MIXED TERRAIN ANKLE ASSISTANCE AND MODULARITY IN WEARABLE ROBOTICS

By

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A Thesis

Submitted in Partial Fulfillment

of the Requirements for the Degree of

Master of Science

in Mechanical Engineering

Northern Arizona University

August 2023

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ABSTRACT

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Having been proven effective in reducing the metabolic cost of walking over the past 10 years, exoskeletons are leaving laboratory environments. In the real world, these systems will face new problems related to durability, assistance, and terrain variance. Systems capable of providing meaningful assistance during all-terrain ambulation will enable researchers to inspect exoskeletons' role in helping mobility-compromised individuals regain activities of daily living. By designing these systems in a modular manner, future work may be accelerated and improved. This thesis takes meaningful steps in the development of a modular platform for wearable robotics, the design, and validation of a high-power ankle exoskeleton using this modular platform and presents a novel controller capable of dynamically adapting its assistive profile in all-terrain conditions. The modular platform is shown to accelerate the development of unique wearable robotic systems. The high-power exoskeleton is shown to provide meaningful assistance during activities previously not possible for predicate designs. The controller is shown to reduce the muscle activity of a primary plantar flexor muscle at various terrain slopes.

Acknowledgments

I would first like to thank my advisor, Dr. Zachary Lerner, for cultivating a fun, inquisitive, and collaborative work environment. You gave me the opportunity to work on exciting problems with amazing people, and I loved my time in the lab. Thank you.

Special thanks to Dr. Paul Pridham with whom I developed the modular system. Your expertise was invaluable, and I enjoyed working with you.

To the past and present members of the Biomechatronics Lab at NAU, it was a blast. Thank you for being so supportive and fun-loving. Most importantly, thank you for being worthy adversaries in foosball.

I would also like to thank my friends and family for their endless love, support, and encouragement. You have always been there to push me towards my best self, and to make sure that I enjoy the process. You are a fun group of wonderful people, and I am glad that you are on my team.

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Introduction

Lower limb exoskeletons have the potential to improve an individual's mobility [19]. A lack of mobility is associated with decreased cognitive function, reduced independence, and an increased risk of fractures, falls, and death [18]. To improve mobility, exoskeletons must impact the effort it takes to walk, quantified by the metabolic cost of walking. Many exoskeletons have been shown to reduce the metabolic cost of walking and running [1]. This is done by providing assistive torques at critical moments in the gait phase and minimizing the weight of the exoskeleton.

Researchers have been developing exoskeletons to improve mobility since the early 2000s. Much progress has been made and many researchers are beginning to inspect the role that these systems have outside of a laboratory [6]. As the capabilities of exoskeleton systems improve, they will begin to be deployed in the real world, and the environments that they operate in will not be controlled. These environments include varying slopes, stairs, obstacle navigation, and varying speed. Outside of a laboratory testing environment, the exoskeleton cannot rely on large equipment or sensors. If the system is going to be used as a mobility aide, work must be done to improve usability and durability. For exoskeletons to be successfully integrated into society, these problems need to be solved.

The people tasked with inspecting how these wearable robots interact with the body, biomechanists, require a large breadth of knowledge. Having high integration and reliability requirements, wearable robotic systems are challenging to design and require domain knowledge that most biomechanists do not possess. Even worse, many exoskeleton systems are created as one-off designs and are only used to collect the data required by a single experiment, forcing the work to be duplicated by future researchers.

By solving these problems in a modular manner, future work will be accelerated, device use can be standardized, and the required domain knowledge will be significantly reduced. Once all these problems are solved, researchers can begin to inspect how these systems could be used in the real world.

The first goal of this thesis was to develop a modular exoskeleton system, consisting of electromechanical and firmware architectures, that would allow for the changing of actuators, joint configurations, and control strategies. To accomplish this goal, we architected a firmware system that utilizes runtime polymorphism for modularity and selected an actuator platform with a networked interface. The second goal was to utilize the modular design to build and validate a high-power cable-driven ankle exoskeleton capable of assisting adults during hiking. To fulfill this goal, we designed an untethered exoskeleton that assists the ankle joint in uneven terrain for the length of a typical hike. The third goal of this thesis was to develop a controller that varies ankle joint assistance with terrain slope. To fulfill this goal, we created a controller that exploits terrain slope differences in joint angles to better match biological joint moments. All these goals were met, accelerating the development of wearable robotics and enabling researchers to inspect the role that these systems have in all-terrain walking.

Design Criteria

Design criteria are the goals that a project must achieve to be successful. The criteria, target, and purpose are tabulated below. The remainder of this section will further detail each of these criteria.

Table 1: Summary of Design Criteria

Criteria	**Purpose**	**Target**
	Goal 1	
Modularity	Accelerate future work	Validate
	Goal 2	
Weight	Reduce metabolic cost	Less than 4 kilograms
Ergonomics	Comfortable operation	Near predicate designs
Speed	Assist during dynamic motions	Do not impede joint angular velocity at 1.5 meters per second
Peak Torque	Provide substantial assistance	0.35 Nm/kg for the average adult male
Low-Level Control	Achieve desired torque	RMSE near predicate designs
Operation Time	Enable data collection	30 minutes
	Goal 3	
High-Level Control	Provide meaningful assistance	Adjust output for all terrains, and reduce target muscle activation

Predicate Designs

Figure 1: Predicate Design [14]

Many metrics for the current design are relative to predicate designs. These devices mounted the motors, power supply, and control electronics at the waist; the small black box on the wearer's waist in *Figure 1*. The motors were connected to the sagittal plane of the ankle joint by a Bowden cable transmission system that terminated at a pulley. The torques from the motors react on the wearer and the environment through custom carbon fiber components at the shank, and within the shoe. The proposed work will utilize the same design architecture as the predicate design.

The predicate system was originally designed for pediatric resistive and assistive interventions and proved underpowered for more demanding adult use cases. The proposed work will address this by using more powerful motors.

Using a force-sensitive resistor (FSR) predicate controllers discretized the gait phase for a single leg into stance and swing. In swing the device would provide a constant dorsiflexion torque to aid in raising the wearers foot. In stance, the device provides plantarflexion assistance that is proportional to the normal ground reaction force. The proposed work will address issues that arise when this controller is used with more powerful actuators, and on sloped terrain.

Goal One: Modularity

Modularity

By solving problems in a modular manner, designers do not have to duplicate work or learn implementation details outside of their domain of expertise. In robotics, modularity can be achieved through hardware or software. Many modular designs, such as the FlexSEA [5], shown in *Figure 2,* rely on multiple printed circuit boards (PCBs) to create a hierarchy of abstraction. In these approaches, certain components are responsible for interfacing with sensors and actuators, while other components perform high-level control.

Figure 2: FlexSEA Printed Circuit Board [5]. A modular design for wearable robotics composed of high-level computation modules and low-level sensing and actuation modules.

While this approach is more modular, it is very complex. This distinction presents a tradeoff between complexity and revisability. A complex system may be very adaptable to new designs, but it is not trivially revisable. Whereas a simple system may be easily revised but is not very modular. As stated previously, wearable roboticists are typically not experts in electronics or software engineering. Moreso, an easily revisable system is more likely to be iterated on rather than expired, reducing workloads. This is why a simpler architecture should be chosen.

Unfortunately, wearable robots must be worn, and when being worn they are swung about the world, smashing and flailing into things. When they are being smashed and flailed, the connectors and sensors are the first to go. A wearable robotics platform should take special care to improve connector strain relief and error detection. To reduce the risk of unintended interactions with the environment, the profile and weight should be minimized.

To be used as a research tool for biomechanists: most of the robotics domain knowledge should be abstracted away, the system should report and collect enough data, the system design should be amenable to revision, and its control should be intuitively parameterized. When all these criteria have been met, the modular design may reduce workloads, standardize training, and persist.

Goal Two: Design

Weight

For untethered devices, weight is one of the most important factors in evaluating a design. As weight increases, the cost of transport increases [2]. Another study found that for every kilogram of mass added to a backpack the metabolic cost of walking increased by 3.28% [25]. Moreso, the distribution of weight matters [20]. To reduce the metabolic burden of carrying a device, the weight should be as close to the wearer's center of gravity as possible [2]. Despite this, the predicate design was shown to reduce the metabolic cost of incline walking by 9.9 +/- 2.6% in unimpaired adults [24]. The device proposed in this thesis and the predicate device, assist the ankle joint so it is essential to reduce distal mass. For these reasons, a cable-driven design should be used. While a cable-driven design decreases the transmission efficiency, it places the heavy actuators and power source near the wearer's center of gravity.

The design presented in this thesis requires more powerful actuators and will be heavier than the predicate design. The distal mass of this design should not increase much from predicate designs, so only the total weight of the device will be used in the following section. Using the metabolic reduction of the predicate design and the relationship between metabolic increase as a function of added mass, the predicate design mass may be increased by ~3kg while still maintaining a net zero change in metabolic cost. This is a loose approximation that is only being used to demonstrate a potential weight budget. The design should not utilize all this weight budget but rather use it to inform high-level design decisions, such as actuator and power supply selection.

Predicate designs weighed ~2.2 kg, the 3 kg weight budget would generate a design goal of 5.2 kg. However, this will likely be too heavy, and a design goal of 4 kg should be used instead.

Device Ergonomics

Device ergonomics may be separated into two categories: device comfort, and device geometry. There are two primary concerns with device comfort: friction and pressure points. Prolonged friction and pressure points can cause rashes and damage to the skin, which may harm the wearer's interaction with the device, and pose a health risk. The device geometry may also cause the wearer to interact negatively with their environment and the device. Certain designs, such as the Collins Exo [6], utilize a transmission system that protrudes from the wearer’s posterior and may preclude the device from assisting during certain tasks, such as stair descent. Significant medial protrusion causes the wearer's legs to interfere with each other during walking, which may increase step width and reduce walking efficiency [13]. Ultimately, poor ergonomics harms the interaction between the wearer and the device, reducing the rate of adoption. The device should minimize protrusions, friction, and pressure points.

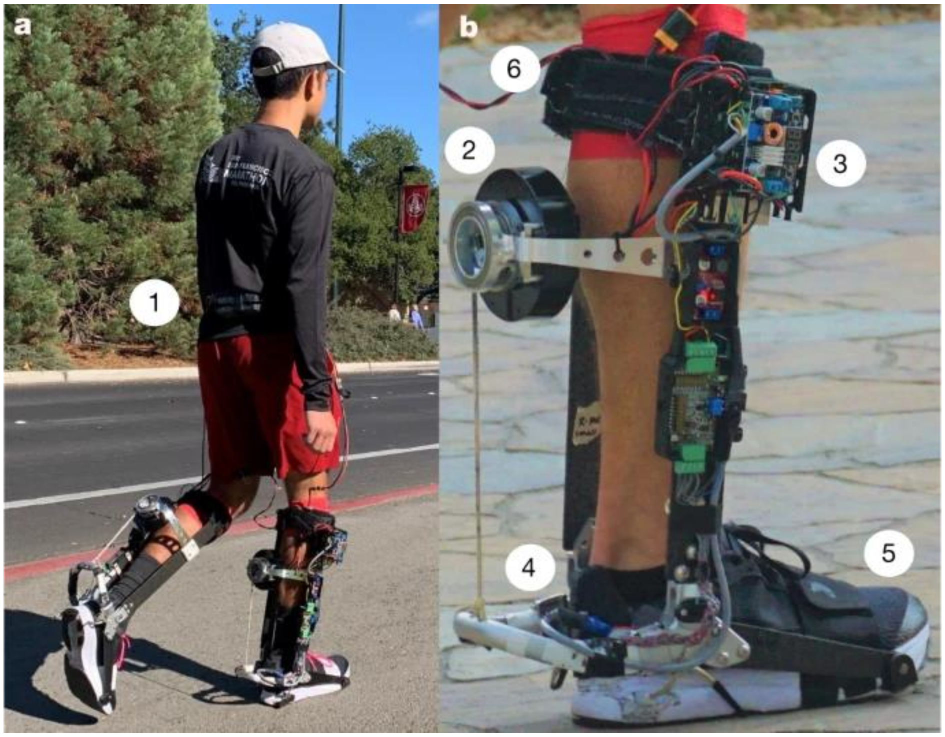

Figure 3: A similar high-power ankle exoskeleton [6]. Displaying significant posterior protrusion and non-negligible medial protrusion.

Speed

If the device is to be assistive it should not limit the wearer's joint speed. When walking at 1.5 meters per second, which is a fast-hiking speed for unimpaired individuals, the ankle's angular velocity can reach peaks of 331 degrees per second [8]. Predicate designs utilized large gearing reductions to achieve peak torque, which limited the joint speed during fast walking. To account for the gearing ratio and time delays introduced by the transmission system, the actuator's angular velocity must be greater than the joint's angular velocity at this peak target speed. The device should not limit the ankle joint speed when walking at 1.5 m/s.

Peak Torque

Determining the peak torque for a joint is complex, it requires knowledge of the task and the wearer. To reduce the metabolic cost of walking, the assistive torque magnitude should be some percentage of the biological joint torque. To provide meaningful assistance, a target peak torque of 0.35 Newton meters per kilogram of body weight is used [24]. The average American male weighs about 86.6 kg [9], and at 0.35 Nm/kg would require 30.31 Nm of peak torque. During faster motion, the actuator power may limit the actuator's ability to generate its static peak torque. Therefore, the device should be able to produce 30.31 Nm of torque at 1.5 m/s; thereby satisfying the peak power requirement.

Low-Level Control

A low-level controller must be used for the device to accurately provide torque at the wearer's joint. The performance of the low-level controller is directly related to the actual torque applied at the ankle. This low-level controller accounts for many unmodeled transmission effects, such as the varying cable bend, tension, and wear. These transmission effects were not modeled due to unobservable state variables. Performance is calculated using the root mean square error (RMSE) between the desired and measured torques. The RMSE is used because it maintains the same units of the original data and the error accumulates over time. Predicate systems had an RMSE of 2.448 Nm. The device should be able to track with improved RMSE than predicate designs.

Operation Time

To inspect the benefits of the device and controller, the device’s operating time should allow for the collection of steady-state metabolic data. Many data collection protocols test the effects of multiple conditions, each requiring the observation of steady-state metabolic data. Metabolic data collections typically last from 6-7 minutes [21],[22]. When collecting metabolic data outside of the lab, walking conditions vary, and steady-state metabolic data may not be observed. To allow for the successive collection of steady-state metabolic data and collections outside of the lab, the device should be able to operate for at least 30 minutes; enabling 5 data conditions.

The battery capacity is directly proportional to operation time and should be selected to enable continuous operation. For a given battery chemistry, the capacity is proportional to the weight of the battery. Lithium polymer batteries should be used because they have very high specific energy and maximum discharge rate. Current data should be collected to estimate the power use of the device during its peak power condition. During this peak power condition, the actuators will generate a significant amount of heat. Actuator temperature affects system wear, comfort, efficiency, and operation time. The actuator temperature should be monitored during the device’s operation. After data collection, the device may need to be serviced. This servicing should be easy or infrequent, further simplifying the interaction between researchers and the device.

Goal Three: Mixed-Grade Control

High-Level Control

The high-level controller is responsible for generating torque trajectories that are informed by the biological joint moment. A torque trajectory is defined as the joint torque as a function of the gait phase, and it is useful for visualizing the exoskeletons' contribution during a single step. The requirements of a high-level controller depend on the requirements of the device. The device proposed in this thesis is designed to assist unimpaired adults in sloped terrain.

A comprehensive review of exoskeleton control was performed by R. Baud, et al [10]. In this review, methods that estimate the biological joint moment were explored; some rely on the synchronization of repetitive biological signals, and others estimate the joint torque using only the current state. Methods that can estimate the joint torque with only the current state of the system, or time-invariant methods (TI), are preferred because they do not rely on the steady gait cycle assumption. Errors in the gait cycle estimate cause improper device torques, and these errors typically happen during modality and speed transitions.

For the swing phase of walking, predicate systems have provided a constant dorsiflexion torque. Many systems do not provide dorsiflexion assistance, relying on the wearer to prevent tripping. These methods simplify design, but many impaired populations require dorsiflexion assistance to prevent tripping [23]. To allow the controller to possibly assist these populations, dorsiflexion assistance should be administered during the swing phase.

Globally optimal methods converge on trajectories that administer ankle assistance at the very end of the stance phase [6]. While these methods have driven the field toward unified and optimal torque trajectories, globally optimal controllers are not step-adaptive and instead rely on small terrain slope variance and few modality transitions, effectively averaging out odd steps. More so, it has been shown that during decline walking the ankle moment is higher earlier in the gait phase [15]. It has also been shown that the ankle joint contributes more negative work during decline walking than other inclinations [16], particularly during early stance. Opposing the findings of globally optimal methods for level walking, sloped joint moment estimates suggest that assisting the ankle during the early stance phase of decline walking may be beneficial.

Relying on the normal component of the estimated ground reaction force, predicate systems provide less assistance during incline walking. However, sloped ankle moment estimates from Nuckols et al [15] have found that the ankle power slightly increases from level. This problem should be addressed.

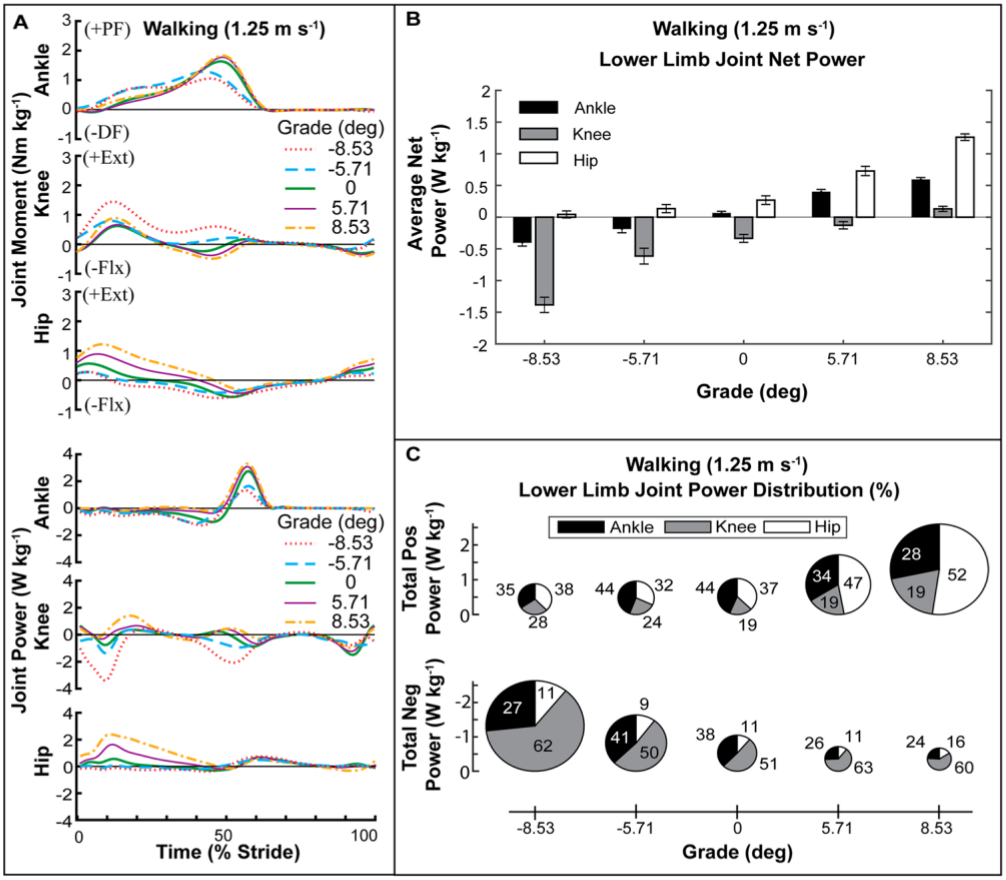

Figure 4: Joint Moment and power at various grades [15]. Figure A plots the joint moment and power normalized by body mass for the ankle knee and hip joints for various grades versus the gait phase. Figure B plots the net power of the ankle, knee, and hip joints for various grades. Figure C shows each joint's contribution of total positive power and total negative power per stride for various grades. Focusing on the ankle joint during negative grades; the joint moment occurs earlier in the stride and the net power of the joint is negative.

Some methods for assisting during sloped walking rely on estimating the slope or walking modality to switch or parameterize the control law; these methods require terrain or modality estimation. If wearable robots are to be used in unsupervised settings, they must be very reliable at predicting the terrain or modality during use cases not anticipated by the designer. Ensuring such reliability, or reducing the consequences of a false modality estimate, is very difficult and likely requires computationally expensive techniques. Other methods require the terrain slope to be provided by the wearer, requiring too much manual input and ruining device usability. It is for these reasons that methods reliant on explicit slope estimation or manual slope input were not explored.

TI controllers are preferred due to their ability to quickly adapt to the wearer’s needs. Therefore, the controller should only be a function of the current state, provide dorsiflexion assistance during the swing phase of gait, and modify its assistance based on the slope of the terrain without directly estimating the terrain.

Summary

Each goal of this thesis has an independent set of goals to indicate success or failure. The modularity work should accelerate development and reduce workloads for operation and design iteration. The high-power ankle exoskeleton should permit prolonged data collection of unimpaired adults in all terrain conditions. The controller should adapt its assistance as a function of terrain slope and reduce EMG activity in a primary plantar flexor muscle.

Methods

The following sections will detail the development of a modular platform for wearable robotics, a high-power ankle exoskeleton built using that platform, and an all-terrain controller designed using that high-power ankle exoskeleton.

Figure 5: High Power cable-driven ankle exoskeleton constructed using the modular platform for wearable robotics.

Device Overview

Using sensors placed at each ankle, control electronics, and a novel sloped terrain controller, the device sent torque commands to waist-mounted actuators to assist the wearer throughout the gait cycle. The actuator torques were transmitted using Bowden cables to a pulley that rotated with the ankle joint in the sagittal plane. The torque applied to the ankle joint reacted against the wearer's shank and the ground through custom carbon fiber components. The device interfaced with a smartphone running a proprietary graphical user interface (GUI), which was used for data acquisition and operation.

Modularity

To balance modularity and revisability, much of the modularity was achieved in software. This simplified the electrical architecture, which relied on unused or networked electrical connections to support multiple configurations. A Teensy 4.1 Development board (teensy) was used to perform data acquisition, control, and actuator communication. The teensy sent state information to an Arduino Nano 33 BLE development board (nano) which was responsible for communicating with a proprietary GUI via Bluetooth Low Energy: Figure 5 shows the interaction of these systems and their peripherals.

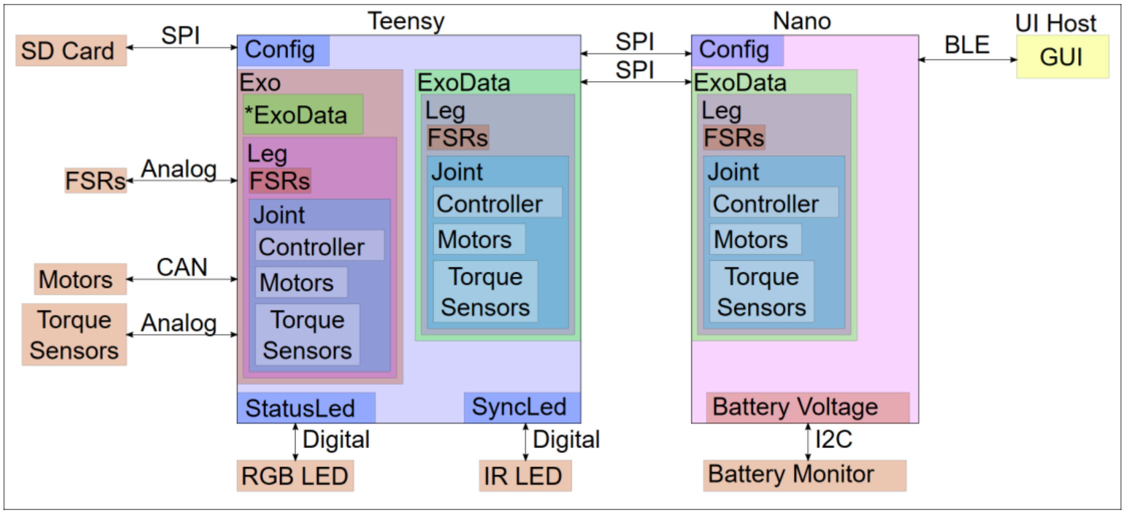

Figure 6: Modular platform software architecture representing external connections as arrows, labeled by their communication methods. The Teensy performs control, sensing, and actuator communication. The nano communicates with a GUI. Configuration data is stored in the SD card and read at startup.

The PCB provided power to the actuators, conditioned the analog sensor signals, used the Controller Area Network (CAN) protocol to communicate with the actuators, and accepted the microcontroller development boards. A simplified graphic of the PCB is displayed below.

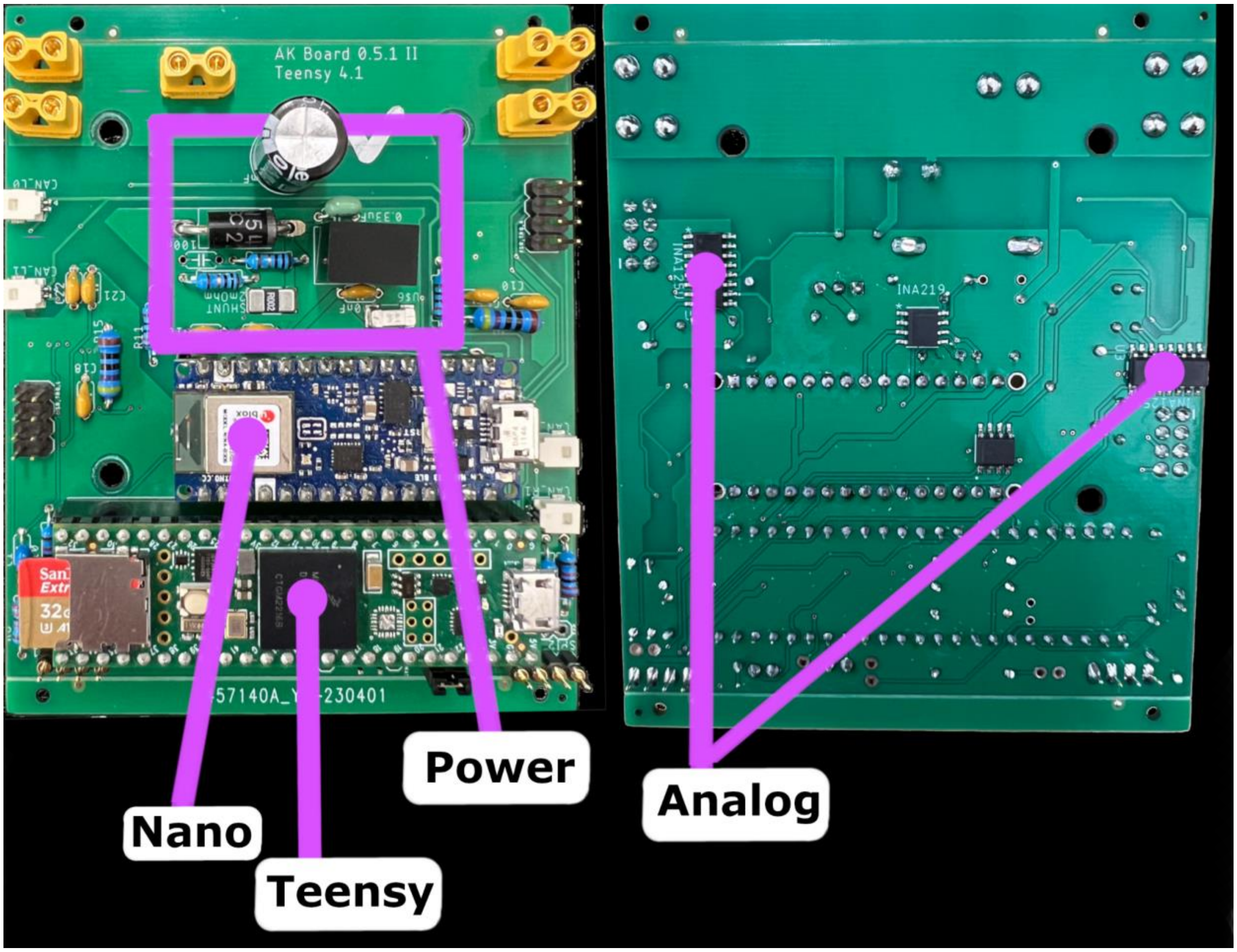

Figure 7: Custom Printed Circuit Board for the modular platform. Depicting the microcontrollers, power circuitry, and analog conditioning circuitry.

Most of the modularity was performed in the software architecture and was achieved through Inheritance-Based Polymorphism (IBP). Other methods used to achieve software modularity, such as opaque types, were inspected but IBP is supported as a core feature in modern C++. As a core language feature, compiler errors and keywords simplify debugging and implementation workflows. C++ offers IBP through the use of abstract classes. A parent abstract class is created to define a standardized interface, this abstract class leaves method implementation to its child classes. The child classes inherit from the abstract class and implement the abstract methods. Each class that inherits from the abstract class then guarantees that it follows a given interface, which was exploited by other parts of the program. This method was used for the actuator, joint, and controller classes. As displayed in Figure 5, the software was structured in a compositional hierarchy composed of Exo, Leg, and Joint classes. This hierarchy features mirrored state classes that were used to store state data. The Exo class was responsible for system timing, and communication with the nano. The Leg class was responsible for storing state data and for sampling and calibrating the FSRs. The Joint class was composed of Controller, Motor, and Torque Sensor classes. The Joint class aggregated and conditioned state data which it passed to the Controller class; it would then pass the controller output to the Motor class.

The software was initially configured at boot via several files stored in a Secure Digital (SD) card. During this initial configuration, the board, actuators, direction modifiers, and sensor suite were selected. The system was monitored and operated by a mobile GUI. The GUI sent user commands to the exoskeleton, displayed data in real-time, and stored data in the cloud where it was retrieved later for analysis.

To promote and standardize future software development the source code was packaged into a self-contained and buildable system, referred to as a codebase. Changes to the codebase were performed through a version control system, which assigned unique numbers to each version of the codebase. Version control systems were used during debugging to revert back to functional codebase versions, and to track changes in the codebase throughout its lifetime. This version control system was hosted on the internet to enable collaboration between software developers. Each developer's changes were managed and integrated into a common codebase, where the versions were thoroughly tested before being released.

Actuators

Research has shown that for optimal human augmentation, the actuators were required to be light, and powerful [6]. To achieve this, we selected the TMotor AK80-9 actuators (TMotors). The TMotors were a commercially available version of the actuators used in the MIT Mini Cheetah 3 [12], which had proven to be highly effective in vital areas such as peak power and angular velocity. The TMotors were brushless direct current (BLDC) actuators with an integrated controller and gearbox. BLDC actuators were selected because of their very high specific power. These actuators used a sensor-based field-oriented commutation scheme. The TMotors had 21 Pole-Pairs, a peak torque of 18Nm, a rated continuous torque of 9Nm, and a no-load speed of 1,710 degrees per second, and after accounting for the gear reduction of the exoskeleton exceed the requirements for this system by 14.8%. All real-time TMotor communication was performed using the CAN protocol; a fault-tolerant, networked, and bus-based communication protocol relying on differential signaling for noise rejection.

Figure 8: TMotor AK80-9 Actuator. This is a BLDC actuator employing a field-oriented commutation scheme with an integrated motor driver and gearbox.

The CAN network was accessed using the teensy, and the SN65HVD230M-EP 3.3V CAN transceiver. The actuators accepted position, velocity, torque, position error gain, and velocity error gain setpoints. Which resulted in the following motor torque equation 1,

$$(1)\ \tau_{Act} = K_p(P_{Des} - P_{Act}) + K_d(V_{Des} - V_{Act}) + \tau_{Ref}$$

By setting the position and velocity error gains to zero, the actuators were placed in the torque setpoint mode. This mode was selected to better standardize the motor interface used in the modular codebase. After control commands were sent, the actuators responded with position, velocity, and current.

Sensor Suite

Most of the sensors were located at the wearer's ankle joint. These included FSRs located at the ball of each foot to estimate the ground reaction force, hall effect angle sensors at the ankle for position and velocity estimates, and a custom, strain-based, torque transducer [14] used to perform closed-loop control. There were many challenges in estimating wearer-centric parameters such as ground reaction force and ankle angle. One of the primary challenges was related to device fit variance, which contributed to dead space, hysteresis, and zero-drift sensing issues. Moreso, device fit variance contributed to these sensing issues in a seemingly non-stationary manner. To compensate for these sensing issues, the device performed one-time range-based calibrations during every session, and control was performed using conditioned state data. Each sensor required a separate calibration process. The angle sensor was calibrated once the device was constructed by moving the device through its full range of motion, the sensor would then return a ratiometric value. After the device was donned, the torque transducers were tared to remove any static offsets. Peak and trough data were collected to normalize the FSR signals during a walking calibration period. During the walking calibration, the device was placed in ‘Zero Torque’ mode, to limit the impedance felt by the wearer.

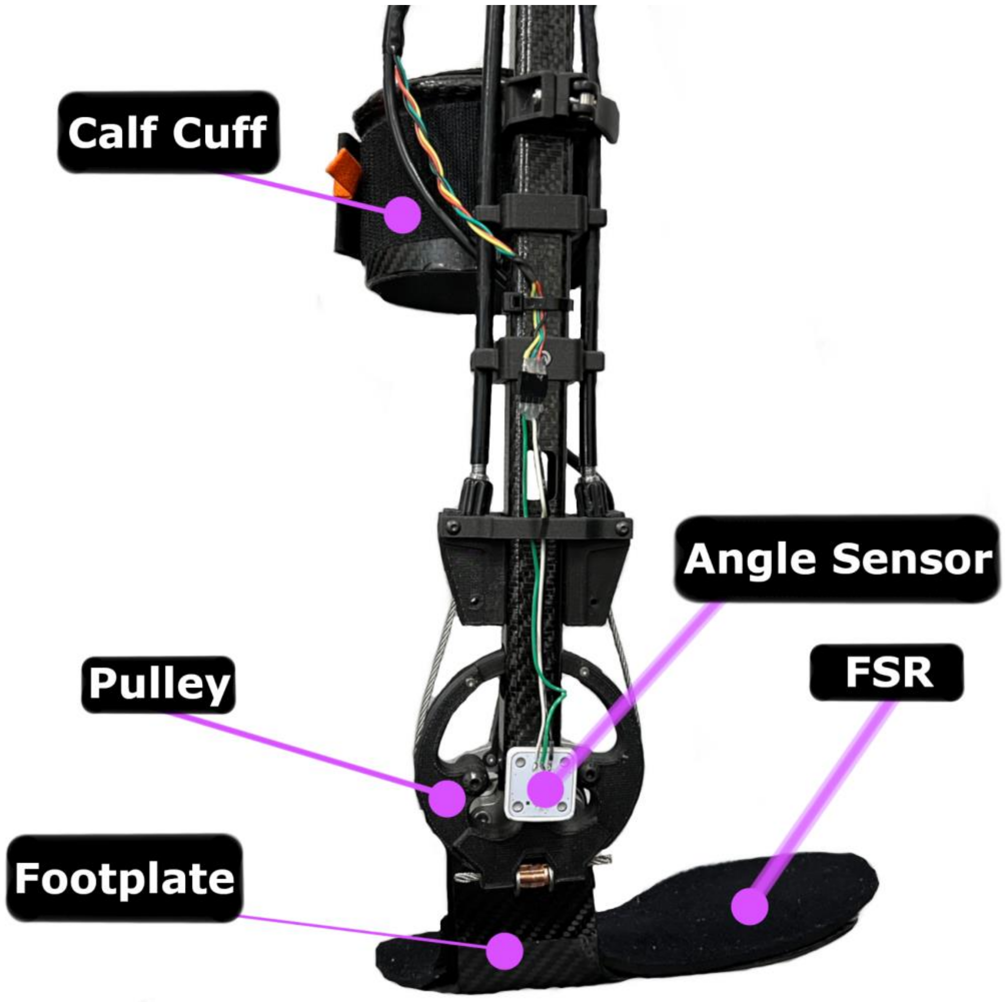

Figure 9: The lower leg assembly is responsible for Bowden cable termination, joint alignment, exoskeleton torque reaction, and sensing. Much of this assembly is created with carbon fiber to minimize distal mass.

Sensor data was used to perform fault detection in real-time, which ensured the device operated safely. The fault detection system was responsible for detecting sensor, actuator, and mechanical failures. When simple thresholding was not enough to reliably capture sensor failures, variance-based bounding was performed to detect sudden changes in sensor readings. To detect actuator and mechanical failures, the low-level controller performance was estimated online.

Once errors were detected, one of three things happened depending on error severity. In order of increasing error severity: the device would warn the operator, place itself in a low-impedance mode, or disable the actuators.

Control

Certain biomechanical features were exploited to assist the ankle joint during a range of slopes. Strutzenberger et al. [11] showed that the ankle angle at weight acceptance, and ankle angular velocity were proportional to the slope. Rather than explicitly estimating the slope, these features were used to modify the assistance for each step.

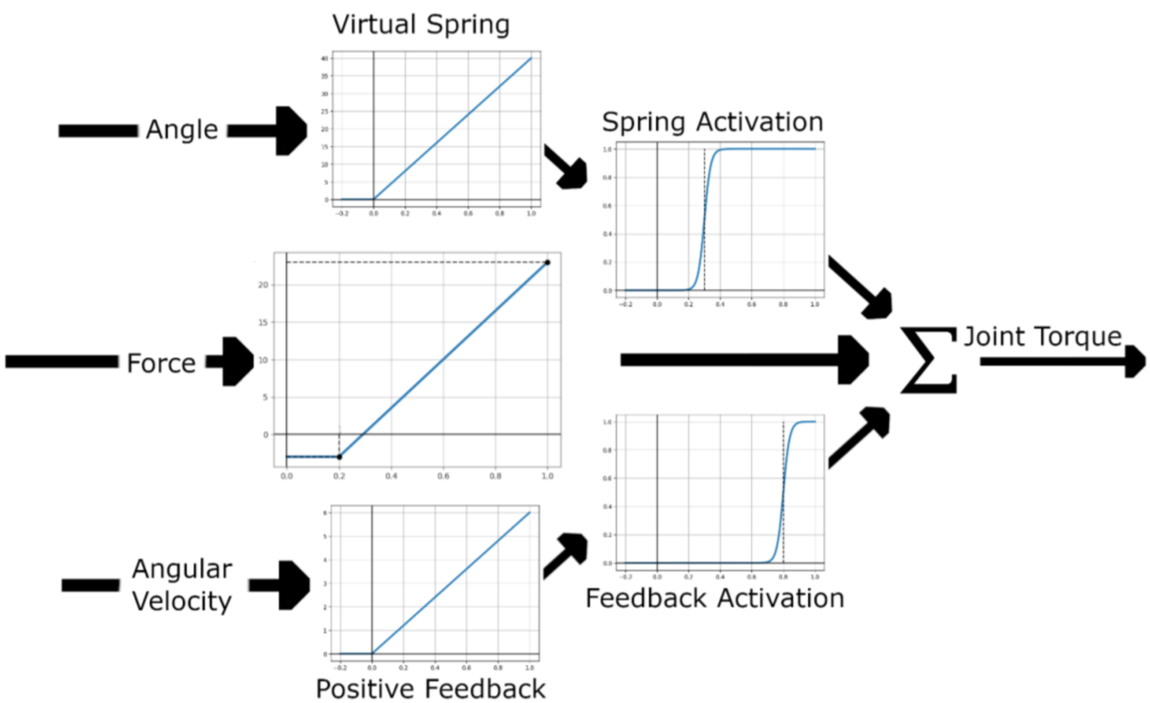

Figure 10: The Terrain Responsive Exoskeleton Controller uses normalized angle, force, and angular velocity data to provide grade adaptive joint torques. The controller modifies assistance for decline walking by modeling the acute intra-step angle range as a virtual spring. The controller modifies assistance for incline walking by performing velocity-based

energy shaping at the end of the stance phase. The controller also uses the normal component of the ground reaction force to provide torque.

The Terrain Responsive Exoskeleton Controller, or TREC, was developed to use the ground reaction force and joint angle to dynamically modify the assistance provided to the wearer during sloped terrain. TREC was a summative controller consisting of three terms: the generic term, the assistive term, and the propulsive term. Using an FSR to approximate the ground reaction force, the generic term mapped a normalized ground reaction force to a scalable assistance profile.

The generic term was calculated with equation 2

$$(2)\ G = max(\frac{PSP-DSP}{1-THR} * (\%GRF - THR) + DSP, DSP)$$

Where %GRF was the normalized FSR signal, PSP was the plantar flexion scaling factor, DSP was the dorsiflexion scaling factor, and THR was the %GRF to saturate the dorsiflexion assistance. The supportive and propulsive terms were used to modify the controller's behavior during decline and incline walking, respectively. During decline walking the ankle joint increasingly dorsiflexes as the stance phase progresses. By modeling the ankle moment as a spring, additional support was provided during more dorsiflexed, or acute, joint angles. The spring stiffness was user-defined.

The assistive term was defined by equation 3

$$(3)\ As = max(k * \theta_D - b * \omega, 0)$$

Where θ_D was the change in angle from the spring rest length defined in equation 4

$$(4)\ \theta_D = \theta_R + (\theta_R - \theta_L) + \theta_0 - \theta$$

Where θ_R was the dynamic rest length, θ_O was a user-defined offset, θ_L was a calibrated angle captured during level walking and θ was the current angle reading. If the spring had a static rest length the assistive torque would be engaged during any dorsiflexion, resulting in plantar flexion assistance during the swing phase of gait and excessive assistance during early stance incline walking. The rest length was updated at toe contact and expired at the entrance of the swing phase. The user-defined offset was used to engage the spring at the entrance of the stance phase and resulted in immediate spring assistance for every step. The term $\theta_R - \theta_L$ was added to increase the angle offset during decline walking. To do this the weight acceptance angle is calibrated during level walking and subtracted from the reference angle. Referred to as the 'dynamic offset' term, this term increased spring engagement during decline walking, and disabled it during level and incline walking. The instantaneous angle ranged from [0,1] where 1 was the device's maximum plantarflexion angle and zero was the device's maximum dorsiflexion angle. To ensure that the spring was not active during the swing phase, the output of the supportive term was passed through a tuned sigmoid that activated when the ground reaction force was greater than THR.

To assist the wearer during incline walking, and more generally add energy to their gait, the propulsive term was modeled as a gain on the plantarflexion velocity. The ankle joint moves through a larger range during incline walking, and for the same walking speed, the joint velocity will increase with the terrain slope.

The propulsive term was defined by equation 5

$$(5)\ P = max(kP * \omega, 0)$$

Where kP was a user-defined gain and ω was the joint angular velocity. To only assist during late stance, P was also passed through a tuned sigmoid activation function which engaged when the ground reaction force was greater than a threshold value. By using tuned sigmoid functions, the controller eliminated discontinuities that would arise from overly simplistic thresholding.

The sigmoid function used was defined by equation 6

$$(6)\ \sigma_{ACT} = \frac{e^{50(\%GRF+ACT)}}{e^{50(\%GRF+ACT)}+1}$$

Where ACT was the %GRF that the sigmoid would begin passing. Finally, equation 7 defined the entire TREC controller.

$$(7)\ \tau = G + (\sigma_{THR} * As) + (\sigma_{1-THR} * P)$$

Low-Level Control

The torque setpoint generated from the high-level controller was passed to a proportional derivative (PD) controller to account for unmodeled transmission dynamics. The PD controller was defined by equation 8

$$(8)\ \tau_R = K_P * (\tau - \tau_{DES}) - K_D * (\tau - \tau_{PREV}) * \Delta t$$

Where τ_R is the torque sent to the motors, K_P is the proportional gain, $(\tau - \tau_{DES})$ is the torque error, K_D is the derivative error gain, and$(\tau - \tau_{PREV}) * \Delta t$ is the numerical derivative of the torque error.

The transmission dynamics were not modeled because the Bowden cable dynamics changed in an unobservable manner throughout the gait cycle. The PD gains were tuned by hand and iteratively improved throughout the use of the device. This simple controller is easy to implement, but it is not adaptive or very robust. The controller also required the sign of the state variables to be properly set. To do this each device had direction modifiers set in its SD card. These direction modifiers only needed to be carefully set after the device was assembled.

Design Validation

Modularity

At the time of writing, four separate systems had been created using the modular architecture proposed in this work, while the predicate system had taken over three years to produce a new generation. By changing a few values in an SD card, the modular electronics and firmware system could be moved from an elbow exoskeleton to a hip-ankle exoskeleton with different actuators. This showed that the modular framework was indeed capable of accelerating the development of new wearable robotic devices. The electrical and software architectures were revised with minimal training and effort, showing that the architecture was intuitive in nature and capable of being further developed.

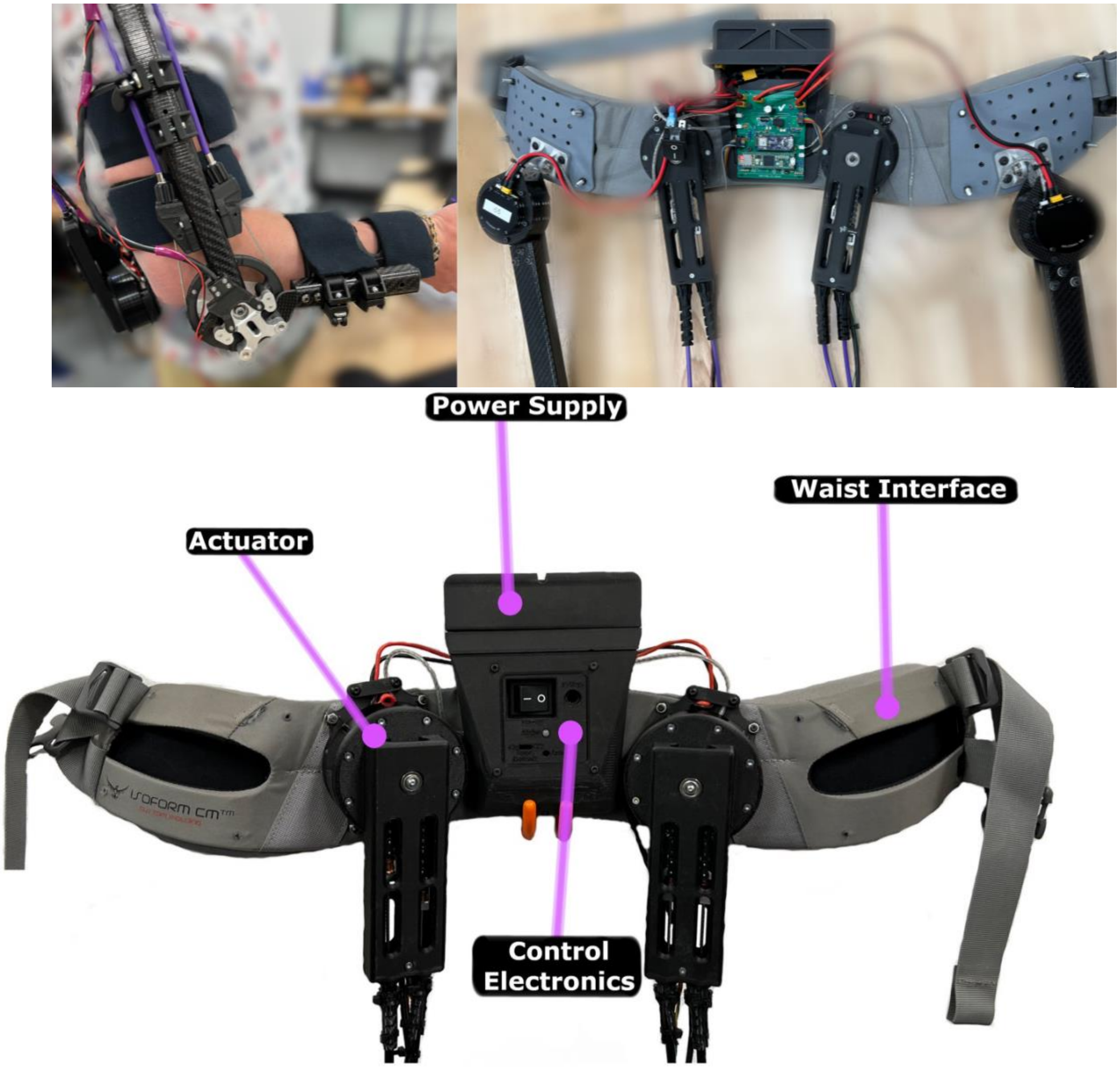

Figure 11: Elbow, Hip Ankle, and High-Power Devices all built with the modular platform for wearable robotics. The High-Power graphic has labeled actuators, power supply, control electronics, and waist interface components. As a demonstration of modularity; so long as values are changed in an SD card, the PCB from any of these devices may be swapped with another. This swapping may be performed for different actuators, configurations, and controllers.

Weight

Not including the weight of the Bowden tubes, the leg assembly weighed 435 grams. The Bowden cable termination blocks, pulley, torque sensor, and Bowden cable strain relief were redesigned to improve stiffness and strength, increasing the mass of the lower leg assembly. The additional angle sensor and its ancillary components also increased the mass of the lower leg assembly.

It is standard practice to not include the battery mass in the total device mass to enable the comparison of designs. Not including the weight of the battery, the total device weighed 3.5 kg. This weighs more than predicate designs, which was expected. The weight increase was caused using heavier actuators and a heavier waist interface. The waist interface was required to affix the heavier actuators, electronics, and power supply. This mass is below 4 kg, satisfying the design requirements. Device mass is tabulated with the predicate device below.

Table 2: Predicate and New (High-Power) Device Masses. The high-power device has a slight increase in distal mass and a notable increase in total mass. The total mass increase is caused by heavier actuators and waist interface components. When normalizing the total mass by actuator power, the benefits of the added mass may be seen.

	Predicate	**New**
Actuators	209 grams	490 grams
Waist Interface	177 grams	392 grams
Distal Mass	385 grams	435 grams
Total Mass	2.2 kilograms	3.5 kilograms
Total Mass Normalized by Actuator Power	24.4 grams per Watt	6.48 grams per Watt

Ergonomics

Device ergonomics were validated qualitatively by wearers. There were comments on the device's ergonomics relative to predicate designs. The device used thicker Bowden cables than predicate designs, in order to improve performance and reliability. There were no reported pressure points, excessive friction, or protrusions.

Speed

The speed of the device was validated by walking at the device's maximum target speed, 1.5 m/s, on level ground at the device's maximum target torque, 30.31 Nm. Figure 12 plots the angular velocity for the ankle joint and the actuator as a function of gait phase. By inspecting Figure 12, it is evident that at no point in the gait phase did the actuator velocity saturate, or reach the rated peak velocity, implying that the actuator velocity was not a limiting factor. This satisfied the design requirement of not impeding the wearer.

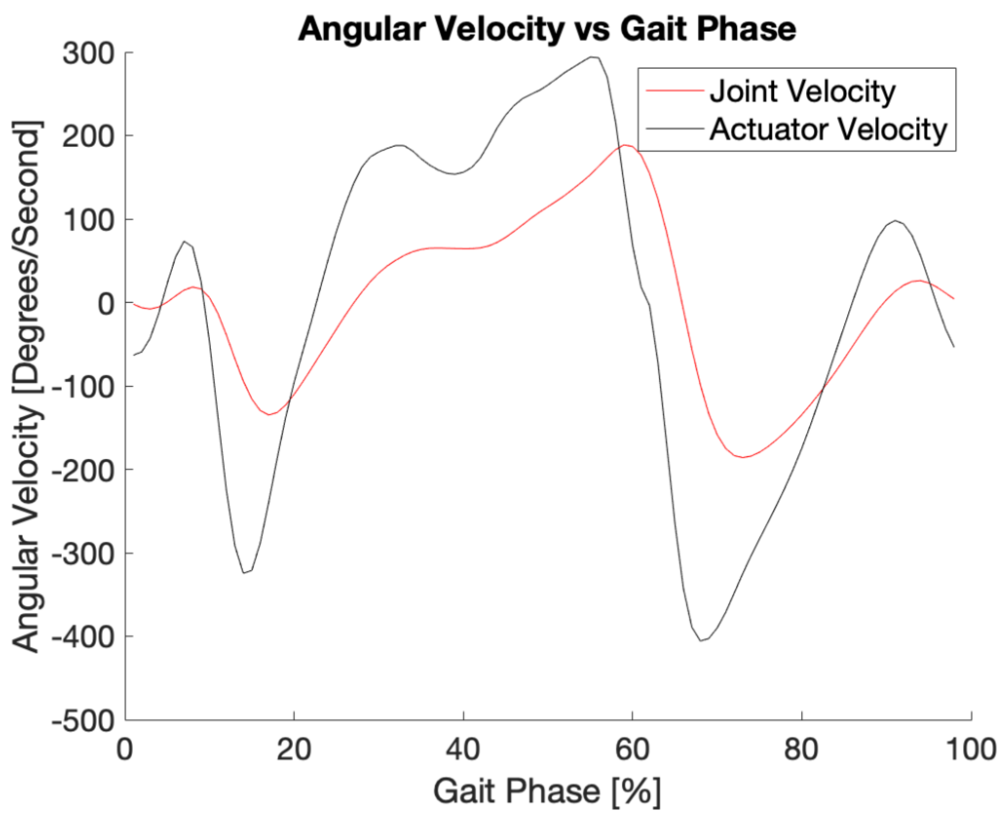

Figure 12: Angular Velocity Vs Gait Phase during level walking at 1.5 m/s. The Angular velocity of the motor does not saturate and is leading the joint velocity. Implying that the joint was not impeded. The maximum actuator angular velocity is not near the actuator's rated maximum. This implies that the actuator may be better geared for energy efficiency.

Peak Torque

To evaluate the peak torque capabilities of the device and low-level controller, data were collected at 1.5 m/s with a plantar torque setpoint of 31 Nm and a dorsi torque setpoint of 3 Nm. These setpoints and walking speeds were selected to test the device at the peak power condition. Ankle angular velocity is positively correlated to walking speed [17], making the fastest walking speed the peak power condition. The peak power condition also tested the abilities of the transmission system and its ancillary components. Under these conditions, the device was able to consistently reach a peak torque of 27.1 Nm. This equates to a -11% error in peak tracking and a 2.75 Nm

RMSE for the entire signal. The peak error can be seen in Figure 13 as the deviation of the desired and actual joint torques.

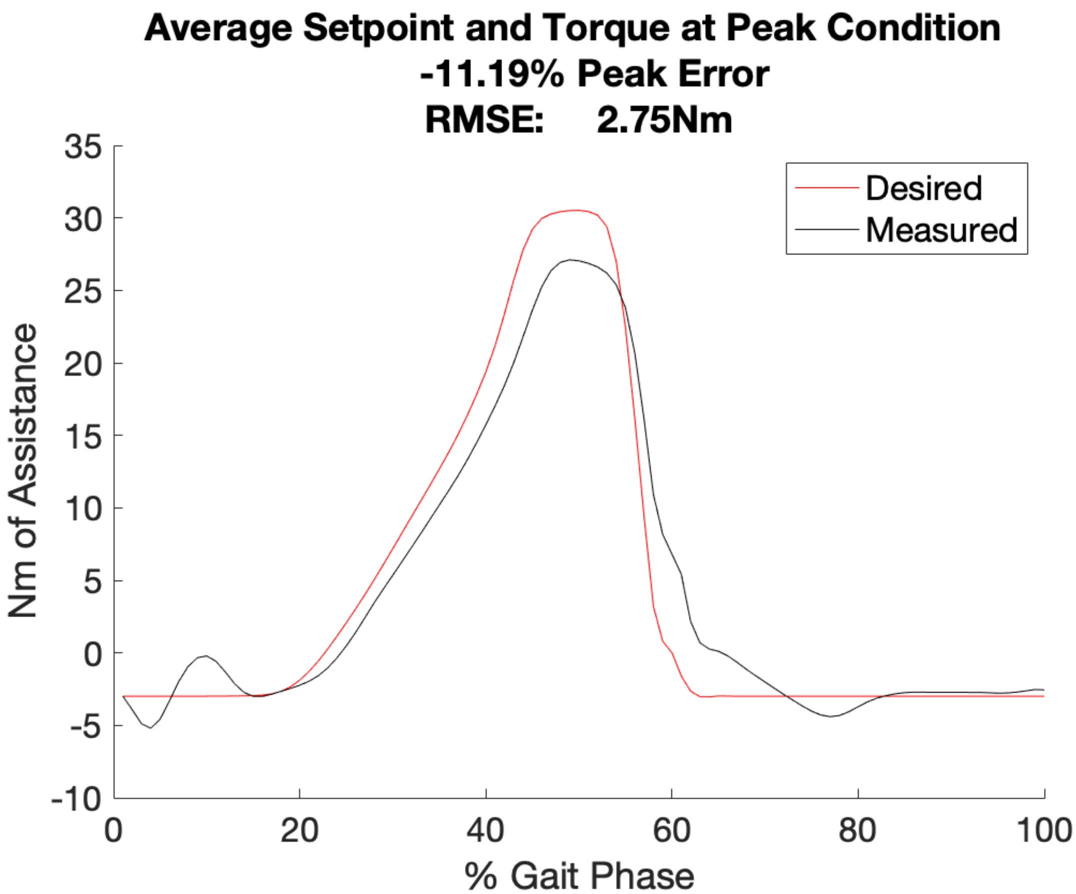

Figure 13: Torque Tracking at the Maximum Specified Device Condition. With a peak tracking error of -11.2%, the device reached peak torques of 27.1 Nm. It was later found that by artificially increasing the setpoint the device was capable of generating torques greater than 31 Nm, implying that this limitation is caused by the low-level controller.

Further testing revealed that when the setpoint was artificially increased, the actuators were able to generate torques greater than 31 Nm. This implied that with continued work the low-level controller could be further optimized to improve tracking and demonstrate targeted peak torque during fast, level walking.

Low-Level Control

The low-level control was validated by collecting desired and measured torque data for a variety of walking conditions. Data were collected at 1 m/s with a setpoint of

23 Nm for several controller configurations and slopes. The average RMSE for the torque tracking was approximately 1.62 Nm. Predicate designs had a torque tracking error of 2.448 Nm during similar conditions. Data were also collected at the device's peak desired performance. Peak torque conditions for the device presented in this thesis and the predicate design are tabulated below. During the peak torque and velocity requirements the root mean square error was 2.75 Nm. Most of these errors result from poor peak tracking. These findings satisfied the design goal for accurate torque tracking.

Table 3: Torque Tracking Comparison. Predicate designs were tested at various torque setpoints at 1.25 m/s. The High-Power design was tested at higher torque setpoints and walking speeds, revealing that the new design has both improved torque tracking and peak performance.

Predicate Design		
Set Torque [Nm]	RMSE [Nm]	Walking Speed [m/s]
15	1.902	1.25
20	2.448	1.25
25	2.984	1.25
High-Power Design		
Set Torque [Nm]	RMSE [Nm]	Walking Speed [m/s]
23	1.62	1.0
31	2.75	1.5

High-Level Control

TREC was validated for its ability to produce different torque setpoints during different terrain slopes. Walking at 1m/s with a 0.35 Nm/kg setpoint, data was collected at 0°, +/- 5°, +/- 10°. For each slope condition, data was collected with only the Generic term enabled, the Generic and Propulsive terms enabled, and the Generic and Supportive terms enabled. For each condition, the average torque setpoint was calculated as a function of the gait phase.

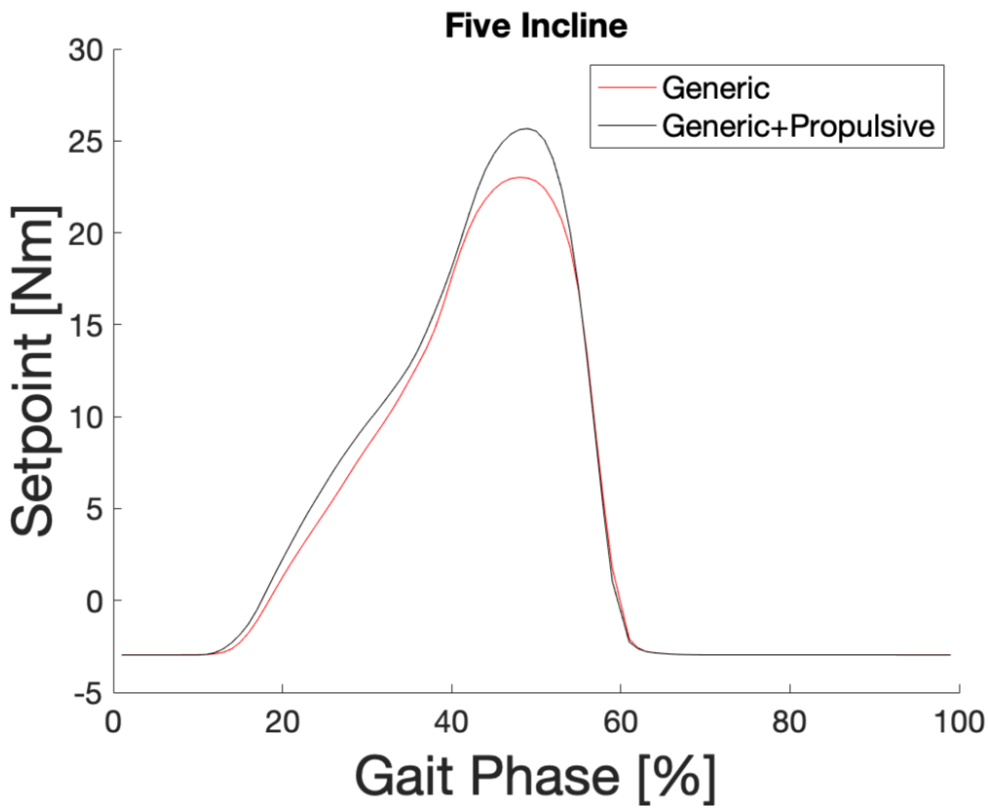

Figure 14: Generic with and without Propulsive Term at five degrees incline, Setpoint Vs Gait Phase. The propulsive term increased the setpoint by 8.1%. It should be noted that even though this percentage may seem small, the torque is occurring at a critical moment in the gait cycle, further approaching optimal assistance.

The propulsive term was designed to add energy to the end of the wearer's stance phase. As shown in Figure 14, the propulsive term added more late stance assistance during level walking. This made the torque profile more like profiles found with globally optimal human-in-loop methods. The propulsive term was also able to add more assistance during incline walking and did not change the assistance during decline walking.

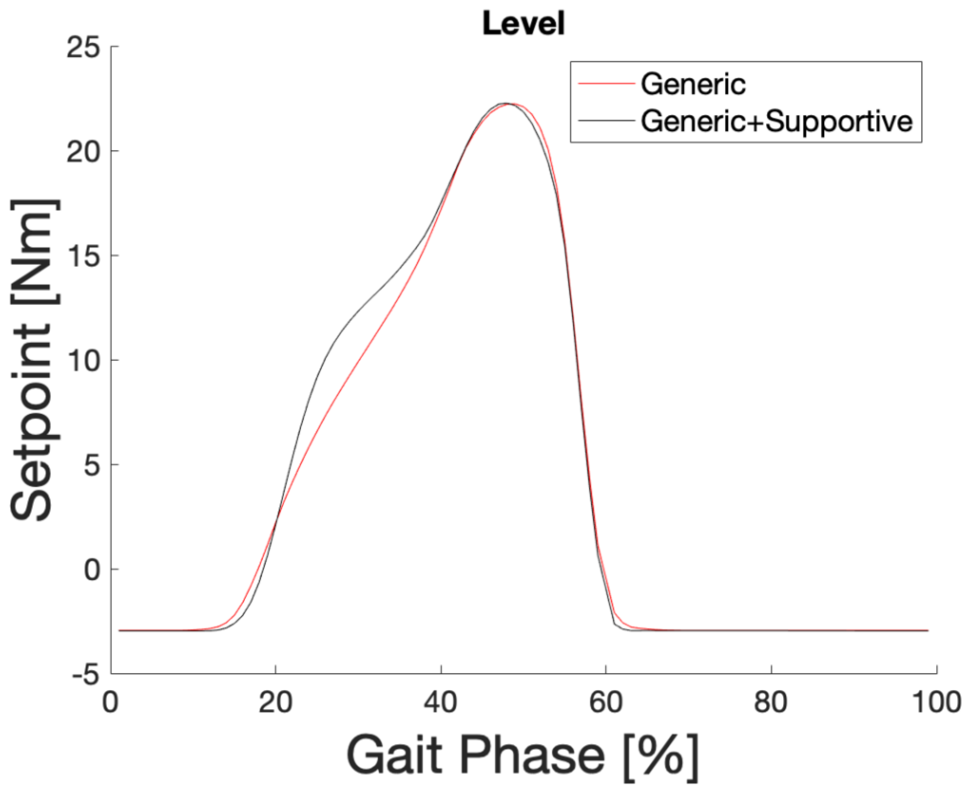

Figure 15: Generic with and without Supportive Term, Setpoint Vs Gait Phase for level walking. The spring term has only modified the setpoint by ~3%. This is beneficial as the spring should not be significantly increasing torque setpoints during level walking. If the spring term were overactive it would result in poorly timed assistance during level walking.

The supportive term was designed to add assistance during the early stance phase of decline walking. Excessive early stance phase assistance during level walking may cause knee hyperextension and should be avoided. Initially, the supportive term was overactive during level walking. Further work was done to add the dynamic offset described in the methods section, and from now on all mentions of the supportive term will include the dynamic offset. During decline walking, the supportive term increased early stance assistance. Ideally, the black and red lines in Figure 15 should be very similar.

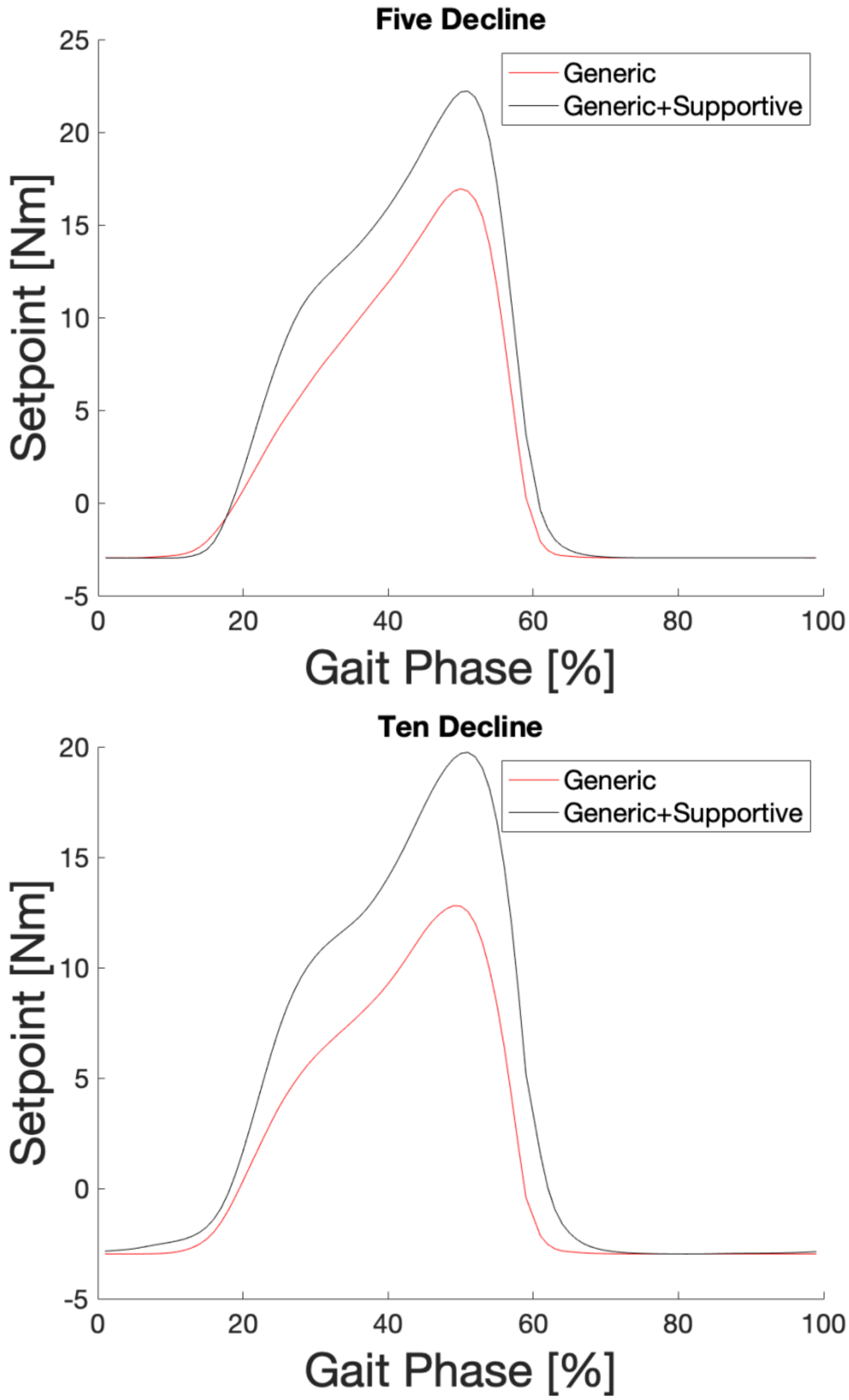

Figure 16: Generic with and without Supportive Term, Setpoint Vs Gait Phase. At five degree decline, the spring term increased the torque setpoint by 37.1%. At ten degree decline, the torque setpoint was increased by 57.7%. This assistance is also occurring further towards the beginning of the stance phase, in line with sloped joint moment data.

Both the supportive term and the propulsive term increased the torque setpoint during their intended condition and gait phase. The supportive and propulsive terms increased assistance as a function of slope, implying that support adjusted dynamically with slope. This can be seen by the changing assistance for five and ten degrees decline in Figure 16. These results are tabulated by their percent change from the generic condition. This satisfies the design requirement for better matching joint moments during sloped walking.

Table 4: Percent change in torque setpoint relative to generic term only for various slopes. The propulsive term should increase assistance for level and positive slopes. The assistive term should increase assistance for negative slopes. Any negative values are likely caused by inter-condition calibration variance.

Slope [degrees]	**Generic + Supportive [%]**	**Generic + Propulsive [%]**
+5	-2.8	8.1
0	3.0	6.1
-5	37.1	-3.7
-10	57.7	-4.1

The controller was also tested for its ability to reduce the activation of a primary plantar flexor muscle, the gastrocnemius. Data were collected using electromyography, or EMG, during level and decline walking. For all slopes data were collected without the exoskeleton (Shod), without the spring engaged, and with the spring engaged at two stiffnesses.

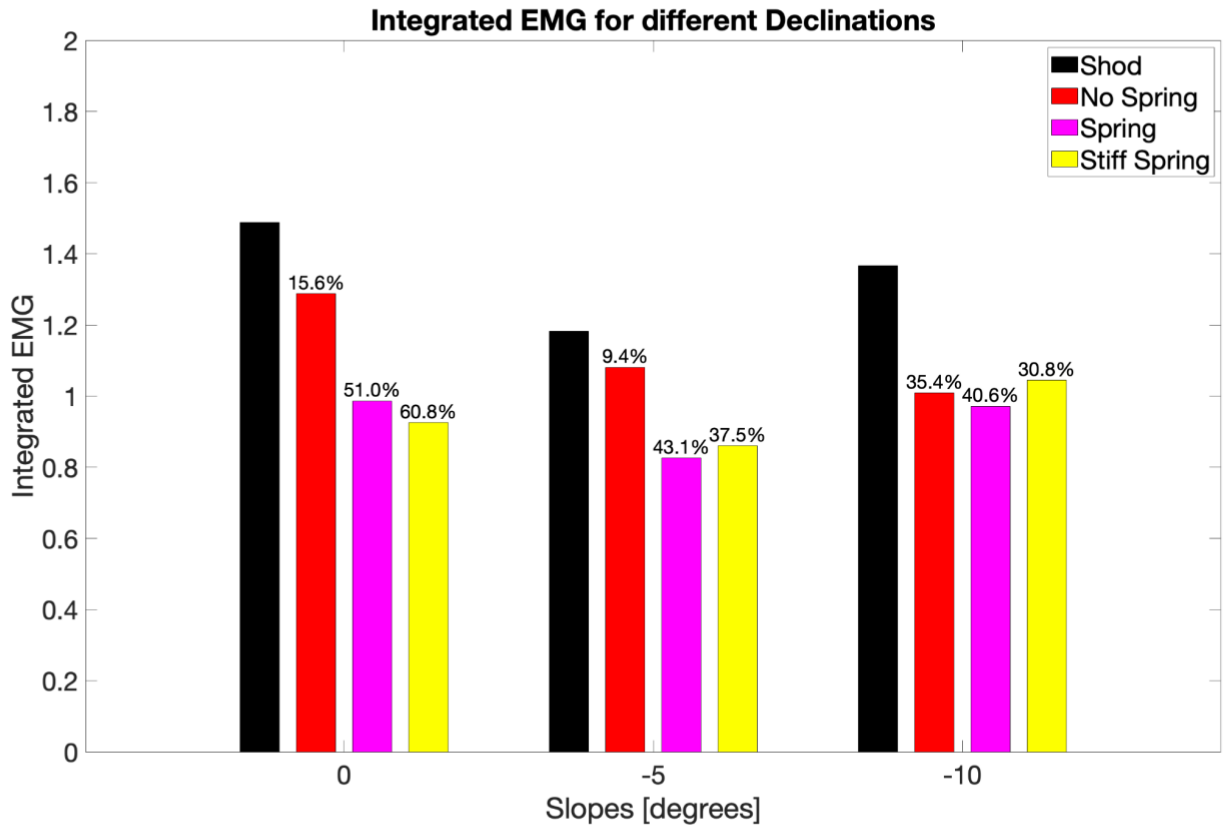

Figure 17: TREC validation, EMG data plotted relative to the shod condition for various grades. Until ten degrees decline, the spring term is further decreasing plantar flexor muscle activation. This suggests that further ankle assistance may not be required for steep grades.

The bar chart displays the percentages for all conditions relative to their respective shod values. For all slopes, there was a decrease in the integrated EMG. During level and negative five-degree walking, engaging the spring term further reduces the EMG activation. However, during the -10° conditions the spring term does not affect EMG activation. This implies that there may be uncaptured joint dynamics or that the joint does not require further assistance. This satisfies the design requirement for reducing plantar flexor muscle activation.

Operation Time

The operation time was validated during the collection of controller validation experiments. After 32 minutes of walking at the peak power condition, the actuators were approaching their rated maximum temperature. Current data were also collected during the peak power condition and were estimated to be ~7.3 amps, permitting a 49-minute operation time when using a 6 Ah battery. This suggests that actuator thermals are limiting the operation time.

The device has been used to collect steady-state metabolic data for five conditions, satisfying the within-lab operation time goal. However, these steady-state data collections call for 10-minute periods of rest between conditions. The outdoor collection of metabolic data does not require these periods of rest and will be affected more by the system's operation time. The device was able to operate for 32 minutes near its peak power condition. However, the trial was stopped when the actuators began approaching their rated peak temperature. This satisfies the design goal of a 30-minute continuous operation.

Summary

The design validation section is necessary to ensure that the work accomplished a set of high-level goals. The three goals of the thesis had separate criteria for success. The modular platform work was validated for its ability to accelerate development. The high-power exoskeleton was validated for its ability to assist the average adult male at 1.5 m/s for 30 minutes. The terrain adaptive controller was validated for its ability to adjust assistance as a function of terrain and for its ability to reduce plantar flexor muscle activation. In a pilot test with one participant, the controller succeeded in both goals.

Table 5: Design Criteria Results. All goals were met other than peak torque. The peak torque failed due to limitations in the low-level controller.

Criteria	Target	Result
Modularity	Accelerate Development	Four new wearable devices
Weight	Less than 4 kilograms	3.5 kilograms
Ergonomics	Near predicate designs	No reported pressure points or friction
Speed	Do not impede intended movement	Does not impede at peak target speed
Peak Torque	30.31 Nm	27.1 Nm, limited by low-level controller
Low-Level Control	Near predicate designs	Improved by ~24%
High-Level Control	Adjust setpoint for all terrains and reduce targeted muscle activation	Increased assistance during decline walking up to 54%, and increased assistance during incline walking by 8%. A reduction in the activation of a primary plantar flexor by ~44%.
Operation Time	30 minutes	32 minutes

Future Work

Actuator Thermals

At the device's most extreme operating conditions, depending on battery capacity, the actuator thermals limit the system runtime. Future designs should better manage actuator heat. This could be done by optimizing the transmission system, changing the actuators, or including a heatsink.

Negative Work Energy Capture

During decline walking the ankle joint exhibits periods of negative power. This power could be harvested to improve the efficiency of the system. The power could be harvested using the natural impedance of the actuators, back driving them to generate electricity. This electricity would be stored in the battery for later use. To do this the system would need to use actuators that can modulate this back drivable impedance.

Hyper-Parameterization

The controller proposed in this thesis could be better parameterized by hyperparameters. If the system were hyper parameterized by the wearer’s weight, the controller would only need to define relationships between parameters. This method may better generalize for different wearers. This method would also be more easily tuned, making the system easier to use for novice operators.

Auto Parameterization

The positive feedback gain for the controller may be auto tuned using only the online system data. This is possible because the system can collect torque and acceleration data online, which can be used to estimate the moment of inertia of the ankle. By performing simple numerical gradient estimates at every step, this metric could be maximized for walking conditions. This method would effectively estimate the shear capacity of the foot-ground system. By ensuring that this auto parameterization can drive the feedback gain negative, this approach has the added benefit of acting as slip detection. Special care, such as asymmetrically weighting the sign of the gradient, may be required for effectively handling slip events. This would improve the ability of the system to generate maximal late stance assistance in terrains with varying shear capacity.

Low-Level Controller Improvements

It may be possible to use an adaptive controller, such as Model-Free Adaptive Control or Extremum Seeking Control, to improve the peak tracking, convergence, and robustness of the low-level controller. The tuning of these controllers may generalize better than a PD controller.

Shear Sensing Control

The current controller uses FSRs placed at the ball of the foot to estimate the ground reaction force (GRF). These sensors are only able to estimate the component of the GRF that is normal to the footplate, which decreases during sloped walking. By additionally placing a thin film shear sensor at the ball of the foot, a more complete estimate of the ground reaction force could be obtained, better-informing control in sloped conditions. This method may also generalize to inversion and eversion assistance of the ankle, which may be required during the most extreme activities.

Conclusion

The first goal of this thesis was to develop a modular exoskeleton system to accelerate the development of future devices. This system has already been used to create four unique devices, and the system has been iterated on several times with minimal training. Many control strategies have been implemented and tested by biomechanists. These new devices have already assisted wearers during tens of thousands of steps. All of this displays its modularity, ease of modification, domain knowledge reduction, and ruggedness.

The second goal of this thesis was to utilize the modular system to build and validate a high-power cable-driven ankle exoskeleton. The design was validated according to the engineering requirements stated in the design criteria section. Failing only the peak tracking criteria. The exoskeleton is also capable of assisting adults during more dynamic activities. Its minimal protrusion, high power, and lightweight make it one of the first ankle exoskeletons that may be able to reduce the metabolic cost of hiking.

The third goal of this thesis was to develop a controller that dynamically varies assistance as a function of terrain slope. To do so, the TREC controller uses the ankle joint angle to modify intent-based torque profiles. TREC's ability to adapt assistance as a function of slope, as well as reduce muscle activation in a primary plantar flexor muscle, was validated during multiple walking conditions. The TREC controller is parameterized to allow for the tuning of its assistance, which will allow researchers to continue to explore solutions for all-terrain exoskeleton assistance.

The work presented in this thesis takes meaningful steps in bringing a class of ultra-lightweight wearable robotics into the real world. Hopefully, the modular platform will continue to accelerate the development of these systems, the high-power ankle exoskeleton will continue to enable research in assistance and control, and the all-terrain controller will provide meaningful assistance to those that require it.

Citations

[1] Sawicki, G. S., Beck, O. N., Kang, I., & Young, A. J. (2020). The exoskeleton expansion: improving walking and running economy. *Journal of Neuroengineering and Rehabilitation*, *17*(1). https://doi.org/10.1186/s12984-020-00663-9

[2] Browning, R. C., Modica, J. R., Kram, R., & Goswami, A. (2007). The Effects of Adding Mass to the Legs on the Energetics and Biomechanics of Walking. *Medicine and Science in Sports and Exercise*, *39*(3), 515–525. https://doi.org/10.1249/mss.0b013e31802b3562

[3] Taylor, C. B., Srinivasan, S., Yeon, S. C., O'Donnell, M., Roberts, T. M., & Herr, H. M. (2021). Magnetomicrometry. *Science Robotics*, *6*(57). https://doi.org/10.1126/scirobotics.abg0656

[4] Zhao, G., Grimmer, M., & Seyfarth, A. (2021). The mechanisms and mechanical energy of human gait initiation from the lower-limb joint level perspective. *Scientific Reports*, *11*(1). https://doi.org/10.1038/s41598-021-01694-5

[5] Duval, J., & Herr, H. M. (2016). FlexSEA: Flexible, Scalable Electronics Architecture for wearable robotic applications. *IEEE International Conference on Biomedical Robotics and Biomechatronics*. https://doi.org/10.1109/biorob.2016.7523800

[6] Slade, P., Kochenderfer, M. J., Delp, S. L., & Collins, S. J. (2022). Personalizing exoskeleton assistance while walking in the real world. *Nature*, *610*(7931), 277–282. https://doi.org/10.1038/s41586-022-05191-1

[7] Gasparri, G. M., Luque, J., & Lerner, Z. F. (2019). Proportional Joint-Moment Control for Instantaneously Adaptive Ankle Exoskeleton Assistance. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, *27*(4), 751–759. https://doi.org/10.1109/tnsre.2019.2905979

[8] Mentiplay, B. F., Banky, M., Clark, R. A., Kahn, M., & Williams, G. (2018). Lower limb angular velocity during walking at various speeds. *Gait & Posture*, *65*, 190–196. https://doi.org/10.1016/j.gaitpost.2018.06.162

[9] Ogden, C. L., Fryar, C. D., Carroll, M. D., & Flegal, K. M. (2004). Mean body weight, height, and body mass index, United States 1960-2002. *Advance Data*, *347*, 1–17.

[10] Baud, R., Manzoori, A. R., Ijspeert, A. J., & Bouri, M. (2021). Review of control strategies for lower-limb exoskeletons to assist gait. *Journal of Neuroengineering and Rehabilitation*, *18*(1). https://doi.org/10.1186/s12984-021-00906-3

[11] Strutzenberger, G., Leutgeb, L., Claußen, L., & Schwameder, H. (2022). Gait on slopes: Differences in temporo-spatial, kinematic and kinetic gait parameters between walking on a ramp and on a treadmill. *Gait & Posture*, *91*, 73–78. https://doi.org/10.1016/j.gaitpost.2021.09.196

[12] Katz, B. (2018b). A low cost modular actuator for dynamic robots. *MASSACHUSETTS INSTITUTE OF TECHNOLOGY*. https://hdl.handle.net/1721.1/126619

[13] Donelan, J. M., Kram, R., & Arthur D., K. (2001). Mechanical and metabolic determinants of the preferred step width in human walking. *Proceedings of the Royal Society of London. Series B: Biological Sciences*, *268*(1480), 1985–1992. https://doi.org/10.1098/rspb.2001.1761

[14] Orekhov, G., Fang, Y., Cuddeback, C. F., & Lerner, Z. F. (2021). Usability and performance validation of an ultra- and versatile untethered robotic ankle

exoskeleton. *Journal of NeuroEngineering and Rehabilitation*, *18*(1). https://doi.org/10.1186/s12984-021-00954-9

[15] Nuckols, Richard W., et al. “Mechanics of Walking and Running up and Downhill: A Joint-Level Perspective to Guide Design of Lower-Limb Exoskeletons.” *PLOS ONE*, vol. 15, no. 8, 28 Aug. 2020, p. e0231996, https://doi.org/10.1371/journal.pone.0231996

[16] Montgomery, J. R., & Grabowski, A. M. (2018, August). The contributions of ankle, knee and hip joint work to individual leg work change during uphill and downhill walking over a range of speeds. *Royal Society Open Science*, *5*(8), 180550. https://doi.org/10.1098/rsos.180550

[17] Mentiplay, B. F., Banky, M., Clark, R. A., Kahn, M. B., & Williams, G. (2018, September). Lower limb angular velocity during walking at various speeds. *Gait & Posture*, *65*, 190–196. https://doi.org/10.1016/j.gaitpost.2018.06.162

[18] Shafrin, J., Sullivan, J., Goldman, D. P., & Gill, T. M. (2017, August 21). The association between observed mobility and quality of life in the near elderly. *PLOS ONE*, *12*(8), e0182920. https://doi.org/10.1371/journal.pone.0182920

[19] Jayaraman, C., Embry, K. R., Mummidisetty, C. K., Moon, Y., Giffhorn, M., Prokup, S., Lim, B., Lee, J., Lee, Y., Lee, M., & Jayaraman, A. (2022, December 30). Modular hip exoskeleton improves walking function and reduces sedentary time in community-dwelling older adults. *Journal of NeuroEngineering and Rehabilitation*, *19*(1). https://doi.org/10.1186/s12984-022-01121-4

[20] Fang, S. (2022). *Biomechanics and Age Group Classification Among Healthy Population with Lower-body Added Mass During Walking* [Doctoral dissertation, University of Dayton]. OhioLINK Electronic Theses and Dissertations Center. http://rave.ohiolink.edu/etdc/view?acc_num=dayton1670839272555752

[21] Fang, Y., Orekhov, G., & Lerner, Z. F. (2022, July). Improving the Energy Cost of Incline Walking and Stair Ascent With Ankle Exoskeleton Assistance in Cerebral Palsy. *IEEE Transactions on Biomedical Engineering*, *69*(7), 2143–2152. https://doi.org/10.1109/tbme.2021.3137447

[22] Seethapathi, N., & Srinivasan, M. (2015, September). The metabolic cost of changing walking speeds is significant, implies lower optimal speeds for shorter distances, and increases daily energy estimates. *Biology Letters*, *11*(9), 20150486. https://doi.org/10.1098/rsbl.2015.0486

[23] Stewart, J. D. (2008, June 1). Foot drop: where, why and what to do? *Practical Neurology*, *8*(3), 158–169. https://doi.org/10.1136/jnnp.2008.149393

[24] Fang, Y., Orekhov, G., & Lerner, Z. F. (2021). Improving the Energy Cost of Incline Walking and Stair Ascent With Ankle Exoskeleton Assistance in Cerebral Palsy. *IEEE Transactions on Biomedical Engineering*, *69*(7), 2143–2152. https://doi.org/10.1109/tbme.2021.3137447

[25] Huang, T. W. P., & Kuo, A. D. (2013, January 1). Mechanics and energetics of load carriage during human walking. *Journal of Experimental Biology*. https://doi.org/10.1242/jeb.091587