modern Ankle-Foot Prosthesis Technology

Most ankle-foot prostheses prescribed today are unactuated, spring-like devices that approximate human ankle function across various locomotor tasks. One of the most commonly prescribed types of ankle-foot prostheses are energy-storage-and-return (ESAR) prostheses, which are built from composite materials like carbon fiber [10, 71]. During walking, ESAR prostheses flex and store energy during midstance and return it to the user during push-off. While they are an impressive clinical and engineering achievement, ESAR feet are fundamentally unable to restore the biological human ankle’s wide range of dynamic behaviors. Passive prostheses have fixed stiffness and alignment properties and cannot produce net positive mechanical work during each stride. In contrast, the healthy ankle modulates its impedance throughout the gait cycle [11], and the ankle musculature produces between 5 and 30 Joules of net mechanical work over every stride [72]. For ankle-foot prostheses to achieve these functions, they require batteries, sensors, actuators, and control systems.

Quasi-passive prostheses, also known as microprocessor-controlled prostheses, are devices

that use sensors, microprocessors, and low-power actuators to adjust the mechanical properties of the prosthesis for different tasks. They allow for some adaptability, such as changing the neutral angle or stiffness for walking at different speeds or on different terrains. Examples of research devices include ankle-foot prostheses that adjust sagittal plane stiffness [73–75], sagittal plane damping [76,77], frontal plane ankle alignment [78], and transverse plane stiffness [79]. Some quasi-passive prostheses are also available for clinical prescription and are used to adapt the neutral angle of the ankle for walking on slopes and improving toe clearance during the swing phase. These devices have been shown to improve walking, balance, and quality of life in multiple independent studies [68,80–84]. Although quasi-passive prostheses may represent an improvement over passive prostheses for some patients, they are still energetically passive in a given setting and cannot fully replicate the dynamic range of human ankle behavior.

Powered ankle-foot prostheses use batteries and powerful actuators to replicate the biological ankle and foot function. Powered prostheses can render specific behaviors across various activities and provide net positive mechanical work like the ankle plantar flexors [85]. These devices are a current research and development focus [86–89]. Open questions remain surrounding the optimal design and control strategies for these devices. Most powered ankle-foot prosthesis designs use electric motors, mechanical transmissions like gears and ball screws, and spring element [87,90,91]. Regarding design, some researchers have explored using series-elastic and parallel-elastic actuators, which combine motors with springs in various configurations to improve torque tracking bandwidth and reduce energy consumption [21,89,92–94]. Control is a significant challenge in developing powered ankle-foot prostheses, with joint torque usually being the controlled variable [18]. However, various methods for generating torque signals, such as impedance control [89,95,96], myoelectric control [97], and neuromuscular control [20], have been proposed. Both design and control impact patient outcomes and satisfaction, with design choices affecting factors like weight,

battery life, and the mechanical capabilities of the prosthesis, and control influencing the ability of the prosthesis to synchronize with the user.

Only one robotic ankle-foot prosthesis with active push-off is commercially available for clinical prescription. The BiOM (now known as the Ottobock Empower) uses batteries, a servomotor, and a composite spring to provide torque and mechanical work to the user [86]. Studies with prototype versions of the BiOM were compelling, with subjects showing improvements in metabolic cost of transport and walking speed relative to their prescribed ESAR prostheses [6, 86]. Further studies by independent researchers confirmed reductions in metabolic cost and found that the device improved biomechanical risk factors for knee osteoarthritis on the sound limb [14, 15]. However, recent studies have been less favorable. One clinical study with ten subjects using the device found no significant differences in metabolic costs of walking or preferred walking speed [16]. Another study with eleven subjects found that the device did not improve performance on functional tests and that significant gait asymmetries persisted [1]. A two-week take-home study of the BiOM with wearable sensors found no significant differences in daily step counts, walking speed, or user perceptions of mobility among twelve participants [17]. Overall, the clinical evidence supporting the prescription of robotic ankle-foot prostheses is meager, and no research group has presented an ankle-foot prosthesis with active push-off that shows a consistent and reliable benefit to end-users.

There are several reasons the inclusion of active push-off in robotic ankles has yet to benefit TTAs as expected. Forward simulation studies of walking have found that virtual transtibial amputees benefit from powered push-off, but the benefits are absent or reduced if whole-body muscle strength is decreased [98, 99], indicating that amputees in poor physical health may not have the strength to incorporate the robotic assistance into their gait effectively. The heavier mass of the prosthesis may be a limiting issue. The added weight of batteries and actuators can irritate the residual limb and increases energy consumption

during walking [100]. It is also challenging to select an appropriate assistance strategy and personalize it to each individual [18, 23].

An analysis of the anatomy of the human ankle musculature offers another potential reason that explains the limited performance of robotic ankle-foot prostheses. The two primary plantarflexor muscles of the calf are the soleus and gastrocnemius. The soleus muscle is a uniarticular ankle plantarflexor, producing only a moment about the ankle. In contrast, the gastrocnemius is a biarticular muscle that crosses the ankle and knee and simultaneously creates moments at both joints. All modern ankle prosthesis systems are uniarticular at the ankle joint and therefore cannot fully mimic the knee flexion moments produced by the gastrocnemius muscle during walking.

2.3 Functional Roles of the Soleus and Gastrocnemius Muscles

The plantarflexor muscles in the human calf are collectively referred to as the triceps surae and are composed of the soleus and gastrocnemius. Both muscles insert at the calcaneus via the Achilles tendon. The soleus originates at the head of the fibula and tibia, while the gastrocnemius has two heads that originate on the medial and lateral condyles of the femur. Therefore, the soleus applies moments to the ankle, and the gastrocnemius applies simultaneous moments to the knee and ankle. This difference in muscle layout causes the two muscles to have different effects on the body during walking.

When reasoning about clinical interventions in populations with impaired gait, it is vital to understand the relative functional roles of the soleus and gastrocnemius during walking. Traditionally, the two muscles were thought to serve similar purposes because both insert into the Achilles tendon and are active at the same portions of the gait cycle. However, using a muscle’s anatomical placement to infer its function during walking can lead to flawed conclusions, as muscles can accelerate joints they do not cross due to dynamic coupling and interactions between the body, gravity, and the ground [101–103]. One common framework

for understanding muscle function in walking involves quantifying how specific muscles contribute to 1) forward propulsion and 2) vertical bodyweight support. Muscles that cause ground reaction forces and center of mass accelerations in the direction of walking are considered to contribute to forward propulsion. In contrast, muscles that cause accelerations that oppose gravity are considered to contribute to vertical bodyweight support. A given muscle generally contributes to both tasks [104], but the relative contributions to each task can help clinicians and scientists reason about the effects of interventions in populations with impairments relating to, e.g., posture or walking speed.

Researchers have used muscle-driven computer simulations to understand the relative contributions of the plantarflexors to bodyweight support and forward propulsion. Liu et al. used forward simulations and a virtual muscle force perturbation method to show that the gastrocnemius and soleus contributed relatively more to accelerating the center of mass forward and upwards, respectively, during the second half of stance [104]. A later study using similar methods, also by Liu, found that soleus was more responsible than gastrocnemius for forward propulsion in late stance at self-selected and fast walking speeds [105]. A study by Neptune et al. analyzed individual plantarflexor contributions to trunk energetics and found that the soleus was more responsible for forward propulsion, while gastrocnemius primarily accelerated the limb into swing [106]. Several other modeling studies have come to similarly contradictory conclusions regarding the functional roles of the gastrocnemius and soleus [107–109].

Other researchers have used experimental manipulations of treadmill walking to understand how gastrocnemius and soleus contribute to forward propulsion and vertical bodyweight support. Gottschall and Kram provided horizontal aiding forces to participants' waists. They used electromyography (EMG) measurements of the plantarflexor response to conclude that the gastrocnemius was primarily responsible for governing forward propulsion [110]. A study by McGowan et al. used a novel experimental setup to manipulate the

body’s mass and weight independently. It used similar EMG measurements to conclude that the soleus was the primary contributor to forward trunk propulsion [111]. More recently, Clark et al. used a combination of horizontal aiding/impeding forces, EMG, ultrasound imaging, and musculoskeletal modeling to find that the gastrocnemius was more responsible for forward propulsion in mid to late stance than the soleus [112]. One other experimental method of studying plantarflexor function involves providing brief perturbations to muscle activation using electrodes during walking and measuring the resulting kinetics and kinematics of the body [113,114]. This method found the gastrocnemius to be more responsible for forward propulsion, with the soleus inducing braking of the center of mass in some instances in the gait cycle [115].

The conflicting conclusions about how the soleus and gastrocnemius contribute to forward progression may be due to various semantic and methodological issues. First, some studies define forward progression differently than others. Specifically, some studies analyze the effects of a muscle on the trunk [106], while others explore the effects on the whole-body center of mass [104]. If a muscle accelerates the leg into swing, it may accelerate the whole-body center of mass but not necessarily the trunk [116]. Second, results from musculoskeletal modeling studies are highly sensitive to models of the underlying geometry. The inferred function of the gastrocnemius is sensitive to assumptions about the ratio of moment arm lengths at the knee and ankle [117]. These parameters are anatomical and vary in the general population, but many musculoskeletal models use a generic set of lever arm parameters. While disagreements about the roles of the plantarflexor muscles persist, there is broad consensus that the muscles have different effects on the body when walking. This insight motivates the need for biarticular prostheses. If people with transtibial amputation do not have gastrocnemius muscles, they are also likely to be missing the functional role the muscle plays during gait and therefore exhibit gait compensations and asymmetric walking patterns.

2.4 Biarticular Prostheses for Restoring Gastrocnemius Function

Biarticular prostheses that aim to replicate gastrocnemius function are an active area of research. In this context, a biarticular prosthesis is an assistive device that combines an ankle-foot prosthesis with a knee orthosis, exoskeleton, or exosuit that provides knee flexion moments. These types of devices have become feasible due to recent advances in exoskeleton technology [70, 118–121]. Biarticular prostheses can be passive, quasi-passive, or powered, each with their own trade-offs and potential target patient populations. Researchers have explored a variety of biarticular prostheses designs.

Researchers at MIT have built two iterations of biarticular prostheses. The first version was a quasi-passive knee exoskeleton that was paired with the BioM powered ankle-foot prosthesis [122]. The exoskeleton used a sensorized spring-clutch mechanism that allowed the device to provide knee flexion moments at appropriate times during stance phase and have a low impedance during swing phase. Preliminary results with two participants found improvements in lower limb kinetics and metabolic walking economy. The second version of the device was a tethered robotic knee exoskeleton paired with the same ankle-foot prosthesis [123]. When tested with six participants, it improved affected-side limb kinetics but had no statistically significant effects on metabolic walking economy – though some participants did have considerable improvements to walking economy when knee assistance was provided. These results are promising and motivate the need for further research on biarticular prostheses.

Researchers at the University of Washington developed a biarticular prosthesis that paired a passive ankle-foot prosthesis with a quasi-passive spring-clutch knee orthosis [124]. The prosthesis had a physical biarticular element that crossed both the ankle and knee. When tested with seven participants, the prosthesis applied knee torques similar to those provided by the intact gastrocnemius and did not affect kinematic symmetry [125].

Researchers at Vanderbilt University developed a biarticular prosthesis emulator that

paired a powered ankle-foot prosthesis with a soft knee exosuit [126]. Both components of the device were driven by commercially available off-board actuators. The device had a mechanically biarticular element, referred to as the "artificial gastrocnemius," and a mechanically uniarticular element, referred to as the "artificial soleus." Although the device was powered, it was programmed to emulate a device with a powered ankle-foot prosthesis and a spring-clutch knee device.

A variety of design embodiments have been used in the development of biarticular prostheses, representing the large design space of potential devices. While preliminary results are promising, more research is needed to evaluate which types of biarticular prostheses can most benefit the lives of people with transtibial limb loss. Further investigation is necessary to determine the optimal design and control strategies for these devices, as well as to understand their potential impact on patient outcomes and satisfaction.

Chapter 3

DESIGN, CONTROL, AND EVALUATION OF AN ANKLE-FOOT PROSTHESIS EMULATOR

Anthony J. Anderson, Yuri F. Hudak, Brittney C. Muir, and Patrick M. Aubin

3.1 Abstract

People with transtibial limb loss suffer from reduced mobility. Intelligent ankle-foot prostheses have the potential to improve quality of life in people with limb loss, but there are scientific, clinical, and commercial barriers that prevent widespread impact. Further research tools and experiments are needed to expand our understanding of how to design and control intelligent prosthetic limbs. We designed and built a robotic ankle-foot prosthesis with off-board actuation and control to serve as a platform for biomechanical lower limb loss research. Our prosthesis fits inside of a shoe during walking, and attaches to standard clinical prosthesis componentry, including carbon fiber prosthetic footplates and pyramid adapters. Our novel mechanical architecture implements a custom torsion spring in parallel with the ankle joint to allow for dorsiflexion and plantarflexion torque control with a single off-board actuator. Benchtop tests show that our prosthesis has peak plantarflexion torques greater than 175 Nm and a torque control bandwidth of 6.1 Hz. Walking experiments with two participants with lower limb loss indicate that the prosthesis can achieve low torque tracking errors and push-off power greater than the biological ankle during walking. This device will enable future experiments on amputee gait biomechanics, human-robot interaction, and prosthesis control.

3.2 Introduction

People with transtibial limb loss have reduced mobility relative to the general population. When walking with conventional passive prosthetic feet, they exhibit gait deviations to compensate for the lack of the ankle muscles. Typical deviations include changes in spatiotemporal gait parameters, significant asymmetries in the forces and motions of the lower limbs, and slower walking speeds [41, 42, 127]. Over time, compensatory walking patterns can lead to high rates of secondary musculoskeletal diseases like chronic lower back pain and knee osteoarthritis [2, 4]. Balance during walking and standing are also disrupted for people with transtibial limb loss, leading to increased rates of falling and fear of falling [50, 51]. Together, these mobility issues can lead to a decreased quality of life [3, 65]. Lower limb amputation is on the rise in the United States, largely due to increases in vascular diseases like diabetes and peripheral artery disease [5, 36]. One way to improve mobility in people with lower limb loss is through improvements to ankle-foot prosthesis technology.

Three classes of ankle-foot prostheses are available to people with limb loss. The first and most common class consists of passive ankle-foot prostheses, which are typically made from composite materials and have fixed joint alignment and stiffness properties. The second class consists of semi-active prostheses, which use adjustable mechanisms and low-power electronics to vary the mechanical properties or joint alignment for different walking conditions [74, 75, 77, 84]. While semi-active prostheses have a wider range of possible behaviors than passive devices, they do not aim to replicate the active push-off provided by the plantarflexor muscles in the intact ankle. The third class consists of active robotic prostheses, which use powerful actuators and batteries to replace muscle work over each stride [86, 87, 89, 128]. Of the three classes, only powered robotic ankle-foot prostheses have the potential to replicate the full range of capabilities of the biological limb. Robotic prostheses have the potential to improve mobility outcomes beyond the current state of clinical care, but further research is needed to maximize impact and adoption.

Across the three classes of ankle-foot prostheses, there are open questions and barriers that limit clinical impact. For conventional passive prostheses, there are many subcategories with different features but little clinical evidence to guide optimal device prescription [71]. For semi-active prostheses, there are design questions regarding which mechanical properties are most consequential to end users and would therefore make good candidates for adjustability. For active robotic ankles, two primary challenges are mechanical design and control strategy. Mechanical design choices, e.g., the inclusion of series or parallel springs, can greatly affect mass, battery life, and controller performance [90, 92]. Researchers have developed many control strategies for active robotic prostheses, but experiments comparing outcomes between strategies are rare and no powered ankle-foot prosthesis has been shown to consistently and reliably improve mobility outcomes [6, 14, 16, 17]. Overall, there are many open scientific and clinical questions about the prescription, design, and control of ankle-foot prostheses, and the field will benefit from more tools to help us understand how to improve mobility in people with limb loss.

Laboratory-based robotic testbeds, or emulators, provide a mechanism to rapidly test hypotheses about wearable assistive device behavior. Emulators use powerful off-board servomotors to transmit forces to wearable devices through flexible actuation tethers. These systems typically use wall power and high-speed measurement and control hardware to achieve impressive mechatronic performance while keeping body-worn mass low, and have been used extensively in lower-limb exoskeleton and prosthesis research [29, 30, 129, 130]. Placing actuation and control hardware off-board greatly simplifies the design of wearable end-effectors, with the trade-off of limiting experiments to treadmill walking. Prosthesis emulators can be programmed to mimic passive prostheses in order to understand and improve the clinical prescription process, emulate the mechanical behavior of proposed devices, and evaluate novel control strategies for robotic ankles [23, 126, 131].

Three noteworthy prosthesis emulator designs have been described in the literature. Ca-

puto and Collins developed a universal prosthesis emulator capable of providing a wide range of ankle behaviors during walking [29]. Their wearable end-effector consisted of a sagittal plane joint with built-in series elasticity and plantarflexion torque control. Kim et al expanded on this work by developing and testing an end-effector with two degrees-of-freedom for plantarflexion and inversion-eversion torque control [132]. Both devices used a passive leaf spring in parallel with the prosthetic pylon to produce dorsiflexion moments in early stance. Chiu et al developed an emulator with two torque-controlled forefoot digits and one torque-controlled heel digit, enabling control of the center-of-pressure trajectory during stance [133]. All three systems can produce high torques and joint powers, have lightweight end-effectors, and achieve low torque tracking error during walking.

Design choices for prosthesis emulator end-effectors likely affect gait biomechanics and influence scientific conclusions about optimal prosthesis behaviors. Shoes and flexible prosthetic footplates can influence how forces and torques are transmitted from the ground to the residual limb during walking, but existing emulator designs include neither. This paper presents the design, development, and evaluation of a custom robotic ankle-foot prosthesis with off-board actuation and control that attaches to clinically standard carbon fiber prosthetic footplates and can fit inside of a cosmesis and shoe during walking. We built an off-board actuation and control unit that actuates the ankle-foot prosthesis through a Bowden cable tether. Our prosthesis design has a novel mechanical architecture that incorporates a parallel torsion spring and allows for bi-directional control of the sagittal plane ankle torque with a single off-board motor. We conducted benchtop tests of the robotic performance of the prosthesis and evaluated its performance during walking tests with two transtibial amputees. This device will enable future research on amputee biomechanics and the design and control of ankle-foot prostheses.

3.3 *Mechatronic System Design*

We designed and built a custom ankle-foot prosthesis and off-board actuation and control unit for use in walking experiments with transtibial amputees. Our intent for this work was to 1) develop a universal off-board actuation and control unit capable of providing robotic assistance to a variety of future wearable devices, and 2) develop a lightweight prosthesis end-effector capable of producing high ankle torques and power. The off-board actuation and control unit consists of a motor, controller, and experimenter interface that rest on a cart (Figure 3.1). The prosthesis generates plantarflexion moments via a Bowden cable actuation tether, and dorsiflexion moments with a pair of torsion springs in parallel with the Bowden cable. These dorsiflexion moments can be further modulated by the applied torque from the off-board actuation unit. Sensor measurements from the prosthesis and instrumented treadmill are measured by the controller and are used for closed-loop torque control. This section describes the details of the design of both the Configurable Off-Board Robotic Actuator (COBRA) and the prosthesis end-effector (COBRA Ankle).

3.3.1 *Off-board Actuation and Control Hardware*

The primary actuator for the COBRA system is a brushless servomotor (AKM74L, Kollmorgen, Radford, VA). The actuator is powered by a dedicated three-phase 208 VAC circuit that rests on the bottom of the cart. The actuator is commutated by a servo drive (AKD-P02406) that can operate in closed-loop position, velocity, or current control mode. The servo drive can measure and report motor position, velocity, and q-axis current. The torque and power output of the system are limited by the servomotor, which is rated for a continuous torque of 49.7 Nm, a peak torque of 143 Nm, and a continuous power output of 5.47 kW. The servomotor was selected using an optimization procedure and dynamical simulations of human walking with robotic assistance at the knee and ankle [134]. Our algorithm iterated over motors from the Kollmorgen catalog and varied mechanical design parameters while minimizing

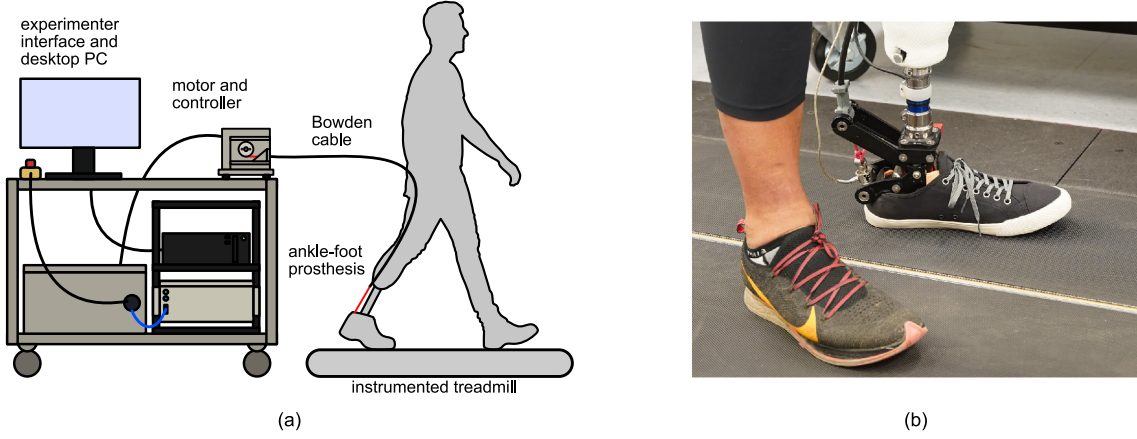


Figure 3.1: (a) Schematic of the hardware layout for the COBRA system. Control and actuation hardware are placed on a stationary cart. Flexible Bowden cable tethers transmit forces to the prosthesis end-effector while a participant walks on an instrumented treadmill. (b) Photograph of the ankle-foot prosthesis end-effector attached to a research participant with transtibial limb loss. The end-effector consists of a custom ankle joint that attaches to standard clinical prosthesis hardware, including a pyramid adapter and various sizes of carbon fiber prosthetic footplates.

reflected inertia at the prosthesis. Low reflected inertia is beneficial for human-robot interaction and is usually lowest in electromechanical systems that use high torque motors and no gearbox, as reflected inertia increases with the square of the gear ratio. Our system does not use an explicit gear box, but some degree of mechanical advantage is achieved through differential Bowden cable leverage at the actuator and end-effector.

We designed and built a custom transmission that delivers motor power and torque to a Bowden cable actuation tether (Figure 3.2). The frame of the transmission was machined from 6061 aluminum (Sound Machine Products, Kent, WA). The stator of the motor bolts directly to the frame. The motor's output shaft is coupled to a carbon steel drive shaft through a flexible coupler. The drive shaft is supported by a pair of ball bearings and was sized to resist large radial loads from the Bowden cable. The Bowden cable consists of a hollow sheath made of helically coiled steel (Lexco Cable, Norridge, IL) and a sliding inner

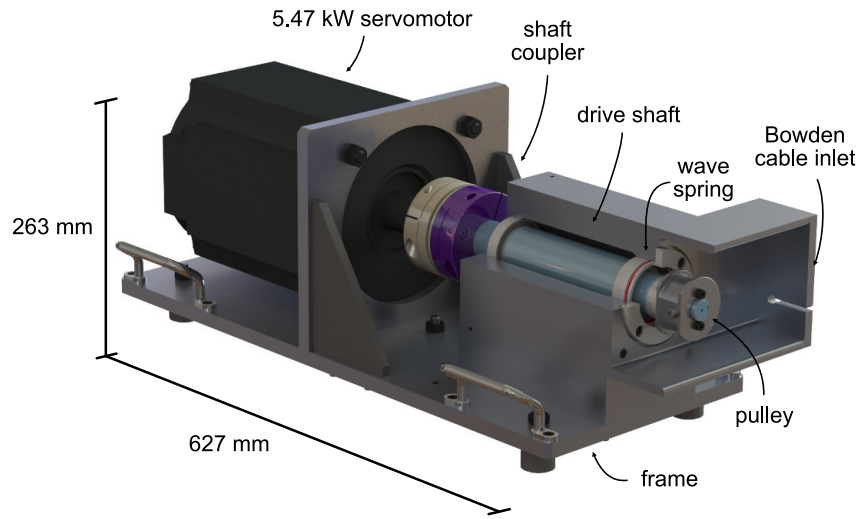


Figure 3.2: Primary COBRA actuator with components labeled. Motor torque and power are delivered to a Bowden cable through a transmission consisting of a flexible shaft coupler (purple/gold), a drive shaft supported by bearings (blue), and a pulley that is fixed to the shaft. A wave spring (red) is included to apply axial preload to the bearings and reduce shaft vibration. The bearing hub block is shown as a partial section view to expose the drive shaft.

synthetic rope cable (West Marine, Fort Lauderdale, FL). The outer sheath clamps to the transmission frame, while the inner rope wraps around a 2.51 cm radius pulley fixed to the drive shaft. Given the peak motor torque and pulley radius, peak cable forces at the motor frame could reach 5.7 kN, but peak forces at the end-effector are reduced due to frictional losses between the outer sheath and inner rope.

The measurement and control unit for the off-board system is a PXIe-8880 (NI, Austin, TX). A desktop PC functions as an experimenter interface and communicates with the PXIe-8880 (controller) through a local area network. Both the desktop PC and controller run custom LabVIEW (NI, Austin, TX) 2018 programs. During experiments, the controller streams lossy data from the prosthesis and torque control system to the desktop PC for data visualization. Lossless data is logged to a dedicated hard drive on the controller.

High-level prosthesis behavior is governed by parameters that are periodically sent from the PC to the controller, while real-time control algorithms are implemented on the controller. The controller communicates with the servo drive through an ethernet cable and etherCAT protocol. The controller has multiple configurable high-speed data acquisition cards for various analog and digital wearable sensors. Both the PC and controller are mounted on a server rack on the bottom of the cart.

3.3.2 Ankle-foot Prosthesis Design and Characterization

Our design intent for the COBRA Ankle was to develop a prosthesis capable of replicating the sagittal plane biomechanics of the biological ankle for a 115 kg person walking at roughly 1.25 m/s. We used biomechanical ankle trajectories from [1] to set design criteria (Table 3.1) and determine mechanical design parameters for the prosthesis. We aimed to minimize prosthesis mass where possible.

The COBRA Ankle consists of a custom rotary joint that bolts to a carbon fiber prosthetic footplate (LP Vari-flex, Össur, Reykjavík, Iceland). The ankle joint is a two-part structure that pivots about a carbon steel shaft on two flanged ball bearings. The joint housing structure was machined from high-strength 7075 aluminum alloy. The Bowden cable outer sheath attaches to a lever arm on the top side of the joint, and the inner rope attaches to a lever arm on the bottom side of the joint. When the servomotor rotates, it pulls the inner rope through the outer sheath, which pulls the lever arms towards each other and generates a plantarflexion moment about the joint. Equal and opposite forces in the outer sheath of the Bowden cable oppose the forces from the inner rope, so no net force is applied to the prosthesis, only a joint moment [29]. Dorsiflexion moments are provided by a pair of custom parallel torsion springs within the ankle joint. A clinically standard pyramid adapter bolts to the top of the joint for attachment to the user's prosthetic pylon. Figure 3.3 shows the design of the ankle-foot prosthesis.

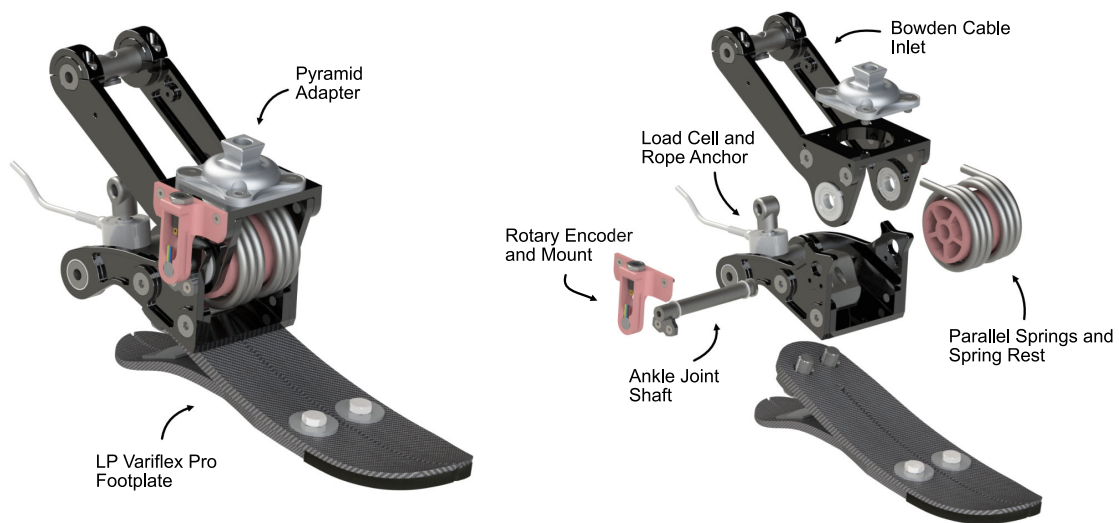


Figure 3.3: Ankle prosthesis as assembled (left) and as an exploded view (right). Primary components are labeled. The custom prosthesis joint attaches to standard clinically-available prosthesis componentry, including the pyramid adapter and Ossur LP Variflex footplate. 3D printed components are shown in red. An additional cover for the encoder mount and 3D printed component that clamps the Bowden cable to the Bowden cable inlet are not shown for clarity.

Table 3.1: Ankle-foot Prosthesis Design Specifications

	Target	Achieved	Units
Peak Plantarflexion Moment	164	178	Nm
Peak Dorsiflexion Moment	26	20-30*	Nm
Peak Dorsiflexion Angle	15	30	deg
Peak Plantarflexion Angle	14	16	deg
Peak Power at Push-off	280	368	W

*Peak dorsiflexion moment varies with ankle angle.

The LP Vari-flex footplate was selected due to its low stack height and split toe design, which provides compliance between the user and the ground in the frontal plane. Our custom joint attaches to multiple different footplate sizes, from as low as 23 cm to 30 cm in length. During walking, the prosthesis fits inside of a cosmetic foot shell and shoe, which provide additional compliance in the frontal and transverse planes. Compliance in the frontal and transverse planes can improve balance and shear loading at the interface between the prosthetic socket and residual limb during walking [79, 135].

Mechanical hard stops limit the sagittal plane range of motion between 16 degrees of plantarflexion and 30 degrees of dorsiflexion. The stack height, as measured from the bottom of the prosthetic footplate to the top of the pyramid adapter, is 13.5 cm. With a size 27 cm footplate, the total ankle-foot prosthesis mass is 1.19 kg, 18% of which is due to the prosthetic footplate. The cosmesis and shoe add an additional 0.47 kg, for a total worn mass of 1.67 kg. Worn mass varies with footplate and shoe size, and the prosthesis can be used without the shoe and cosmesis if a lower mass is desired.

The purpose of the parallel torsion springs is to allow for the prosthesis to achieve both

net plantarflexion and dorsiflexion moments with a single servomotor. The Bowden cable can only apply plantarflexion moments, while the parallel springs bias the ankle consistently into dorsiflexion. With no cable force, the springs drive the ankle into the dorsiflexion hard stop. The springs and Bowden cable are in an agonist-antagonist configuration, so the net ankle torque is the sum of the torques provided by both elements. Therefore, the servomotor can control both plantarflexor and dorsiflexor moments by modulating cable tension across the gait cycle. We chose the torsion spring stiffness and rest angle based on ankle biomechanics data, simple estimates of actuator work across a stride, and manufacturing considerations, with a goal of achieving suitably high dorsiflexion moments while minimizing actuator work. Detailed spring design and manufacturing were carried out by Precision Coil Spring in El Monte, CA.

There are three sources of significant series elasticity between the servomotor and the load. The first source of series elasticity is the Bowden cable [34], the second is the prosthetic footplate, which is designed to flex significantly during walking, and the third is the soft-tissue interface between the residual limb and prosthetic socket. The amount of series elasticity provided by the prosthetic footplate is variable across the stride and depends on the anterior-posterior distance between the ground reaction force vector and the ankle joint center. As there are mechanical elements that provide joint torques to the prosthesis both in series and parallel with the servomotor, the actuator is a series-parallel elastic actuator [136]. The equation describing the net torque applied to the ankle is:

$$\tau_a = F_c r(\theta_a) + \tau_s(\theta_a) \quad (3.1)$$

where τ_a is the net applied prosthesis ankle torque, F_c is the applied Bowden cable force, r is the lever arm distance between the Bowden cable line of action and ankle joint center, τ_s is the torque provided by the torsion springs, and θ_a is ankle angle. Both spring torque and lever arm vary with ankle angle.

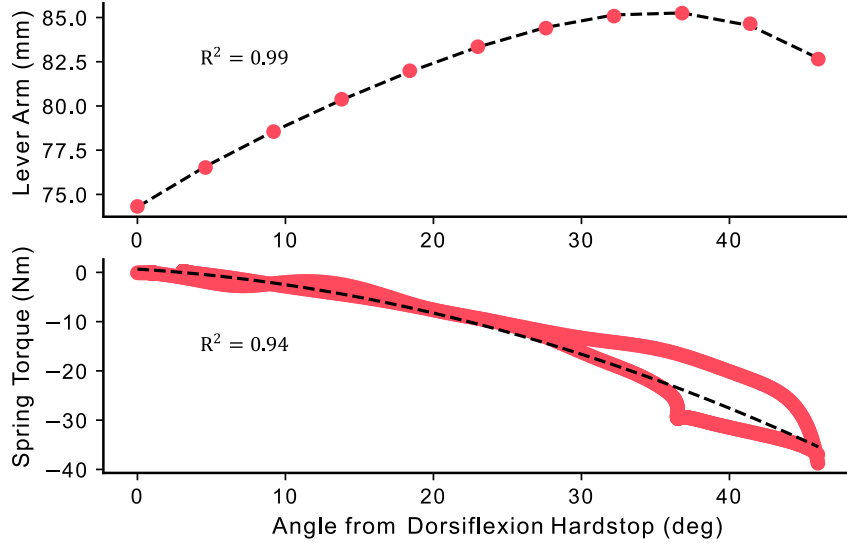


Figure 3.4: Mechanical characterization of the prosthetic ankle. Experimental data is red and best-fit lines are black. Curve fit equations are run in real-time on the prosthetic ankle torque controller. Top: Distance from the Bowden cable line of action to the prosthetic ankle joint center as measured by the 3D solid model used to design the joint. Given that the pulley at the transmission has a 25.1 mm radius, the transmission ratio between the motor and the ankle joint ranges from 3 to 3.4. Bottom: Dorsiflexion torque provided by the parallel torsion springs.

To control net ankle torque during operation, all terms in (3.1) must be measured or estimated in real-time. Force in the inner rope of the Bowden cable is measured with an inline load cell (FSH03887, Futek, Irvine, CA). Ankle angle is measured with a magnetic encoder (RM08, Renishaw, West Dundee, IL) and magnet recessed within the ankle joint shaft. We measured the line of action between the Bowden cable and ankle joint center, i.e., lever arm, at 11 equally spaced points across the range of motion within the COBRA Ankle CAD model. We fit a 4th order polynomial to the data that maps ankle angle as measured from the dorsiflexion hard stop to the instantaneous lever arm of the prosthesis (Figure 3.4).

We estimated the torques provided by the torsion springs through simple experiments.

The prosthesis was mounted sideways in a rigid frame so that the footplate mass did not cause a gravitational torque about the joint. The prosthetic footplate was not in contact with anything, so the only torques about the joint were those applied by the Bowden cable and the torsion springs. During the experiment, the motor pulled the ankle slowly through the range of motion. The equation of motion for the joint during the experiment was:

$$I\ddot{\theta}_a = F_cr(\theta_a) + \tau_s(\theta_a) \quad (3.2)$$

where I is the moment of inertia of the lower section of the ankle joint and footplate about the rotation axis. We assumed negligible inertial torques due to the low acceleration and mass of the footplate, reducing (3.2) to:

$$\tau_s(\theta_a) = -F_cr(\theta_a) \quad (3.3)$$

We used (3.3), measurements of cable force and ankle angle, and our lever arm model to compute spring torque across the range of motion. A second-order polynomial was fit to the angle-torque profile for use in the real-time control system (Figure 3.4).

3.4 Control System Design

We developed a hierarchical torque control system that prescribes and tracks ankle torque signals at 1000 Hz. All measurements used in the control system are low-pass Butterworth filtered at 100 Hz. A real-time gait cycle segmentation algorithm uses signals from the instrumented treadmill to measure foot-ground contact and provide a gait phase estimate used in the tracking controller. The prosthesis can operate in two high-level modes – passive emulation mode and active push-off mode. In passive emulation mode, the COBRA Ankle emulates an idealized passive ankle-foot prosthesis. In active push-off mode, additional torque is added in late stance to provide push-off work. Both modes were developed to evaluate the mechatronic performance of the prosthesis and were not necessarily intended to

be optimal for walking. Torque setpoint signals are sent to a closed-loop tracking controller that uses classical feedback control and iteratively learned feedforward control (Figure 3.5). Participant safety is ensured using software-level safety limits, an emergency stop button that cuts motor power, and a fall harness.

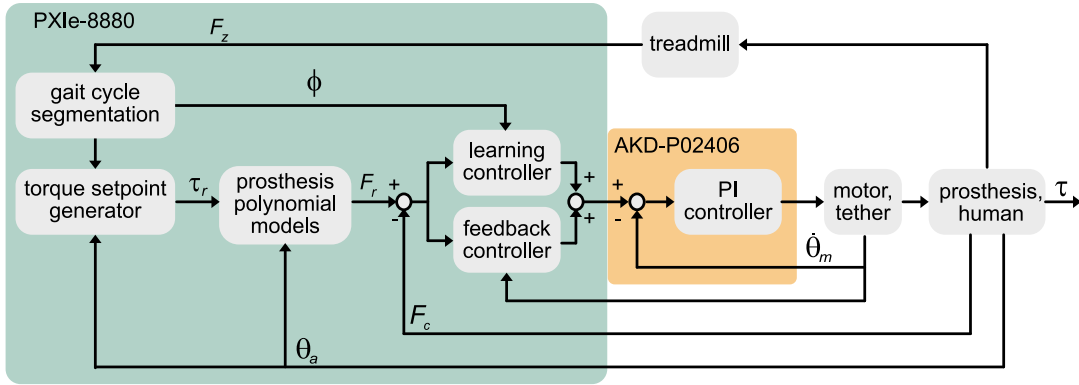


Figure 3.5: Torque control system for the COBRA Ankle. Torque setpoints, τ_r , are generated using ankle angle, θ_a , and ground reaction forces, F_z . Torque setpoints are transformed into cable force setpoints, F_r , using models of the ankle’s lever arm and spring torque. A cascaded control architecture is used to track cable force setpoints, where an outer force tracking loop generates motor velocity commands, $\dot{\theta}_r$, for an inner motor velocity loop. The force tracking loop uses feedback on measured cable force error with a damping term on measured motor velocity, $\dot{\theta}_m$. The learning controller generates a feedforward trajectory as a function of gait phase, ϕ .

3.4.1 Real-time Gait Cycle Segmentation

Heel-strikes are identified in real-time with a threshold on the rate of the ground reaction force. Vertical ground reaction force, F_z , is measured with a Bertec split-belt instrumented treadmill on the prosthesis side. The signal is then saturated at 200 N. The rate of force development for the saturated signal is computed using a finite-difference derivative. Prosthesis-ground contact is detected on the first loop iteration during swing phase where the vertical

loading rate is higher than a preset threshold. This threshold is tuned for each participant. The ground reaction force signal is saturated at 200 N to prevent loading rate fluctuations in midstance from triggering an additional heel-strike. We found this method of detecting heel-strike to be slightly more responsive than a threshold on the ground reaction force magnitude in pilot testing. Toe-off is detected when the ground reaction force magnitude drops below 20 N.

The control system uses heel-strike information to estimate the instantaneous stride percentage, ϕ . To compute ϕ , a timer is started at each heel-strike. At each loop iteration, the current time is divided by the average stride duration over the previous five strides. ϕ is reset to zero on loop iterations where a heel-strike is detected. If ϕ reaches one, it is clamped there until the next heel-strike.

3.4.2 Ankle Torque Setpoint Generation

Both the passive and active torque setpoint generators use finite state machines to govern prosthesis behavior during walking (Figure 3.6). In passive mode, the ankle emulates a virtual torsion spring with two linear stiffness regions that meet at a virtual rest angle. One virtual stiffness region generates plantarflexion moments and the other generates dorsiflexion moments, which approximates the differences in stiffness between the heel and forefoot of a passive ankle-foot prosthesis. The virtual rest angle is set near the pylon vertical angle and is tuned during experiments to prevent toe drag during swing phase. The prosthesis produces torques proportional to the angular deviation from the virtual rest angle, with a stiffness constant that switches depending on whether the angle is plantarflexed or dorsiflexed relative to the virtual rest angle:

$$\tau_r = -K_i(\theta_a - \theta_o) \quad (3.4)$$

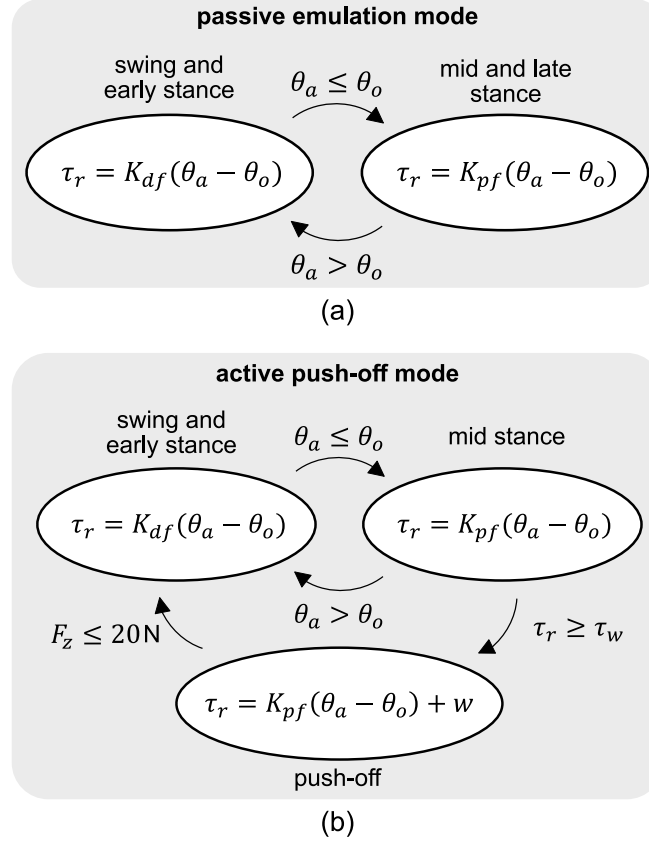


Figure 3.6: Finite state machine diagrams for the COBRA Ankle control system. (a) Finite state machine for ankle torque setpoint generation when the ankle is in passive mode. The passive torque setpoint generator switches between two linear stiffnesses depending on the plantarflexion angle of the prosthesis. (b) Finite state machine for ankle torque setpoint generation when the ankle is in passive mode. The active torque setpoint generator is similar to the passive setpoint generator, provides push-off in late stance when the requested torque exceeds a tunable threshold. When the threshold is exceeded, an additional constant offset torque is added to the passive stiffness, which is equivalent to an instantaneous change in the rest angle of the virtual spring.

In (3.4), τ_r is the torque setpoint, K_i is a stiffness constant that switches depending on ankle angle, θ_a , and θ_o is the virtual rest angle. K_i switches to K_{pf} when the ankle is dorsiflexed and to K_{pf} when the ankle is plantarflexed. to K_{pf} , to K_{df} , and θ_o can be changed through the experimenter interface during walking experiments.

Active push-off mode adds an additional state to the passive emulation mode in late stance. When the torque setpoint from the passive emulation mode exceeds a threshold, τ_w , the controller switches to the push-off state:

$$\tau_r = -K_{pf}(\theta_a - \theta_o) + w \quad (3.5)$$

where w is a constant torque offset. The sudden addition of w is equivalent to instantaneously changing the rest angle of the virtual spring, which adds energy to the virtual spring and causes the prosthesis to push-off in late stance. The prosthesis stays in the push-off state until toe-off is detected. Both τ_w and w are tunable parameters. Changing τ_w influences push-off timing and changing w influences the magnitude of the mechanical ankle work done in push-off. The torque setpoints in (3.5) are not intended to be followed precisely by the tracking controller. The setpoint changes instantaneously, but the true rate of torque development is gradual and dependent on the tracking controller behavior.

Both controller modes limit their dorsiflexion torque setpoints based on the maximum amount of torque the parallel spring can provide at a given angle. If the dorsiflexion torque setpoint exceeds the torque that can be provided by the springs, the setpoint is clamped to the maximum spring torque (i.e., very low cable force). Both the passive and active mode control laws are variants of methods described in [87] and [137].

3.4.3 Torque Tracking Controller

The purpose of the tracking controller is to drive the ankle torque to the commands sent by the torque setpoint generators in real-time. The tracking controller first transforms the

desired torque setpoint to a desired cable force setpoint using the polynomial prosthesis models developed in Section II. The tracking controller uses a cascaded architecture with an outer Bowden cable force control loop and an inner motor velocity control loop. The cable force control loop is implemented on the NI PXIe-8880 and uses a feedback controller with an iteratively learned feedforward term that corrects for tracking errors that happen at repeatable points in the gait cycle. The motor velocity controller is implemented on the Kollmorgen AKD-P02406 and uses a proportional-integral controller with feedforward friction compensation. Controller gains for the motor velocity loop were set using an auto-tuning procedure when the Bowden cable was detached from the motor and transmission. The controller sampling frequency is 1,000 Hz for the outer force control loop and 1.5 MHz for the inner velocity control loop.

Torque setpoint signals are transformed into cable force setpoints using measurements from the ankle encoder and the polynomial models that describe spring torque and lever arm. Eq. (3.1) is solved for the cable force required to achieve the desired ankle torque:

$$F_r = \frac{\tau_r - \tau_s(\theta_a)}{r(\theta_a)} \quad (3.6)$$

where F_r is the reference cable force that is required to drive the prosthesis to the desired net ankle torque. Requested cable force is saturated within the control system to be between 5 N and 2224 N. The lower limit is to prevent cable slack, and the upper limit is the capacity of the load cell that measures cable tension.

The force tracking loop uses parallel feedback and feedforward control algorithms to send a command to the inner motor velocity loop. The feedback controller uses proportional control on the cable force error and a damping term on the motor velocity measurement. The feedforward controller is an iterative learning controller (ILC) that leverages the cyclic nature of walking to reduce systematic tracking errors at consistent points in the gait cycle. The command sent to the motor velocity controller at every loop iteration is:

$$\dot{\theta}_r(\phi, n) = K_p e_F - K_d \dot{\theta}_m + \dot{\theta}_L(\phi - D, n) \quad (3.7)$$

In (3.7), K_p is and K_d are positive scalar feedback gains, e_F is the difference between desired and measured cable force, and $\dot{\theta}_m$ is the motor velocity measurement. $\dot{\theta}_L$ is the iteratively learned feedforward trajectory and is a function of gait phase, ϕ , and the current stride number, n . D is a phase delay parameter. $\dot{\theta}_L$ is initialized with an empty trajectory and is updated with the following equation at every heel-strike:

$$\dot{\theta}_L(\phi, n) = K_L e_F(\phi, n - 1) + \beta \dot{\theta}_L(\phi, n - 1) \quad (3.8)$$

In (3.8), K_L is a learning gain that scales the cable force error trajectory from the previous stride and β is a forgetting gain ($0.95 \leq \beta \leq 0.99$) that prevents instability. In effect, the feedforward motor velocity trajectory for a given stride is a weighted average of the cable force tracking error and the feedforward trajectory from the previous stride. The delay, D shifts learned trajectory in phase so that changes in motor velocity preempt systematic errors. ILC is a common solution for challenges in force control of Bowden cables due to their highly nonlinear friction characteristics [30, 123], and our algorithm is an independent implementation of the ‘‘PD-LRN’’ controller described in [30]. The ILC can be activated and deactivated through the experimenter interface, so the system can operate with feedback control only when desired. When the prosthesis is in active push-off mode, the ILC is ‘clamped’ to the stiffness-based feedforward trajectory, because the sudden discontinuity in ankle torque setpoint at push-off caused high frequency ripples to form in the learned signal.

3.5 Experimental Methods

We conducted experiments with the COBRA Ankle that evaluated 1) the robotic performance of the system in isolation and 2) the biomechatronic performance of the prosthesis during walking with two transtibial amputee participants. We evaluated the peak torque

and controller bandwidth in benchtop experiments. During walking experiments, we evaluated the tracking controller performance and the ability of the prosthesis to produce push-off power and work during walking.

3.5.1 Benchtop Experiments

We built a rigid test frame for benchtop experiments with the prosthesis. The top of the prosthesis joint was bolted to the upper section of the test frame and the footplate pressed into a reaction bar bolted to the bottom section of the frame. The prosthesis ankle joint was not explicitly locked during the benchtop tests and could pivot slightly when the carbon fiber footplate deflected. All tests were conducted with a 27 cm footplate inside of a cosmetic foot shell and shoe. The reaction bar was placed at the midpoint between the ankle joint center and the toe of the footplate.

Torque step tests were conducted to evaluate the maximum plantarflexion torque for the prosthesis. Step input tests were conducted with peak torques at 75 Nm, 125 Nm, and 175 N, with a 25 Nm preload. The highest torque value was limited by the capacity of the prosthesis load cell. For each test, the system was held at the preload for one second, commanded to the intermediate torque value for one second, and then returned to the preload for one second. Three tests were conducted for each magnitude. We averaged the measured torque trajectories at each magnitude and computed the 90% rise and fall times and percent overshoot.

We conducted bandwidth tests to evaluate the responsiveness of the torque controller with sinusoidal torque setpoint signals. The torque setpoint signals had a constant offset of 100 Nm and oscillated between 50 Nm and 150 Nm at fixed frequencies. We tested eight discrete oscillation frequencies ranging from 0.4 Hz to 6 Hz. Experiments were stopped at 6 Hz due to excessive vibrations at the off-board motor and cart when higher frequencies were tested. We computed the gain and phase lag of the closed-loop control system at each

Table 3.2: Participant Demographics

	Participant I	Participant II
Body Mass (kg)	104.8	114.2
Leg Length (m)	0.89	0.95
Walking Speed (m/s)	1.18	1.22
Prosthetic Side (L/R)	R	L

frequency and fit a transfer function model to the frequency response data for future analyses of the controller performance. We defined the closed-loop control bandwidth as the frequency where the magnitude response crossed -3 dB.

3.5.2 *Walking Experiments*

We recruited two participants with unilateral transtibial limb loss to walk on the COBRA Ankle for an IRB approved research study at the Center for Limb Loss and Mobility (CLiMB) within the Veterans Affairs Puget Sound Health Care System in Seattle, WA. The purpose of the experiment was to evaluate the mechanical and robotic performance of the prosthesis during walking and was not to measure or improve the participants' walking patterns in response to the prosthesis.

Before data collections, the prosthesis was configured to match the side of the participant's amputation and was fit with a footplate that matched the size of their prescribed prosthesis. Upon arrival, the participants were weighed, and their leg lengths were measured as the distance from the head of the greater trochanter to the floor on the prosthetic-side limb. A certified prosthetist then fit and aligned the experimental prosthesis to the participant. For all walking experiments, both participants walked at a non-dimensionalized Froude walking

Table 3.3: Prosthesis Settings for Passive Emulation Mode

	Participant I	Participant II	Units
Low K_{pf}	9.73	10.60	Nm/deg
Medium K_{pf}	12.16	13.25	Nm/deg
High K_{pf}	14.59	15.90	Nm/deg
K_{df}	3.93	4.28	Nm/deg
θ_o	-5	-2	deg

speed of $Fr = 0.16$, where $Fr = v^2/gl$, v is walking speed, and g is the gravitational constant. We elected to have participants walk at this Froude speed because it is near self-selected walking speeds for people without limb loss, and it allows for comparisons to other studies of powered prosthetic ankles [1, 14]. Table 3.2 shows the participant demographics and walking speeds.

For experiments with passive emulation mode, we collected data at low, medium, and high virtual forefoot stiffness values (K_{pf}), corresponding to values of 0.093, 0.116, and 0.139 Nm/deg-kg. The virtual stiffness values are near the quasi-stiffness of the biological ankle during walking and were based on [137]. For all walking trials, the virtual heel stiffness (K_{df}) was held at 0.038 Nm/deg-kg. The virtual rest angle of the prosthesis, θ_o , was set so that the prosthesis was comfortable during standing and the toe had significant ground clearance during swing phase. We collected data in an additional condition with the medium forefoot stiffness and the iterative learning controller disabled to evaluate the effect of the ILC on the tracking error. Table 3.3 shows the settings of the prosthesis across passive emulation mode conditions.

Initially, both participants were given between five and ten minutes to adapt to the experimental prosthesis on the medium stiffness condition. The participants then walked on

the low, medium, and high stiffness settings for several minutes each. Two ten-second trials were logged at each condition.

During experiments with active push-off mode, the participants walked with the medium virtual forefoot stiffness. The push-off torque threshold, τ_w , was initially set so that push-off would be triggered slightly before the peak ankle torque in passive emulation mode, and then conversationally tuned with feedback from the participant. The additional push-off torque, w , was initially zero and increased over several minutes, until the experimenter observed peak ankle power values near values reported in the literature. Ten-second trials were logged at several increasing values of w .

We evaluated the torque tracking performance of the prosthesis when it was in passive emulation mode, and the average peak sagittal plane ankle torque and power when the prosthesis was in active push-off mode. All results were computed using the instrumented treadmill and sensors within the COBRA Ankle. Ankle angles and moments were filtered with a fourth-order zero-lag Butterworth filter at 12 Hz and 20 Hz, respectively. We computed the ankle angular velocity using finite-difference derivatives and computed ankle power by multiplying ankle velocity with ankle torque. In passive emulation mode, we evaluated both total and systematic RMS tracking error. Total RMS error computes error between the torque setpoint and measurement across the concatenation of all trials for a given condition. Systematic RMS error was computed using the torque setpoint and measurement for the average stride within a condition [30]. We did not evaluate torque tracking performance in active push-off mode, because tracking error is not a meaningful metric for our push-off controller. A large, instantaneous torque tracking error is imposed by our setpoint generator, and the response of the tracking controller response is what causes push-off.

3.6 Results

3.6.1 Benchtop Results

The peak ankle plantarflexion torque achieved in the step response tests was 178.3 Nm (Figure 3.7). For the 175 Nm test, the 90% rise and fall times were 146 ms and 139 ms, respectively. The percent overshoot was 5.3%. A time delay in torque development of approximately 48 ms was noted for all three step test conditions, which was not present in the motor current or velocity signals. The highest frequency successfully evaluated in the bandwidth tests was 6 Hz, which had a magnitude response of -2.83 dB and phase delay of -249.3 degrees. The frequency response was well characterized by a second-order transfer function with a time delay:

$$G(s) = \frac{1.001}{0.001s^2 + 0.024s + 1.000} e^{-0.048s} \quad (3.9)$$

The time delay of 48 ms observed during the step responses was fixed during model fitting. The -3 dB crossing of the model fit is only slightly outside of our tested frequency range, so we used the model to estimate our torque control bandwidth of 6.1 Hz. The closed-loop system has a resonant frequency at approximately 4 Hz.

3.6.2 Walking Results

During the walking tests, both participants walked comfortably on the COBRA Ankle (Figure 3.8). For both participants, the total RMS error across all passive emulation conditions was 4.12 Nm, while the RMS torque tracking error for an average stride (systematic error) was 2.71 Nm, both of which are less than 5% of the peak torques during the walking experiments. Table 3.4 shows tracking errors across all stiffness conditions for both participants. During tests of the medium stiffness condition where the ILC was disabled, the total and systematic tracking RMS errors were 8.49 Nm and 6.14 Nm, respectively, indicating that the

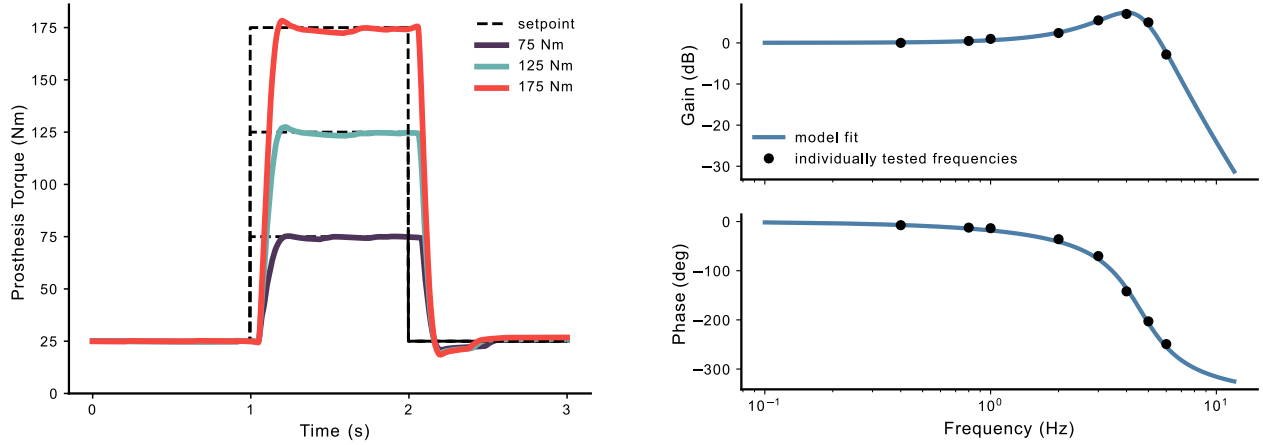


Figure 3.7: Benchtop test results of the prosthesis and control system. Left: Step responses collected with the foot mounted in a test stand. Right: Frequency response collected with the foot mounted in a test stand. Discrete sine waves of torque setpoint were sent to the controller with an offset of 100 Nm and amplitude of 150 Nm. A second-order transfer function model with a time delay was fit to the experimental gain and phase data.

ILC reduced tracking errors by roughly 50%. Figure 3.9 shows ankle biomechanics for the three stiffness conditions.

During active push-off trials, the peak ankle powers achieved by the COBRA Ankle were $351.6 \text{ W} \pm 33.88 \text{ W}$ and $368.03 \text{ W} \pm 12.67 \text{ W}$ for participants one and two, respectively. Body mass normalized, these values are $3.08 \text{ W/kg} \pm 0.30 \text{ W/kg}$ and $3.51 \text{ W/kg} \pm 0.12 \text{ W/kg}$. As a point of comparison, Russell Esposito et al found peak ankle powers of $3.15 \text{ W/kg} \pm 0.73 \text{ W/kg}$ for the BiOM and $2.29 \text{ W/kg} \pm 0.42 \text{ W/kg}$ for intact ankles when walking at the same Froude speed [14]. Figure 3.10 shows how ankle power in active push-off mode increased with the parameter w .

3.7 Discussion

We designed, built, and evaluated a robotic ankle-foot prosthesis and off-board actuation and control unit for walking experiments with transtibial amputees. The COBRA Ankle uses a

Table 3.4: Stance Phase Torque Tracking Errors in Passive Emulation Mode

	Low Stiffness RMSE		Medium Stiffness RMSE		High Stiffness RMSE		All Conditions	
	Total*	Systematic	Total	Systematic	Total	Systematic	Total	Systematic
Participant I	4.15	3.40	4.05	2.91	4.77	3.65	4.32	3.32
Participant II	3.43	2.28	3.62	2.55	4.81	1.91	4.12	2.10
Mean	3.79	2.84	3.84	2.73	4.79	2.78	4.22	2.71

*Total RMSE is computed over all trials for a given condition. Systematic RMSE is calculated using torque error for the average stride for a condition.

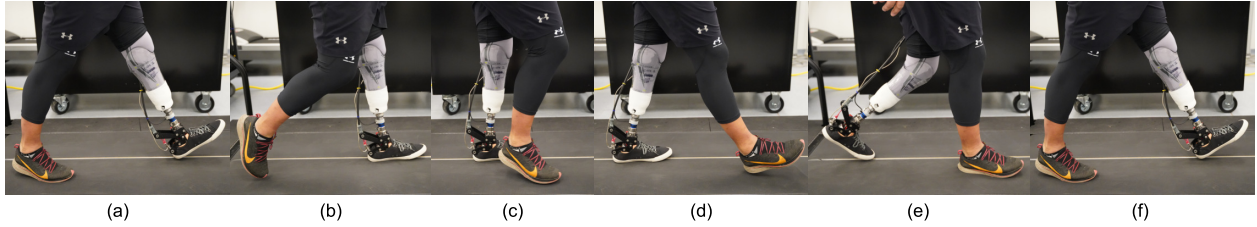


Figure 3.8: One gait cycle on the COBRA Ankle. (a) Initial contact between the prosthesis and treadmill is detected in real-time. (b) Foot-flat occurs. Between (a-b), the prosthesis produces a net dorsiflexion torque due to the parallel torsion spring. Near (b), the ankle angle moves through the virtual neutral angle and the prosthesis switches to a net plantarflexion torque. (c) During mid-stance, the prosthesis bears the full bodyweight of the participant. (d) The peak dorsiflexion angle corresponds with the peak plantarflexion torque. When the prosthesis is in active push-off mode, push-off is triggered near this moment. (e) The prosthesis leaves the ground and returns to the virtual neutral angle during swing. (f) Heel-strike for the next stride occurs.

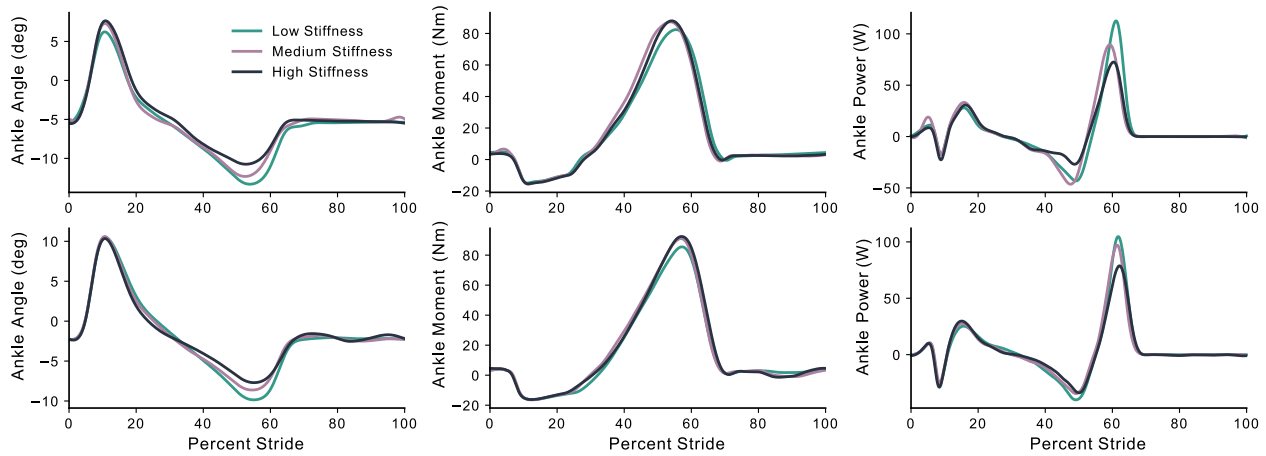


Figure 3.9: COBRA Ankle biomechanics across three stiffness conditions. The top row shows data from Participant I and the bottom from Participant II. Peak ankle torques remained relatively constant across conditions, while peak ankle dorsiflexion angles decreased as stiffness increased. Peak ankle power was lowest for the stiffest condition.

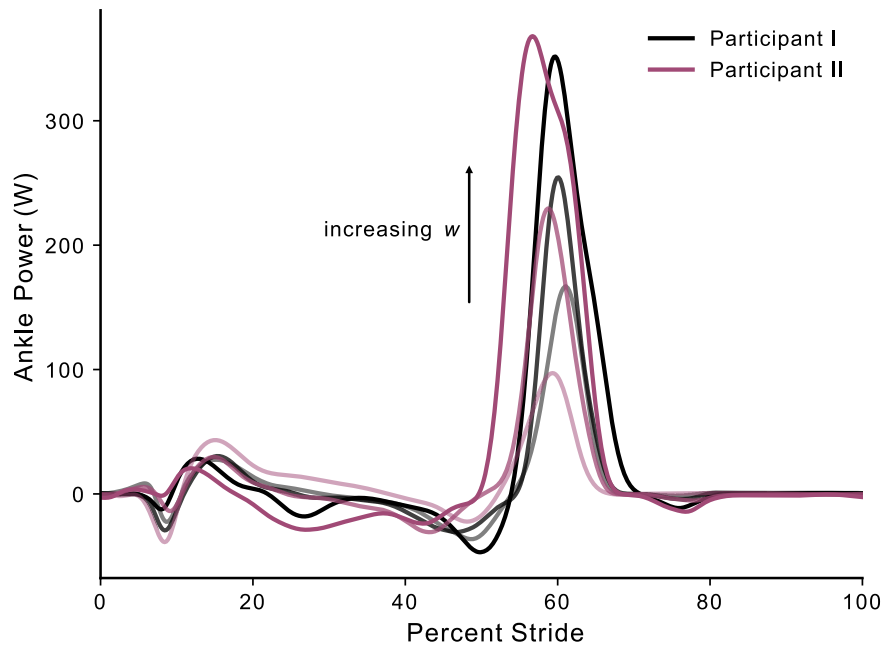


Figure 3.10: Ankle power during active push-off mode tests. As the push-off torque parameter w from (5) was increased, peak ankle power increased.

novel mechanical layout that pairs an off-board motor and Bowden cable tether with a parallel spring to provide bi-directional torque control during walking. We performed benchtop tests that showed that the prosthesis can provide high plantarflexion torques and that the torque feedback controller is sufficiently responsive for walking. Walking experiments indicated that the system achieves low tracking error and can provide peak ankle push-off power greater than that of the biological ankle and similar to other robotic prostheses. Our experiments indicate that COBRA and the COBRA Ankle are highly performant experimental platforms that will enable a wide variety of future research in prosthesis design and control.

One of the biggest differences between the COBRA Ankle and other research prostheses with off-board actuation and control is that it can estimate and control the torque about the ankle joint throughout the entire gait cycle, from heel-strike to heel-strike. Other emulator designs provide dorsiflexion moments via passive elements in parallel with the prosthetic pylon [29, 132]. These dorsiflexion moments are not measured or estimated in real-time, so total control of the net ankle torque can only be achieved after the passive heel element lifts off the ground in mid-to-late stance. The key difference for our design is that our dorsiflexion spring is embedded within the mechanical joint and its deflection is measured with an encoder. These capabilities allow for different types of experiments to be carried out with the COBRA ankle, e.g., experiments that explore the effects of systematic variation of dorsiflexion moments in early stance.

Inclusion of the parallel torsion spring allowed for active modulation of dorsiflexion torque but required several design trade-offs. In terms of benefit, torsion springs are relatively easy to design and are readily available by a variety of manufacturers. Our design is relatively compact and simple compared to prosthesis designs that achieve parallel elasticity using compression springs [21, 93]. Our parallel spring is biased into dorsiflexion, such that the rest angle of the spring is outside of the range of motion of the ankle. In this configuration, the Bowden cable must overcome the spring torque before it can develop a net ankle

plantarflexion torque. This configuration would require more consideration for a battery powered prosthesis, as energy is expended biasing the spring. However, the spring torque and energy storage is small relative to the capabilities of the off-board motor, so the trade-off is reasonable. Torsion springs can exhibit more hysteresis than leaf springs or compression springs due to inter-coil friction during loading and unloading. Hysteresis, as seen in Figure 3.4, makes it challenging to model the spring torque as a function of ankle angle and affects our ability to estimate total joint torque. Issues with our spring could be improved by winding the torsion spring with more spacing between adjacent coils to avoid friction losses or redesigning the joint to use a composite leaf spring instead. Alternatively, a more advanced real-time model could be developed that more accurately characterizes the dynamic torque output of the spring.

Inclusion of the carbon fiber prosthetic footplate also involved design trade-offs. The compliance in the footplate protects our mechanical joint from shock loading, e.g., at heel-strike. The engineering design process was also made substantially simpler by incorporating the widely available commercial part that is well-suited for human walking. We are also able to swap footplates to match the participant's shoe size, which allows us to match foot length between the amputated and intact sides. Foot length is known to affect the mechanics and energetics of walking [138]. The split-toe design of the LP Variflex also provides beneficial compliance in the frontal plane.

The footplate deflects due to loading during walking, which is not measured. Our encoder measures the angular displacement of the mechanical ankle joint, whereas many of ankle angle measures from the biomechanics literature are derived from ankle-foot models that combine angular displacements from the talocrural, subtalar, and forefoot joints [139]. Therefore, comparing the presented kinematics and power of the COBRA Ankle to values from the literature must be done with caution. For example, depending on whether the footplate is absorbing/returning energy during late stance, the total ankle-foot power would

be lower/higher than the ankle power measured using onboard sensors. This phenomenon is minimized in emulator systems with very stiff forefoot designs, such as [29, 132, 133].

The shoe and cosmetic foot shell provided significant transverse plane compliance to our prosthesis. Shoes are known to change the stiffness and other relevant mechanical properties of prosthetic feet during walking [140], though this effect has not been quantified in the transverse plane to our knowledge. An appropriate amount of transverse compliance is important during walking, as can reduce painful shear forces between the residual limb and prosthetic socket [79]. Prosthetic foot emulators with low transverse plane stiffness could cause amputees to walk differently than they would in other scenarios to avoid uncomfortable loads on the residual limb. Emulators that use shoes may therefore have more external scientific validity.

Our torque controller bandwidth of 6.1 Hz is low relative to other ankle-foot prostheses with off-board actuation, with one system reporting bandwidths up to 33 Hz [133]. This is partly due to the experimental setup for our benchtop tests. It is common to lock the ankle joint during bandwidth tests [87, 131–133], but this method would likely overestimate our controller’s ability to develop torque during walking because it does not account for series elasticity between the ankle, ground, and user. Both the footplate and the interface between the socket and residual limb contribute to additional series elasticity. Even with a relatively compliant test setup, our bandwidth is similar to other robotic ankles [86, 87] and is clearly high enough to systematically alter prosthesis behavior during walking (Figure 3.9). Given our controller bandwidth, the COBRA Ankle currently cannot haptically emulate other devices with high fidelity. The controller is not fast enough to emulate passive devices when large external disturbances occur, e.g., at heel-strike. Future work to improve controller bandwidth will include 1) improving the interface between the motor and the cart to reduce vibration and 2) improving the signal-to-noise ratio in our ankle encoder measurements, which will allow us to use more aggressive feedback gains. However, not all research

applications require high fidelity emulation, and our systematic torque tracking errors were similar to other published work [29, 132, 133] due to the iterative learning controller.

3.8 Conclusion

We developed an off-board actuation and control system for wearable devices, as well as a tethered ankle-foot prosthesis with bidirectional torque control. Benchtop tests and walking experiments show that our device is well-suited for experimental research with transtibial amputees. These devices will be used in future studies on wearable robotic device design and control for people with limb loss and other clinical populations.

Chapter 4

DESIGN AND EVALUATION OF AN KNEE FLEXION ASSISTANCE EXOSKELETON FOR PEOPLE WITH TRANSTIBIAL AMPUTATION

Anthony J. Anderson, Yuri F. Hudak, Kira A. Gauthier, Brittney C. Muir, and Patrick M. Aubin

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4.1 Abstract

People with below-knee amputation walk with asymmetric gaits that over time can lead to further musculoskeletal disorders and decreased quality of life. While prosthesis technology is improving, prosthetic ankles may be fundamentally limited in their ability to restore healthy walking patterns because they do not assist the residual knee joint. The knee on the residual limb has muscular deficits due to the loss of the gastrocnemius, a biarticular muscle that crosses both the ankle and knee. Here we present the design, development, and preliminary evaluation of a robotic knee exoskeleton for people with transtibial amputation. The device is intended to restore gastrocnemius-like flexion moments to the knee on the residual limb. The exoskeleton uses a custom offboard actuation and control system to allow for a simple and lightweight design with high torque capabilities. A preliminary walking experiment with one person with transtibial amputation was conducted. The exoskeleton provided a range of knee flexion torque profiles and had an RMS tracking error of 1.9 Nm across four assistance conditions. This device will be used in future studies to explore the effects of providing knee

flexion assistance to people with transtibial amputation during walking. Long term, findings from studies with this exoskeleton could motivate future assistive device designs that improve walking mechanics and quality of life for people with limb loss.

4.2 Introduction

Lower limb amputation is a growing problem in the United States, largely due to increases in vascular diseases like diabetes and peripheral artery disease [36]. People with transtibial amputation exhibit slower walking speeds [43], increased metabolic energy expenditure during walking [45], and asymmetric walking patterns [1] relative to the general population. Over time, cumulative loading asymmetries lead to high levels of musculoskeletal comorbidities like knee osteoarthritis in the sound limb and chronic lower back pain [4]. Novel robotic ankle-foot prostheses promise to improve walking mechanics, but initial clinical outcomes are mixed [6, 16].

Over the two decades, there has been a large research and development focus on improving prosthetic ankle technology [74, 87]. However, transtibial amputees may also benefit from devices that assist other joints during walking. The residual knee joint exhibits muscular deficits due to the transection of the gastrocnemius muscle during amputation. The gastrocnemius is a large calf muscle that originates on the femur and inserts at the calcaneus via the Achilles tendon and produces coupled ankle plantarflexion and knee flexion moments. This function is not present in people with transtibial amputation, and the consequences of this impairment for walking are largely unknown. It is possible that the benefits of the gastrocnemius at the residual knee joint could be restored to people with transtibial amputation with an assistive knee exoskeleton.

Simulation studies have found that clinically standard energy-storage-and-return prostheses provide the vertical bodyweight support of the soleus muscle during walking, but not the forward propulsion provided by the healthy gastrocnemius [141, 142]. Other simulation

and experimental evidence has shown that the gastrocnemius and soleus have synergistic but distinct roles during walking in healthy populations [112]. Powered prosthetic ankles address some of the limitations with passive ankles, e.g., powered push-off, but they actuate the ankle joint only and are therefore fundamentally limited to emulating the behavior of the soleus.

During push-off, plantarflexor moments can induce knee extension through intersegmental dynamics, while knee flexion moments can stabilize the joint and prevent hyperextension [101]. In transtibial amputees, knee flexion moments during push-off are reduced [1], and must be provided entirely by muscles in the hamstrings rather than the gastrocnemius. Many of the muscles in the hamstrings are biarticular at the knee and hip, e.g., the biceps femoris. If biarticular muscles in the hamstrings are used to provide compensatory knee flexion moments, they may create maladaptive hip moments and contribute to asymmetric gait. An exoskeleton that provides knee flexion assistance during push-off may allow for a healthier gait that leads to fewer musculoskeletal comorbidities over time.

Several research groups have investigated the effects of pairing knee flexion assistance with prosthetic ankles. Previously, we built and tested a clutched spring biarticular prosthesis to replicate gastrocnemius function [125]. Other research groups have paired active robotic knee exoskeletons with robotic ankles [123,126]. Initial results have been mixed, with some participants showing large increases in metabolic walking economy [123]. Further research is needed to understand optimal assistance strategies for an artificial gastrocnemius.

This work introduces the design, development, and preliminary evaluation of a robotic knee flexion exoskeleton for people with transtibial amputation. The device is lightweight and can produce high torques with a custom offboard actuation and control system. The exoskeleton can be used with either a passive or powered prosthesis. The purpose of this study was to evaluate the mechatronic performance of the exoskeleton while it assisted a participant with transtibial amputation during walking. Our experimental results indicate that the

device can successfully provide a variety of knee flexion torque profiles during walking with low torque-tracking error. The exoskeleton described here will be used in future experiments to determine how knee flexion assistance may benefit people with transtibial amputation.

4.3 *Mechatronic Design and Control*

The purpose of the exoskeleton is to provide knee flexion moments to the wearer during walking (Figure 4.1). Our design intent was to build a lightweight device capable of delivering large moments and forces to the leg. We were inspired by other exoskeletons that achieved these aims with offboard actuation and planar carbon fiber componentry [30,143]. We made several design decisions that were specific to our intended population of transtibial amputees. For example, we required of our design that the posterior side of the leg was unobstructed when wearing the device so that our subjects could sit in a chair and rest during breaks in the data collection procedure. Additionally, the interface between the exoskeleton and the shank accounted for the prosthetic socket and pylon. We designed and built an offboard actuation system and prototype exoskeleton with a Bowden cable actuated pulley on the lateral side of the knee.

4.3.1 Offboard Actuation and Control System

The knee exoskeleton is actuated and controlled by a custom offboard actuation system, which is shown schematically in Figure 4.2. The offboard system, developed at the Center for Limb Loss and Mobility (CLiMB), is called the Configurable Off-Board Robotic Actuator, or COBRA. The primary components of COBRA are 1) a powerful servomotor and flexible transmission tether, 2) a high-speed measurement and control system, and 3) a desktop PC that runs an experimenter interface. These components sit on a cart that is positioned behind an instrumented treadmill during experiments.

The primary actuator for COBRA consists of a 5.47 kW three-phase servomotor (Koll-

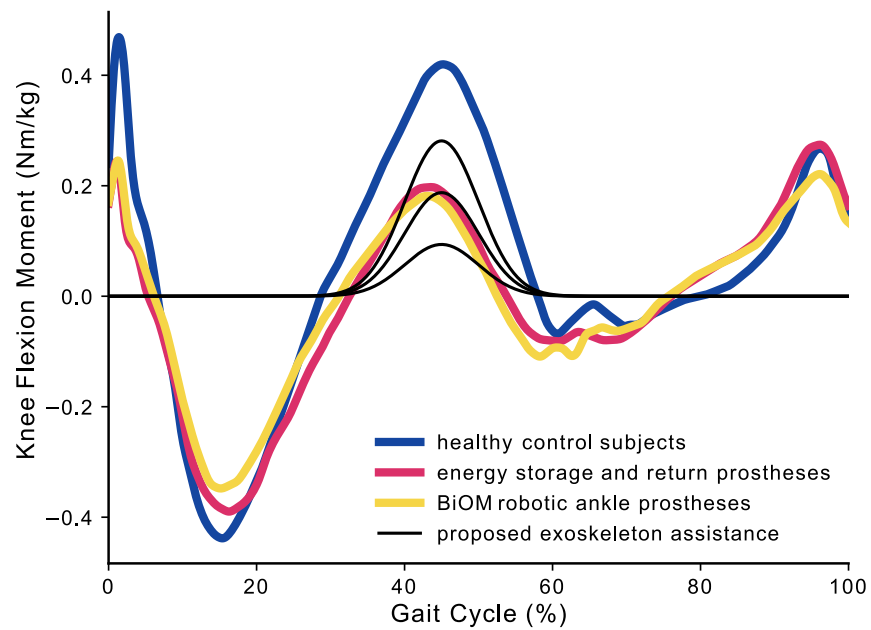


Figure 4.1: Knee flexion moments for healthy control subjects and people with limb loss walking on passive and powered prosthetic ankles, digitized from [1]. Flexion moments are notably lower on the residual limb as compared to those from the control subjects, regardless of whether the prosthesis is active or passive. Proposed knee flexion exoskeleton assistance strategies are overlaid in black.

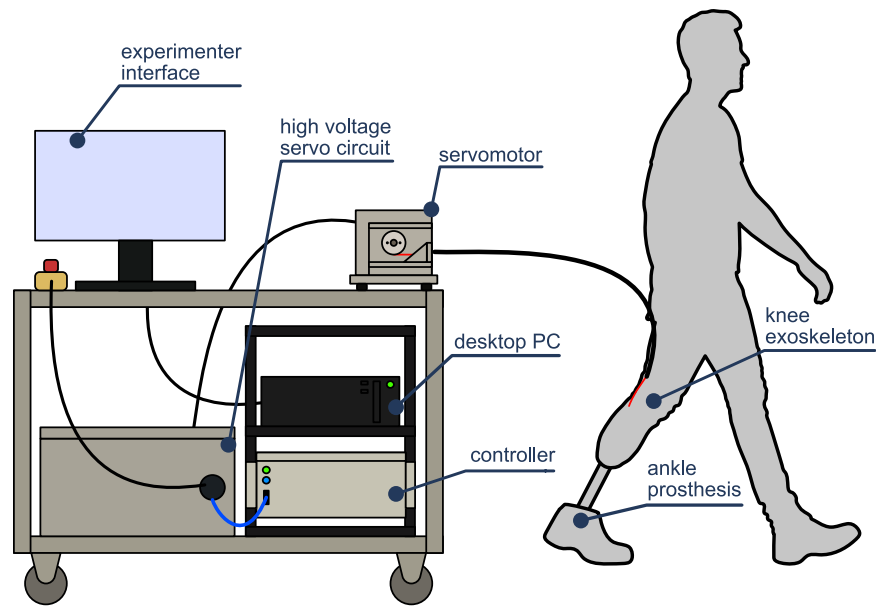


Figure 4.2: Schematic of the experimental setup. The participant walks on a treadmill as a stationary motor delivers torque to the knee exoskeleton through a flexible Bowden cable tether. The cart also contains a high-speed computer for taking sensor measurements and sending motor commands. The experimenter interface is a desktop PC that streams data to and from the controller.

morgen, AKM74L), an industrial servo drive (Kollmorgen, AKD-P02406), and a custom transmission to transfer motor torque to Bowden cable force. The transmission uses a shaft supported by bearings and pulley to pull on the inner rope of the Bowden cable. The outer Bowden cable sheath is fixed rigidly to the motor chassis. The actuators and open design parameters of COBRA were chosen based on a modeling and optimization framework to minimize the reflected inertia of the system [134].

A high-speed real-time controller (NI, PXIe-8880) measures signals from sensors on the wearable device, computes controller outputs, and sends digital motor commands to the servo drive through an etherCAT connection. The controller communicates with a desktop PC that allows the researcher to change the device behavior, observe sensor measurements

from the knee exoskeleton, and log experimental data.

4.3.2 Knee Exoskeleton Prototype

The exoskeleton frame is made of a thigh section and a shank section, each of which is composed of flat carbon fiber struts on the medial and lateral sides of the leg (Figure 4.3). For both sections, the medial and lateral struts are rigidly connected to each other through a hollow carbon fiber crossbar on the anterior side of the leg. The thigh and calf sections are connected through simple rotary joints on both sides of the knee that are comprised of ball bearings and machined 7075 aluminum components. Thin strips of foam padding line the inside faces of the carbon fiber struts for comfort and to prevent scuffing on the user's prosthetic socket.

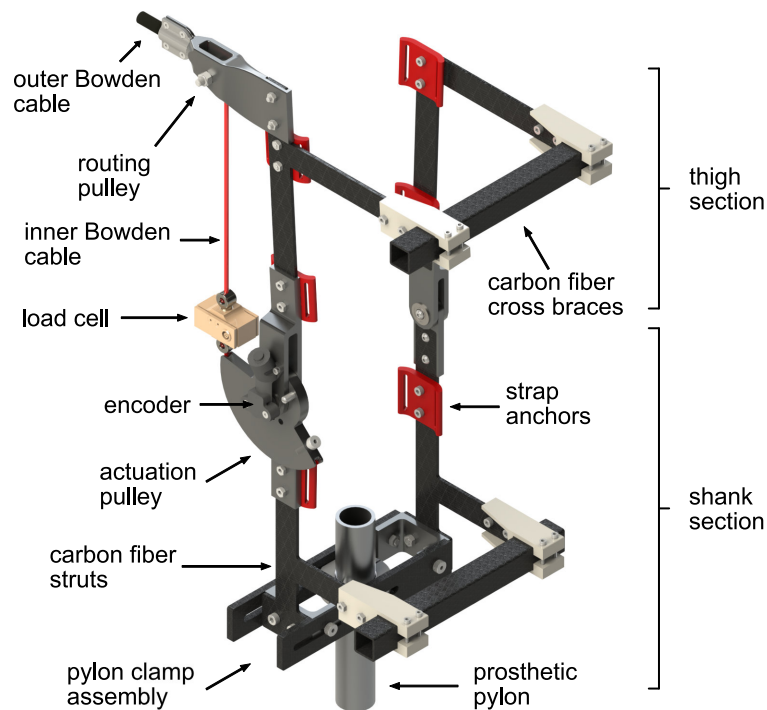


Figure 4.3: Rendering of the knee exoskeleton with labels for the primary components described in the text.

The exoskeleton frame delivers forces to the thigh through two sets of straps that weave through thin sheets of foam-backed plastic to distribute load. The straps are simple strips of 1" wide hook-and-loop and connect to the carbon fiber struts through 3D-printed anchor points. The frame connects to the shank section with a similar strap directly below the knee joint that is tightened against the prosthetic socket. Instead of a strap at the distal portion of the shank, the exoskeleton is fixed directly to the prosthetic pylon with carbon fiber crossbars and a 30 mm shaft collar clamp.

The Bowden cable actuator enters the exoskeleton from the posterior direction and is clamped to the lateral carbon fiber strut on the thigh. A pulley routes the inner cable down the side of the thigh to an anchor point on the shank section of the device. The anchor point is built into the lateral section of the rotary joint and constrains the rope to wrap around a constant 0.055 m radius that is machined into the joint structure. We chose to route only the inner cable down the thigh to keep the outer Bowden cable sheath as straight as possible between the motor and the device. Bowden cable friction increases greatly with the total bend angle of the cable, and high levels of friction can degrade controller and actuator performance.

An encoder (RLS, RM08A) on the lateral knee joint measures exoskeleton rotation, and a load cell attached in line with the Bowden cable (Futek, LRF350) measures tension in the rope at the anchor point. The exoskeleton has hard-stops at 5 degrees of hyperextension and 90 degrees of flexion, allowing for a total range of motion of 95 degrees. A smaller rope selected specifically for its tensile strength couples the Bowden cable to the load cell and acts as a safety breakaway to prevent excessively high cable forces.

The exoskeleton fits a variety of leg sizes, as we have multiple distinct sizing options and adjustability within each size. We accommodate different leg lengths by swapping different lengths of carbon fiber struts and can adjust for leg diameter by increasing the length of the carbon fiber cross bars. Mass optimization was a focus of the design, as adding mass to the

Table 4.1: Mass Distribution of Wearable Exoskeleton Components

Component	Mass (g)	(%)
Structural Carbon Fiber	354	25
Metal Joints	152	11
Bowden Cable Inlet & Pulley	218	16
Sensors & Mounts	151	11
Straps & Strap Anchors	216	16
Pylon Clamp Assembly	143	10
Fasteners	160	11
Total:	1394	100

lower limbs of prosthesis users can reduce their metabolic walking economy [39]. Table 4.1 shows a breakdown of the exoskeleton mass.

4.3.3 Torque Control System

The knee exoskeleton operates in joint torque control mode. The primary functions of the control system are to detect heel strikes, estimate gait phase, compute torque setpoints, and send motor velocity commands. All controller functionality described here was implemented in a custom NI LabVIEW 2018 program.

Vertical ground reaction forces measured by an instrumented treadmill are used to detect heel strikes in real-time. The ground reaction force signal is smoothed with a digital Butterworth filter, and the force rate of change is computed with a finite difference derivative. When the force rate of change exceeds a preset threshold and the magnitude of the ground reaction force is below a separate threshold, heel-strike is identified. Pilot tests indicated

that this method more quickly identified heel strikes compared to a single threshold on the ground reaction force magnitude.

Heel-strike data is used to estimate gait phase. In this context, gait phase is defined as a monotonically increasing value that represents the percentage of time that has passed between successive heel strikes, with both zero and 100 indicating a heel strike. The exoskeleton uses gait phase as a real-time domain for torque setpoint signals. Gait phase is computed using a timer, where the instantaneous duration since the previous heel-strike is divided by the total duration of the previous five strides.

The desired knee flexion moment profile is defined using the equation for a Gaussian curve, similar to methods in [144]. The following equation is used to generate the torque setpoint in real-time:

$$\tau = a \cdot \exp(-(\phi - b)^2/2c^2) \quad (4.1)$$

In (4.1), τ is the torque setpoint and ϕ is the real-time phase estimate. Parameters a , b , and c scale the peak, midpoint, and width of the Gaussian curve, respectively. These parameters can be changed by the experimenter to explore various knee flexion moment profile shapes during walking. Figure 4.1 shows the effect of increasing a while keeping b and c constant.

The tracking controller uses a cascaded control architecture to minimize error between the torque setpoint and the applied exoskeleton torque. An outer torque control loop computes motor velocity setpoint commands that are sent to an inner high-bandwidth motor velocity tracking loop implemented on the industrial servo-drive. This method has been shown to be successful in Bowden cable driven exoskeletons [30].

The torque tracking loop uses a combination of feedback and feedforward control. For feedback, the controller uses a proportional term on measured torque error and a damping injection term on measured motor velocity. The feedforward controller uses iterative learning methods to overcome systematic tracking error due to the nonlinear effects of Bowden cable

friction. Specifically, the torque tracking controller is an implementation of the “PD+LRN” described in detail in [145]. The command sent to the servo-drive at each loop iteration is defined as:

$$\omega_c = K_p \cdot e_\tau - K_d \cdot \omega_m + \omega_{L,n}(\phi - D) \quad (4.2)$$

In (4.2), ω_c is the reference velocity command sent to the servo-drive. K_p and K_d are positive scalar controller gains, and e_τ is the instantaneous torque error. Torque error is the difference between the setpoint torque and the torque estimated from the load cell and known exoskeleton geometry. ω_m is the motor velocity as measured by a digital encoder. $\omega_{L,n}$ is the iteratively learned feedforward motor velocity signal for the current stride, n , gait phase, and a parameter, D , that shifts the signal in time to account for dynamic delays between changes in motor velocity and changes in applied torque. The feedforward signal is updated at each heel strike with the following equation:

$$\omega_{L,n}(\phi) = K_L \cdot e_{\tau,n-1}(\phi) + \beta \cdot \omega_{L,n-1}(\phi) \quad (4.3)$$

In (4.3), K_L is a learning gain that scales error from the previous stride and β is a “forgetting” term that is less than one and is used to prevent divergence of the signal. The effect of (3) is that the torque errors and the feedforward signal from the previous stride are scaled and combined to produce the feedforward signal for the current stride. During operation, the iterative learning controller can be activated and deactivated by the experimenter through the user interface.

4.4 Experimental Methods

The mechanical and control system performance of the knee exoskeleton system were evaluated with one subject with transtibial amputation in an IRB-approved research study at CLiMB within the Veterans Affairs Puget Sound Health Care System. The purpose of this experiment was to perform an initial evaluation of the prototype during walking. The partic-

ipant was a traumatic unilateral transtibial amputee with no other neurological or orthopedic comorbidities that prevented them from walking.

After an informed-consent process, a certified prosthetist fit the participant with a new ankle prosthesis (Ossur, LP Variflex Foot). This was done because the participant’s prosthesis had a built-in torsional adapter that would have interfered with the exoskeleton’s pylon attachment. After fitting, the participant was given five minutes to walk at a self-selected pace and adapt to the new ankle. The participant’s body mass, leg length (distance from greater trochanter to floor), and height were then recorded as 82.1 kg, 0.87 m, and 1.65 m, respectively. The prosthetist then fit the exoskeleton to the participant. During the fitting process, the prosthetist aligned the rotation axes of the exoskeleton and the knee as closely as possible. Between fit adjustments, the participant flexed and extended their knee and reported on comfort and perceived shear forces at the strap interfaces. The combined mass of the participant and exoskeleton was recorded as 83.6 kg.

For all walking trials, the participant walked on an instrumented treadmill at a non-dimensionalized Froude (Fr) speed of 0.16, where $Fr = v^2/gl$, v is walking speed, l is the measured leg length, and g is the gravitational constant of 9.81 m/s^2 . A Fr speed of 0.16 was selected as it is close to self-selected walking speed for people without amputation and allowed us to choose exoskeleton torque magnitudes based on values from the literature [1]. For this participant, the nondimensionalized walking speed corresponded with a treadmill speed of 1.17 m/s.

First, the participant walked on the treadmill for several minutes with the exoskeleton set to ‘transparent’ mode, meaning only a constant small torque setpoint was commanded to prevent excessive cable slack. Ground reaction force data was collected and visualized to set the initial timing parameters for the exoskeleton assistance. Initial parameters were set such that the peak of the assistance profile occurred at 45% of the gait cycle, turned off completely before swing phase, and had a magnitude of 10 Nm. The participant then walked with the

exoskeleton providing a knee flexion moment with these settings for three minutes. At the end of three minutes, the participant was allowed to request changes to the exoskeleton assistance timing parameters (b and c from (4.1)) to be more comfortable. Parameter selection was achieved through a conversational tuning process between the experimenter, participant, and prosthetist. Once the timing parameters were established, they were held constant throughout the rest of the experiment.

A parameter sweep of the assistance magnitude, a , was performed. The tested magnitudes included 0 Nm/kg through 0.375 Nm/kg in 0.09375 Nm/kg increments, resulting in five total conditions. Body mass while wearing the exoskeleton was used to scale the assistance magnitude. This range was selected as 0.375 Nm/kg is near the peak knee flexion moment for healthy controls walking at a similar speed (Figure 4.1). The participant walked at each setting for roughly three minutes, and two 10-second data collections occurred at each setting. Figure 4.4 shows the participant walking in the exoskeleton.

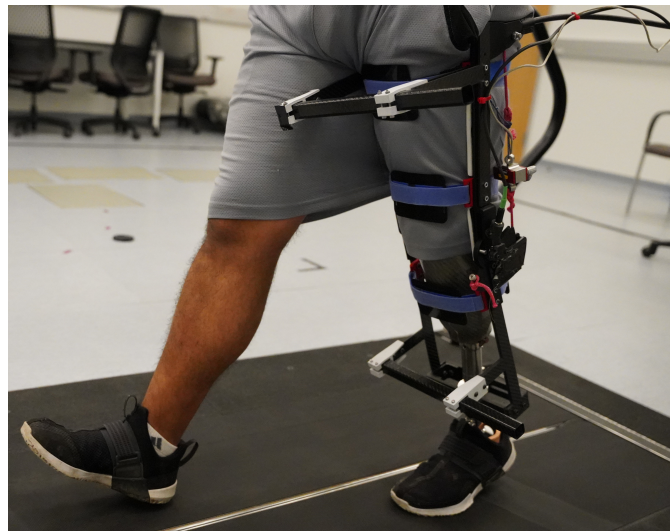


Figure 4.4: Photo of the participant walking with the robotic exoskeleton.

During data analysis, torque tracking errors were evaluated for each condition by con-

catenating the two 10-second data collections and computing the RMSE between torque command and measurement across the entire gait cycle. Gait cycles were then segmented across all conditions using heel-strike data and were normalized in time. Mean and standard deviation torque signals were computed by averaging all normalized strides within each condition. Stance phase peak torque was identified from the average torque signal for each condition. Exoskeleton range of motion was computed using the encoder signal for each stride and averaged within conditions.

4.5 Results

The prosthetist successfully fit the exoskeleton to the participant. Slight lateral bending of the struts allowed for the device to conform to the participant’s leg geometry. Clamping the device directly to the prosthetic pylon prevented the device from migrating down the leg during walking, which is a common problem for knee braces and exoskeletons. Measurements of the participant with and without the exoskeleton indicated that the device mass was 1.5 kg. This is slightly more than we found through measurements of the device (Table 4.1) due to the added mass of the Bowden cable actuator.

The final timing parameters selected by the participant, b and c , were 0.44 and 0.07, respectively. The participant completed the first four walking conditions but could not coordinate their gait with the exoskeleton at the highest torque setting (31.35 Nm peak), so data was not collected at this setting. The participant was unable to maintain a steady gait and stated that the treadmill speed would need to be increased for them to walk with the highest setting. It is possible that the participant could have learned to walk with the highest setting with additional time or training, or the assistance level may have simply been too high for the controlled walking speed.

The exoskeleton hardware and software performed as expected during the walking experiment. Figure 4.5 shows several of the signals measured from the participant and device

Table 4.2: Exoskeleton Performance Metrics Across Conditions

Assistance Magnitude, a, (Nm)	RMS Tracking Error (Nm)	Peak Torque (SD) (Nm)	Range of Motion (SD) (deg)
0	1.43	3.02 (0.46)	57.2 (1.4)
7.84	1.67	8.37 (0.34)	56.16 (0.33)
15.68	2.61	16.34 (0.46)	57.87 (0.85)
23.51	1.88	21.44 (0.68)	59.06 (0.95)

during a representative 10-second trial. Table 4.2 shows how torque tracking error, peak torque, and exoskeleton range of motion vary with assistance magnitude. The peak torque during stance increased with the assistance magnitude setpoint as expected. Variability in exoskeleton range of motion within and across conditions was under 3 degrees, which is 5% of the average range of motion across all trials.

Torque tracking errors were generally low, except for during swing phase. During late swing, the knee extends rapidly to position the foot for the upcoming heel strike. At this instant, the motor must unspool the Bowden cable quickly avoid applying a flexion moment. Though the motor unspools rapidly (roughly 800 degrees/second), friction in the cable housing causes some force to develop in the cable. However, we consider this tracking error to be of relatively little concern, as the biological knee also exhibits a knee flexion moment in late swing to decelerate limb. Figure 4.6 shows the tracking performance of the exoskeleton over the average stride for the four conditions.

The participant’s feedback on the exoskeleton was generally positive. They stated that

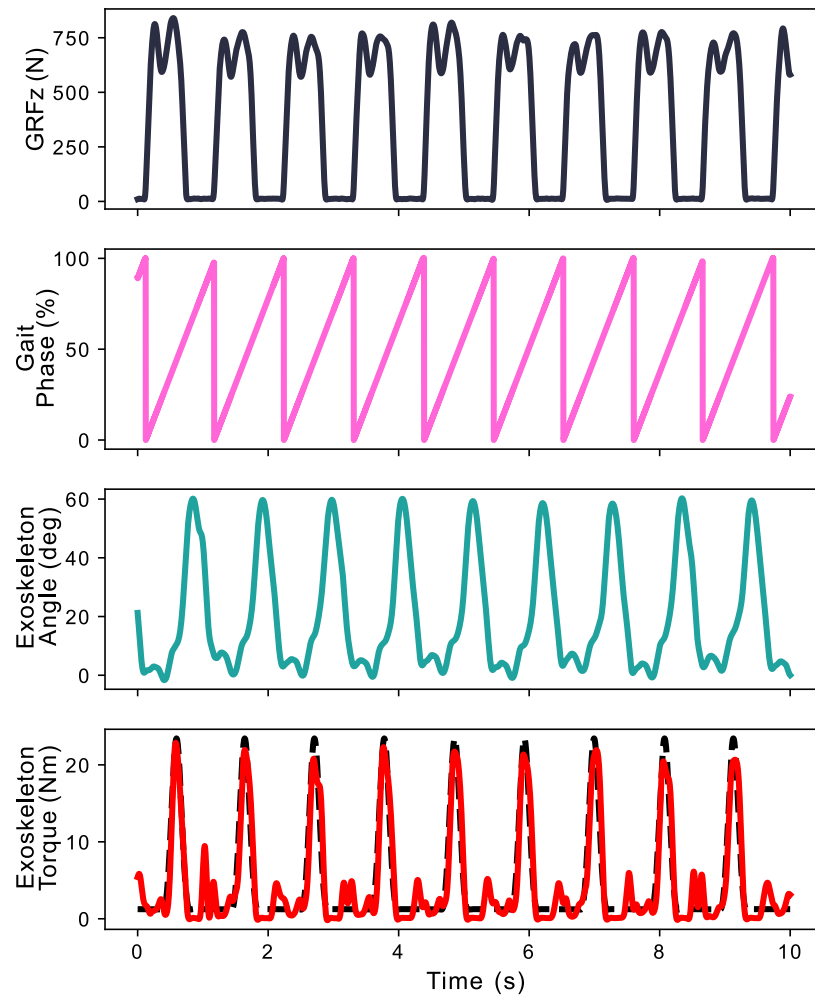


Figure 4.5: Plots of relevant real-time controller signals for a representative trial. Vertical ground reaction forces are measured as an analog input from the split-belt instrumented treadmill. Gait phase is estimated in real-time from ground reaction force data. The knee exoskeleton encoder measures rotation of the biological knee joint as well as additional rotation due to soft tissue compression during actuation. In the exoskeleton torque plot, the blacked dashed line is the setpoint and the red line is the measured exoskeleton torque.

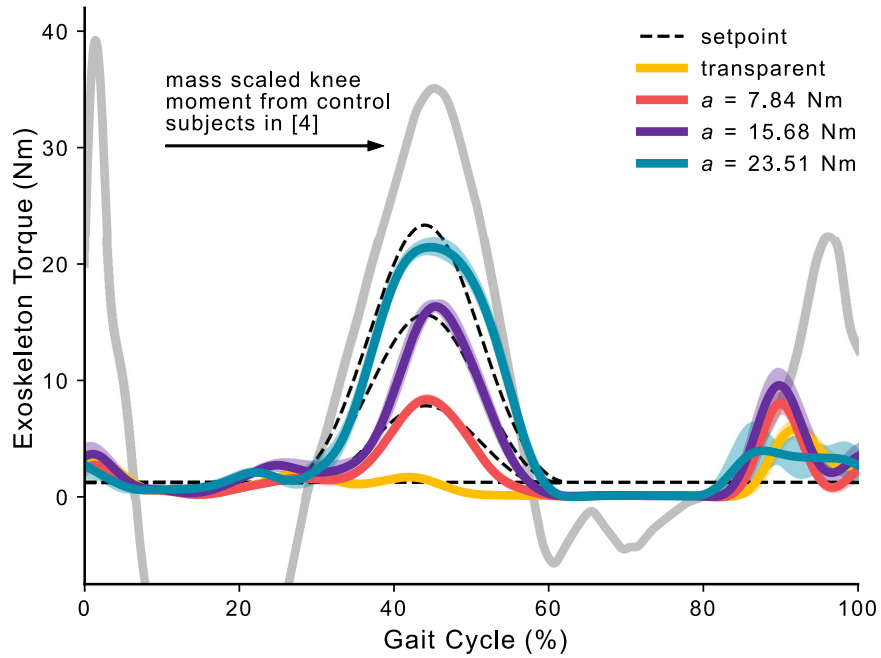


Figure 4.6: Torque tracking performance for the knee exoskeleton. Signals show mean strides for each condition $\pm$ one standard deviation. The average biological knee moment from control subjects in [1] is overlaid in grey to give context to the timing and magnitude of the exoskeleton torque profiles.

the device “made their leg feel more stable” and gave them a “more powerful stride.” They also noted that the device altered the way forces were transmitted to their residual limb through the prosthetic socket but had no feedback on whether the change was positive or negative.

4.6 Discussion

In this paper, we presented the design and control of a knee exoskeleton for transtibial amputees and a preliminary experiment with one participant to evaluate the robotic performance of the device. We found that the exoskeleton could systematically vary the assistance

magnitude between nearly-transparent and 65% of the peak biological knee flexion moment with low tracking error. For this preliminary system integration work, we evaluated a torque assistance strategy that is low-dimensional and unlikely to be optimal for transtibial amputee gait, but was relatively easy to tune and understand. Future work with this device will focus on the development of a more flexible and robust assistance strategy and clinical experiments with more subjects.

We chose to develop an exoskeleton with offboard actuation and control hardware, which had several major benefits. Our exoskeleton is lightweight and can produce a broad range of torque profiles. If necessary, future design changes can be made to the exoskeleton end-effector independently of the offboard actuation system. This platform will allow us to rapidly iterate and experiment with knee exoskeleton hardware and assistance strategies to improve walking outcomes and set the design specifications for a future untethered exoskeleton that can be worn outside of the lab. We envision an assistive device that integrates mechanically with the prosthetic socket.

It is currently unknown how the loss of gastrocnemius actuation at the knee affects amputee gait mechanics. Though the neuromuscular system is highly redundant, simulation and experimental work has shown that dysfunction in a single muscle can significantly reduce the feasible set of endpoint forces that a limb can produce [146]. Further biomechanical and clinical experiments are needed to help us understand how assistive technology can improve amputee mobility and health, and exoskeleton testbeds like the one described here are valuable tools for this effort.

Chapter 5

A ROBOTIC EMULATOR FOR EXPLORING BIARTICULAR PROSTHESIS DESIGNS FOR PEOPLE WITH TRANSTIBIAL LIMB LOSS

Anthony J. Anderson, Kira A. Gauthier, Mathew Sunil Varre, Kimberly Nickerson,
Brittney C. Muir, and Patrick M. Aubin

5.1 Introduction

People with transtibial limb loss face two main challenges that are related to the loss of ankle and foot musculature. The first challenge is a decrease in mobility, as transtibial amputees (TTAs) have slower walking speeds, lower daily step counts, and higher rates of falling compared to the general population [44, 50, 147]. Additionally, they have asymmetric walking patterns and relatively low scores on clinical measures of functional mobility [1, 41]. The second challenge is the high prevalence of secondary musculoskeletal diseases among TTAs, likely due to cumulative asymmetric limb loading during walking. TTAs frequently experience knee and hip osteoarthritis and chronic lower back pain [2, 61]. Both of these challenges are rooted in ankle-foot dysfunction, so advances in ankle-foot prosthesis technology have the potential to alleviate these issues. However, even when using state-of-the-art robotic ankle-foot prostheses, TTAs continue to have relatively low functional mobility [1]. Their daily step counts and walking speeds remain low and asymmetric gait patterns persist [1, 17].

One potential factor that may explain the limitations of powered ankle-foot prostheses is that modern designs do not reflect the anatomical layout of the human plantarflexor muscles. The triceps surae muscles, which are the primary ankle plantarflexors, consist of

the soleus and gastrocnemius. The soleus is a uniarticular muscle that crosses the ankle joint only, while the gastrocnemius is biarticular and crosses both the ankle and the knee. Uniarticular ankle-foot prostheses, regardless of whether they are powered or passive, are able to mechanically emulate soleus function, but cannot provide the coupled knee flexion moments that are normally provided by the gastrocnemius.

It has been well-established through simulation and experimental studies that the soleus and gastrocnemius play synergistic but distinct functional roles during locomotion. Both muscles contribute to both vertical bodyweight support and forward propulsion, with the soleus being relatively more responsible for bodyweight support and the gastrocnemius being relatively more responsible for forward propulsion of the center of mass [104, 106, 111, 112, 115]. This has been demonstrated through simulation-based induced acceleration studies of normal walking, which have found that the gastrocnemius is responsible for accelerating the center of mass in the direction of walking [104, 106]. Experimental work has also supported these findings, including studies that apply external forces to the body and measure muscle responses with electromyography [111, 112], as well as delivering perturbations to muscle activation with external electrodes and measuring the induced ground reaction forces during walking [115]. These findings imply that the gastrocnemius has a unique role in providing forces that accelerate the body in the forward direction during normal walking, partly by accelerating the stance limb into swing [106, 116].

These findings are particularly relevant for TTAs, who do not have functional gastrocnemius muscles. One simulation study of TTAs walking with passive energy-storage-and-return prostheses found that the prostheses provided the vertical bodyweight support of the soleus but not the forward propulsion of the gastrocnemius [141]. Another gait analysis study of TTAs walking on passive and powered prostheses found knee flexion moments on the amputated-side limb to be significantly lower than the intact limb during midstance, regardless of prosthesis type [1]. The muscles of the lower limb must decelerate and accelerate the

body in the vertical and forward directions during each step [148]. Without the propulsive forces usually provided by the gastrocnemius, TTAs must accelerate the body in the walking direction with other muscles on the amputated side limb [142]. This asymmetric gait strategy could contribute to long term musculoskeletal issues. It is possible that a new class of assistive device for TTAs that emulates the biarticular gastrocnemius could improve their mobility and long-term musculoskeletal health.

Researchers have started exploring assistive devices that combine ankle prostheses with devices that provide knee assistance – which we will refer to here as biarticular prostheses (BPs). There have been a variety of designs, including devices that incorporate both passive and powered ankle prostheses, as well as quasi-passive and powered knee exoskeletons or exosuits [122–126, 149]. Some preliminary studies have found promising results in terms of improving metabolic walking economy and reducing hip compensations [122, 123], but no BPs are currently clinically available. Overall, BPs are in their infancy and more research is needed to assess if and how they can reliably lead to improvements in mobility for TTAs.

This work introduces the design of a robotic biarticular prosthesis with two actuated degrees of freedom. The device consists of a powered ankle prosthesis and a knee flexion exoskeleton, both of which are driven by a custom off-board actuation and control system. We developed this BP as a platform to explore the basic biomechanics of walking with biarticular assistance, as well as to identify optimal designs for future untethered biarticular prostheses. This paper describes the design and control of the biarticular prosthesis, presents benchtop tests that characterize its robotic performance, and presents results from an initial walking demonstration with one person with transtibial limb loss. The purpose of this study is to provide a detailed description of our biarticular prosthesis and demonstrate its functionality and potential for improving mobility in people with transtibial limb loss.

5.2 *Mechanical Design*

We designed and built a robotic biarticular prosthesis that couples a robotic knee exoskeleton with a powered prosthetic ankle. Both devices are driven by a custom offboard robotic actuation system with two servomotors and with Bowden cable transmissions. This section describes our design intent, the design of the offboard actuation system, and the design of the two assistive devices.

5.2.1 *Design Intent*

The design space for a biarticular prosthesis is vast – several different categories of assistive device could potentially improve walking patterns for TTAs. During walking, the gastrocnemius behaves largely isometrically, i.e., the muscle actively maintains a nearly constant length while the Achilles tendon stretches and stores energy [150], indicating that a quasi-passive spring-clutch BP could replicate much of the functionality of the gastrocnemius without requiring heavy batteries or actuators. However, powered devices may be necessary, because the gastrocnemius does deliver 3-5 J of net positive work during each stride [107,141]. With these possibilities in mind, we aimed to build a prosthesis emulator capable of behaving like a variety of device embodiments (Figure 5.1).

Future BPs could be built with a true mechanical biarticular element, i.e., a spring or actuator that crosses the knee and ankle, or from two uniarticular elements (e.g., a uniarticular ankle plantarflexor and a uniarticular knee flexor). The two uniarticular elements could operate independently, or render torques from a common virtual actuator signal. We decided to develop two independent devices so that we can emulate biarticular prostheses with virtual lever arms that can be changed in software. Biarticular muscle function is highly sensitive to lever arm ratio [117], so it was desirable to avoid building a mechanical lever arm ratio into our device. Further, decoupling the uniarticular ankle prosthesis allows for future studies of uniarticular ankle prostheses with the same hardware.

Powered Ankle, Passive Knee Eilenberg et al., 2018. Powered ankle prosthesis with quasi-passive spring-clutch knee orthosis Ziemnicki et al., 2021. Emulated powered ankle prosthesis with emulated passive biarticular knee-ankle actuator	Powered Ankle, Powered Knee Eilenberg et al., 2018. Powered ankle with powered robotic knee exoskeleton
Passive Ankle, Passive Knee Willson et al., 2021. Passive ankle prosthesis with quasi-passive spring-clutch knee orthosis	Passive Ankle, Powered Knee Anderson et al., 2022. Passive ankle prosthesis with powered knee exoskeleton

Figure 5.1: The potential design space for biarticular prostheses, partitioned by the devices’ ability to deliver net positive work at each joint over a stride. Examples of each device embodiment from the literature are shown for each combination. The robotic prosthesis introduced in this work was designed with the ability to emulate devices in all four quadrants.

Prosthesis and exoskeleton emulators have become a popular tool for wearable robotics research [29–31, 132, 133]. We developed a robotic BP emulator so that we can efficiently explore the design space on a single hardware platform (Figure 5.2). We can find out what devices are most helpful and what trade-offs they require (e.g., need for batteries, number of sensors, etc.). For example, we can directly compare emulated quasi-passive designs to emulated powered designs, find potential device designs that are helpful, and then build those devices for further experiment.

The design requirements for an emulator capable of rendering devices in all quadrants of Figure 5.1 are as follows. The BP requires an ankle-foot prosthesis capable of both emulating passive devices and providing a powered push-off. The BP also requires a knee exoskeleton with the ability to render, passive, quasi-passive, and powered knee flexion moments. For example, in a quasi-passive emulation mode, the exoskeleton might use gait event detection and a measurement of the joint angle to provide spring-like torques during stance. When emulating a robotic knee exoskeleton, the device may send a pre-defined torque signal as

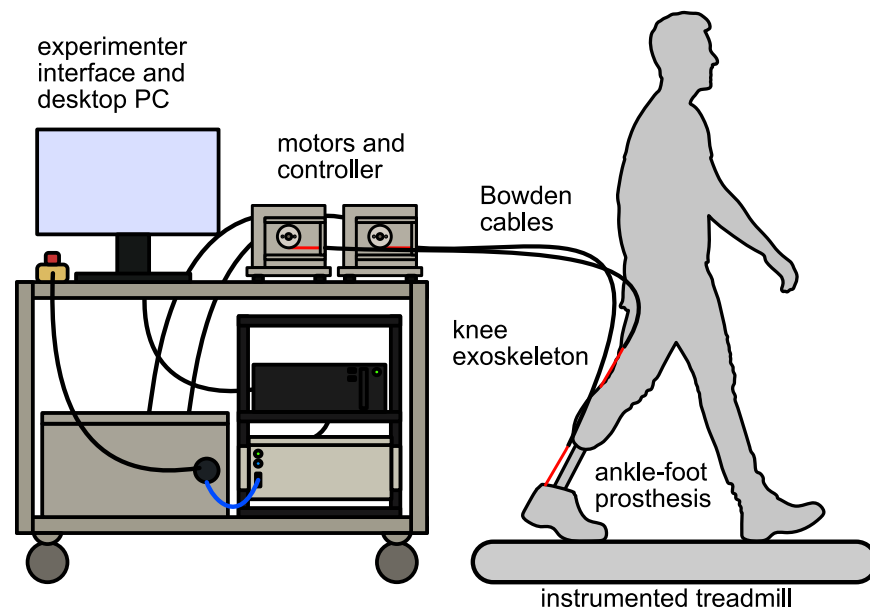


Figure 5.2: A schematic of the biarticular prosthesis emulator with primary components labeled. A research participant with transtibial limb loss walks on an instrumented treadmill while wearing the robotic ankle-foot prosthesis and exoskeleton. The stationary cart holds two servomotor actuators, a high-voltage servo circuit, a high-speed industrial controller, and a desktop computer that acts as an experimenter interface. Sensors from the prosthesis, exoskeleton, and treadmill are read by the industrial controller.

a function of gait phase, independent of the joint kinematics. These requirements indicate that the system must have sensors for measuring real-time joint angles at each joint, as well as a method of estimating gait phase. During the design process, we aimed to maximize modularity of the hardware and control software to enable a wide range of experiments and device behaviors.

5.2.2 Off-board Actuators and Transmission Design

The BP emulator uses a custom Configurable Off-Board Robotic Actuator (COBRA) and control platform to provide torques to the ankle-foot prosthesis and knee exoskeleton. The COBRA platform was previously introduced in [151] for a single-actuator configuration, and we have added a second actuator to provide independent multi-joint torques to the BP.

Each wearable device is actuated by a 5.47 kW AKM74L servomotor (Kollmorgen, Radford, VA) and a custom Bowden cable transmission. The custom transmission includes Bowden cables and hardware to convert torque from the motor to Bowden cable force. The Bowden cable transmission consists of a synthetic rope and an outer flexible sheath made of coiled steel, with a PTFE liner to reduce friction. The forces in the rope and sheath are equal and opposite at each end, allowing for pure torques to be transmitted to the wearable devices without applying a net force to the user [143]. In terms of the custom hardware, the motors are attached to steel drive shafts supported by ball bearings, with a 2.51 cm radius sheave mounted to the output of each drive shaft to pull on the inner sheath of the Bowden cable. The outer sheath is rigidly clamped to the aluminum chassis of the transmission. The continuous and peak torques of the servomotors, as listed by the manufacturer, are 49.7 Nm and 143 Nm, respectively.

The servomotors are commutated by two AKD servo drives and a custom high-voltage circuit, which is powered by a three-phase 208 V wall outlet with a peak current of 30 A. The transmissions, drive circuit, and control computers are all placed on a stationary cart

with wheels.

5.2.3 *Ankle-Foot Prosthesis Design*

The BP emulator uses a robotic ankle-foot prosthesis end-effector previously developed by our research group, the COBRA Ankle [151]. The important features of the design are recounted here for completeness.

The COBRA Ankle is a Bowden cable driven prosthesis consisting of a custom pivot joint and an off-the-shelf carbon fiber footplate. The footplate can be sized to each participant and fits inside a wide range of shoe sizes. Plantarflexion moments are provided by the Bowden cable and dorsiflexion moments are provided by a pair of parallel torsion springs that are nested within the joint. Ankle joint angle and Bowden cable force are measured in real-time with a magnetic rotary encoder and load cell, respectively. Ankle torque is estimated in real-time using the encoder angle, load cell force, and models of the lever arm geometry and spring torque-angle relationship. The ankle has a peak torque of over 175 Nm, a range of motion of 46 degrees, and a torque control bandwidth of 6.1 Hz.

5.2.4 *Knee Exoskeleton Design*

The knee exoskeleton described here is a second-generation design of the device described in [149]. The exoskeleton provides a knee flexion moment with a Bowden cable that enters the lateral posterior portion of the exoskeleton and is routed down the thigh with a pulley (Figure 5.3). The exoskeleton has rigid, planar fiberglass struts on the medial and lateral sides of the residual limb. The struts are connected with custom components machined from aluminum 7075 – including rotary joints that consists of steel shafts and ball bearings on each side of the knee. Though only the lateral side of the exoskeleton is actuated by the Bowden cable, early experiments with prototype devices indicated that exoskeleton migration was reduced and comfort was improved if a medial strut was included. Hard stops limit the

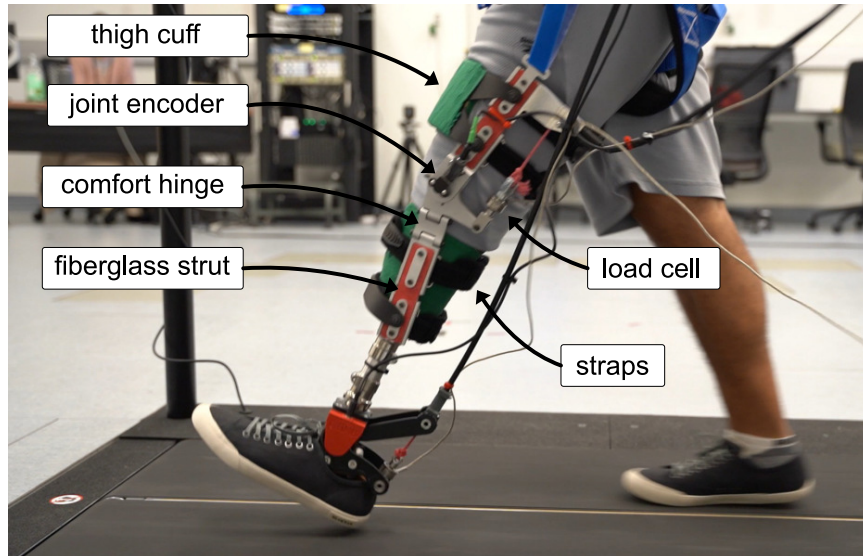


Figure 5.3: A photograph of a participant walking in the biarticular prosthesis. Key components of the exoskeleton hardware are labeled.

exoskeleton range of motion between 5 degrees of hyperextension and 90 degrees of flexion. A component with two hinges was included below the knee joint to allow for leg tapering and frontal plane misalignment between the thigh, residual femur, and prosthetic socket, based on the design described in [152].

The exoskeleton’s sagittal plane joint angle is measured with a contactless magnetic encoder on the lateral joint (RM08, Renishaw). Bowden cable force is measured with an in-line tension load-cell (FSH03887, Futek) that is connected to a pivot at the end of a lever arm on the exoskeleton joint. The effective lever arm of the Bowden cable varies with the joint angle, so a polynomial model was created that maps the joint angle measurement to the lever arm length across the range of motion. Data from the exoskeleton CAD model was used to create the polynomial model (R^2 of > 0.99). The joint torque applied by the exoskeleton is estimated in real-time by multiplying the Bowden cable force measurement with the lever arm estimate.

A rigid thigh cuff connects the two sides of the exoskeleton and delivers forces to the user's thigh. The thigh cuffs are customized to each user's leg dimensions by taking measurements of thigh circumference at two points above the knee joint. These dimensions are used to design a thigh cuff component that is laser cut from a flat sheet of thermoplastic. The thermoplastic sheet is then heat formed to either the participant's leg or the leg of a mannequin. This workflow was initially proposed in [152]. The thermoplastic cuff is lined with neoprene foam for comfort. A second heat formed cuff is used to apply forces to the anterior side of the prosthetic socket at the distal end of the exoskeleton. The exoskeleton is tightened to the user's leg with adjustable straps taken from a commercial orthotic knee brace. Downward migration of the exoskeleton is prevented by wrapping the prosthetic socket in self-adhering Coban wrap and a hook-and-loop strap that attaches the upper portion of the exoskeleton to either a belt or fall harness. The fiberglass struts lengths can also be customized to patient measurements when necessary.

5.3 Control System

Both the knee exoskeleton and ankle-foot prosthesis operate under closed-loop torque control. The controller architecture was designed to be as modular as possible to facilitate future experiments with a wide variety of torque controllers. The controllers currently operate independently from each other, though future controllers could couple the knee and ankle torque on the software level, e.g., to simulate a virtual gastrocnemius muscle. A schematic of the control system and signal flow is shown in Figure 5.4.

5.3.1 Heel Strike Identification and Phase Estimation

During operation, heel strikes are identified in real-time using the vertical ground reaction force (GRF) from the instrumented treadmill and a tunable threshold. Heel-strikes are identified when the GRF measurement is higher than the threshold and the previous measurement

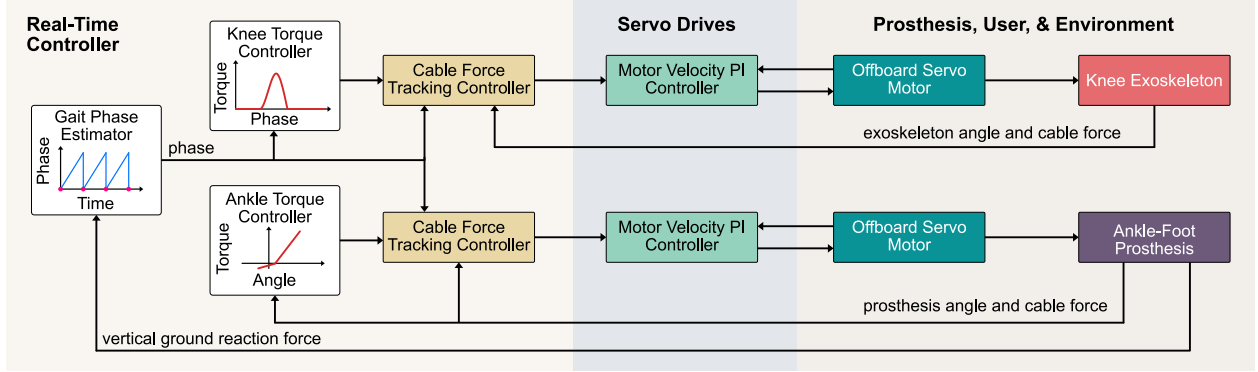


Figure 5.4: A diagram of the controller and signal flow for the biarticular prosthesis. On the real-time controller, heel strikes times are used to estimate gait phase. Desired torque signals are generated for the knee and ankle using a phase-based and angle-based controller, respectively. These desired torque signals are transformed to desired cable forces and are tracked with feedback controllers that output motor velocity commands for each motor. Bowden cable transmissions apply torques to the devices, and wearable sensor data is fed back into the real-time controller.

was below the threshold. The threshold is generally set between 10 and 20 N to avoid false positives due to noise in the GRF measurement. Heel strike identification is used to compute the stride period and estimate gait phase. Following each heel strike, a timer is started. On each loop iteration, the time since the most recent heel strike is divided by the average stride period over the previous five strides to compute the phase estimate. The phase estimate is scaled between zero and 100 over each stride. The phase estimate is used to generate torque command signals for the knee exoskeleton and is used in a cable force tracking controller for both wearable devices.

5.3.2 Ankle-Foot Prosthesis Torque Controller

For the experiments described in this paper, torque commands for the ankle were generated using a linear quasi-stiffness controller, as described in [151]. The ankle has a virtual rest angle and independent quasi-stiffness parameters for when the prosthesis angle is on each

side of the virtual rest angle, which is intended to emulate independent ‘heel’ and ‘forefoot’ stiffnesses. Both the stiffness parameters and rest angle are tunable by the experimenter to approximate the quasi-stiffness profile of the biological ankle. Previous work showed that the ankle-foot prosthesis is capable of tracking torque profiles generated by this controller with low error [151].

5.3.3 *Knee Exoskeleton Torque Controller*

The knee exoskeleton torque controller implemented for this work was a phase-based controller adapted from [23, 153]. The controller uses four parameters and a Hermite-spline technique to create a flexible torque profile that can approximate the biological knee moment during stance (Figure 5.5).

5.3.4 *Low-Level Torque Tracking Controllers*

The Bowden cable transmissions have nonlinear and time-varying friction characteristics that require an adaptive compensator to overcome [154]. A cable force tracking controller was implemented for each joint that uses proportional-damping control to reject disturbances and compensate for stride-to-stride variability, and a feedforward iterative learning controller (ILC) to adapt to systematic torque errors across strides. The cable force tracking controller outputs a motor velocity command that is sent to a high-bandwidth motor velocity controller. Details of the controller implementation can be found in [149].

The cable force tracking controller operates at 1 kHz and the motor velocity controller operates at 1.5 MHz. The nested controller architecture is beneficial as it allows the faster motor velocity controller to sense and break the static friction in the Bowden cable by increasing motor current, and therefore makes the outer cable force tracking controller easier to tune [155]. This method of Bowden cable torque control, incorporating classical feedback and iterative learning control, has found to be highly effective for torque tracking with

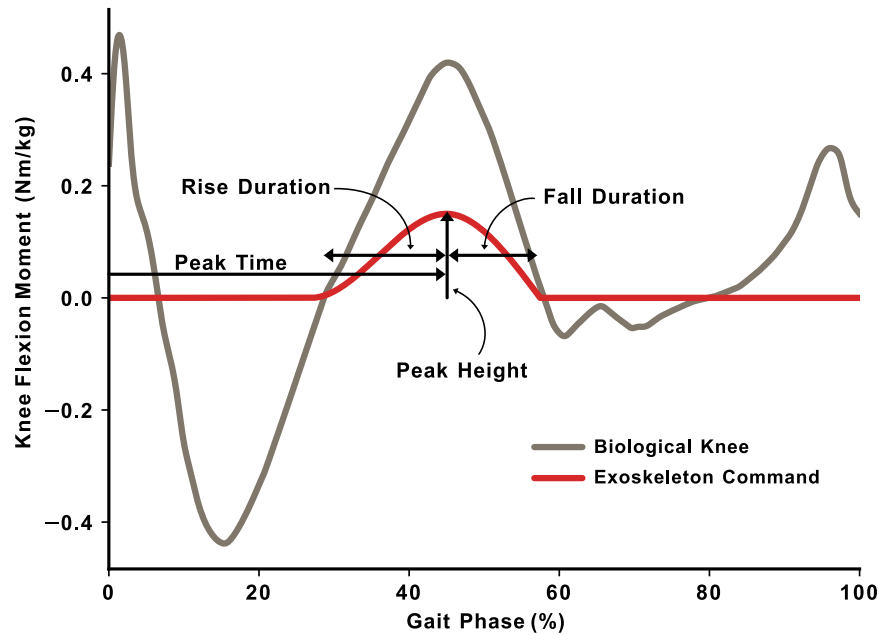


Figure 5.5: Knee torque command parameterization for the knee exoskeleton and the average biological knee flexion moment from healthy subjects, digitized from [1]. Gait phase is estimated in real-time from historical heel-strike data and is used as the domain for the commanded exoskeleton torque signal.

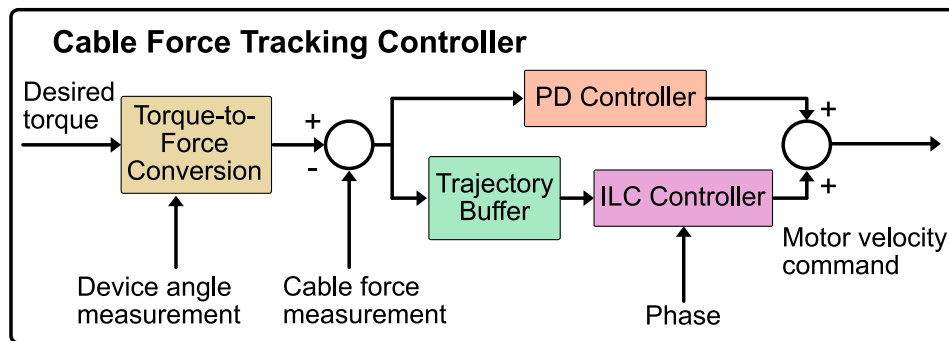


Figure 5.6: Expanded schematic of the cable force tracking controller. A desired torque is sent to the cable tracking controller from an upstream torque controller and transformed into the Bowden cable force required to achieve the requested torque. For the knee exoskeleton and ankle-foot prosthesis, the encoder measurement is used to compute the instantaneous lever arm of the device. For the ankle-foot prosthesis, the encoder measurement is also used to estimate the torque provided by the parallel spring.

wearable robots [145].

5.3.5 *Controller Hardware and Software Implementation*

The control system for the BP is distributed across several computers. High-level prosthesis settings, e.g., controller parameter and states, are controlled through a user interface via a LabVIEW program that runs on a desktop PC running Windows 10. Lossy data is streamed from the prosthesis and is displayed on the user interface for monitoring during experiments. Communication between the host PC and real-time controller is managed through an ethernet router and local area network (LAN).

The real-time controller for the BP is a PXIe-8880 (NI, Austin, TX). The real-time controller runs a LabVIEW program that reads sensor inputs, communicates with the host PC and servo drives, and logs data. Separate parallel while-loops manage host PC communication, data logging, and control. The primary control loop is hardware-timed and deterministic, while the communication and logging loops only run intermittently. The control loop on the RT-controller implements the gait phase estimation, torque setpoint generation, and cable force tracking algorithms, and passes motor velocity commands to the servo drives. The servo drives communicate with the controller via etherCAT, a real-time communication protocol for industrial devices. The servo drives are connected to the controller in a daisy-chain setup as shown in Figure 5.7.

5.4 *Methods*

We conducted benchtop experiments to characterize the parallel torque controllers of the knee exoskeleton and demonstrate the ability of the system to provide synchronous joint torques at the knee and ankle. We also collected walking data with one person with transtibial amputation to demonstrate the behavior of the BP.

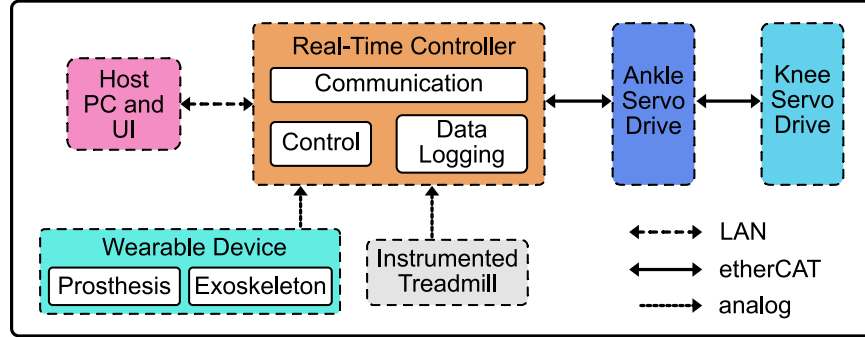


Figure 5.7: Biarticular prosthesis communication and connectivity diagram. The host PC sends control parameters to and receives telemetry from the real-time controller via a local area network (LAN) and ethernet router. Signals from the wearable sensors and instrumented treadmill are read by an analog-to-digital converter on the real-time controller. Control algorithms are implemented on the real-time controller and motor velocity setpoint commands are sent to the servo drives via an etherCAT protocol.

5.4.1 Benchtop Characterization

The purpose of the benchtop experiments in this study was to characterize the hardware, software, and controller performance of the BP emulator. To do this, the ankle-foot prosthesis and knee exoskeleton were mounted in rigid test stands with the joints locked in a neutral position. For the knee exoskeleton, only the actuator portion on the lateral side of the thigh was mounted, along with the actuator output and the frontal plane hinge. To improve the rigidity of the ankle-foot prosthesis mount, the carbon fiber footplate was removed and the top and bottom of the joint were both bolted directly to the test stand frame. Figure 5.8 shows both test stands.

The frequency response of the knee exoskeleton's torque feedback controller was evaluated using sinusoidal torque setpoints. These setpoints had a static torque offset of 20 Nm and a peak-to-peak amplitude of 10 Nm, oscillating between 15 Nm and 25 Nm. The test was performed at eight oscillation frequencies ranging from 0.1 Hz to 5 Hz, with each frequency tested three times for 30 seconds. The iterative learning controller was disabled during this

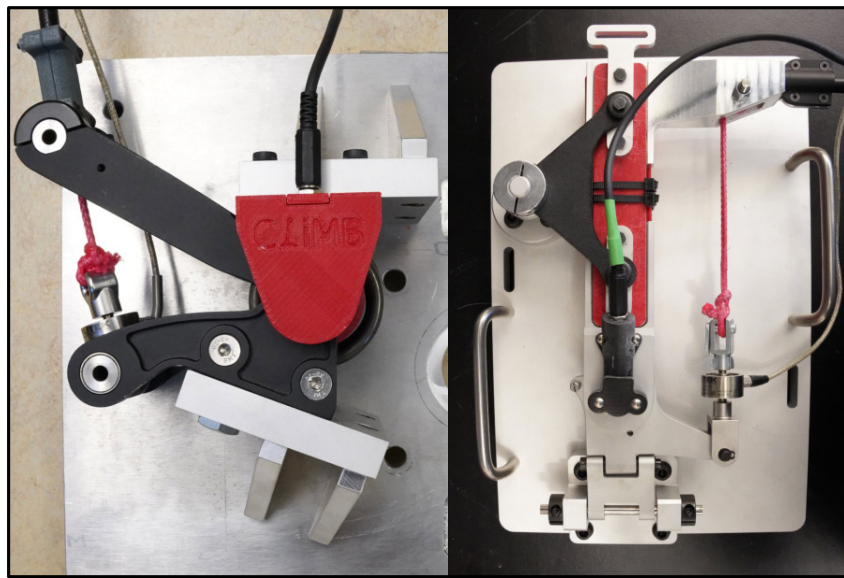


Figure 5.8: Photograph of the test stand setup for the ankle-foot prosthesis (left) and knee exoskeleton (right). The prosthetic footplate was removed from the prosthesis in order to efficiently lock the ankle joint. The lateral, actuated joint was removed from the knee exoskeleton for benchtop tests in order to simplify the design of the test stand.

characterization. To calculate the gain and phase metrics during data analysis, the output signals were first centered at zero and fit with a sinusoid curve of the form:

$$F(t) = A \cdot \sin(2\pi ft + \phi) \quad (5.1)$$

where A and ϕ were open optimization parameters and f and t were the input frequency in Hz and time samples for the experiment, respectively. Controller gain was computed by dividing A by 5 Nm, the input oscillation magnitude, and converted to decibels. Controller phase was defined as ϕ , as the input signal had zero phase offset.

The frequency response of the motor velocity tracking controller was evaluated using the method described above, but with no velocity offset, a peak velocity of 300 deg/s, and frequencies ranging from 0.1 Hz to 128 Hz. The motor's velocity controller was evaluated with the Bowden cable detached.

To characterize the influence of the iterative learning controller on exoskeleton torque, a hardware-in-the-loop experiment was performed using a spoofed ground reaction force signal to trigger the gait phase estimation algorithm and torque setpoint generation. In the first part of the experiment, the iterative learning controller was disabled to evaluate the performance of the feedback controller alone. In the second part, the iterative learning controller was turned on until tracking was significantly improved, and then clamped, meaning the learned signal was sent as a feedforward command on each step but did not continue to update. Without clamping, the output of the iterative learning controller eventually develops high frequency "ripples" [156]. The experiment recorded 25 strides in each condition, and the average output signal was computed as a function of gait phase. The root mean squared error was computed across the 25 strides in each condition.

A series of step torque inputs were commanded to both the exoskeleton and prosthesis at the same time to demonstrate the ability to provide time-synchronized torques at the knee and ankle. Step command experiments were separated into three batches. In the first

batch, the knee exoskeleton was commanded to 10 Nm and the ankle-foot prosthesis was commanded to 50 Nm. In the second batch, the exoskeleton was commanded to 20 Nm while the prosthesis was commanded to 100 Nm. In the final batch, the exoskeleton was commanded to 30 Nm and the prosthesis was commanded to 150 Nm. The step response for each setting was collected five times. Each step command started with a slight preload to remove any cable slack. In order to capture joint loading and unloading behavior, the commanded torque was held at the maximum value for one second and then returned to the preloaded value. The average step response was computed, as well as the 90% rise time, 90% fall time, and percent overshoot for each joint.

During all benchtop experiments, the PD controller gains were tuned to deliver a rapid step response with minimal overshoot and oscillation on the test stand. The controller gains appropriate for walking are too aggressive for the benchtop tests because the joint range of motion is fixed at zero. The exact ILC gains used in the iterative learning experiment are the same gains that are used during walking and were tuned by hand in pilot experiments.

5.4.2 Walking Demonstration

One person with unilateral transtibial limb loss participated in an IRB-approved experiment at the VA Puget Sound Health Care System to demonstrate the hardware and software behavior of the biarticular prosthesis in a walking environment. During a specific fitting visit, measurements were taken of the participant's thigh and socket to design custom thigh and socket cuffs for the exoskeleton.

During the walking experiment, a certified prosthetist fit the COBRA ankle-foot prosthesis to the participant's socket. Initial ankle control parameters were set based on the participant's body mass and able-bodied data from the literature. The participant walked on an instrumented treadmill with the robotic prosthesis for five minutes and the treadmill was occasionally stopped so the prosthetist could adjust the alignment of the prosthesis for

comfort and walking ability. The participant then walked on the treadmill and requested a stiffer or less stiff prosthesis based on preference, and the device parameters were changed accordingly. The participant walked at 1 m/s for the experiment.

After fitting and tuning the prosthesis, the prosthetist and experimenters fit the knee exoskeleton to the participant. Initially, the participant walked with the exoskeleton in transparent mode, where the exoskeleton attempted to apply no torque to the joint. After ten minutes of walking, three ten-second trials of data were logged. The exoskeleton was then turned on with a low peak torque and default timing settings, and the torque and timing parameters were tuned with the participant until they were comfortable with the exoskeleton behavior. The participant walked with a constant torque profile for 10 minutes and then three ten-second trials of data were logged. During data analysis, signals from the vertical GRF were used to identify heel-strikes and gait cycles were segmented, timed normalized, and averaged. Signals from the wearable devices were not filtered in post processing, as they are low-pass filtered in real-time.

5.5 Results

The benchtop experiments found that the biarticular prosthesis is capable of applying sufficient joint torques at sufficient bandwidth for future walking experiments. The walking experiment showed a proof-of-concept where the prosthesis provided simultaneous knee and ankle assistance to a participant with transtibial limb loss in two modes – the knee exoskeleton behaving transparently and the knee exoskeleton providing an active torque profile.

5.5.1 Benchtop Experiments

The exoskeleton torque controller and motor velocity controller's -3 dB bandwidths were approximately 5 Hz and 100 Hz, respectively. Figure 5.9 shows the Bode plot for both controllers across the tested frequencies.

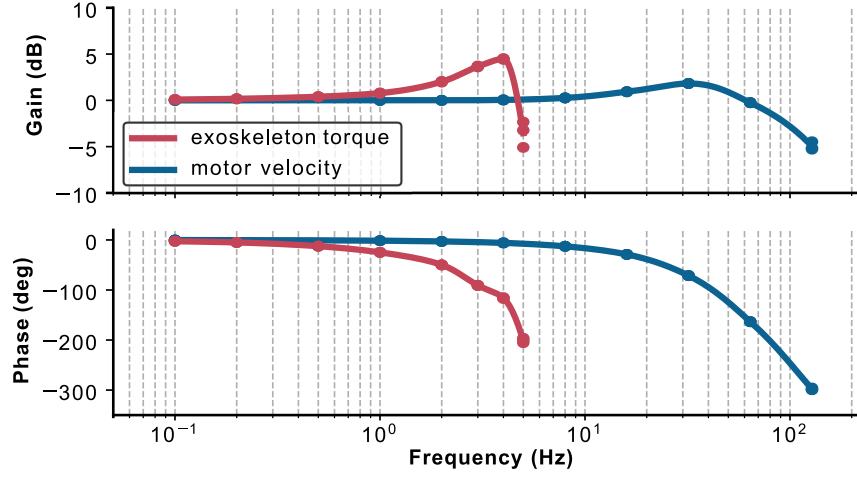


Figure 5.9: Bode plot for exoskeleton torque and motor velocity tracking controllers. The exoskeleton torque controller shows a resonant peak near 4 Hz. Points show individually tested frequencies, and a spline connects the points for visualization.

Both controllers show characteristics of a second-order dynamical system. The exoskeleton controller shows a resonant peak near 4 Hz, which is dependent on the controller gains used in the benchtop experiments. The motor velocity shows a resonant peak at 33 Hz. The controller gains for the motor velocity controller are kept constant between the benchtop and walking environments.

Two of the individual frequencies are shown in a time series plot in Figure 5.10 to show the effects of the Bowden cable nonlinearities on the cable force tracking performance. Errors are most prevalent when the commanded torque signal changes directions, e.g., from loading to unloading. This is a well-known phenomenon in Bowden cable transmissions [157] and can be compensated for with iterative learning control.

Inclusion of the iterative learning controller significantly improved tracking performance in the benchtop test. Figure 5.11 shows the mean torque signals over 25 strides with and without the iterative learning controller turned on. The root-mean-squared torque tracking error for the average stride was reduced from $6.04 \text{ Nm} \pm 0.16 \text{ Nm}$ to $0.99 \text{ Nm} \pm 0.12 \text{ Nm}$ –

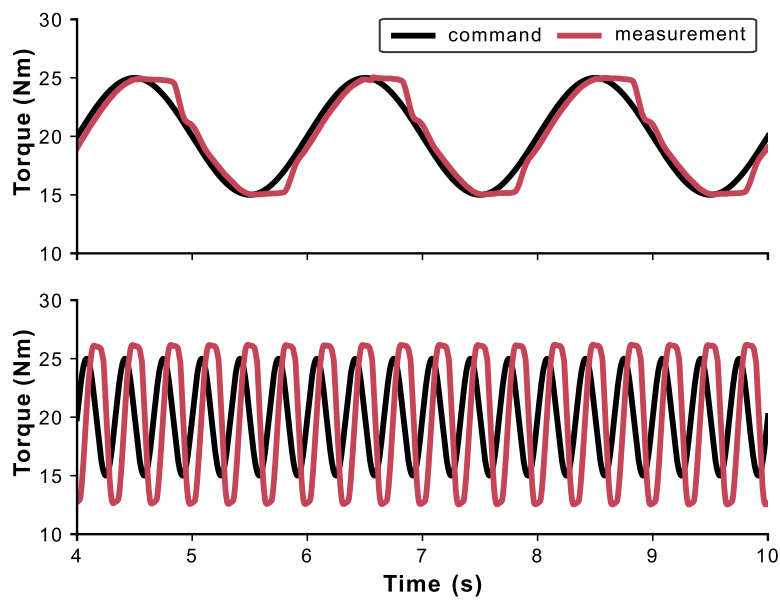


Figure 5.10: Torque tracking performance with two of the sinusoidal torque commands at 0.5 Hz (top) and 3 Hz (bottom). Nonlinearities due to Bowden cable friction cause higher tracking errors when the loading rate changes directions.

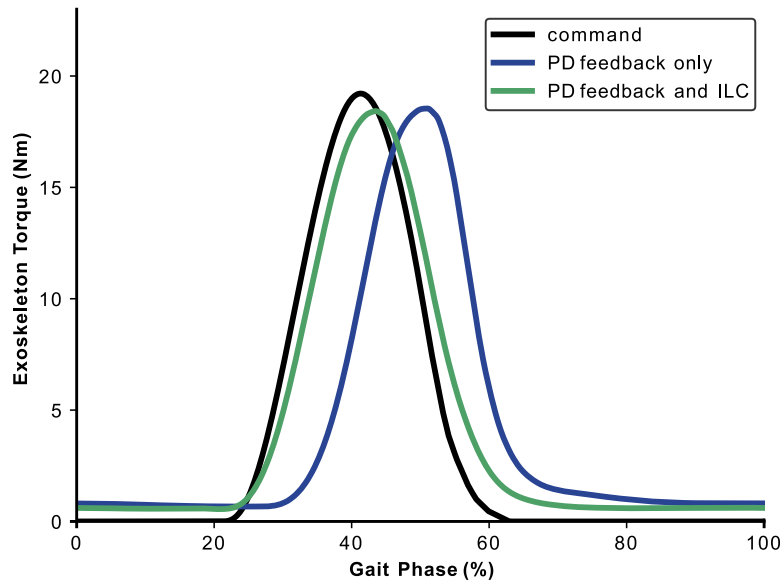


Figure 5.11: Influence of the iterative learning controller as shown in the benchtop experiment. The torque command is shown in black, the proportional-damping controller (PD) in blue, and the proportional-damping controller with iterative learning control (ILC) in green. The significant phase lag in the PD controller is largely addressed by the feedforward nature of the ILC algorithm.

an 84% reduction in tracking error.

The results of the time-synchronized step response tests for the knee exoskeleton and ankle prosthesis are shown in Figure 5.12. The step responses were highly repeatable and showed low variability, as can be seen in Table 5.1. One consistent feature in the step responses for both devices is an approximately 50 ms delay between the onset of the torque command and the onset of the torque development. This delay is not observed in the motor motion following the torque command, so the delay in torque production is likely due to static friction in the Bowden cable transmission.

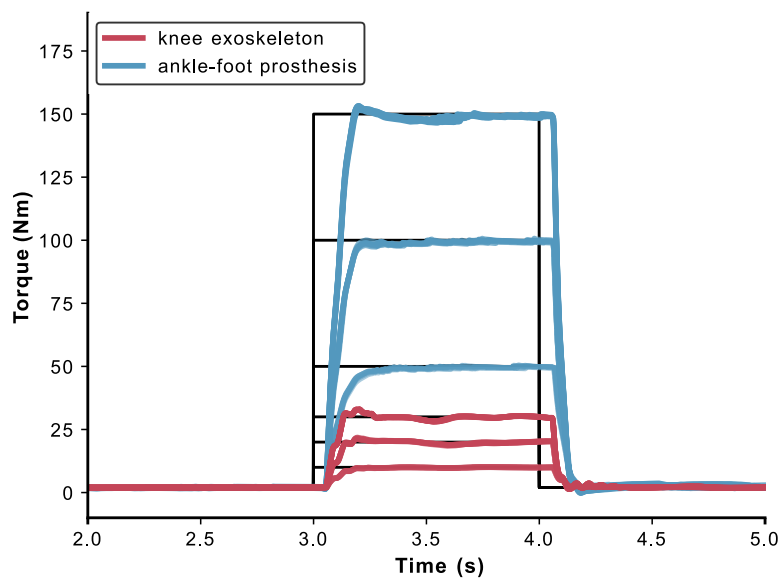


Figure 5.12: Coordinated step responses between the ankle-foot prosthesis and knee exoskeleton. Step commands were conducted in three batches, corresponding to low, medium, and high magnitudes for each device. For a given batch, torque steps were commanded to the knee and ankle simultaneously. All five step responses are shown for each batch at both joints.

Table 5.1: Step Response Metrics

Setting	Low	Medium	High
Ankle Magnitude (Nm)	50	100	150
Knee Magnitude (Nm)	10	20	30
Ankle Rise Time (ms)	194 (8)	169 (0)	154 (2)
Knee Rise Time (ms)	178 (0)	131 (0)	125 (0)
Ankle Overshoot (%)	0.66 (0.13)	0.49 (0.16)	1.80 (0.38)
Knee Overshoot (%)	1.00 (0.06)	6.88 (1.09)	9.82 (0.28)

*Values in parentheses are standard deviations.

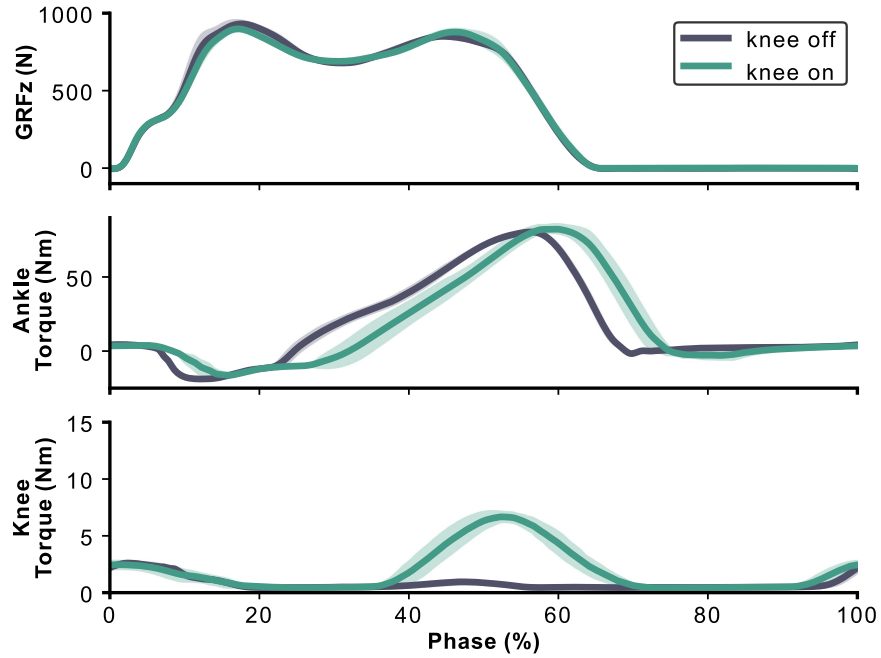


Figure 5.13: Average measurements of the biarticular prosthesis signals during walking. The ground reaction force is shown on the top row, the prosthesis torques in the middle row, and the exoskeleton torque on the bottom row.

5.5.2 Walking Results

Torques from the biarticular prosthesis are shown with the vertical ground reaction force in Figure 5.13. The virtual heel and forefoot stiffnesses selected by the participant were 9.56 Nm/deg and 3.10 Nm/deg, respectively. The peak knee exoskeleton torque during the transparent mode was 2.60 ± 0.23 Nm, while the peak knee exoskeleton torque during the active mode was 6.67 ± 0.45 Nm. This peak value is approximately 20% of the peak biological knee flexion moment for an able-bodied 80 kg subject walking at approximately 1.25 m/s [1]. The peak exoskeleton torque timing chosen by the participant occurred at 53% of the gait cycle. This is later than the biological knee flexion moment, which usually occurs around 45% of the gait cycle.

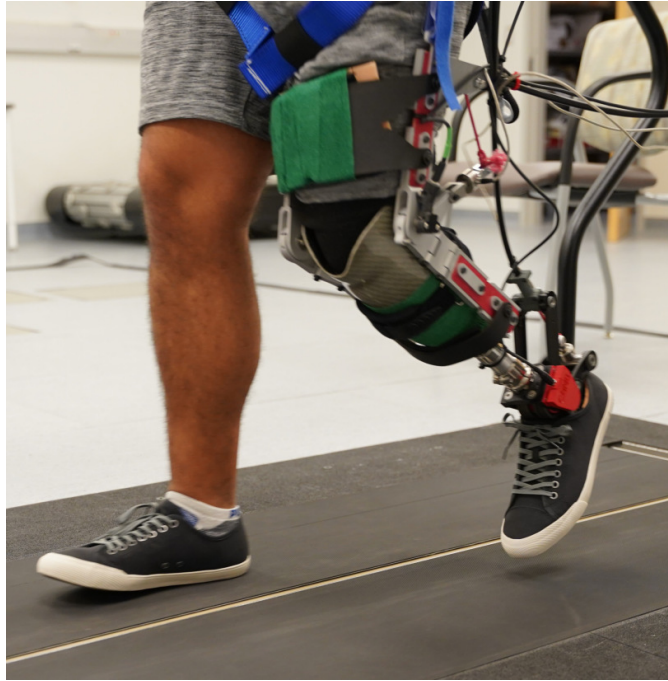


Figure 5.14: The participant walking with the robotic biarticular prosthesis. As the participant moves into swing phase, the knee exoskeleton cable is nearly slacked, allowing for low joint impedance during swing.

The participant gave positive feedback about the performance of the knee exoskeleton and preferred the behavior of the BP when the exoskeleton was active. Specifically, the participant claimed that their knee felt more stable with the assistance and that forces from the device provided them with additional proprioceptive information about the state of their limb. Figure 5.14 shows the participant walking in the BP.

5.6 Discussion

This work presented the design, control, and characterization of a robotic biarticular prosthesis emulator for people with transtibial limb loss. We detailed our design choices, including the rationale for developing an emulator with two independently actuated joints, and the

torque control system for the device. We also presented benchtop data that characterized the actuators and control systems of the BP emulator and provided data from a simple walking demonstration with one participant.

Design choices play a critical role in the success of any lower-limb assistive device. Our BP design incorporates both standard and customizable components to allow for a high level of flexibility in sizing and fit. The strut lengths and thermoformed cuff geometries can be tailored to each user, while adjustable straps provide additional fine-tuning of the fit. Additionally, our simple and lightweight end-effectors can be easily reconfigured or even redesigned if necessary, as they are attached to an offboard actuator rather than being integrated into an autonomous, untethered prototype. This approach allows us to focus on the design and functionality of the BP through rapid experiments rather than the more complex challenges of building an untethered device. Overall, these design choices contribute to the versatility of our BP emulator.

The results of the benchtop experiments and walking trial demonstrate the feasibility of the BP emulator as a tool for studying new classes of assistive devices that incorporate gastrocnemius-like assistance for people with transtibial amputation. The benchtop experiments showed that the BP has sufficient torque and bandwidth for future walking experiments but requires the iterative learning controller to compensate for nonlinear Bowden cable friction, similar to other Bowden cable-based devices [30,123]. The results of the walking trial showed that the BP was capable of providing simultaneous knee and ankle assistance to the participant during walking. The participant reported feeling more stable and having additional proprioceptive information about their limb while using the BP, suggesting that the device could be helpful in improving walking performance and stability. However, the walking results in this paper are only a preliminary demonstration of the device’s capabilities rather than an exploration of how a BP might affect the gait of people with limb loss. Further studies will be needed to fully understand the impact of the BP on walking performance

and stability.

Using an emulator allows for efficient exploration of the design space of BPs. This is because the emulator can be programmed to test a wide range of designs and behaviors, including fully robotic devices, fully passive devices, and quasi-passive devices at each joint. By testing these various designs using an emulator, we can more easily identify the most effective design and control strategies for an untethered BP, rather than spending time and resources building and testing a variety of untethered BPs. Our prior work has shown that the COBRA ankle-foot prosthesis can deliver ankle push-off power similar to other powered prostheses [151], indicating that the BP emulator is capable of emulating devices in all quadrants of the design matrix in Figure 5.1. For example, future work may emulate quasi-passive BP devices similar to those in [122,124] using our device’s joint encoder signals within an impedance controller and virtual clutch that changes its behavior based on gait events. Overall, using a BP emulator allows for a cost- and time-effective and efficient way to develop and test biarticular prostheses.

Other research groups have also developed BPs to provide knee flexion assistance to TTAs. For example, our research group previously developed an autonomous BP that combined a knee orthosis with a passive ankle prosthesis [124,125]. The device used a clutched biarticular spring to provide gastrocnemius-like torque to the ankle and knee during stance and minimal torque during swing. However, the device was not optimized for comfort, and it was difficult to make changes to the device behavior without substantial hardware redesigns. Another research group at MIT developed two versions of a BP – the first was quasi-passive and also used a clutch, but with an autonomous powered ankle prosthesis [122]. The second device used a powered knee exoskeleton with the same ankle prosthesis [123]. Both devices showed promising results with preliminary walking experiments, including substantial decreases in metabolic cost in some participants. Ziemnicki et al. developed a BP using two Humotech offboard actuators – one ‘artificial soleus’ and one ‘artificial gastrocnemius’, the latter of

which physically crosses both the knee and ankle [126]. Their device is unique in that it uses soft exosuit technology instead of a knee exoskeleton, which is lightweight and comfortable, but does not have a joint encoder at the knee. The lack of real-time measurement of the knee angle limits the types of control schemes that can be effectively implemented. Overall, each of these approaches are valid ways to design BPs and the diversity of design choices helps to improve the likelihood that the field as a whole will uncover beneficial designs. As with standard uniarticular prostheses, there is no single universally optimal design, as patients have a wide variety of needs that can be best met with a variety of passive, quasi-passive, and powered devices.

One limitation of the current BP is the relatively low torque tracking bandwidth, which may impact the ability to emulate certain device embodiments. For example, emulating a clutch requires rapid changes in joint torque, which may be challenging with the current controller performance. Additionally, the low bandwidth of the classical feedback controller makes the overall torque tracking behavior of the BP more reliant on the iterative learning controller, which is prone to poor behavior in the event of false heel strikes, e.g., in the event of the user's feet crossing the instrumented treadmill belts. It is worth noting that the current study only presented data from one subject in a simple experiment, so further studies with a larger number of subjects will be necessary to understand the potential benefits and limitations of the BP emulator.

In terms of future work, engineering improvements will be made to the BP emulator to improve its torque tracking bandwidth. This could be achieved by choosing wearable sensors with lower signal-to-noise ratios, increasing the sampling frequency of the controller, and optimizing the real-time low-pass filter parameters. These changes would allow for more aggressive gains to be used in the feedback controller. Additionally, further experiments will be conducted with more participants to optimize device performance through human-in-the-loop optimization algorithms. These experiments may involve comparing the optimized

performance of uniarticular prostheses to biarticular ones and using the results to understand the effects of knee assistance. Other experiments may compare different device embodiments, for example, exploring how emulated quasi-passive devices compare to active ones. Finally, further experiments may also be conducted at various slopes and speeds to understand the contexts in which a BP may be more advantageous than a standard ankle-foot prosthesis.

BPs can potentially improve the life of people with limb loss in the short term by making walking easier and less painful and in the long term by reducing secondary musculoskeletal comorbidities like knee arthritis and lower back pain that are associated with gait asymmetries. The biarticular prosthesis emulator presented here will aid researchers in developing new design and control strategies for BPs and may help us understand some of the inherent limitations of our current uniarticular prostheses.

5.7 Conclusion

In conclusion, the development of biarticular prostheses has the potential to greatly improve the gait patterns and musculoskeletal health of individuals with limb loss. The BP emulator presented in this study is a promising tool towards this end, as it allows for the simultaneous assistance of both the knee and ankle during walking. Through the use of an offboard motor and emulator, the BP was able to provide synchronous joint torques with sufficient bandwidth for future walking experiments. While further engineering improvements and testing with additional participants are needed, the BP emulator shows promise as a tool for improving the gait patterns and overall mobility of individuals with limb loss.

Chapter 6

CONCLUSIONS

6.1 *Summary*

This dissertation presents a comprehensive study on the design, control, and evaluation of a biarticular prosthesis emulator. This device combines a robotic ankle-foot prosthesis and knee exoskeleton to facilitate scientific exploration and comparison of different prosthesis designs in a laboratory setting. My work provides an important contribution toward the development of future biarticular prostheses by detailing the design, control, and performance characterization of an example device. While the studies in this dissertation are primarily focused on the engineering development of novel wearable robotics systems, the design and control choices were intimately informed by the mechanical requirements of human walking. They required close collaboration with other engineers, prosthetists, and clinicians.

The first aim of our research was to develop the configurable off-board robotic actuation system (COBRA) and the COBRA Ankle-Foot Prosthesis (Chapter 3). Our powered ankle-foot prosthesis emulator can provide biological levels of torque and power during walking and can emulate idealized passive devices with low torque tracking errors. It is the first cable-driven ankle-foot prosthesis emulator that can modulate both plantarflexion and dorsiflexion moments with a single motor due to a novel actuator configuration that pairs a Bowden cable transmission with a set of custom parallel torsion springs within the prosthesis joint. The prosthesis uses a standard prosthetic foot plate and cosmesis, providing beneficial frontal plane compliance. It also fits inside a shoe, providing compliance in the transverse plane, which could reduce shear loading at the socket. Our work described the design, control, benchtop testing, and preliminary walking performance. This device will serve as a state-of-

the-art platform for research studies of robotic ankle-foot prosthesis design and control.

The second aim of our research was to develop the COBRA Knee Exoskeleton, a lightweight knee flexion assistance end-effector that is actuated by the COBRA system (Chapter 4). We developed COBRA Knee Exoskeleton specifically for people with transtibial limb loss. The device can provide high levels of knee flexion assistance during walking. The assistive torque profile was parameterized with the equation for a gaussian curve and was a function of gait phase. When evaluated in a walking experiment with a single transtibial amputee participant using a passive ankle-foot prosthesis, the device had low torque tracking errors and elicited positive subjective feedback from the participant. The COBRA Knee Exoskeleton may be used in future research studies of uniarticular knee flexion assistance. As ankle-foot prostheses are designed to replicate the torques provided by both the soleus and gastrocnemius muscles at the ankle, it is possible that a stand-alone uniarticular knee exoskeleton could benefit people with transtibial limb loss.

The third and final aim of our research was to develop the COBRA Biarticular Prosthesis Emulator. This unified device combines the COBRA platform, the COBRA Ankle-Foot Prosthesis, and a second-generation design of the COBRA Knee Exoskeleton (Chapter 5). We added a second actuator to COBRA to allow the platform to provide assistive torques to the knee and ankle simultaneously. The second-generation design of the knee exoskeleton included features that improved comfort and a more flexible torque control law. Benchtop tests characterized the controller performance, and a basic walking experiment validated the ability of the device to provide coupled knee and ankle torques.

As a whole, this dissertation provides several significant contributions to the field of bionic prosthesis development. The COBRA Ankle-Foot Prosthesis and COBRA Knee Exoskeleton can be used as stand-alone devices for future studies, and the designs can inform the development of other assistive devices. The COBRA Biarticular Prosthesis Emulator provides a platform for future research on biarticular gait assistance and comparing different hard-

ware embodiments for prostheses that improve mobility and quality of life for people with transtibial limb loss. The potential impact of this work is significant, as the development of more effective prostheses for people with transtibial limb loss can significantly improve their mobility and overall quality of life.

6.2 Future Work

6.2.1 Engineering Improvements

Several engineering improvements may increase the robotic performance of the current off-board actuation system and end-effectors. Emulators lend themselves well to continuous improvement and integration, as the components are necessarily distributed. For example, end-effectors can be completely redesigned without having needing to consider changes to the power supply, computing platform, or Bowen cable transmission, or high-level control system. The following sections detail engineering improvements that could be implemented in the future.

COBRA System

In terms of improvements to the off-board actuation and control system, the interface between the motor and the mobile platform could be reinforced to reduce vibration. When motor velocity changes rapidly, inertial forces cause the platform to vibrate, reducing torque tracking performance. The actuators and transmissions currently sit on damping feet but could be bolted directly to the cart frame instead. Further, a heavier and stiffer platform could help reduce vibration.

One factor that limits the torque tracking bandwidth for both assistive devices is the presence of electrical noise in the load cells and rotary encoders. Electrical noise from the environment limits how highly the gains on the torque and velocity controllers can be without inducing unwanted oscillations. These issues can be addressed through careful optimization

of electrical shielding, minimizing any sensor delays or latency, increasing controller sampling frequency, and reducing delays caused by digital low-pass filters. Fourier analyses of steady-state sensor behavior could also be used to develop digital notch filters that reject electrical noise related to the servo drives or building power. After these issues are addressed, feedback gains for the torque and velocity controllers may be further optimized to improve torque tracking performance.

Another issue that affects the COBRA control system is the need for accurate real-time heel strike identification and gait phase estimation. Currently, heel strikes are identified with a thresholding algorithm on the ground reaction force as measured by an instrumented treadmill. This approach is prone to erroneous controller behavior in the event of false positive heel strikes, for example, during a slight stumble or when the intact foot accidentally steps on the wrong treadmill belt. Future engineering developments could detect heel strikes with a pressure-sensing insole or kinematic sensors like inertial measurement units (IMUs). IMUs would also allow for more flexible estimates of gait phase, as thigh or shank angle could be computed and used within an adaptive oscillator [1] – a method that can more robustly estimate gait phase.

COBRA Ankle-Foot Prosthesis

Several control improvements could improve the behavior of the COBRA Ankle-Foot Prosthesis. For example, a dedicated ankle position controller would be beneficial to dorsiflex the ankle during swing phase rather than relying on the torque controller to drive the prosthesis to an equilibrium angle. Additionally, more advanced torque controllers should be used in future research, e.g., neuromuscular or impedance controllers [2]–[4], as the ankle torque controllers presented in this dissertation were only developed for preliminary hardware and software evaluation purposes.

Mechanical design improvements would benefit the prosthesis. The ankle joint’s top

and bottom aluminum pieces are constructed of several flat pieces that bolt together. We pursued this design as a cost-saving measure, as flat aluminum pieces are easy to manufacture. However, over time, cyclic loading can cause misalignment between the bolted parts, which can corrupt the joint encoder measurements. Frequent encoder calibrations are needed with the current design for reliable joint angle measurement. Machining the top and bottom of the joint out of solid blocks of aluminum would reduce the part count and solve the misalignment issue.

The inclusion of the torsion springs allows for our novel bidirectional torque control functionality, but the steel torsion springs we selected are relatively heavy and exhibit significant hysteresis compared to other potential design options. Future designs could explore replacing these torsion springs with composite leaf springs, as are found in different prosthesis designs [4]–[6]. Further, the parallel spring torque is estimated using a polynomial model that does not account for hysteresis or damping effects. Future work could focus on developing a dynamic model that accounts for hysteresis [7], or a load cell could be built into the device for direct torque measurement [8]. The latter approach would also potentially allow for removing the tension load cell at the Bowden cable interface and greatly simplify computation within the control loop associated with estimating the joint torque.

Knee Exoskeleton

The upgrades that would most improve the knee exoskeleton are rooted in mechanical design and comfort. Future work could focus on improved digital workflows for manufacturing patient-specific thigh cuffs to improve comfort. For example, recent studies have proposed manufacturing pipelines for exoskeleton and orthosis components that use 3D scans of people’s limbs to create custom anatomical interfaces [9], [10]. Removing the inner strut of the exoskeleton would also improve comfort but would likely come with the trade-off of increasing exoskeleton migration. Alternatively, soft exosuit technology could improve comfort

and mass, though alternative methods of knee angle sensing would be required to emulate quasi-passive devices.

6.2.2 Future Areas of Research

The COBRA Biarticular Prosthesis Emulator was developed explicitly to facilitate future research. This section lists several promising areas of future study that can be conducted with the device. Several ideas reference human-in-the-loop optimization, a method of continuously updating device behavior based on measures of the human biomechanical or physiological response to find optimal device behaviors [11], [12]. Several optimization targets would make sense for people with transtibial amputation, including kinematic or kinetic walking symmetry, user satisfaction, metabolic walking economy, or preferred walking speed – all of which could feasibly be measured while using the emulator. The research directions presented here are agnostic to which optimization targets are used, though this is also an interesting area to explore.

Device Design Oriented Studies

- **Informing design and control parameters using human-in-the-loop optimization:** Assistive device emulation and human-in-the-loop-optimization methods could be used to find optimal design and control parameters for future untethered biarticular prostheses. For example, the COBRA Biarticular Prosthesis Emulator could be programmed to emulate a quasi-passive biarticular prosthesis with a uniarticular spring at the ankle and a clutched biarticular spring that crosses the knee and ankle, similar to the device by Willson et al. [13]. Spring stiffnesses, clutch activation timings, and lever-arm geometries could be open design parameters determined by the human-in-the-loop optimization experiment. Parameters could be optimized across a group of participants, and the ranges of optimized parameters could be used as design targets

for a future untethered device.

- **Comparing biarticular prosthesis design architectures via human-in-the-loop optimization:** Similar to the previous idea, human-in-the-loop optimization algorithms could be used to help facilitate even higher-level design decisions, like whether or not to include powered actuators at the knee and ankle. Human-in-the-loop optimization studies could compare the user’s biomechanical and physiological performance while walking with different classes of optimized biarticular prostheses. Examples of classes of BPs that could be emulated include those with active and passive ankle-foot prostheses, active and quasi-passive knee exoskeletons, and any combination of the two. An example hypothesis for this type of study might be “*An optimized quasi-passive knee flexion exoskeleton paired with a powered ankle-foot prosthesis can provide equivalent improvements to kinematic symmetry during walking as an optimized fully robotic biarticular prosthesis.*” If true, it would indicate that the quasi-passive device would be the preferred device to develop due to the added complexity and cost of fully robotic devices.

Basic Science Studies

- **Understanding hip compensations when walking with uniarticular and biarticular prostheses:** It would be interesting to use motion capture, inverse dynamics, and EMG to compare basic gait biomechanics and muscle activation when walking with emulated uniarticular prosthesis and biarticular prostheses. When walking with passive and powered ankle-foot prostheses, people with transtibial limb loss have elevated hip flexion moments during late stance [14]. Modeling studies have found that the gastrocnemius typically accelerates the stance limb into swing [15]. Uniarticular prosthesis users likely use the hip flexors to accelerate the leg into swing as a compensation. It is possible that a biarticular prosthesis could help accelerate the stance leg

into swing and reduce hip flexor demands.

- **Understanding virtual gastrocnemius and soleus contributions to forward propulsion and bodyweight support via perturbation studies:** Prior perturbation studies with healthy participants and surface electrodes have found that the soleus is relatively more responsible for bodyweight support, while the gastrocnemius is relatively more responsible for forward propulsion of the body's center of mass [16]. These studies could be replicated with the biarticular prosthesis emulator using uniarticular (soleus-like) and biarticular (gastrocnemius-like) torque perturbations. If biarticular prostheses are found to uniquely contribute to forward propulsion of the body's center of mass, they may be more well-suited to increase walking speed than uniarticular prostheses.

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