Mechanical Hip Prosthetic

Initial Design Report

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Fall 2025-Spring 2026

HORIZON HIP
EMPOWERING EVERY STEP

Faculty Advisors | Sponsor Mentors: Dr. Dante Archangeli, Dr. Reza Razavian
Instructor: Professor David Willy

DISCLAIMER

This report was prepared by students as part of a university course requirement. While considerable effort has been put into the project, it is not the work of licensed engineers and has not undergone the extensive verification that is common in the profession. The information, data, conclusions, and content of this report should not be relied on or utilized without thorough, independent testing and verification. University faculty members may have been associated with this project as advisors, sponsors, or course instructors, but as such, they are not responsible for the accuracy of results or conclusions.

EXECUTIVE SUMMARY

This Mechanical Hip Prosthetic project aims to develop an active powered prosthetic hip designed to improve the lives of individuals who have undergone a hip disarticulation amputation. This device will allow users to drastically improve their quality of life by providing a more energy-efficient, stable, and supportive prosthesis. The project began in August 2025, led by a team of four mechanical engineering students. The team is guided by sponsors Dr. Dante Archangeli and Dr. Reza Razavian. The main objective is to create a powered prosthesis capable of fully supporting a 90 kg individual during walking, stair ascent and descent, as well as sitting and standing. The design will include a natural range of motion within the sagittal plane and compatibility with all standard types of prosthetic knees and sockets. To accomplish this, the team has partnered with Next Step Prosthetics and participated in the NSF I-Corps Aspire Program to gain valuable insights into user needs and functionality.

The current proposed design integrates multiple ideas generated by the team to create a simple and realistic prototype. This device utilizes a standard rotary motor to drive hip movement. To reduce weight, the design incorporates an industry-standard pylon to extend to the knee joint. The device has been refined through a detailed process involving mathematical modeling, SolidWorks design, and MATLAB analysis to create a prosthesis that benefits the user. The design targets motion within the sagittal plane from -20° to 130°, allowing for a natural gait pattern and smooth walking. The goal is for the device to operate for a minimum of 15 minutes before needing to recharge. The system will be controlled through a combination of motor controllers and sensors that automatically detect and determine when to lift.

Current progress within this design has included benchmarking, background research, and multiple rounds of concept selection and development. These steps allowed the team to break the design into multiple parts using functional decomposition and the black box method. Through these processes, the team identified four main subsystems: actuation, mechanism, power transmission, and attachment style. Mathematical modeling was then performed to determine the most effective options for each subsystem, leading to the final concept now in development.

This thorough design process has given the team confidence as they progress into prototype fabrication. The team is currently finalizing motor selection, motor control systems, and material choices. Multiple mathematical models were developed in MATLAB to compute torques for three potential motors: the AKE90-8 KV35, the AK60-39 V3.0 KV80, and the AK80-64 KV80. Additionally, the team identified aluminum, steel, 3D-printed polymers, and carbon fiber as promising materials for construction. A major design criterion is to make the prosthesis as light and simple as possible, improving usability for users and allowing for future development and refinement.

The next major step for the team is to begin prototype fabrication. This includes using PVC pipe to lay out the design, utilizing a Raspberry Pi and LabVIEW to control the motor and read sensor inputs, and maintaining an updated bill of materials and budget. According to the current schedule, the completed device is expected by Spring 2026.

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1 BACKGROUND

Hip disarticulation amputations account for only 1% of all amputees, and yet it is one of the most debilitating, as after the surgery, many people end up permanently stuck within a wheelchair. There have been attempts to create a prosthetic hip that works well for these individuals; however, they all have their own problems. Most are passive designs, which struggle with things like activation, energy draw, and stability. With this design, we will create an active powered design that solves these problems.

1.1 Project Description

This project has been in the works for just over a month. Through it, we have gone through identifying our requirements, a comprehensive literature review, and multiple client and sponsor meetings. The main goal of this design is to create an active hip prosthetic that allows amputees to experience the freedom of energy-efficient mobility again. Since we began, we have kept in consistent contact with our clients, having weekly meetings to update, go over designs, and receive feedback throughout this process. In addition to our clients, we have also had unique opportunities to work with a local prosthesis clinic, Next Step Prosthetics. We have also been able to take an NSF I-Corps Aspire Course, allowing us to learn additional information on how to connect with individuals connected to this project. We have also achieved our fundraising goals. We were given an initial investment of $4500 with a fundraising goal of $450. With the completion of the NSF I-Corps, we will be receiving $3000 to assist us in this project and design. In addition, Next Step Prosthetics has offered us to use their facilities, which include their 3D printers. These resources will be able to assist us in our prosthesis design to improve the quality of life of these amputees.

1.2 Deliverables

The primary goal of designing, building, and testing an active powered hip prosthesis capable of restoring hip motion for individuals who have lost their full leg during hip disarticulation. For this design, the baseline is to have a functional prototype that has active actuation with similar degrees of freedom to a normal leg. This should also be able to be controlled through some sort of control system that the user could control. As this is an upper leg prosthesis just for the hip, it must integrate with a standard prosthetic knee and a standard hip socket. In addition, it would need to be completed by a final report that includes the design specifications, control logic, safety documentation, and recommendations for continued development of the device. Regarding deliverables outside of the device itself, we will need to fundraise at least 10% of our starting investment and keep a detailed bill of materials outlining what money is spent on and where/what it is. To assist in keeping us on schedule, we will need to keep a detailed record of progress and use a schedule we create to keep us on time. This could also be a part of the NIH DEBUT Challenge under the rehabilitative and assistive technologies, which could include deliverables such as video demonstration, technical abstract, and supporting documents. As not all of these deliverables have been confirmed yet, and the competition does not release until the beginning of 2026, we have only recorded these as tentative deliverables to keep in mind.

1.3 Success Metrics

For our design, our success metrics are based on achieving a controlled, realistic prosthesis. Our metrics are based on quantitative and qualitative expectations. Quantitative metrics are things that are specifically measurable by a numerical value. For range of motion, we will aim for a sagittal plane of motion between -20 and 135 degrees. When looking at the torque required from a motor, we need it to be able to lift a 90kg person. Similarly, we need the angular velocity, which is similar to the steps per second, to follow the natural walking cycle. For the weight we will be aiming for as light as possible, with a max weight of 15lb. As for the usage of time for the device, we aim for it to last at least 15 minutes under normal walking conditions. Our qualitative metrics refer to any metric that is not based on a numerical value. This refers to having increased comfortable geometry within our hip prosthetic to increase the feeling of stability. We also want the device to be reliable and to have a smooth gait cycle when the user moves. It is also vital that the device is safe and can be weight-bearing without collapsing. We plan to test these by using a bypass method to measure and test the gait profiles of the device. We will also use engineering standards and FEA loads to ensure safety factors are achieved.

2 REQUIREMENTS

This chapter will include a listing of the requirements that the customers have for us as engineers. After the customer requirements were outlined, they were used to give us quantifiable engineering requirements. From those quantifiable requirements, we created a house of quality to best select our basis for our future design.

2.1 Customer Requirements (CRs)

Our customer is a hip disarticulation patient who seeks to live a more active lifestyle than what is readily available to them with an unpowered hip prosthetic. These requirements include a stable leg, the ability to walk and climb stairs, easy and comfortable use, a non-cumbersome design, efficient battery life, and adaptable use.

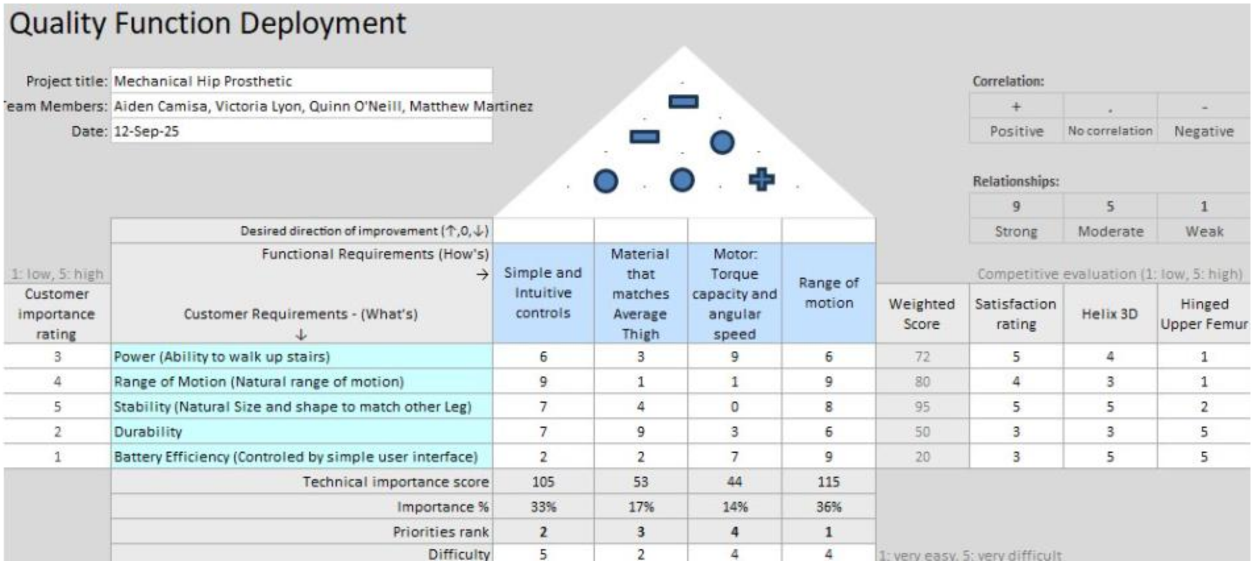
The stable leg implies the leg will be able to hold the weight of a 90kg user in standing and during all stages of the walking gate. The ability to walk and climb stairs implies that the leg will be able to reach a certain range of motion, and the motor will have enough power to allow the customer to ascend stairs. Both easy and non-cumbersome use highly depend on the shape and weight of our design, and is defined as how easy it is for the customer to lift, sit with, and perform everyday tasks with. Efficient battery life means the customer can use the leg throughout the day without a constant need to recharge. Adaptable use will mean that the leg can attach to most other lower leg prosthetics.

2.2 Engineering Requirements (ERs)

With each customer requirement stated, we turn each of these goals into quantifiable goals, or at the very least, better define them within the scope of our project. To make the leg stable to stand and walk on, it will need to be able to regularly withstand the reaction force of walking and the stress that causes. We calculated the maximum stress to be around 64.854 MPa during the walking phase. Our goal for an easy, non-cumbersome design will be defined with a weight that is less than 20lbs and a diameter less than 3in for the primary pylon. Additionally, the length will be around 17in, close to the average upper leg length. For the gate of movement we determined for the leg to best match the necessary gate, it will need a range of -30 to 100 degrees, assuming 0 degrees is standing up straight.

For the Requirements of our motor, we need it to be able to lift the leg to around 100 degrees from standing. The determined minimum torque for this motor will need to be 22Nm, assuming a gear ratio of 1 is used to apply that torque to the leg. Additionally, it would need to be able to reach around .5 rev/s to match walking speeds. The exact battery used will depend on the completed design of the leg, but our goal will be an 8-hour battery life under regular usage.

2.3 House of Quality (HoQ)


Quality Function Deployment

Project title: Mechanical Hip Prosthetic
Team Members: Aiden Camisa, Victoria Lyon, Quinn O'Neill, Matthew Martinez
Date: 12-Sep-25

Correlation: + Positive; . No correlation; - Negative
Relationships: 9 Strong; 5 Moderate; 1 Weak

Desired direction of improvement (↑,0,↓)

1: low, 5: high Customer importance rating	Customer Requirements - (What's) ↓ / Functional Requirements (How's) →	Simple and Intuitive controls	Material that matches Average Thigh	Motor: Torque capacity and angular speed	Range of motion	Weighted Score	Satisfaction rating	Helix 3D	Hinged Upper Femur
3	Power (Ability to walk up stairs)	6	3	9	6	72	5	4	1
4	Range of Motion (Natural range of motion)	9	1	1	9	80	4	3	1
5	Stability (Natural Size and shape to match other Leg)	7	4	0	8	95	5	5	2
2	Durability	7	9	3	6	50	3	3	5
1	Battery Efficiency (Controled by simple user interface)	2	2	7	9	20	3	5	5
	Technical importance score	105	53	44	115				
	Importance %	33%	17%	14%	36%				
	Priorities rank	2	3	4	1				
	Difficulty	5	2	4	4	1: very easy, 5: very difficult			

Competitive evaluation (1: low, 5: high)

Figure 1: Quality Function Deployment

3 Research Within Your Design Space

3.1 Benchmarking

Within the field of powered prosthesis, many different models exist for limbs and joints such as the knee, ankle, shoulder, and wrist. However, there are currently no powered hip prostheses available. Therefore, we selected the most technologically advanced passive hip prosthesis, as well as the baseline, most available model. Further, we include a generic subsystem adapter in our benchmarking process, as our product must be compatible with this system.

The Helix3D Hip Joint by Ottobock [1] is the most advanced system commercially available. Utilizing springs and hydraulics, the system is passively powered by movement and is advertised to create a more even and fluid gait. The system can support up to 100kg and is usable through 130 degrees of motion in the sagittal plane. This system is primarily made up of aluminum components.

Next, the most common type of hip prosthetic available resembles a simple hinge, known as a uniaxial hip prosthetic. While this product is not confined to a specific company, it is a common solution for those who experience hip disarticulation [2]. This model confines motion to the sagittal plane and does not include any kind of passive actuation or mechanism. This system is also able to support 100kg. Additionally, this model is also composed of aluminum.

Lastly, every prosthetic hip joint must have an adapter to connect to a knee joint. It is universally an unofficial standard to implement the use of a pyramid adapter [3]. Able to support up to 181kg and

12N-m of torque, the pyramid adapter is a universal subsystem in the field of prosthetics. The adapter is commonly made of titanium or, occasionally, stainless steel. Our team found it necessary to include this subsystem in our benchmarking to reinforce and ensure that our product will be compatible with the adapter design.

3.2 Literature Review

Each member of our team independently conducted a Literature Review to gain expertise in various topics spanning prosthetics and biomechanics.

3.2.1 Aiden Camisa

[4] "A normative database of hip and knee joint biomechanics during dynamic tasks using anatomical regression prediction methods"

This article talks about the biomechanics of the hip and knee, which will assist in our design by looking at the angles of movement. The authors collected motion capture and force data from healthy adults to establish baseline joint angles, moments, and forces. They used statistical models to predict joint biomechanics based on anatomical parameters. The findings provide standardized data that can serve as a reference for prosthetic design and rehabilitation planning. This database helps clinicians and engineers identify deviations from typical joint mechanics in patients with limb loss or mobility impairments, which can further assist in informing our design.

[5] "GLOBAL STANDARDS FOR PROSTHETICS AND ORTHOTICS"

ISO 7206-8: Endurance performance of stemmed femoral components under cyclic loading.

This is a page based on the general standards of Prosthetics and Orthotics. It is useful to make sure we follow global standards for devices. This resource guides our team to ensure our prosthetic device meets rigorous performance and safety criteria. It is particularly relevant for hip disarticulation prostheses, where mechanical reliability and alignment are critical.

[6] "Hip biomechanics"

This is a page that gives degrees of movement of the hip for all movement angles. The authors highlight how understanding hip mechanics is critical for surgical planning, rehabilitation, and prosthetic design. The paper discusses the kinematics of the hip during various activities, such as walking, running, and jumping. Understanding these biomechanical principles helps optimize gait patterns in amputees, which was identified as a critical aspect by Prosthetist Mentor Mike K.

[7] "Powered ankle-foot prosthesis"

An example of a powered ankle prosthetic, we could use different aspects of a design for a different joint for the hip. The paper highlights improvements in energy efficiency and gait symmetry compared to passive prostheses. It also discusses the challenges of designing robust and lightweight actuators suitable for daily use. Further, it also includes the importance of integrating sensors and control systems in a powered prosthetic.

[8] "The Design, Control, and Testing of an Integrated Electrohydrostatic Powered Ankle Prosthesis"

Another example of an ankle prosthetic, however, utilizes an integrated electrohydrostatic actuation. The paper highlights the importance of precise torque control and real-time adaptability for gait restoration. The findings provide insight into integrating hydraulic and electronic components in lower-limb prostheses, which we have uncovered is no longer a viable option, as advised by our Sponsor Mentors.

[9] "A review of current state-of-the-art control methods for lower-limb powered prostheses"

This is a review of different methods for controlling prostheses, particularly for lower-limb prostheses. The authors discuss the advantages and limitations of each method, focusing on robustness, responsiveness, and energy efficiency. They also highlight human-in-the-loop control strategies that improve gait symmetry and reduce user effort. As our team begins to explore control methods, this provides optimal introductory information.

[10] "Overview of Hip Disarticulation"

A general overview of prostheses for Hip Disarticulation, which we found to be useful for comparing designs and revising work. The author specifically reviews socket designs, suspension methods, and joint mechanisms used in contemporary devices. The paper also discusses challenges faced by amputees, including stability, gait symmetry, and energy expenditure.

3.2.2 Victoria Lyon

[11] "Does the new Helix 3D hip joint improve walking of hip disarticulated amputees?"

A study on comfort and usability for 3 patients, which we found offers insight for design improvements and preliminary preferences. In this study, participants underwent gait analysis before and after receiving the prosthesis. The study highlights the importance of multidirectional motion in restoring functional gait. It also provides empirical data supporting the effectiveness of innovative hip joint designs.

[12] “A pelvic kinematic approach for calculating hip angles for active hip disarticulation prosthesis control”

An in-depth study providing useful information regarding sensors and control systems from the natural body kinematics. Using motion capture and sensor data, the authors developed algorithms to calculate real-time joint angles from pelvic motion. The method enables more accurate control of powered hip joints during swing and stance phases. The findings highlight the importance of integrating kinematic modeling into prosthetic control systems, as opposed to other types of control models.

[13] “Loads in hip disarticulation prostheses during normal daily use”

A static assessment of the prosthetic leg and hip was helpful in mathematical modeling and general comprehension. The authors measured forces and moments during walking, sitting, and stair climbing in a cohort of patients. Findings indicate that prosthetic components are subjected to complex, multi-directional loads that must be considered in design. This was also helpful information to keep in mind for material and mechanism selection.

[14] "Design and prototype validation of a laterally mounted powered hip joint prosthesis"

A Master's student thesis on a laterally mounted power hip, which opposed the common mounting style to the front of a socket. The authors conducted biomechanical tests to validate the range of motion, joint torque, and gait stability. The paper also highlights challenges in actuator integration and weight distribution, which we hope to learn from and integrate into our design thought process.

[15] "Energy expenditure during walking in amputees after disarticulation of the hip. A microprocessor-controlled swing-phase control knee versus a mechanically-controlled stance-phase control knee"

This study is centered on comparing active knee prosthetics in standing or walking. The data collected measured oxygen consumption and gait parameters during treadmill walking. Results showed that microprocessor-controlled swing-phase knees significantly reduced energy expenditure and improved gait efficiency. The study highlights the advantages of integrating smart control technologies into prosthetic limbs, which we again hope to smoothly integrate into our design space as we begin to analyze

[16] “Biomechanical gait analysis for a hip disarticulation prosthesis: power source for the swing phase of a hip disarticulation prosthetic limb”

In this study, the authors analyze where the body draws power to propel movement after hip disarticulation. The results are useful for improving gait symmetry and reducing compensatory movements, which we found to be a major component of patient dissatisfaction from our interview with an individual with Hip Disarticulation.

[17] "Design and optimization of a hip disarticulation prosthesis using the remote center of motion mechanism"

The improvement of walking motion with a prosthetic utilizing the body's center of motion, which is additionally useful in suspension design choices and passive actuation. The authors modeled joint kinematics to reduce the lateral displacement of the prosthetic limb during walking. Testing from this study indicated smoother swing-phase motion and reduced energy expenditure with a remote center of motion.

3.2.3 Matt Martinez

[18] *"Wearable Robotics: Challenges and Opportunities"*

A book spanning wearable robotics, including exoskeletons and powered prostheses. The author discusses engineering challenges such as actuation, control, and human-robot interaction. Case studies illustrate applications in rehabilitation and mobility restoration. The book also addresses opportunities for improving energy efficiency and adaptability in robotic devices, which we have taken into account with design and actuation.

[19] AGMA. "Standards & Emerging Technology *American Gear Manufacturers Association (AGMA)"*

A technical standard with important implications, as our design potentially includes the use of a gearbox. The guidelines ensure durability, precision, and reliability, which are crucial for prosthetic joint mechanisms. It includes specifications for materials, load capacity, and tolerances.

[20] "Overview of the Components Used in Active and Passive Lower-Limb Prosthetic Devices"

A chapter on the technical details of joints, actuators, and sockets to help guide our selection. The chapter also discusses material selection and design considerations, helping us to further understand integration challenges and trade-offs.

[21] "Total Hip Disarticulation Prosthesis with Suction Socket"

A case report that offers practical challenges with sockets, which will help us design a prosthetic that is more socket-adaptable. The study highlights the role of socket fit and suspension in prosthetic performance, which remains relevant in our analysis to determine the best configuration to suspend the joint.

[22] "The biomechanics of trans-femoral amputation"

Important and helps to understand load transfer, gait mechanics, and biomechanical limitations in above-the-knee amputations (hip disarticulations). Findings inform alignment, socket design, and joint selection in prostheses, and by understanding mechanical stresses, we can help prevent overuse injuries and improve mobility.

[23] "A multiple-task gait analysis approach: Kinematic, kinetic, and EMG reference data for healthy young and adult subjects

A study that provides a baseline for gait and force data on healthy subjects. It supports the design of prostheses that restore natural movement patterns. The dataset is particularly useful for validating powered hip and knee devices. In turn, this can also be used to compare against prosthetic gait data to evaluate function quality.

[24] "State of the Art and Future Directions for Lower Limb Robotic Exoskeletons

Highlights current trends and limitations in robotic exoskeletons, which can inform various aspects of our prosthetic design, such as power, weight, and comfort.

3.2.4 Quinn O'Neill

[25] 'EMG Muscle Activation Pattern of Four Lower Extremity Muscles during Stair Climbing, Motor Imagery, and Robot-Assisted Stepping: A Cross-Sectional Study in Healthy Individuals"

This study investigates control strategies for powered transfemoral prostheses during stair ascent. The authors designed and tested algorithms that coordinate knee and hip actuation based on real-time sensor feedback.

[26] "On the design evolution of hip implants: A review"

Study on hip implants, detailing materials and systems best used to replace a hip. This will allow us a basic idea of how to best connect our prosthetic to the hip of a patient. The findings highlight the growing demand for advanced prosthetic technologies. Understanding population trends supports innovation in high-performance prostheses.

[27] "A novel hip joint prosthesis with uni-directional articulations for reduced wear"

The unidirectional hip articulator is the most basic benchmark, as examined in the benchmarking sections.

[28] "Ground reaction forces at different speeds of human walking and running"

This paper examines the biomechanics of gait in individuals with and without prosthetic limbs using ground reaction force analysis. The authors identify characteristic differences in loading, propulsion, and balance between the two groups.

[29] "Shijiazhuang Perfect Prosthetic Manufacture Co., Ltd."

The upper-to-lower leg pyramid adapter, a common adapter found on lower limb prosthetics, is a good benchmark for designing the lower attachment for our prosthetic, as mentioned in the benchmark.

[30] "(PDF) A review of gait cycle and its parameters"

An analysis of the stages of the walking gait cycle, in which subjects demonstrate how a bionic ankle–foot prosthesis can restore near-normal walking biomechanics. The authors used a powered system comparable to biological muscles. Results showed improved gait symmetry, reduced metabolic cost, and enhanced walking speed.

3.3 Mathematical Modeling

3.3.1 Torque Analysis for Hip Flexion – Matt Martinez

To estimate the torque required for hip flexion, the leg was modeled as a rigid body subjected to gravitational loading and joint geometry constraints. Two equations were used to describe the static torque at the hip:

$$\tau = rFsin\theta \tag{1}$$

Where τ is the joint torque (N·m), F is the applied muscle or actuator force (N), and r is the effective moment arm length (m).

The static torque on the hip was then modeled as:

$$\tau_{hip}(\theta,\beta) = m_{thigh}gr_{thigh}sin\theta + m_{shank}g(L_1sin\theta + r_{shank}\sin(\theta-\beta)) + m_{foot}g(L_1sin\theta + L_2\sin(\theta-\beta)) \tag{2}$$

Where m is the mass (kg), g is gravitational acceleration (9.81 m/s²), L is the length (m), θ is the hip flexion angle (°), β is the knee flexion angle (°), and r is the center of mass (m)

These equations were used to calculate the torque required to lift the leg at various knee flexion angles (0, 60, 90), where the hip angle is 90. The plot shows that static hip torque decreases from 58.3 N·m at 0° knee flexion to 45.8 N·m at 90°. MATLAB was used for data processing and curve generation.

3.3.2 Power and Battery Analysis – Aiden Camisa

This analysis outlines the electrical power and battery sizing requirements for an active hip prosthetic. Mathematical modeling using sinusoidal approximations for hip motion and torque interpolations helps to set requirements for the battery. Equations necessary are:

$$\theta(t) = \theta_0 + A\sin(\frac{2\pi t}{T}) \tag{3}$$

$$\omega(t) = \frac{d\theta}{dt} = A \cdot \frac{2\pi}{T}\cos(\frac{2\pi t}{T}) \tag{4}$$

$$E_{step} = \int_0^T \frac{\max(P_{mech}(t),0)}{\eta_{motor}} dt + P_{anc}T \tag{5}$$

Where $\theta(t)$ is the angle with respect to time, θ_0 is the initial angle, $\omega(t)$ is the angular velocity (rad/s), P_{mech} is the mechanical power (W), and E_{step} is the electrical energy per step. The results are in the table below:

Time (Min)	Battery required (Wh)
10	21.73
20	43.46
30	65.19
45	97.79
60	130.38
90	195.57

Table 1: Battery Sizing for Active Hip Prosthetic

3.3.3 Forces on Leg Due to Walking – Quinn O'Niell

This analysis provides critical insights into the peak loads experienced by the leg during normal walking. Reaction force is approximately 1.5 times body weight during heel strike/toe-off, and body weight at mid-stance for a 90 kg mass. – Foot length: 24.16 cm, shank length: 39 cm, foot at 90° to shank, heel/toes at 20° to ground. Reaction forces are modeled by:

$$R_1 = 1.5mg \tag{6}$$

$$R_2 = mg \tag{7}$$

R_1 is the reaction force during heel strike and toe-off (N), R_2 is the reaction force at mid-stance (N), m is the mass of the individual (kg), and g is gravitational acceleration (9.81 m/s²). A static equilibrium analysis is then performed to determine force and moment distributions and a geometric modeling of leg segments during gait phases (heel strike, mid-stance, toe-off):

$$\sum F = 0 \tag{8}$$

$$\sum M_k = 0 \tag{9}$$

where ΣF is the sum of forces (N) and ΣM_k is the sum of moments about the knee (Nm).

$$F_t = F_k \sin\theta \tag{10}$$

$$F_a = F_k \cos\theta \tag{11}$$

F_t is the transverse force (N), F_a is the axial force (N), F_k is the force at the knee (N), and θ is the angle of the force (degrees). - Maximum force: 1.324 kN during heel strike and toe-off

The maximum force of 1.324 kN and moment of 176.65 kNm highlight the need for robust materials and designs capable of withstanding these dynamic loads.

3.3.4 Stress and Strain on a Prosthetic Leg – Victoria Lyon

This is a stress-strain analysis using material properties (Young's modulus), as well as load distribution modeling for standing and walking scenarios, aiding in material selection. Equations used include:

$$\sigma_{axial} = \frac{F}{A} \tag{12}$$

$$\sigma_{bending} = \frac{Mc}{I} \tag{13}$$

$$\varepsilon = \frac{\sigma_{max}}{E} \tag{14}$$

Where σ_{axial} is axial stress (MPa), F is the applied force (N), A is the cross-sectional area (m^2), $\sigma_{bending}$ is bending stress (MPa), M is moment (Nm), c is distance from neutral axis to outer fiber (m), I is moment of inertia (kgm^2), ϵ is strain, σ_{max} is maximum stress (MPa), and E is Young's modulus (MPa).

While standing σ_{axial} = 5.0185 MPa, $\epsilon = 7.169(10^{-5})$. While walking σ_{axial} = 7.528 MPa, $\sigma_{bending}$ = 57.317 MPa, σ_{max} = 64.845 MPa, and ϵ = 0.0009262. The stress and strain calculations reveal that walking imposes significantly higher bending stresses compared to standing, with maximum stress reaching 64.845 MPa. This underscores the importance of using high-strength materials like aluminum with appropriate thickness.

3.3.5 Static Force Analysis on Attachments – Aiden Camisa

The static analysis compares various attachment configurations, showing that dual attachments minimize stress concentrations. Static force and moment equilibrium equations are used:

$$F = mg \tag{15}$$

$$M = Fd \tag{16}$$

Where F is the total force (N), m is the mass (kg), g is gravity (9.81 m/s^2), M is the moment (Nm), and d is the distance between bolts (m). Below is the free-body diagram that was used:

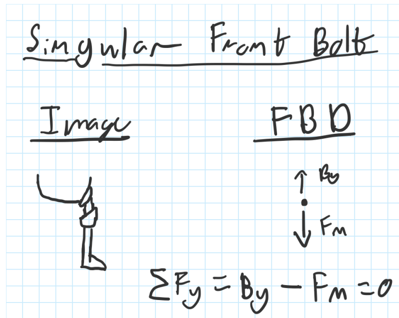


Figure 2: Free-body Diagram of Prosthetic

Then the force is used to find the force per bolt (F_b) and shear stress (σ_{shear}):

$$F_b = \frac{F}{n} \tag{17}$$

$$\sigma_{shear} = \frac{F_b}{A} \tag{18}$$

Shear stress is in MPa, force per bolt is in N, n is the number of bolts, and A is the area (m^2). The table below shows the different attachment used and their respective values.

Type	Force (N)	Moment (N*m)	Force per Bolt (N)	Bearing Stress (MPa)	Shear Stress in the bolt (MPa)
Dual Attachment	882.90	52.974	441.45	13.80	8.78
Laterally Mounted	882.90	70.632	882.90	27.59	17.56
Singular Front Bolt	882.90	88.29	882.90	27.59	17.56
Angled Corner	882.90	105.948	882.90	27.59	17.56

Table 2: Comparison of Attachment Designs

The lowest shear stress of 8.78 MPa in dual setups enhances structural integrity and is optimal for load distribution, reducing failure risk and improving prosthetic stability under body weight loading.

3.3.6 Actuation Static Force Analysis – Quinn O'Neill

This analysis determines the force needed for actuators to achieve a 100° hip motion range by using moment and force balances for selecting actuators (hydraulic, pneumatic, and electronic linear). The equations that model this are equations (8),(9), (15), and (16). Center of mass (COM) percentages are 43% for the thigh, 43% the shank, and 42% for the foot. The results yield a required actuator force of 142.8 N for a 42.854 Nm moment.

The 142.8 N requirement allows selection from various actuator types, with cost as a deciding factor. All three actuator types (hydraulic, pneumatic, electronic) are viable, but the cheapest reliable option should be chosen for cost-effective design without compromising performance.

3.3.7 Joint and Motor – Matt Martinez

The joint and motor analysis models human hip motion to select appropriate motors and gearing. The torque of the hip (τ_{hip}) is given from a data set, and then body weight (BW) is factored in.

$$\tau_{hip} = -BW \cdot \tau \tag{19}$$

Other variables calculated for the hip joint are angular velocity (ω_{hip}) in rad/s and power (P_{hip}) in watts. θ is the hip angle in degrees.

$$\omega_{hip} = \frac{d\theta}{dt} \tag{20}$$

$$P_{hip} = \tau_{hip} \cdot \omega_{hip} \tag{21}$$

Results were calculated and then modeled using MATLAB. Below are the kinematics at the hip joint during normal gait over one cycle (1.2 seconds):

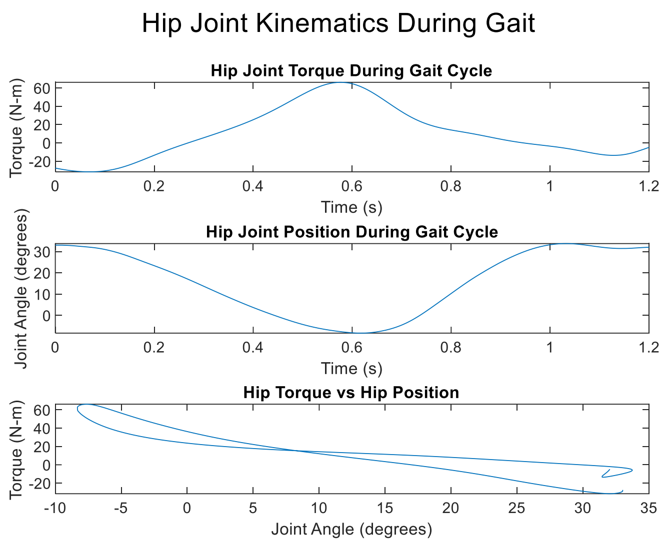


Figure 3: Hip Joint Kinematics During Gait Cycle

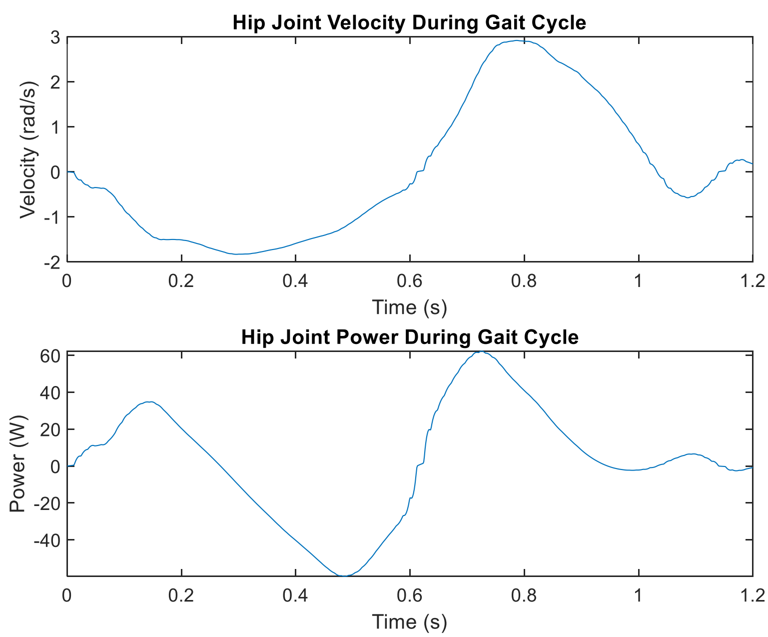


Figure 4: Hip Joint Velocity and Power During Gait Cycle

Next, the specifications from the chosen motor are used to find the current (I) in amps, voltage (V) in volts, and power (P) in watts.

$$I = \frac{\tau_m}{k_t} \tag{22}$$

$$V = IR + k_t \omega_m \tag{23}$$

$$P = V \cdot I \tag{24}$$

τ_m is motor torque (Nm), k_t is the torque constant (Nm/A), R is resistance (Ω), and ω_m is the motor angular velocity (rad/s). The following graphs show these models:

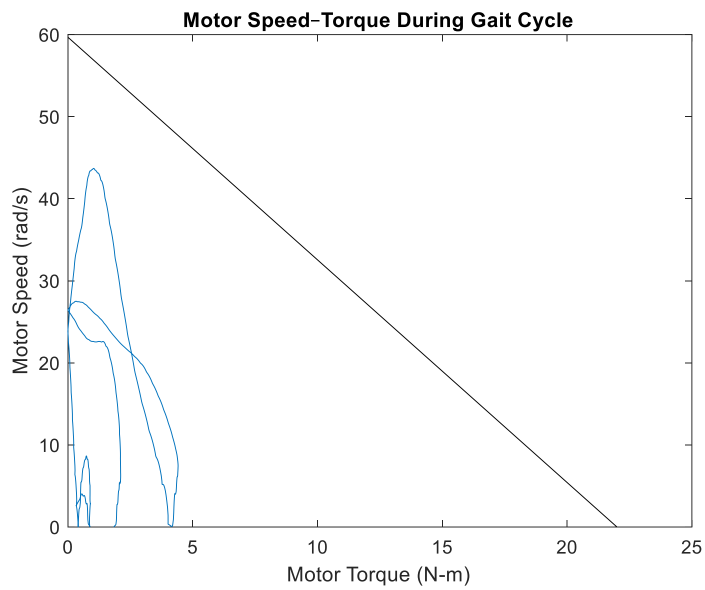


Figure 5: Motor Speed-Torque During Gait

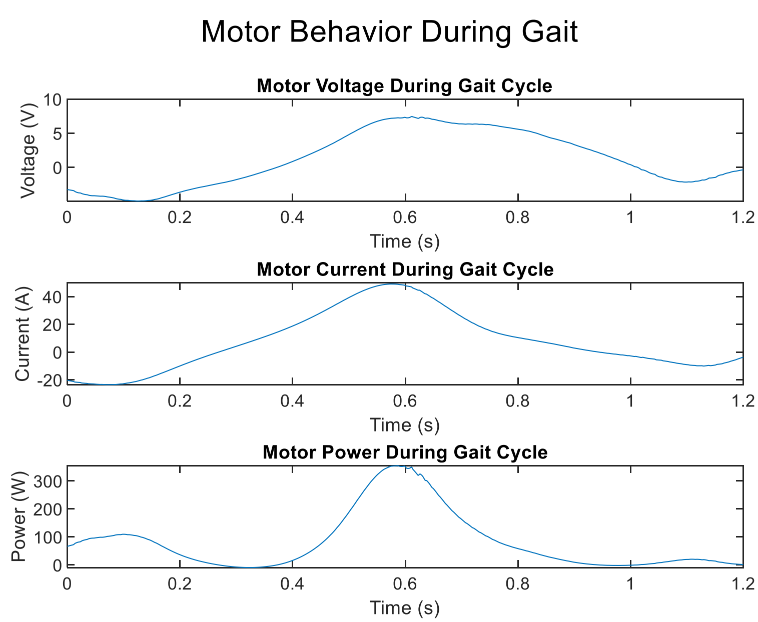


Figure 6: Motor Behavior During Gait

The AK80-9 motor meets torque and power needs, but high gear ratios increase inefficiency.

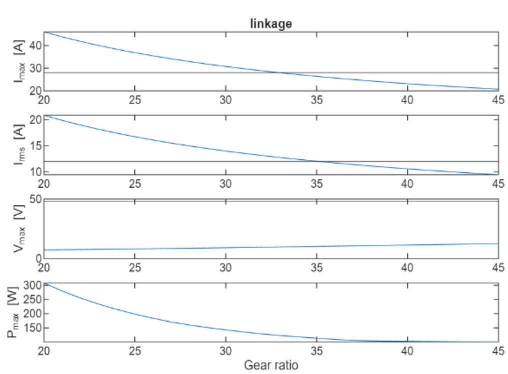
3.3.8 Mechanism Analysis – Victoria Lyon

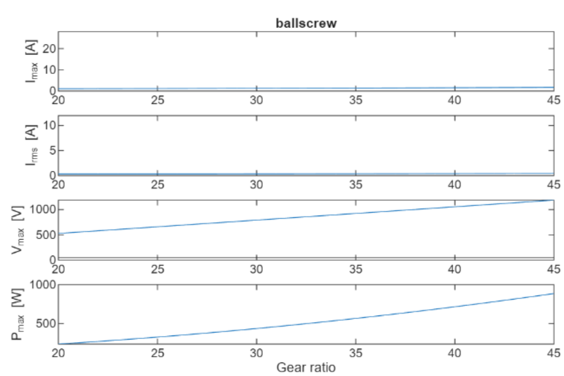
This analysis evaluates mechanisms for transmitting motor torque effectively. Each mechanism's torque formula aids in selecting the best for support and efficiency. The torque calculations are based on applied force, geometry, and mechanism type – the selections being link, ball screw, universal joint, and Stewart platform. The table below shows each mechanism and its corresponding equation:

Mechanism	**Equation Used**
Mechanical Link	$T = F * l$
Ball Screw	$T = (F * l)/2\pi$
Universal Joint	$T = P/n$
2 DOF Stewart Platform	$T = F * 2 * l$

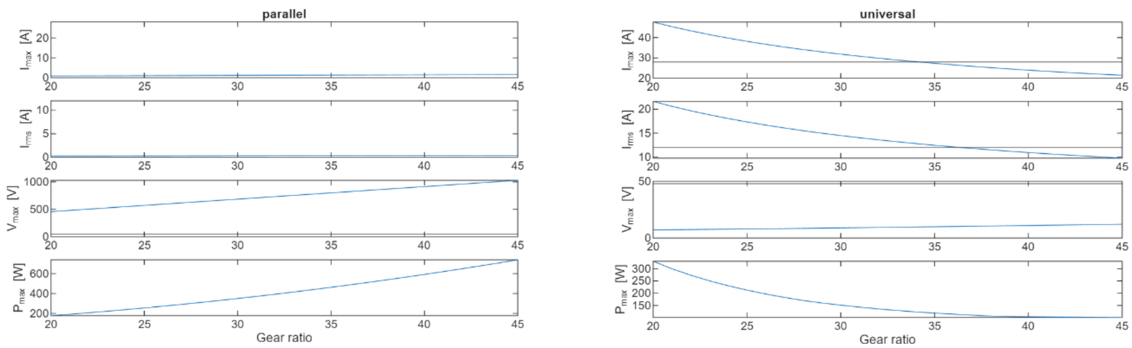
Table 3: Mechanisms and Equations

Where T is torque, F is force, l is length, P is power, and n is speed. The results are modeled below:





Figures 7 and 8: Current, Voltage, and Power of Linkage and Ballscrew Mechanisms with Respect to Gear Ratio



Figures 9 and 10: Current, Voltage, and Power of Parallel and Universal Mechanisms with Respect to Gear Ratio

Ball screws and universal joints offer efficient torque transmission with lower losses, ideal for prosthetic applications requiring precise control and minimal bulk.

4 Design Concepts

4.1 Functional Decomposition

Component	1	2	3	4
Actuation	Electronic Linear actuator	Series Elastic Actuators	Rotary Actuator	Variable Stiffness Actuator
Power Transmission	Gear system	Cable	Electrostatic clutch	Belt
Mechanisms	Stewart platform with 2 Links	Ball Joint	Universal Joint	Rigid Links
Suspension / Attachment Configuration	Dual attachment [2 components bolted to the socket]	Lateral [side socket attachment]	Singular front bolt [typical use]	Angled alignment [lower corner attachment,

Table 4: Functional Decomposition Table

This is a chat detailing each of our important criteria for function, and 4 different possibilities for each criterion. Our first criterion is actuation, a means to extend and contract parts of the leg to best fit the movement of the walking cycle. Next is Power transmission, which is how we will get the motor to connect to and power the movement of the leg. Third is mechanisms, which are means to connect each component of the leg and still allow for freedom of movement within our specified range. Finally, our method of attaching the leg to the prosthetic cast.

4.2 Concept Generation

Concepts were not individually drawn, but were all understood by each team member.

For means of actuation, we found that the hydraulic was the strongest but most expensive, and the electronic linear actuator was the weakest and cheapest. The variable stiffness and rotary were early concepts, and it was decided that they would not work with our design goals, as they would require too much power to regularly activate.

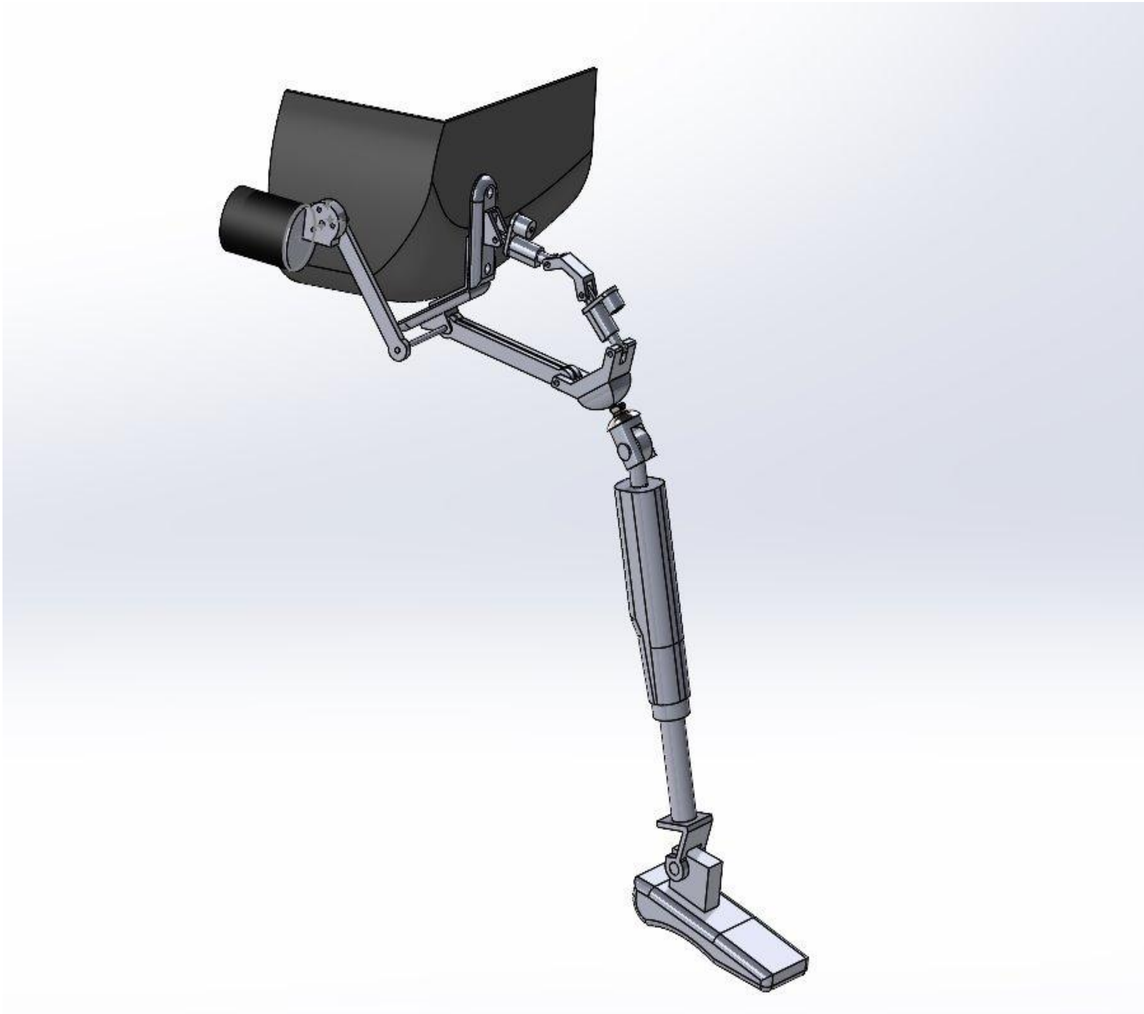
For means of power transmission, we thought belts and gears would both be simple and allow for applicable speed ratios, with the cable being a bit simpler to design but louder. The cable could be used, but would need to be run through pulleys to pull the leg at a point of maximum efficiency.

For the Mechanisms, the Stewart system seemed appealing as it is used in other robotic limbs to move in all 3 dimensions, but was too large to fit into our prosthetic. Both the ball and universal would be best to rotate in all directions, but that isn't the scope of our design. With the mechanical linkages only allowing for 1 direction of rotation, but being the simplest.

Lastly, for means of suspension, the lateral side socket is closest to the sagittal center of rotation, but is awkward to walk around on as it is loaded to the side of where a sound leg would be. The lower corner would technically be the most structurally sound loading of the leg when standing, but it is hard to lift. And both the front and dual attachments are good at extending in the sagittal, but create a slightly awkward balance as it is unaligned from the sound leg.

4.3 Concept Selection

Our decisions on the final components were not made with tables but by discussion between the team and our mentors. We decided to use electronic linear actuators for our means of actuation as they used the least energy and were the cheapest, and were still able to meet the required force of 148.2N. The decided means of power transmission was gears, as the team knows the most about them already due to our experience in machine design. We went with the mechanical linkage's ads as our mechanism due to its simplicity, while still being able to reach our 2D rotational goal. Lastly, we went with the dual attachment as the front loading is used across other prosthetics, as other patients have described it as easy to use, and the dual attachment will allow the stress to be better dispersed across a mold.



Model 1: Most Recent CAD Model of Design

5 CONCLUSION

This report summarizes the progress and outcomes of our semester-long effort to design an active, powered hip prosthesis for individuals with hip disarticulation amputations. Although these amputations account for only about 1% of all limb losses, they present some of the most severe mobility challenges. Many individuals who undergo this procedure lose the ability to walk independently and are confined to wheelchairs due to the lack of effective prosthetic options. Existing hip prostheses are predominantly passive designs that rely heavily on the user's momentum and upper-body control. These systems often fail to provide adequate stability, energy efficiency, or assistance during tasks such as walking, climbing stairs, or rising from a seated position. The primary objective of our project has been to address these limitations by developing a powered hip prosthesis capable of restoring a greater degree of natural, energy-efficient mobility.

Throughout this semester, with the help of our project mentors, we've established clear customer and engineering requirements to guide our design process. Our clients/ sponsor mentors require a prosthesis that is stable, efficient, and comfortable to use. The design must provide a stable leg capable of supporting a user weighing up to 90 kg throughout all phases of gait, including stance, swing, and transitions such as standing or sitting. The prosthesis must enable natural walking and stair climbing, requiring a sufficient range of motion and power output from the actuator. We determined that a hip joint capable of achieving a range of –30° to 100° of motion relative to the neutral standing position would best replicate the necessary gait cycle. Further, the motor supplying power must provide a minimum torque of 22N-m.

Our team's collaboration with Next Step Prosthetics and participation in the NSF I-Corps Aspire Course have been incredible assets for both the technical and financial aspects of the project. Through NSF I-Corps, we have gained insight into customer discovery and evaluation, which has been useful for gaining lots of differing insights to inform our design, as well as many networking opportunities for future potential sponsorships. Further, in working closely with Next Step Prosthetics, we have gained invaluable access to materials, knowledge and mentorship, tools and workspaces, and direct contact with a hip disarticulation patient. The combined financial and material support—totaling approximately $8,000 in funding and resources—positions our team exceedingly well, and we are confident to continue our design and developmental processes.

To assess design criteria and development, a mathematical model was developed, assessing criteria ranging from static force assessments to motor torque and power requirements.

The final proposed solution is a single-plane powered hip prosthesis designed to operate primarily in the sagittal plane, which corresponds to the natural forward and backward motion of the leg during walking. By focusing on a single degree of powered motion, the design minimizes unnecessary weight and complexity while maximizing performance for high-demand tasks such as walking, stair ascent, and transitioning from sitting to standing. The final electrically actuated system will include a sensor-driven control system as well, to help create a customizable gait for natural cycles and improve system adoption.

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7 APPENDICES

7.1 Appendix A: Motor Analysis MATLAB

```
BW = 90;
torque = -BW*bovi.adult.normal.hip.sagittal.torque(:,2);
rms(torque) % for hip joint for continuous / gear ratio
position = bovi.adult.normal.hip.sagittal.position(:,2);
stride_time = 1.2;
time = linspace(0, stride_time, length(torque));
close all
subplot(311); plot(time,torque); % Flexion
subplot(312); plot(time,position); % Extension
subplot(313); plot(position, torque);
pos_rad = deg2rad(position);
vel_rad = dfdx(time,pos_rad);
acc_rad = dfdx(time,vel_rad);
close all
subplot(211); plot(time, vel_rad)
subplot(212);plot(time, vel_rad.*torque)
GR = 15;
motor_inertia = 579/10^7;
kt = .09;
R = .160;
vel_motor = vel_rad*GR;
tor_motor_static = torque/GR;
tor_motor_dynamic = motor_inertia*acc_rad*GR;
current = (tor_motor_static + tor_motor_dynamic)/kt;
volts = current*R + kt*vel_motor;
close all;
subplot(3,1,1); plot(time, volts);
subplot(3,1,2); plot(time, current);
subplot(3,1,3); plot(time, volts.*current)
close all;
% speed torque curve
plot(abs(current*kt), abs(vel_motor)); ylabel("motor speed rad/s"); xlabel("Motor
Torque Nm"); hold on;
w_limit = [570, 0]*2*pi/60;
t_limt = [0, 22];
plot(t_limt, w_limit, 'k'); hold off
GR = 25:45;
motor_inertia = 579/10^7;
kt = .09;
R = .160;
```

```
vel_motor = vel_rad.*GR;
tor_motor_static = torque./GR;
tor_motor_dynamic = motor_inertia.*acc_rad.*GR;

current = (tor_motor_static + tor_motor_dynamic)./kt;
volts = current.*R + kt.*vel_motor;
e_power = current.*volts;

i_max = max(abs(current));
i_rms = rms(current);
v_max = max(abs(volts));
p_max = max(abs(e_power));

close all;
subplot(411); plot(GR, i_max); ylabel("Amps"); hold on; yline(28); % peak current
limit
subplot(412); plot(GR, i_rms); ylabel("Amps"); hold on; yline(12); % continuous
current limit
subplot(413); plot(GR, v_max); ylabel("Volts"); hold on; yline(48); % votlage
limit
subplot(414); plot(GR, p_max); ylabel("Power");
```

7.2 *Appendix B: Modeling Equations*

$$\tau = rFsin\theta \tag{1}$$

$$\tau_{hip}(\theta, \beta) = m_{thigh} g r_{thigh} sin\theta + m_{shank} g(L_1 sin\theta + r_{shank} \sin(\theta - \beta)) + m_{foot} g(L_1 sin\theta + L_2 \sin(\theta - \beta)) \tag{2}$$

$$\theta(t) = \theta_0 + Asin(\frac{2\pi t}{T}) \tag{3}$$

$$\omega(t) = \frac{d\theta}{dt} = A \cdot \frac{2\pi}{T} \cos(\frac{2\pi t}{T}) \tag{4}$$

$$E_{step} = \int_0^T \frac{\max(P_{mech}(t), 0)}{\eta_{motor}} dt + P_{anc} T \tag{5}$$

$$R_1 = 1.5mg \tag{6}$$

$$R_2 = mg \tag{7}$$

$$\sum F = 0 \tag{8}$$

$$\sum M_k = 0 \tag{9}$$

$$F_t = F_k \sin\theta \tag{10}$$

$$F_a = F_k \cos\theta \tag{11}$$

$$\sigma_{axial} = \frac{F}{A} \tag{12}$$

$$\sigma_{bending} = \frac{Mc}{I} \tag{13}$$

$$\varepsilon = \frac{\sigma_{max}}{E} \tag{14}$$

$$F = mg \tag{15}$$

$$M = Fd \tag{16}$$

$$F_b = \frac{F}{n} \tag{17}$$

$$\sigma_{shear} = \frac{F_b}{A} \tag{18}$$

$$\tau_{hip} = -BW \cdot \tau \tag{19}$$

$$\omega_{hip} = \frac{d\theta}{dt} \tag{20}$$

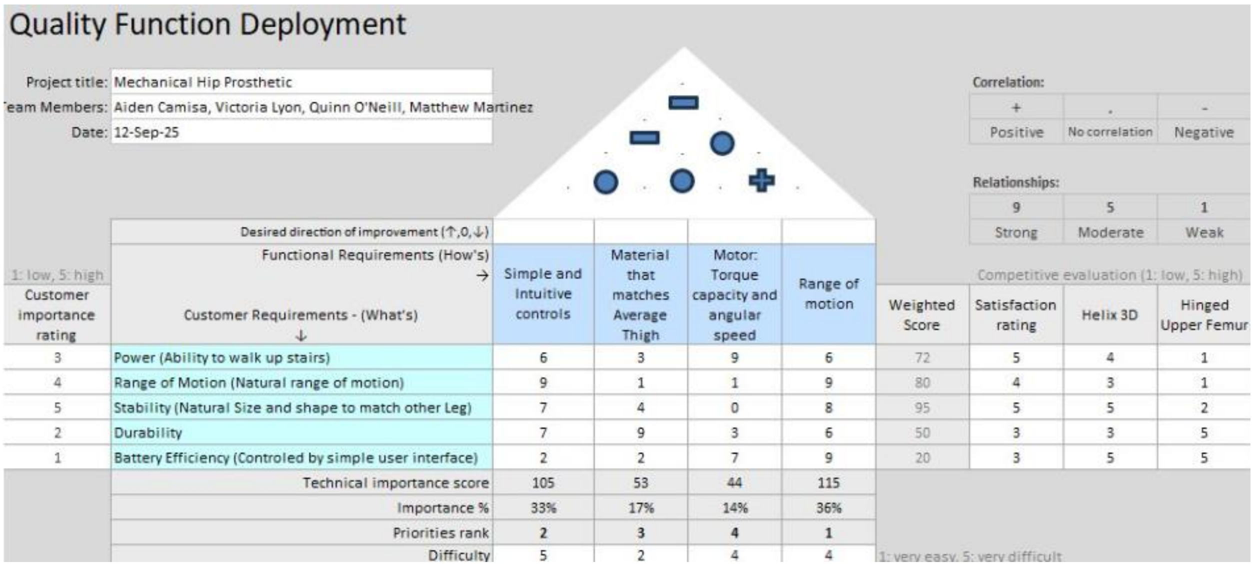
$$P_{hip} = \tau_{hip} \cdot \omega_{hip} \tag{21}$$

$$I = \frac{\tau_m}{k_t} \tag{22}$$

$$V = IR + k_t \omega_m \tag{23}$$

$$P = V \cdot I \tag{24}$$

7.3 *Appendix C: Figures*



Quality Function Deployment

Project title: Mechanical Hip Prosthetic
Team Members: Aiden Camisa, Victoria Lyon, Quinn O'Neill, Matthew Martinez
Date: 12-Sep-25

Correlation:		
+	.	-
Positive	No correlation	Negative

Relationships:		
9	5	1
Strong	Moderate	Weak

1: low, 5: high Customer importance rating	Desired direction of improvement (↑,0,↓) Functional Requirements (How's) → Customer Requirements - (What's) ↓	Simple and Intuitive controls	Material that matches Average Thigh	Motor: Torque capacity and angular speed	Range of motion	Weighted Score	Competitive evaluation (1: low, 5: high) Satisfaction rating	Helix 3D	Hinged Upper Femur
3	Power (Ability to walk up stairs)	6	3	9	6	72	5	4	1
4	Range of Motion (Natural range of motion)	9	1	1	9	80	4	3	1
5	Stability (Natural Size and shape to match other Leg)	7	4	0	8	95	5	5	2
2	Durability	7	9	3	6	50	3	3	5
1	Battery Efficiency (Controled by simple user interface)	2	2	7	9	20	3	5	5
	Technical importance score	105	53	44	115				
	Importance %	33%	17%	14%	36%				
	Priorities rank	2	3	4	1				
	Difficulty	5	2	4	4	1: very easy, 5: very difficult			

Figure 1: Quality Function Deployment

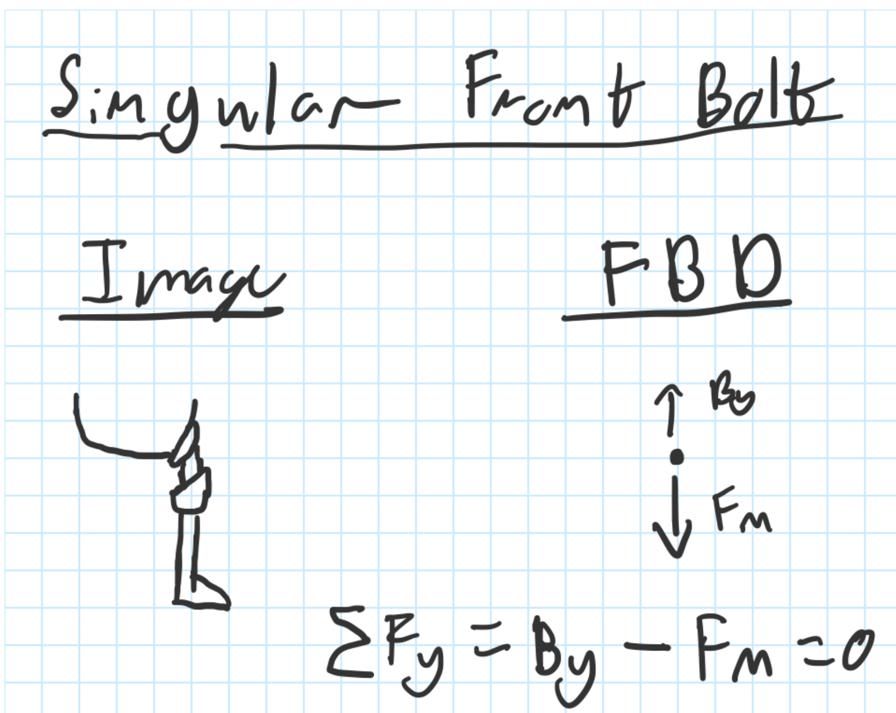


Figure 2: Free-body Diagram of Prosthetic

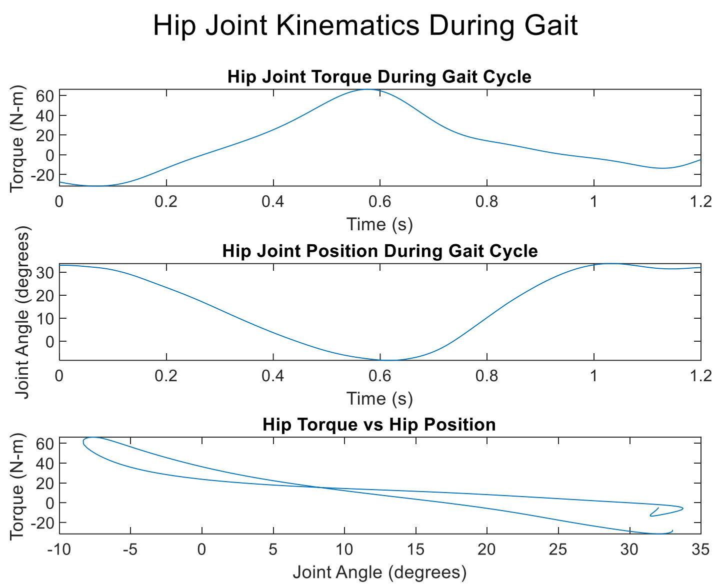


Figure 3: Hip Joint Kinematics During Gait Cycle

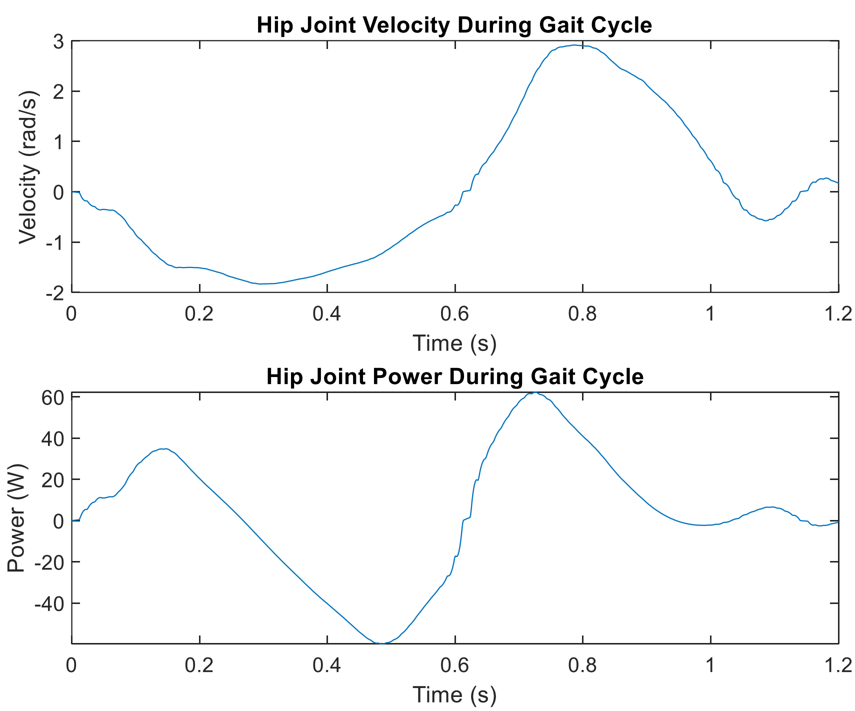


Figure 4: Hip Joint Velocity and Power During Gait Cycle

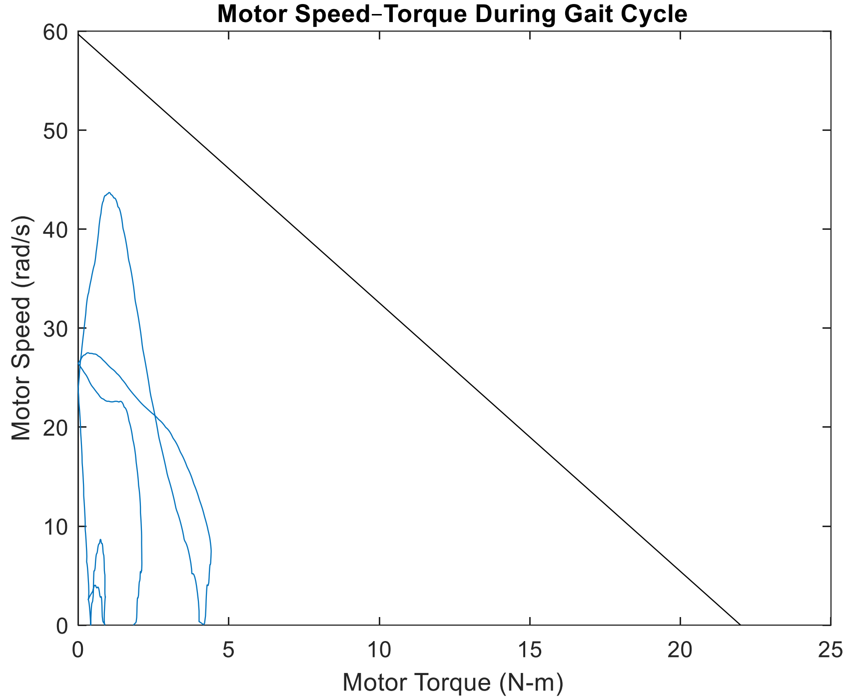


Figure 5: Motor Speed-Torque During Gait

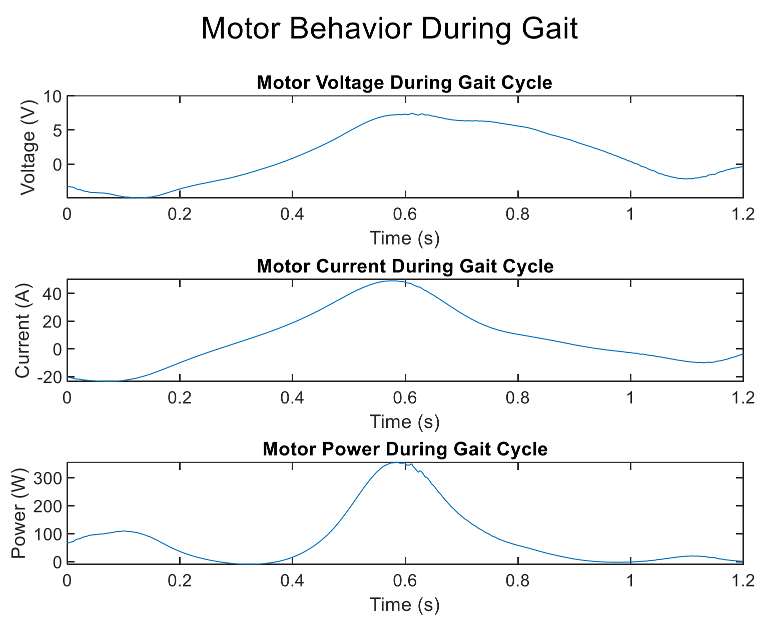
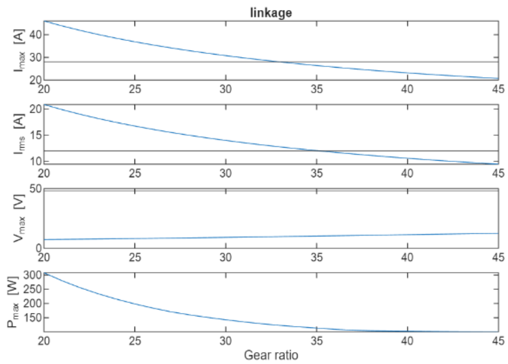
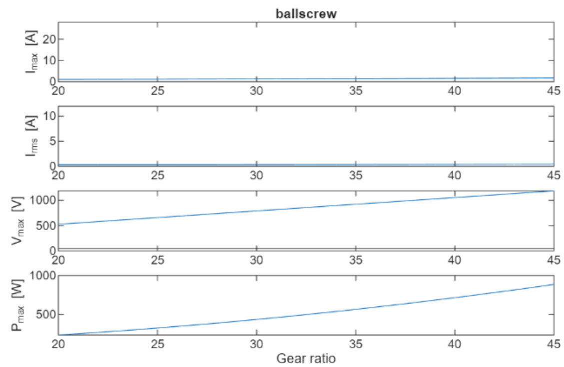
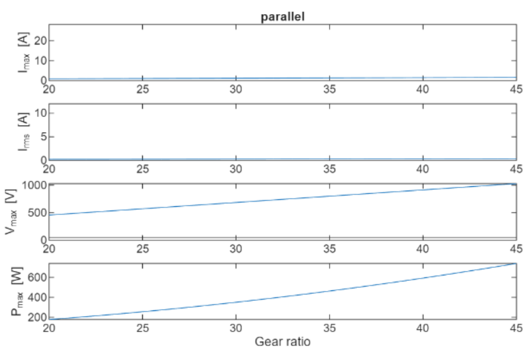


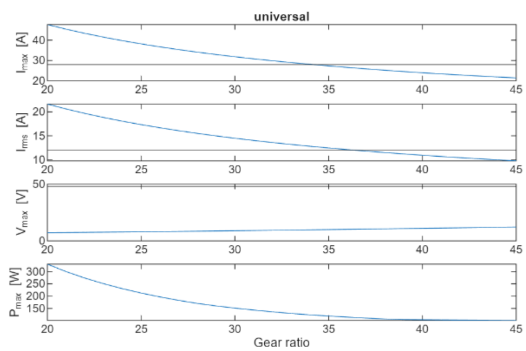
Figure 6: Motor Behavior During Gait





Figures 7 and 8: Current, Voltage, and Power of Linkage and Ballscrew Mechanisms with Respect to Gear Ratio





Figures 9 and 10: Current, Voltage, and Power of Parallel and Universal Mechanisms with Respect to Gear Ratio

7.4 Appendix D: Tables

Time (Min)	Battery required (Wh)
10	21.73
20	43.46
30	65.19
45	97.79
60	130.38
90	195.57

Table 1: Battery Sizing for Active Hip Prosthetic

Type	Force (N)	Moment (N*m)	Force per Bolt (N)	Bearing Stress (MPa)	Shear Stress in the bolt (MPa)
Dual Attachment	882.90	52.974	441.45	13.80	8.78
Laterally Mounted	882.90	70.632	882.90	27.59	17.56
Singular Front Bolt	882.90	88.29	882.90	27.59	17.56
Angled Corner	882.90	105.948	882.90	27.59	17.56

Table 2: Comparison of Attachment Designs

Mechanism	Equation Used
Mechanical Link	$T = F * l$
Ball Screw	$T = (F * l)/2\pi$
Universal Joint	$T = P/n$
2 DOF Stewart Platform	$T = F * 2 * l$

Table 3: Mechanisms and Equations

7.5 Appendix E: Models

Model 1: Most Recent CAD Model of Design