INDIAN INSTITUTE OF TECHNOLOGY, MADRAS

INTERDISCIPLINARY CREATIVE PROJECT - ID4200

Design of Active Transfemoral Prosthesis for Stair Traversal

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Contents

1 Objective

The primary objective of Project Asai is to develop an innovative, low-cost prosthetic knee system that reduces user effort during high-torque tasks such as stair ascent, descent, ladder climbing, and rising from a seated position. The system aims to provide:

1. Powered assistance for knee extension during stance phase
2. Flexion assistance for obstacle clearance during swing phase
3. Intuitive control system that adapts to different environments and user needs
4. Biofeedback mechanism to enhance proprioception and user control
5. Lightweight design promoting all-day usability

2 Design Overview

2.1 Cable Driven Actuation System

Taking inspiration from human biomechanics, specifically the quadriceps muscle group, our system employs a dual cable-driven actuation approach. This bio-inspired design provides several key advantages:

1. Mechanical Simplicity: The cable-driven system elegantly bypasses the complex geometry of the Kadam four-bar polycentric knee joint without requiring additional components around the compact knee structure.
2. Reduced Weight: Compared to alternative transmission systems such as gears or lead screws, cables offer significant weight reduction—a critical factor for user comfort and mobility.
3. Efficient Power Transfer: The high strength-to-weight ratio of Dyneema cables ensures efficient force transmission with minimal elasticity, allowing precise control of joint movement.
4. Bi-directional Control: The dual-cable system enables both extension assistance (for stair climbing and standing) and flexion assistance (for obstacle clearance and swing phase control).

The patella mechanism increases the moment arm for the extension cable, significantly reducing the required motor torque while maintaining a biomechanically appropriate action line.

2.2 Bio feedback System

Many of the mobility challenges experienced by lower-limb amputees result from the lack of proprioception on the affected limb. Proprioception is defined as the ability to sense orientation and movement of a body part, it plays a major role in balance control during walking. In non-disabled individuals, tactile receptors on the lower-limb detect force and position information throughout locomotion and transmit the information through the peripheral nerves to the central nervous system. However with amputees they have impaired peripheral nerves on the affected limb, they utilize the friction and pressure sensations detected by tactile receptors along the prosthesis-limb interface to control the prosthetic limb. However this impaired feedback leads to increased reliance on alternative senses such as vision and hearing to perform the mobility tasks. When the visual feedback is lost, for example if the user is carrying a large box, then this results in suboptimal performance in gait and balance functions. For our feedback system we decided to go for a vibrotactile setup i.e by using vibratory motors, or tactors, to provide tactile feedback. This was decided since it was minimally invasive, easy to setup and will not hinder with the other senses.

2.2.1 Tactile Sensitivity in Skin

The skin has 4 types of mechanoreceptors corresponding to different sensations like flutter, stroking, pressure, texture, vibration and skin stretch. The pacinian corpuscle responds to vibrations at the skin surface and nearby joints and muscles. When compared to other mechanoreceptors this one is larger and less numerous and hence the receptive fields are larger with indistinct boundaries. Skin was found to be most sensitive to vibration ranging from 40 Hz to 1000 Hz with maximum sensitivity at 250 Hz. We also found that human skin cannot differentiate well between stimuli of different frequencies for our tactors we decided to keep frequency constant and vary the amplitude. The sensory homunculus shows how much cortical area is dedicated to sensory input from each part of the body, here the hip is right next to the leg so it made sense to put our tactors on the thigh for easy sensory mapping. Phantom-limb pain is cured in a similar way where the areas of the brain associated with the amputated limb get reassigned to face and areas of the residual limb. The thigh is non-glaborus skin, therefore normal stimulation is preferred here as compared to tangential stimulation.

2.2.2 Setup

A configuration of 4 tactors was used on the thigh. The tactors used were eccentric rotating mass (ERM) motors and they were placed on a memory foam belt inside the liner. For the thigh the two point discrimination threshold is 15 mm so we have followed this minimum distance.

On the foot we have placed force sensing resistors (FSRs) to measure the force applied on localized areas of the foot. We have used Tekscan FlexiForce A301/1 Force Sensors for this purpose. To determine the knee angle we have made use of two MPU-6050 units on the prosthetic leg. One is on the socket (thigh) PCB, and one on the pylon (shin) PCB.

2.2.3 Audio Feedback

Another mode of biofeedback that was explored was audio-based feedback. In this mode, auditory cues were provided to convey information about the prosthetic limb's state. For example, distinct tones or patterns were used to indicate events such as heel strike, excessive knee flexion, or force thresholds detected by the FSRs. The volume, pitch, or repetition rate of the audio signals could be modulated to reflect the magnitude of the parameter being monitored. An app was made on the phone which could output the necessary sounds based off of angle and force triggers.

The biofeedback system was primarily designed to assist users during the initial phase of adapting to the prosthetic limb. The aim was to enhance proprioceptive awareness and motor learning by providing supplementary sensory information. However, to prevent over-reliance on the feedback system, the feedback intensity and frequency were intended to be gradually reduced over time. This progressive reduction strategy encourages the user to internalize control of the prosthetic and develop natural movement patterns without becoming dependent on external cues.

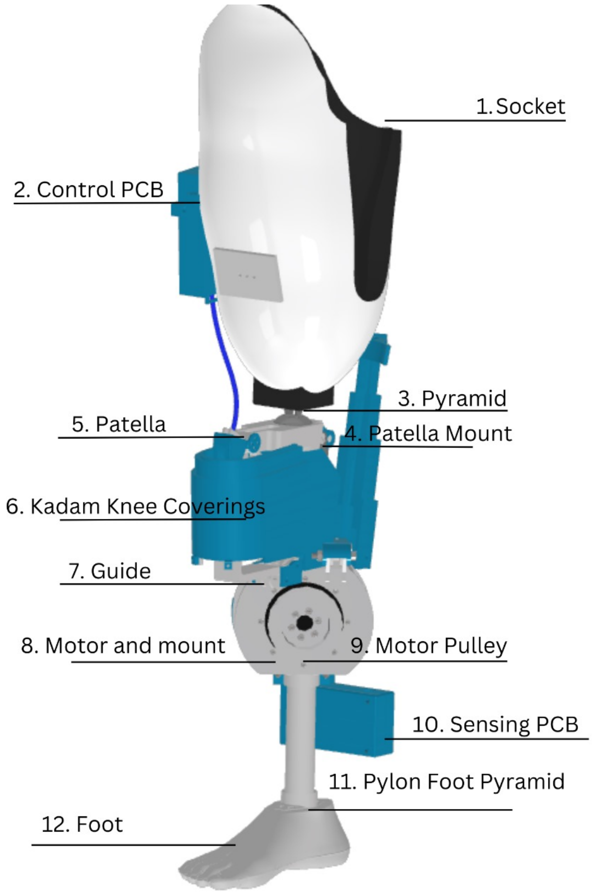
3 System Architecture and Integration

3.1 Drive System

To reduce user effort in high-torque tasks such as stair ascent, ladder climbing or getting up from a seated position, etc. an actuation system is required. Taking a bio-inspired approach based on the contraction of the quadriceps muscles in the human leg, a dual cable-driven actuation system has been designed for knee extension assistance during stance phase and flexion assist for obstacle clearance. The tension in the cable pulls the thigh up, rotating the thigh (top link in the fourbar polycentric knee) about its instantaneous center. The cable is fixed onto the socket, and it is driven by the motor on the pylon.
The cable-drive transmission system was adopted for the following advantages:

- **Simplicity:** A cable-driven system is the simplest option to actuate the fourbar knee as its design does not depend much on the geometry of the knee, which acts as a geometric constraint. The ability to bypass the fourbar also avoids the need to add complicated parts around the compact Kadam knee.

- **Weight Reduction:** Cables are significantly lighter than other transmission systems like gears or lead screws. This is an important factor for user comfort and mobility.



(a) Full assembly

(b) Full Assembly Manufactured

3.1.1 Specifications

The following physical parameters were considered while designing the cable drive system:

- Patella {x, y}: The point of contact of cable as seen in the sagittal plane
- Motor pulley position
- Attachment point on the socket
- Motor pulley radius
- Patella pulley radius

MSC Adams, a multibody dynamics simulation software, was used to optimise parameters ADAMS simulations, explain tension requirements, cable selection - We have modeled the knee joint along with the foot and socket, incorporating the appropriate weight distribution according to the inertia of each component to simulate the practical implementation of our design. Our objective was to optimize the positions of the patella, motor, and attachment point on the socket to minimize the torque required for knee extension. All calculations were performed assuming a body weight of 100 kg, ensuring that the selected motor provides a factor of safety of at least 2 for the torque requirement.

For the ADAMS simulations, the foot was grounded, and tension was applied along the cable. An initial velocity of 1 rad/s was applied to the thigh to account for the influence of the contralateral limb during extension. We utilized the design exploration feature in ADAMS to analyze the impact of various parameters on the minimum tension required to lift the leg. The results indicated that the positions of the motor and the attachment point on the socket had minimal impact, provided they remained within a specific range of values. However, the x-position of the patella had a significant influence on the torque requirement; specifically, the further the patella was positioned outward, the lower the torque needed. By projecting a line vertically downward from the socket, we determined that it intersects the horizontal line passing through the top link of the Kadam knee at a distance of 12 cm outward. Consequently, the minimum x-position of the patella should be 12 cm. On running some iterations we found that at a distance of 14 cm the torque requirement was 1200 N which was optimal. In determining the radius of the motor pulley, we balanced the trade-off between minimizing torque and minimizing the time required for knee extension.

3.1.2 Cable Selection

The cable we selected for the system needed to meet the following criteria:

- Capable of withstanding high tension (1500 N)
- Exhibits very low elasticity
- Not be too thick

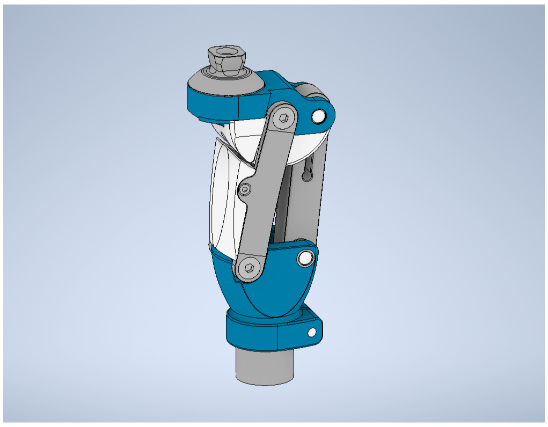
A dyneema cable was chosen for the extension assist systems because:

- It has a high strength-to-weight ratio ensuring reliable force transmission and high energy efficiency.
- It exhibits minimal elongation under load, and can transmit forces more precisely without significant stretching.
- It offers resistance to abrasion and fatigue, making it durable over long periods of use.

3.2 Kadam Knee

The Kadam knee is a four-bar polycentric knee joint with the following key features:

- The link lengths are optimized to provide a larger region of stance stability (larger flexion angle up to which the instantaneous centre of rotation is behind the load line for stable heel contact and load bearing), as well as maximized toe clearance during swing.
- It allows higher knee flexion (range is limited by the socket).
- Frictional swing control adjustment allows for selecting different walking speeds.
- Hard chrome plated EN8 pins, high fatigue polymer bushings and PU bump-stops for durability



(a) Kadam Knee

3.2.1 Parts

The Kadam knee is composed of:

- **Bottom link** (Al 6061 T6): The bottom link (or shank link) in the fourbar assembly clamps onto a 30mm diameter pylon below.
- **Top link** (Al 6061 T6): The top link (or thigh link) has a standard pyramid adapter which can be adjusted in the anterior-posterior direction.
- **Anterior link** (SS 304): The anterior link connects the top and bottom links.
- **Posterior link** (SS 304): The posterior link completes the fourbar geometry. It limits the motion of bump stop thus preventing hyperextension. It also allows friction adjustment for different walking speeds.
- **Cap**: A lightweight polymer part that press fits onto the top link. It comprises of a polyurethane bumpstop which prevents hyperextension of the knee.
- **Pyramid adapter** (Stainless steel): A standard pyramid adapter that connects to the pyramid receiver in the socket, thus attaching the knee to the residual limb.
- **Joint pins**: Hard chrome plated EN8 pins to connect two links
- **Bushings**: High fatigue life polymer bushings for minimizing friction at joints

3.2.2 Tests

- ISO 10328 compliant for proof strength, ultimate strength and fatigue strength for P4 condition
- 3 million cycles static test performed with no failure: P4 level (60-80kg) test loading condition I (heel loading)
- 3 million cycles static test performed with no failure: P4 level test loading condition II (forefoot loading)

3.3 Socket

3.3.1 Standard Specifications

A socket attaches to the knee through pyramid components: a male pyramid adapter and the female pyramid receiver, which help in aligning the prosthesis in all planes. Our test sockets are made of polypropylene, and the final device may include a carbon fibre socket if deemed necessary. The positive mold was made using a digital scan of the residual limb and the fitting and alignment was done by a professional prosthetist. Our two pilots have different mechanisms to attach the socket due to varying levels of amputation:

- P1 wears a silicone liner, and uses a vacuum suspension by pumping out the air between the socket and liner.
- P2 wears a sock over his residual limb and uses an elastic suspension belt which wraps around the pelvis and closes with velcro, holding the socket in place.

3.3.2 Vibrotactile Setup

As part of our biofeedback system, we have incorporated a vibrotactile setup inside the socket in contact with the skin on the residual limb. This consists of a memory foam belt worn inside the liner/sock, and has four eccentric rotating mass (ERM) motors distributed along the thigh. The total added thickness due to this setup including the wiring is 5 mm.

3.3.3 Cable-drive Attachment

For the extension assist system, we have a cable attached to a motor in the pylon, wrapping over the patella, and attaching onto the socket. This cable attachment is done externally on the socket using a pair of aluminium rings padded with memory foam that clamp onto the socket with a velcro cinch strap assembly. A flat plate with a hole is welded onto the rings using some attachments, and this hole is used to mount the cable. A canvas strap between the ring and a waist belt prevents the ring from sliding down the tapering figure of the socket. To provide more support, abutments are attached to the socket under the ring by use of adhesives.

3.4 Foot

A Jaipur foot is used in our device. It is attached to the pylon via a pyramid assembly, making it a solid ankle device. This is a permanent joint as the female pyramid adapter has been riveted onto the pylon and the screws are held by an adhesive such that they cannot be loosened.

3.4.1 Assembly

Part	Material	Assembly Instruction
Thigh brace	Aluminium 6061	Attached to the socket with a Velcro belt
Patella mount	Aluminium 6061	Screwed between the topmost link of the Kadam knee and the pyramid adapter that attaches to the socket.
Patella pulley	Aluminium 6061	Bolted at four points on the patella mount
Motor mount	Aluminium 6061	Attached to the bottom part of the Kadam Knee.
Load cell top	Aluminium 6061	Attached below the motor mount, above the load cell
Load cell bottom	Steel (SS-304)	Attached between the load cell and the foot

3.4.2 Parts

1. **Patella**
To operate the cable drive system for extension assist, a patella has been designed to increase the moment arm of the tension force in the cable and thus reduce the torque requirement from the motor.

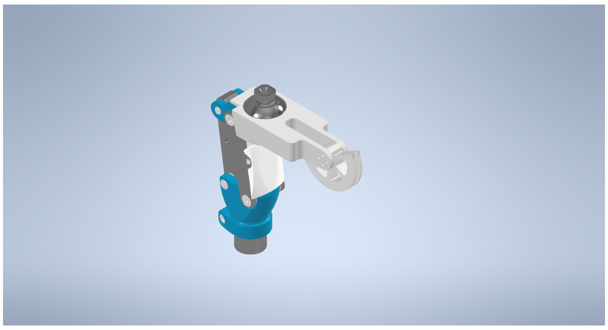
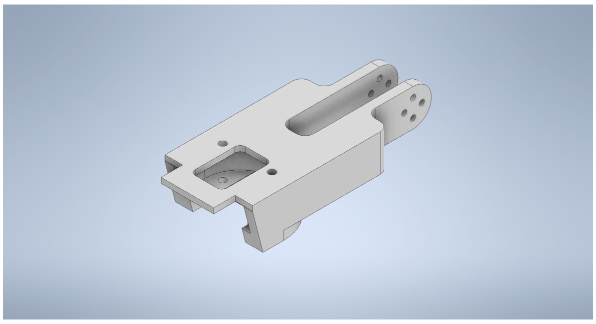


Figure 3: The Kadam Knee with the patella

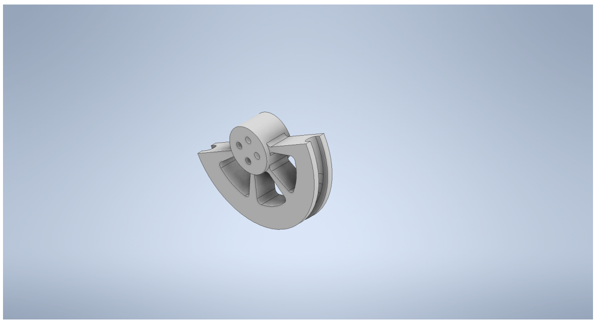
Assembly: The patella consists of two parts:

- Mount: The mount holds the pulley in place, at the optimized distance from the Kadam knee.
- Pulley: The cable from the cable-driven actuation system passes over the pulley.

Both parts were made from Aluminium 6061 owing to its easy machinability, low weight, and high yield strength.

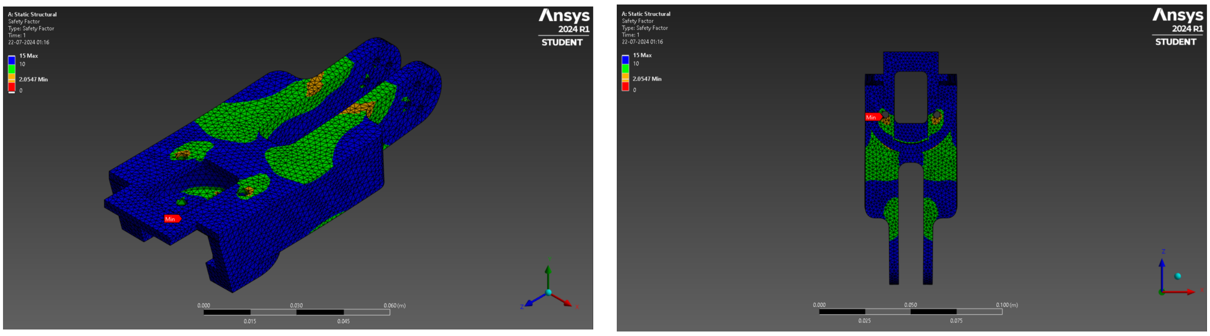


(a) Patella Mount



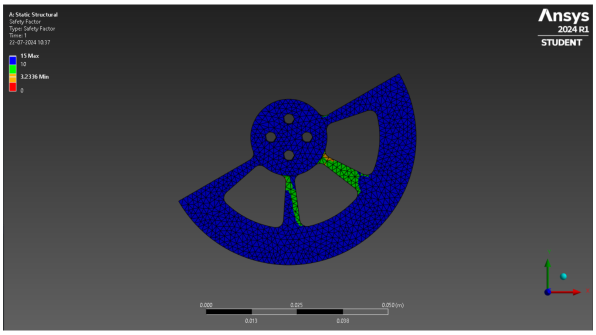
(b) Pulley

The tension in the cable does not exceed 1500 N at any point during the actuation. At maximum flexion, the angle of wrap of the cable around the pulley will be maximum, and hence, the pulley will experience a maximum force at maximum flexion. This maximum angle of wrap is less than 90 degrees, and is taken to be 90 degrees, for the purpose of safety analysis.



(a) Factor of safety for the mount (1)

(b) Factor of safety for the mount (2)



(c) Factor of safety for the pulley

2. **Motor mount**
   The motor T-Motor (CubeMars) AK80-64 was selected for the cable drive system due to its high peak torque to weight ratio. The motor mount fixes the motor into the prosthetic pylon, while also providing a safe covering around the motor. The guide setup ensures that the cable starts from the plane of the pulley attached to the motor, and then goes into the plane of the patella.

   Assembly Definition: The motor mount is a machined part consisting of the following features:

   - Rear plate: The rear plate is a 5mm laser-cut aluminium sheet that is rigidly attached to the front plate by welded ribs, and has attachment points to the rear connector plate.

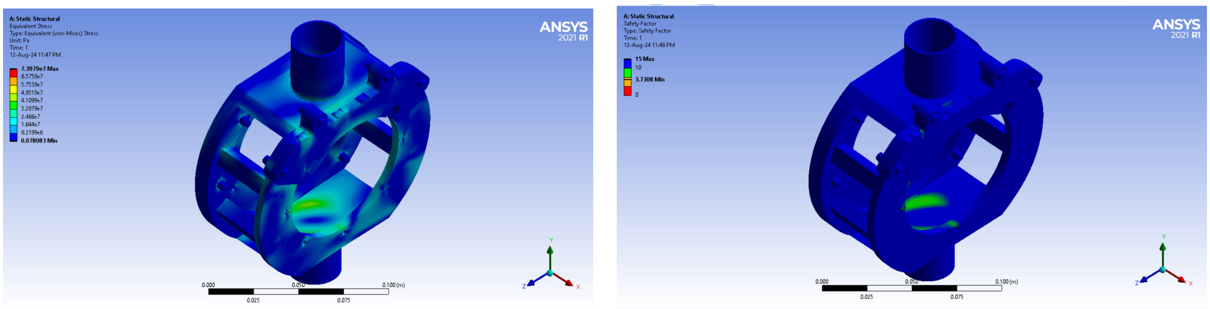
- Body: The body of the mount is an enclosure around the motor, with bolts in the front to attach onto the motor. The rear plate bolts onto this. There are ribs running across to transfer load.
- Pylon connectors: Two identical machined plates form the top and bottom faces of the motor mount which allow connections to pylon tubes. These plates have slots into which aluminium tubes of 30mm outer diameter are welded. These pylon tubes go respectively to the Kadam Knee and the top cover of the load cell mount.
- Front plate: The front plate serves as the connection between the motor mount and the a guide pulley mount.
- Guide pulley arms: The guide pulley arms are designed to guide the extension and flexion cables into the plane of the kadam knee.
- Guide pulley: The cable in the cable drive system passes from the motor pulley to the patella, both of which are in different planes. Due to this, the cable may brush against the walls of the motor and patella pulleys, which would reduce the efficiency of the actuation system. To eliminate this problem, we introduce an intermediate pulley aligned such that the cable enters the grooves of both the aforementioned pulleys parallel to their walls thereby eliminating any brushing and ensuring optimum performance.

Specifications: All the parts of the motor mount assembly are made up of aluminium 6061 owing to its durability, low density, and high yield strength. To reduce costs, the mount was manufactured mainly by laser cutting and welding.

Safety

- The ribs connecting the front connector plate and the rear plate must be inserted and welded carefully as this is crucial for the precise fitting of the motor.
- The pylon connectors and pylon tubes must be aligned perfectly to ensure the stability of the resulting pylon assembly.
- The screws attaching the pulley and the plates to the motor must be tightened sufficiently.

Structural Simulations: The motor mount was simulated to bear forces due to the weight of the user, the weight of the motor and the force acting due to rope interaction with the guide pulley. The maximum rope tension was assumed to be 1500N as obtained from MSC Adams. The results below depict the equivalent stress distribution and the safety analysis.



(a) Equivalent Stress distribution **(b)** Safety Analysis

3. **Motor Pulley**
   To operate the cable drive system for extension assist, a pulley is connected to the motor. The pulley has been designed such that the actuation for stair climb is successful with 85 per cent of the maximum available angular speed output from the motor. The pulley has two concentric grooves, one is used for flexion assist cable drive, while the other, is for extension, thereby making our system under-actuated. This was done in order to keep the cost low, while still providing the necessary features needed to complete all the Cybathlon tasks.

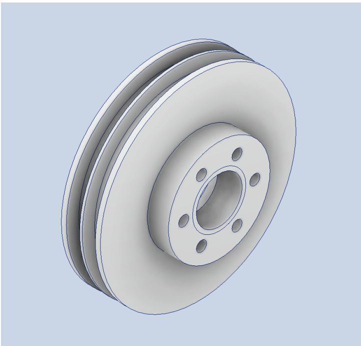


Figure 7: The Motor Pulley

The pulley is made from Aluminium 6061 owing to its easy machinability, low weight and high yield strength.
The tension in the cable does not exceed 1500N at any point during the actuation. The radius of pulley at contact is 31 mm. This means that the maximum moment on the pulley about its axis does not exceed 46.5 Nm. The pulley was thus simulated at this critical condition.

For actuating the knee in flexion, another groove of the same radius has been made. This cable will have a maximum tension of about 200 N in normal operation. The radius of this groove is the same as the extension cable groove.

The safety factor is greater than 15 at all nodes according to the Equivalent stress theory.

3.5 Mechanical Alignment

For comfortable use of the prosthetic knee, the alignment of the leg should be done properly by a professional prosthetist. Since it was not possible for a prosthetist to be present at every user session, we learnt some helpful checks and fixes:

- **Static alignment**:
  1. **Sagittal plane**: There are three main points of alignment that should be in a straight line:
     (a) the midline of the socket (in the mid-sagittal plane)
     (b) centre of rotation of the knee
     (c) a line 3cm posterior from the midline of the foot
  2. **Frontal plane**: The central axis of the pylon should align with the second toe or a point between the first and second toe.
- **Dynamic alignment**:
  1. If there is knee-buckling during heel strike, the socket alignment may be too far ahead.
  2. If there is buckling during foot flat, it is likely that the foot alignment is not proper.
  3. If the user is circumducting the affected limb, it is possible that the socket fitting is not proper, or the height of the prosthetic leg is too much.

A practice that we followed during disassembly so as to minimize misalignment was to only loosen and tighten the posterior and any one lateral grub screw, without touching the other two grub screws (at any pyramidal attachment).

3.6 Electronic Control Unit

3.6.1 Safety

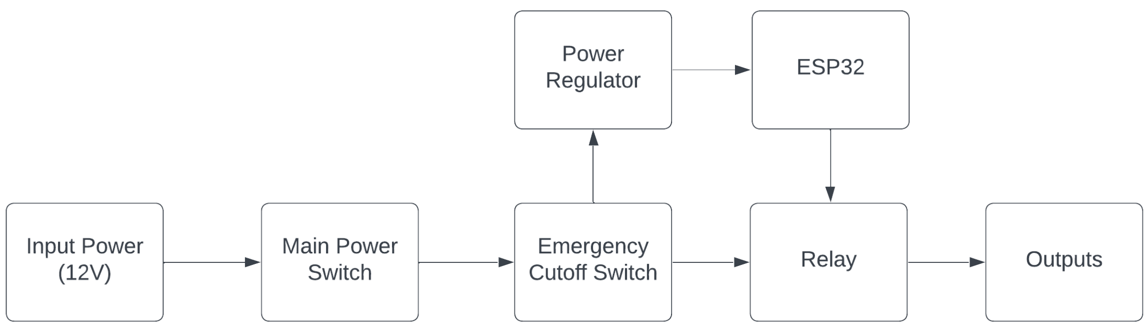


Figure 8: Schematic of Shutdown circuit

Shutdown circuit The Shutdown circuit (SDC) is designed to enable the prosthetic to be powered off, both manually and automatically, in case of an unsafe condition.

- The main power switch is for turning the prosthetic on or off during normal operation.

- The emergency button is easily accessible and is supposed for manual power shutdown in case of emergencies. The emergency button is located on top of the control PCB mounted on the socket.
- The Microcontroller on the Control Unit controls a relay (NO) so that the power can be shutdown immediately if an unsafe condition is detected.

Water-proofing The connectors used in the battery enclosure are panel mounting type with o-rings for extra protection against splashes.

3.6.2 Schematic

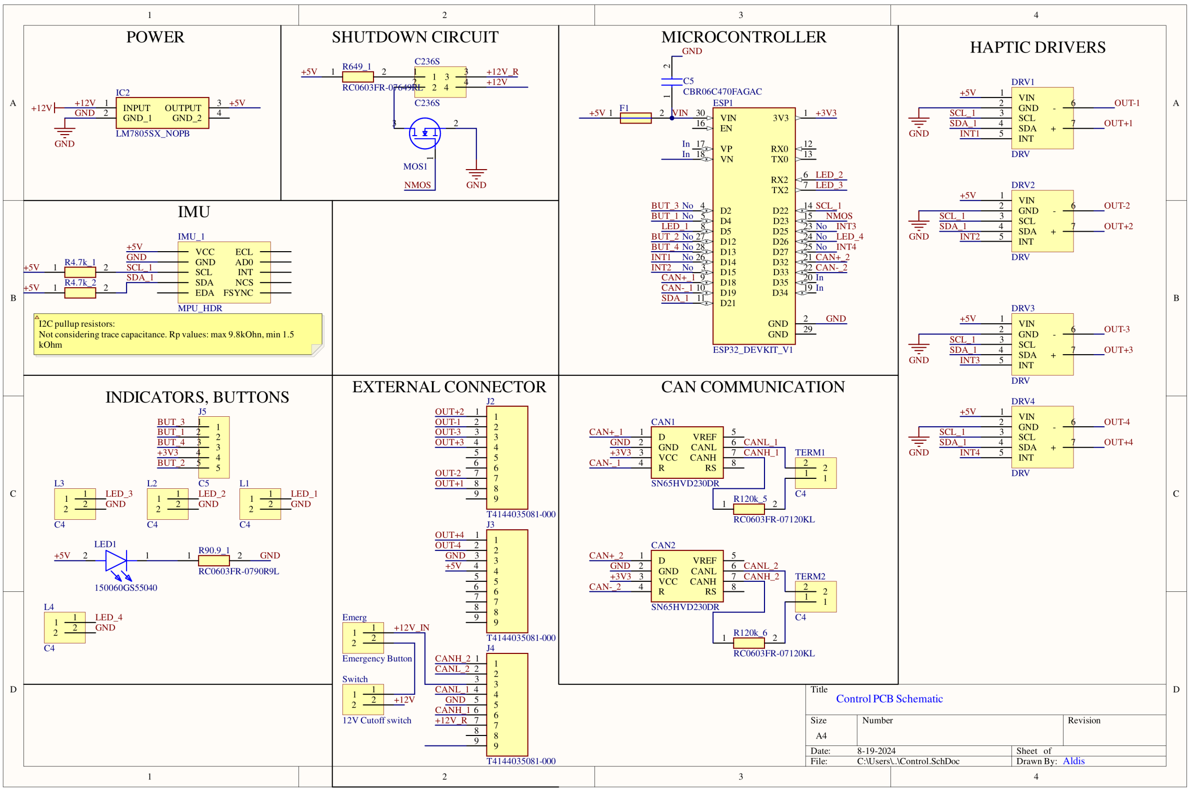


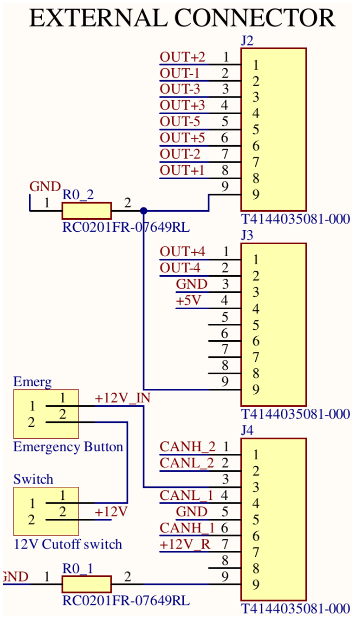
Figure 9: Circuit schematic of Control unit

Connectors - The Control Unit features three T4144035081-000 8 pin connectors to connect with the rest of the electronics. This connector was chosen because of its compactness and water-proof nature.
The connections are:

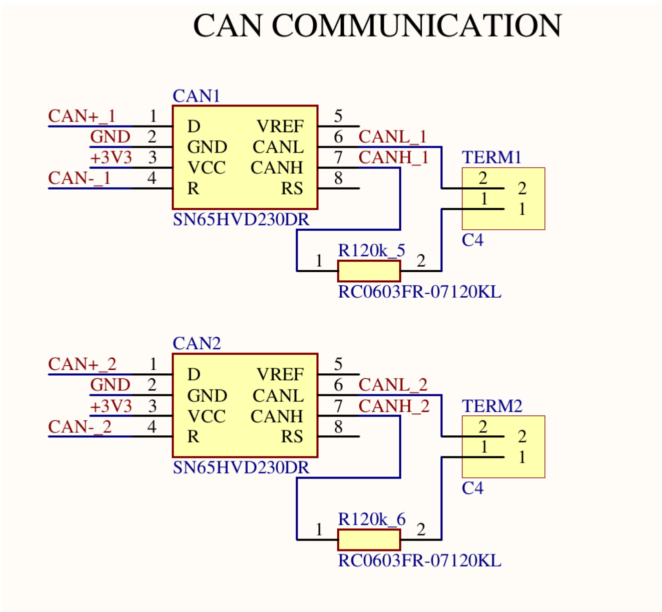
- Connector 1
  - Tactor leads for tactors 1, 2, 3 and 5
- Connector 2
  - Tactor 4 leads
  - +5V and ground for the Sensor unit

- Connector 3
  - CAN bus-1 for motor - control unit communication
  - CAN bus-2 for sensor unit- control unit communication

Communications - The Control unit features 2 CAN buses for external communication: one for the motor and one for the sensing PCB. SN65HVD230DR CAN transceivers are used as the ESP32 does not feature the required hardware. The board allows for optional 120kΩ termination resistances on both CAN buses, which can be enabled by shorting the provided Berg connector switch a jumper.



(a) Connectors of the control unit.



(b) Communication module of the control unit.

Buttons - The Control unit features 4 membrane buttons for user input. They were chosen for their compactness and ease of mounting. Their functions include resetting the ESP and selecting the mode of the leg.

IMU - A MPU6050 is used to measure the relative orientation of the linkages as part of gait control. The IMU features 6 axes: 3 axis accelerometer and 3 axis gyroscope, which are used in combination to increase the reliability of the readings. It communicates to the ESP via a I2C bus that it shares with the tactor drivers.

Haptic driver - DRV2605 haptic drivers are used to control the ERM motors. The driver boards share a single I2C bus.

3.6.3 Housing

The housing is comprised of 3 parts, the main enclosure, the top plate and back plate. The main enclosure has holes that mate with the panel-mount connectors on the Control PCB to create a water-proof seal. It also has the interface buttons on its side. The

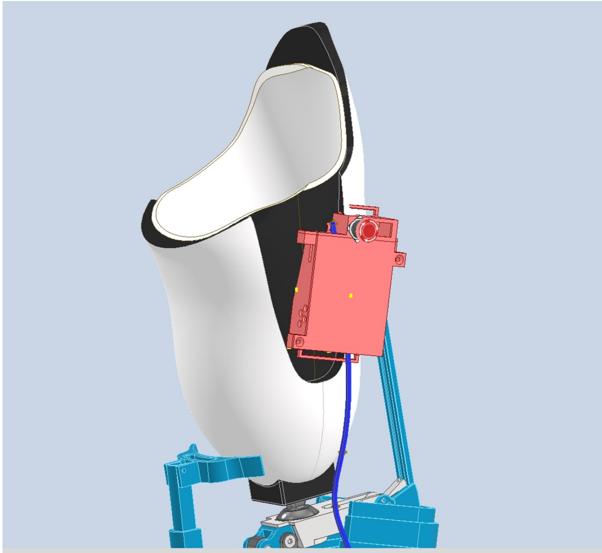
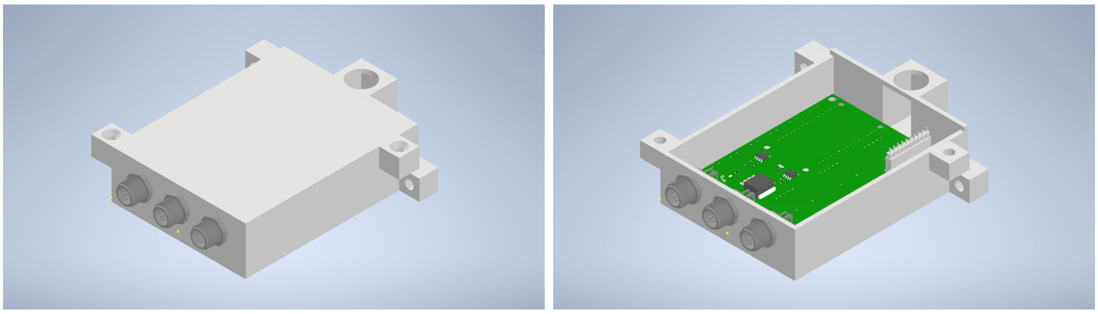


Figure 12: Emergency Stop button on the side of the leg

back plate houses the main power switch, the emergency cutoff button and the indicator LEDs. The top plate completes the enclosure and provides structural integrity. The components of the housing are joined with countersunk screws and threaded inserts in the 3d-printed enclosure.



(a) External view of control unit enclosure. **(b)** Internal view of control unit enclosure.

Figure 11: Views of the control unit enclosure: external (left) and internal (right).

3.7 Electrical Power Unit

3.7.1 Powerstages

- **Battery (HV)**- We are using two batteries (Orange 22.2V 2500mAh 35C 6S lithium polymer battery) in series, each of optimal voltage 22.2V giving a voltage of 44.4V. The battery is placed in a casing mounted on the lower back of the user using a clipper on a waist belt. The casing also contains a Battery PCB where power distribution begins. High voltage (HV) wires from the battery PCB are routed along the waist belt and supply the motor.
- **12V**- The battery PCB has an isolated voltage converter which converts 44.4 volts to 12 volts. The 12V output is stabilized using a 1:1 voltage converter to supply stable voltage to other components and PCBs.

- **5V**- On the main control PCB, a 12-volt to 5-volt converter which supplies 5-volt to power up specific components.

3.7.2 Power Distribution

- **HV wiring**: 13 AWG stranded wires have been selected for the HV connections. Stranded nature of the wire allows for greater flexibility to accommodate the anticipated movement. Braided insulation has been chosen because of its increased mechanical resilience.

- **LV wiring**: 22 AWG solid core wires with PVC insulation is used for LV connections.

- **Communication wiring**: 2 pair twisted 22 AWG wires, shielded are used for CAN communication. Twisted pair wires help mitigate external electromagnetic interference (EMI). This serves as an additional level of safety over the differential bus technology employed in the CAN protocol chosen.

3.7.3 Safety

- **Relay** – We are using a relay on our battery PCB which acts as a switch and restricts high voltage or high current. The relay we are using is a Double pole single throw (DPST) relay with a DC voltage rating of 24 volts.

- **ESD protection** We are placing a grounded metal plate on the casing of the battery which is responsible for providing ESD protection by ensuring that any electrostatic charge accumulated in the shields goes into the metal plate which provides a low impedance discharge path, protecting the circuit.

- **PCB isolation** The high voltage circuit and the low voltage circuit on the battery PCB are isolated and it is ensured that there is no contact between the user and the high voltage part as it may be unsafe for the user.

3.7.4 Battery PCB

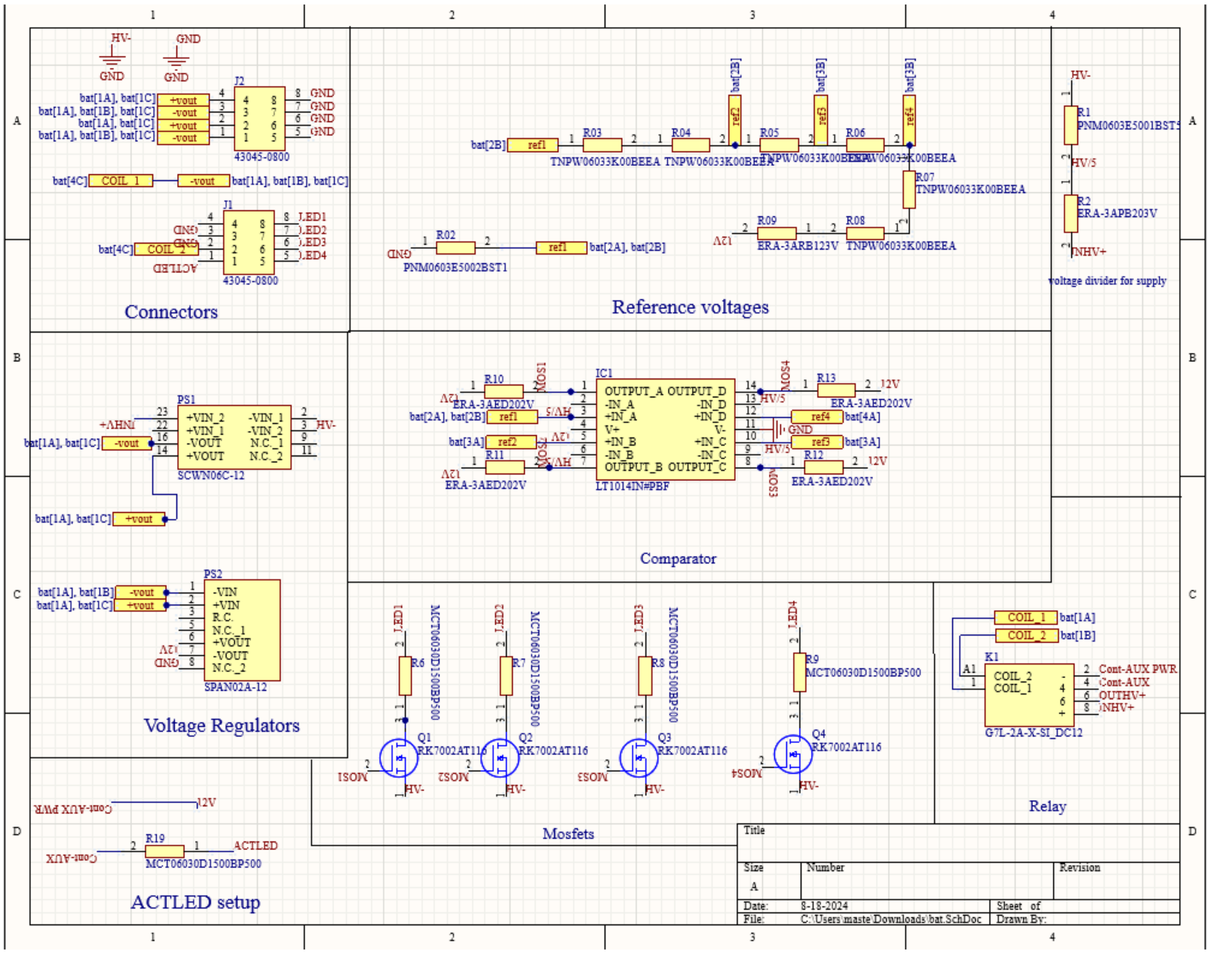


Figure 13: Schematic of Battery PCB

The battery PCB is responsible for safe power distribution to the motor and rest of the device.

- The PCB takes an input voltage of 44.4V using XT-60 connectors.
- A 36-72 V_in to 12 V_out isolated DC-DC converter (SCWN06C-12) is used along with a DC-DC regulated convertor (SPAN02) to stabilize the 12V output.
- A relay (G7L-2A-X DC12) is used for safety which is used to break the circuit if required.
- An analog comparator circuit is used to display battery level using 4 LEDs.

3.7.5 Battery Housing

We have a battery housing located on the pilot's lower back. The battery housing is 3D printed and made of PLA material. The casing acts as an insulator and prevents any electrical contact between the driver and the power supply wires. The batteries are placed in such a way that they are easily accessible, removing and changing of batteries can be done conveniently.

Mounting and unmounting The battery is secured to the housing by Dual Lock™. The enclosure is designed with appropriate clearances to minimise rattling and thus ensures proper securing of the battery. The battery enclosure is mounted onto the belt on the lower back of the user by a custom designed compliant clip.

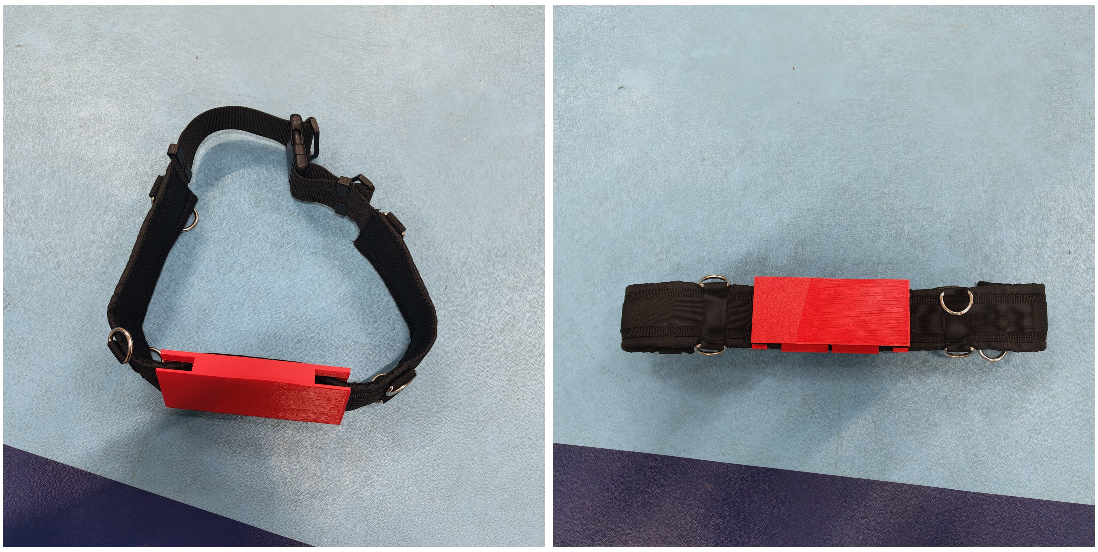


Figure 14: Different views of the clip used to mount the battery enclosure on the belt.

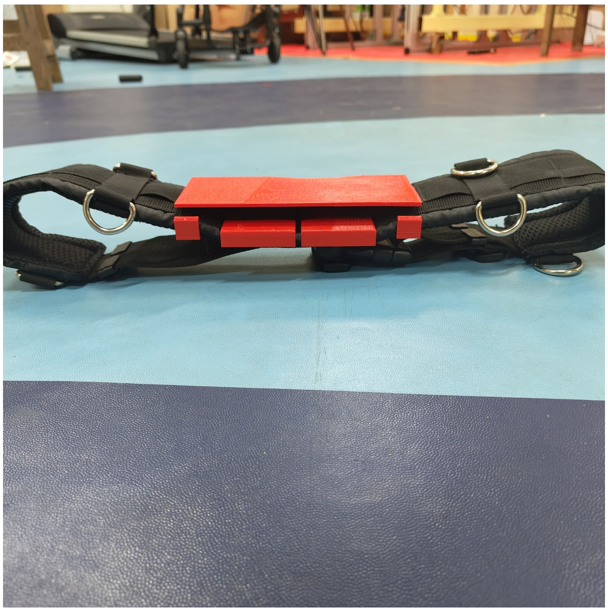


Figure 15: View of clip used to mount battery enclosure on belt

The clip is designed to arrest movement in all three axes with respect to the belt. The mounting of the clip onto the belt is shown in Figure 14 and Figure 15.

Construction of Battery housing The battery housing can be divided in two parts, the side part containing the safety PCB, Fuse and connectors while the main part contains the energy source. As an energy source we are using two 22.2 volts batteries which are connected in series giving a total of 44.4 volts supply to the battery PCB. The

two batteries are separated by a 3D-printed divider (PLA) to avoid any contact. The datasheet of the used PLA has been linked here. .
The maximum temperature that PLA can handle before getting distorted is 53 degrees Celcius. This is well above the operation temperature of the battery. The battery will be providing an average current of approximately 4A in the stair mode, and lower in all other modes, with a peak current of about 20A. This corresponds to less than 2C continuous current for the battery (rated for 2.5 Ah). For a usage of a maximum of 15 mins (considering continuous execution of stair task during training), the battery will not heat up by a considerable amount, not coming close to 53 degrees.
The battery housing has been provided with venting holes on either side of the battery enclosure, to prevent any sort of pressure buildup or hot air to build up inside the enclosure.
Figure 20 depicts the upper part of the housing where each of the two slots are for batteries and these parts are separated by an insulated wall. Also, the upper holes are meant for passing the wire connection from batteries to the PCB in lower part.
In order to charge or discharge the battery, the same connector used to connect the wires to the motor can be used.

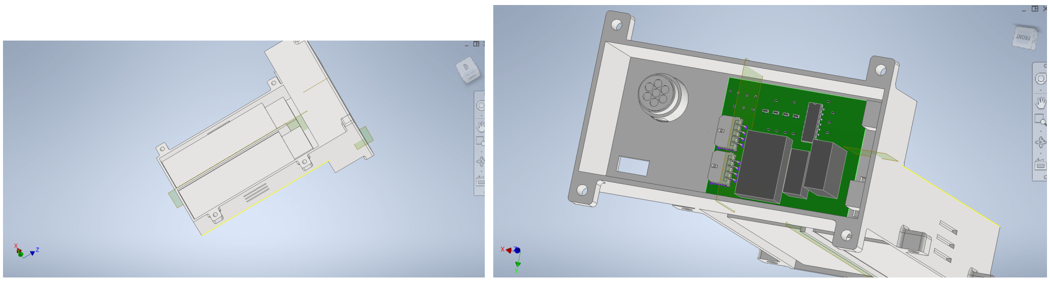


Figure 16: Different views of the battery housing:view with batteries in enclosure with vents for heat dissipation (left) and battery PCB (right).

3.8 Sensing Unit

3.8.1 Schematic

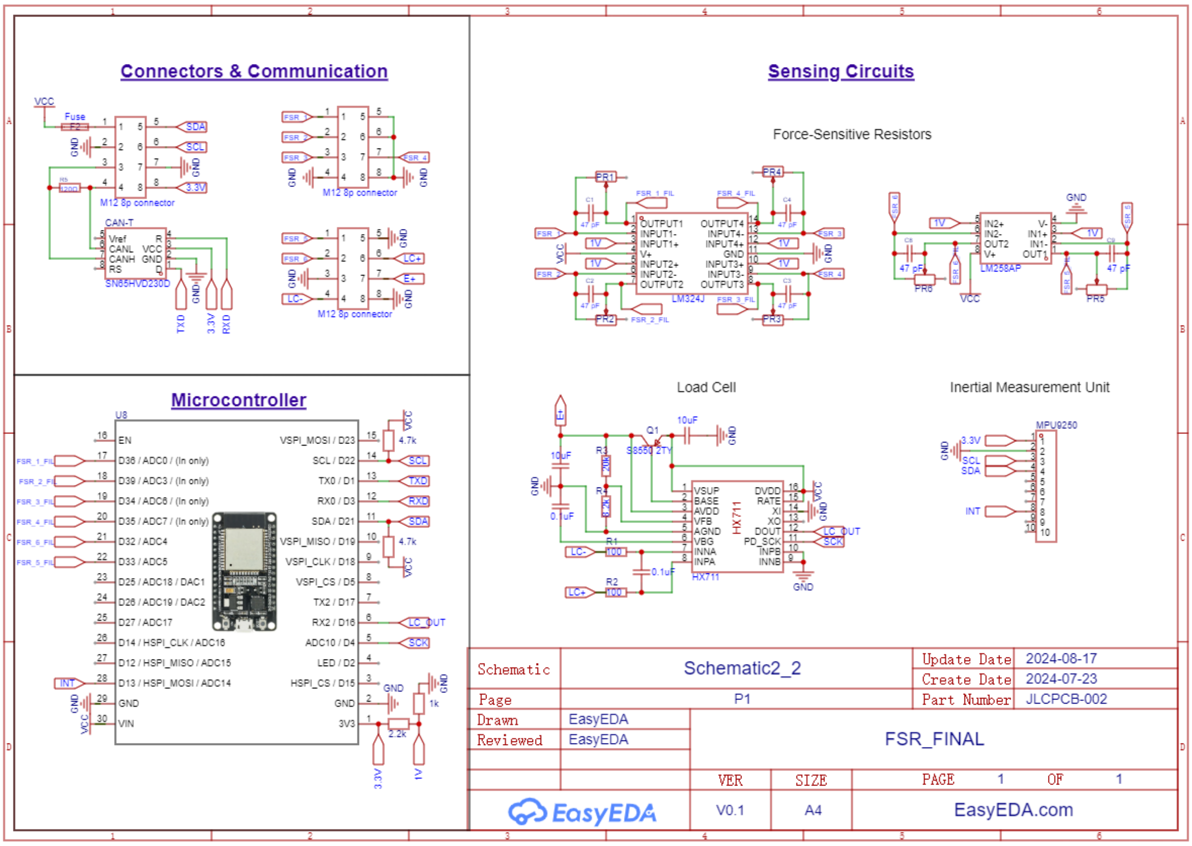


Figure 17: Sensor PCB Schematic

The sensing PCB is composed of the sensing sub-circuits, three 8-pin panel mount connectors, one CAN transceiver to communicate with the control unit and an ESP32 microcontroller. The power to the PCB is a 5V input from the control unit through a fuse rated 1.5A. The sensing sub-circuits include:

1. **Force-Sensitive Resistor (FSR) Circuits:** FSRs have a non-linear (inverse) relation between force-output voltage. A non-inverting op-amp circuit is used to linearize the input force to output voltage relation. A potentiometer is used as the feedback resistor, to allow for adjusting the force sensitivity for different ranges of user weight. A voltage divider is used to supply the reference voltage of 1V from the ESP 3.3V output. We have provisions for 6 FSRs by using LM324 (quad op-amp) and LM258 (dual op-amp). The filtered outputs from the op-amps are read by the ESP32 ADC.

2. **IMU:** A MPU9250 is incorporated in the sensing unit to compute the shank angle and hence the knee angle. It is powered using 3.3V from the ESP32 and communicates with the microcontroller through I2C.

3.8.2 Sensors

- **Force Sensing Resistors -**
  The FSRs are Tekscan Flexiforce A301/1 sensors. Four of them are routed to the sensing PCB from the sole. The resistance of the FSR decreases with increasing force. A low-pass filter is applied on the raw values to refine the data.
  The data of the FSR values is sent to the control PCB. The control PCB uses these values for an initial Gait phase estimation.
- **Inertial Measurement Units** -
  There are two MPUs on the prosthetic leg. There is one on the Control PCB to measure the thigh angle, and one on the Sensing PCB to measure the shank angle. The shank angle data being sent to the Control PCB helps find the knee angle. This data helps further refine the phase estimation and perform actuation.

3.8.3 Housing

The housing is comprised of two parts, the top half casing and the bottom half casing. Both the enclosure halves have holes that align with the holes on the PCB. The components of the housing and the PCB are held in place using M4 bolts.

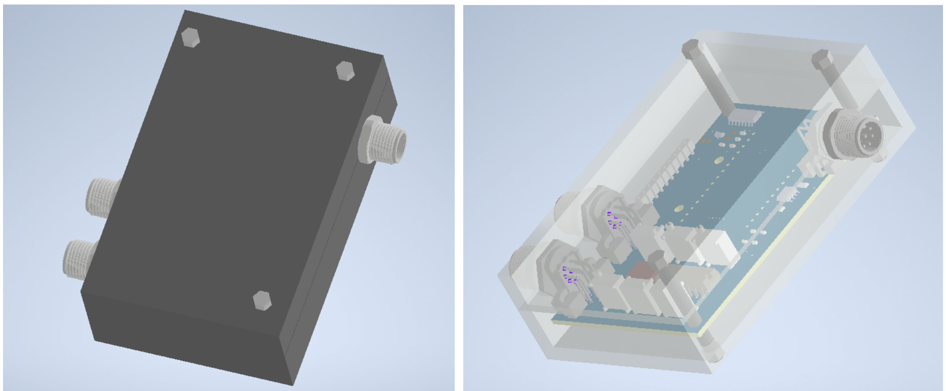


Figure 18: The sensing PCB assembly

3.9 Wireless Interface

The ESP32 is connected via Bluetooth to a mobile phone, with an application that can generate audio panned to either direction. This is conveyed to the user through wired headphones.

3.10 Motor

3.10.1 Overview

The motor used in the prosthetic is a AK80-64 BLDC motor. It features a operating voltage of 48V, a rated torque of 48Nm, a peak torque of 120Nm and a 14 bit single loop encoder.

3.10.2 Interface

We have utilised the CAN ports of the motor for communication with the Control PCB. The motor is supplied 48V power from the power unit. The relay is used to switch power to the motor on/off.

3.10.3 Control

The motor has a on-board KV-80 driver board. Control is achieved by a combination of the MIT control mode made available by the KV80 and custom control loops implemented in the control unit.

4 Software

4.1 Gait Phase Estimation

For the purpose of control on the motors and tactors, a real-time estimation (guess) of the gait phase is desired. We designed our estimation model considering a gait cycle divided into 4 sections:

Phase	Abbreviation
Heel Strike	HS
Mid-Stance	MS
Toe-Off	TO
Swing	Sw

Table 1: The Gait cycle divided into 4 phases

This module provides an initial estimate of the gait phase using only the FSR values (roughly proportionate to the ground reaction force). Based on the data collected from user trials of standing and walking, thresholds were fixed for the FSR values in each phase, thus enabling an estimation of the current gait phase as follows:

- If the sum of all FSR values is below $tFSR\_Sw$, then change the current phase to Sw.
- If the sum of all FSR values is above $tFSR\_St$, then:
  - If the value from the FSR on the front of the foot (ball) accounts for atleast $tFSR\_TO$ (=0.8) times the sum of all FSR values, current phase is set to TO.
  - If the value from the FSR on the rear of the foot (heel) accounts for atleast $tFSR\_HS$ (=0.8) times the sum of all FSR values, current phase is set to HS.
  - If neither is true, current phase is set to MS.

Here, $tFSR\_Sw < tFSR\_St$. The dual threshold for phase switching allows for a bandgap (selected based on error range) within which the gait phase does not switch.
This phase estimation, however, is still error-prone, as adding the noise from all the FSRs could give a large variance in measurements. Actuation does not depend solely on these values, but occurs after further safety checks involving knee angle, vertical ground reaction force, user intent as well as the previous state.

4.2 Finite State Machine

User can access different modes by accessible buttons to switch between the control modes-

1. Walking Mode
2. Stair Mode
3. Balance State

4.2.1 Walking State

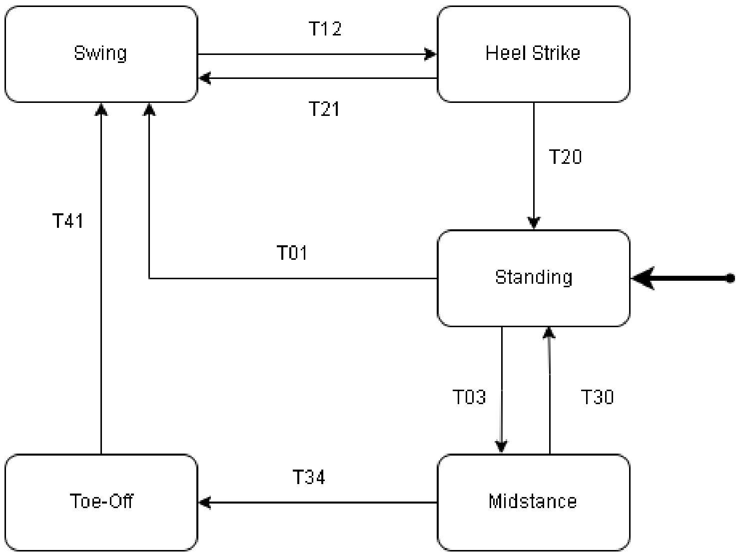


Figure 19: Gait phases in normal walking state

Transition	Condition
T12	FSRs- HS, θ1 < θt < θ2, θk ≈ 0
T20	FSRs- MS, w1<GRF<w2, θk ≈ 0, θt ≈ 0
T03	FSRs- MS, GRF ≈ bw , θk ≈ 0
T34	FSRs- TO, θt < 0
T41	FSRs- Sw, GRF ≈ 0
T01	FSRs- Sw, GRF ≈ 0
T21	FSRs- Sw, GRF ≈ 0
T30	FSRs- MS, w1<GRF<w2, θk ≈ 0, θt ≈ 0

Figure 20: Transition Conditions for Walking

Each state corresponds to a phase of the gait cycle, and the transitions (T01, T12...) represent the possible switching from one phase to another based on sensor inputs and logic gates.

- **Standing**: This is the initial or resting state where the user is standing still. The transition T01 leads from the standing phase to the Swing phase, which occurs when the leg begins to move forward in preparation for the next step.
- **Swing**: In this phase, the leg is in the air, moving forward to prepare for the next heel strike. The swing phase ends when the heel makes contact with the ground, transitioning into the Heel Strike phase through T12. If the user stops mid-swing, the system may transition back to the Standing phase via T10.
- **Heel Strike**: This phase occurs when the heel first contacts the ground after the swing. This is a critical phase for stability as it marks the beginning of the stance phase. After heel strike, the foot progresses through Midstance (T20), where the entire foot is flat on the ground, supporting the body's weight.
- **Midstance**: The midstance phase is when the body passes over the planted foot. It's a balanced phase where the foot remains flat, and the bodyweight shifts forward. The transition to the next phase, Toe-Off (T34), occurs when the heel lifts off the ground, and the push-off begins, preparing the leg for the swing phase again.
- **Toe-Off**: This is the phase where the foot pushes off the ground, propelling the body forward and preparing the leg for the swing phase. After toe-off, the leg returns to the Swing phase via T41, repeating the cycle.

Safety features: The prosthetic leg is designed with the ability to exit the gait cycle at various stages, providing the user with flexibility and enhancing overall safety. These exit points allow the leg to transition out of the expected sequence of gait phases and return to a stable position when necessary.

4.2.2 Stair Mode

The Stair Mode is intended to be used for the Stair Climb, High Step and Ladder events.

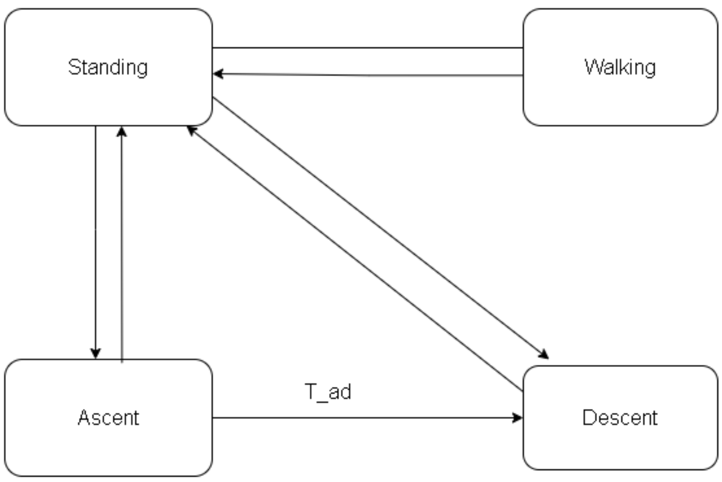


Figure 21: Supervisory State Machine

The Supervisory State Machine in Stair Mode switches between three states - Walking, Stair Descent and Stair Ascent, through an 'idle' state - Standing. From Standing, based on the nature of the consecutive Heel strike, the state is uniquely identified. Additional entry points into the Stair Descent State machine have been incorporated to make the

Overall State Machine more generalised for Stair Climbing tasks.

1. Stair Ascent:

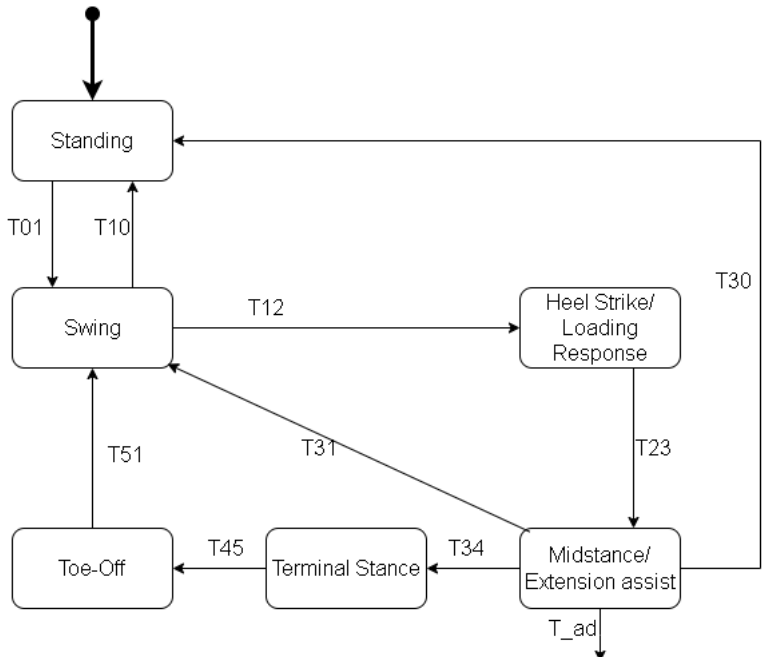


Figure 22: Stair Ascent State Machine

Transition	Condition
T01	FSRs- Sw, GRF ≈ 0
T12	FSRs- HS, GRF >0, θk > θ3
T23	FSRs- MS, GRF > w3 , θk >θ3
T34	FSRs- MS, θk ≈ 0
T45	FSRs- TO, θt < 0
T51	FSRs- Sw, GRF ≈ 0
T10	FSRs- MS, w4<GRF<w5, θk ≈ 0
T30	w1<GRF<w2, θk ≈ 0
T31	FSRs- Sw, GRF ≈ 0
T_ad	FSRs- MS, θ3 < θk < θ4

Figure 23: Transition Conditions for Stair Ascent

The state machine for the Stair Ascent mode consists of the following key states and transitions:

- **Standing**: The initial state where the user can activate stair ascent or switch to other modes.
- **Swing**: The state where the prosthetic leg moves forward, preparing for the next step.

- **Heel Strike/Loading Response**: The state where the prosthetic foot contacts the step and begins to bear weight.
- **Midstance/Extension Assist**: The phase where the knee is fully extended, and the leg supports the user's weight.
- **Terminal Stance**: The phase following midstance, where the motor prepares for the next action based on the user's intent.
- **Toe-Off**: The state where the foot leaves the step, preparing for the next swing phase.

Stair Ascent Flow:

- **Activation of Stair Mode:**
  The pilot initiates the Stair Ascent mode by pressing the corresponding button while in the Standing state. The system enters the Stair mode, readying itself for either ascent, descent or walking depending on the pilot's input.
- **Starting the Ascent:** From the Standing state, the pilot may begin stair ascent directly or start walking and then enter stair ascent mode. The system detects the pilot's intention and transitions from Standing to Swing state (T01) to initiate the gait cycle.
- **Heel Strike and Loading Response:** As the prosthetic foot strikes the step, the system transitions to the Heel Strike/Loading Response state (T12). The system monitors the loading on the prosthetic to ensure it is safe to proceed to the next phase.
- **Midstance and Extension Assist:** After sufficient load is detected during the Heel Strike/Loading Response phase, the system transitions to the Midstance/Extension Assist state (T23). During this phase, controlled knee extension is performed through motor control, stabilizing the user as they continue the ascent.
- **Terminal Stance:** Once knee extension is complete, the system transitions to Terminal Stance (T34), where the motor prepares the cable for the next action. Based on the pilot's actions, the system can transition to one of three possible states:
  - Directly transition to descent: If the pilot initiates descent, the system adjusts to facilitate this transition.
  - Continue ascent: The system continues to support the pilot through additional stair steps.
  - Enter Standing and Exit Stair Ascent Mode: The system can transition back to Standing (Tad) and enter Walking mode if the pilot decides to exit the stair ascent.
- **Toe-Off and Swing:**
  After the foot leaves the step, the system transitions to Toe-Off (T45), and then back to Swing (T51), repeating the cycle as the pilot continues the ascent. This process continues until the pilot reaches the top, at which point the system transitions back to Standing (T10).
- **Completion of Ascent:**
  Upon reaching the top of the stairs, the pilot re-enters the Standing state. From here, the pilot can either switch to Walking mode or prepare for descent. If the pilot initiates descent, the system detects this based on the nature of the heel strike and transitions accordingly, ensuring controlled lowering during the descent phase.

- **Return to Passive Mode:**
  After completing the Stair task, the pilot may press the button to exit the Stair mode and return to Passive mode, where the prosthetic behaves in a passive manner.

Safety checks

- **Unsuccessful Knee Extension in Midstance:**
  Reattempting Extension: If the knee extension is unsuccessful during the midstance phase, the pilot has the option to reattempt the extension. The system remains in the Midstance/Extension Assist state, allowing the pilot to exert additional force or adjust their posture to complete the extension.
- **Restarting Gait by Entering Swing Phase**:
  Alternatively, the pilot may choose to restart the gait cycle by transitioning back into the Swing phase. This allows for a reset of the gait process, providing another opportunity for a successful ascent.
- **Aborting Stair Ascent:**
  If the pilot feels that the stair ascent is unsafe or infeasible, they can choose to abort the process entirely by transitioning to the Standing phase. This action immediately halts the stair ascent, allowing the pilot to regain balance and evaluate the next steps safely.

2. Stair Descent:

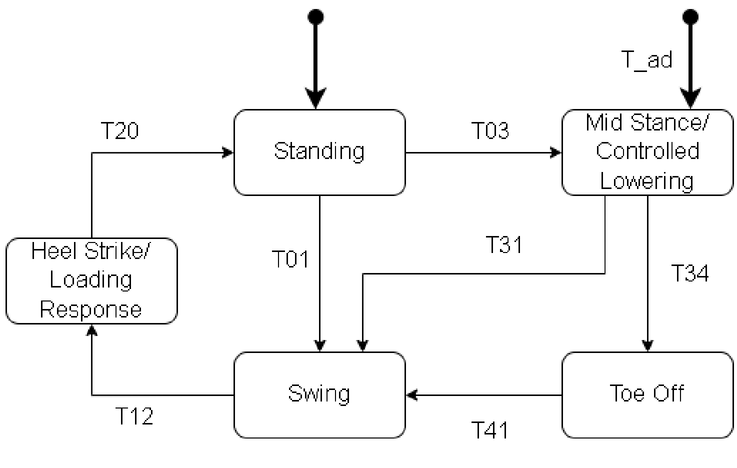


Figure 24: Stair Descent State Machine

Transition	Condition
T01	FSRs- Sw, GRF ≈ 0
T12	FSRs- HS, θk ≈ 0
T20	FSRs- HS, w1<GRF<w2, θk ≈ 0
T03	FSRs- MS, GRF ≈ bw , θ3 < θk < θ4
T34	FSRs- TO, θk > θ5
T41	FSRs- Sw, GRF= 0
T31	FSRs- Sw, GRF ≈ 0
T_ad	FSRs- MS, θ3 < θk < θ4

Figure 25: Transition Conditions for Stair Descent

Stair Descent Flow:

(a) **Activation:**
The user initiates stair mode by pressing the designated button while in the Standing state. This prepares the prosthetic leg for stair descent.

(b) **Transition to Heel Strike/Loading Response:**
Upon stepping forward with the prosthetic leg to begin the descent, the system transitions from the Standing state to the Heel Strike/Loading Response state (T20). In this state, the prosthetic leg prepares to absorb the impact of the user's weight as it contacts the stair. The sensor system ensures that the leg is positioned correctly before proceeding forward.

(c) **Transition to Mid Stance/Controlled Lowering:**
After successful Heel Strike and loading, the system transitions to the Mid Stance/Controlled Lowering state (T03). Here, the prosthetic leg carefully manages knee flexion through precise motor control, ensuring that the user's descent is gradual and controlled. The controlled lowering action is crucial for maintaining balance and preventing sudden drops due to buckling of the knee, which could lead to falls.

(d) **Transition to Toe Off:**
As the leg reaches the end of the controlled lowering phase, it transitions to the Toe Off state (T34). In this phase, the prosthetic leg prepares for the next step by pushing off the stair, ensuring a smooth transition to the subsequent descent cycle. The push-off is executed with enough force to move the leg into the next position without compromising stability.

(e) **Return to Standing:**
Once the user reaches the bottom of the stairs, the system transitions the prosthetic leg back to the Standing state.. This state serves as a neutral position, allowing the user to stabilize before moving into walking mode or performing other actions.

(f) **Mid Stance to Swing:**
This option is present keeping in mind the requirement for the Ladder task, where the pilot would have to descend backwards on the ladder.

(g) **Entry from Ascent to Descent (Tad):**
This alternate entry point into descent is present to enable the pilot to complete the High Step task successfully.

Safety checks

- **Real-Time Sensor Monitoring:**
  The system continuously monitors data from force sensors and inertial measurement units (IMUs) to confirm that the prosthetic leg is correctly positioned and loaded before initiating each transition. This monitoring helps prevent premature or unsafe transitions.
- **Freedom to decide when to descend:**
  During Stair descent, the pilot has control over when he wants to begin controlled lowering (flexion of the prosthetic knee) by bending the knee by a small angle, He can also initiate it from standing.
- **Controlled Motor Response:**
  The motor controlling the knee flexion provides gradual, controlled movements during the descent avoiding sudden, jerky motions that could destabilize the user. In case of motor failure, the system defaults to a locked position to avoid uncontrolled leg movement.

4.2.3 Balance State

The Balance Mode is intended to be used for the Balance Beam and Cross Country events

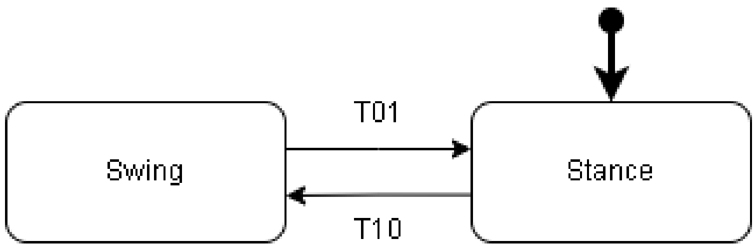


Figure 26: Balancing Mode State Machine

Transition	Condition
T01	FSRs- Sw, GRF ≈ 0
T10	FSRs- HS, GRF > 0

Figure 27: Transition Conditions

Vibrotactile Biofeedback:
Vibrotactile feedback is present throughout the Stance phase as corrective feedback. There are 4 tactors/vibration motors mounted on the thigh at the anterior, posterior, medial and lateral positions. Directional corrective feedback about the CoP is given if it goes outside a predetermined stable zone.

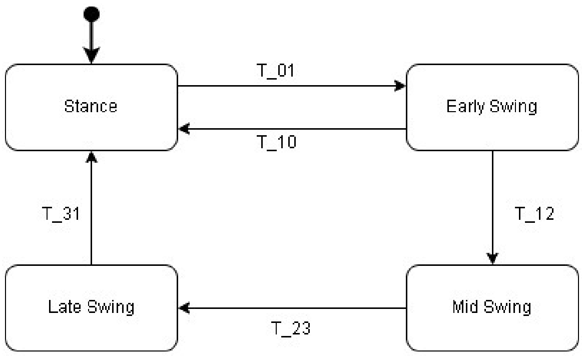


Figure 28: Flexion Control State Machine

Transition	Condition
T_01	FSRs- Sw,
T_12	FSRs- Sw, $\theta t<\theta 6$
T_23	FSRs- Sw, $\theta t>\theta 6$, $\theta t<\theta 7$
T_31	FSRs- Sw, $\theta t<\theta 8$, $\omega t<0$
T_10	FSRs- HS or FSRs- MS

Figure 29: Transition Conditions for Flexion Control

4.2.4 Flexion Control State

The Flexion control state is to be used in the hurdles and the step-over tasks, to ensure that the pilot is able to clear the height of a barrier in front of him.
Flexion Control Flow

1. Walking/Stance: This state is used when the pilot has placed his prosthetic leg on the ground, and the sound leg is in swing phase.
2. Early Swing: As toe-off takes place, the transition to this state takes place. We stay in this state until the user raises his leg above a threshold, and then the transition to mid-swing takes place. If the user does not lift his leg enough, and starts to lower it again then the transition back to stance takes place.
3. Mid-Swing: In this state the motor is wrapped to first make the flexion cable taut, and then the cable wraps to ensure that the toe does not hit the wall in front of the pilot. The toe position is estimated using the IMU angles. The user controls this motion using his hip to control thigh angle. When he has raised his prosthetic leg to the highest desired point, then the transition to Late swing takes place.
4. Late-Swing: As this state is entered, the flexion cable is unwrapped slowly, ensuring no pendulum movement when the leg is in air. As he lowers his leg and touches it on the ground again, it returns back to walking/stance phase.

4.3 Calibration

For repeatable and reliable trials, we developed a calibration procedure for the sensors, as described below.

4.3.1 Force Sensor Calibration

The op-amp circuit for FSR readings, as described in Section 3.8, is used to linearize the relation between input force to output voltage. The potentiometers in this circuit used as feedback resistors can be adjusted to change the force sensitivity.

- As the FSRs are placed on the insole of the user's shoe, the force is transferred from the ground to the sensors through the cushioned shoe sole, effectively decreasing the sensitivity.
- The main purpose of our FSRs were to detect when the user is bearing weight on his prosthetic foot, so we could afford to have very high sensitivity at the cost of force measurement range, as low sensitivity can be afforded at higher forces.
- The sensors should have a range high enough so that they do not saturate when the shoe is worn but in the air.

For the above reasons, our FSR calibration procedure was as follows:

1. Place the FSRs (attached on the insole) on a hard surface, such as a table (or on the floor).
2. Connect them to the op-amp circuit and power it on. Measure the voltage as a digital output using the ADC pins on the sensor board.
3. Note the mean of 20 consecutive readings as the resting voltage output of the FSR.
4. On just touching the FSR lightly with a finger (or placing the foot gently on the sensor, without loading), the voltage should not fluctuate from the resting value.
5. On pressing as hard as possible with a finger (or applying full body weight on the sensor), the digital measurement should not fully saturate. It could ideally go upto 75-80% of the saturation value.

4.3.2 IMU Calibration

1. Ensure that the IMUs are mounted securely, powered and the measurements are being logged.
2. Straighten the prosthetic knee completely - record the readings of both IMUs. The difference between the relevant angles on both IMUs should read zero. If there is a slight offset between the values, it can be accounted for in software.
3. As the knee angle is increased (the knee is further bent), the difference between the IMU angles should increase. If this is not the case, it may be an issue with the orientation of the IMU. Again, this can be accounted for in software by applying the required transformations.

4.3.3 FSM Threshold Calibration

Force Thresholds: The FSRs attached on the insole of the user's shoe, which had to be removed frequently for minor adjustments. As a result, the distribution of forces between the sensors and the rest of the foot could change, thus making it necessary to calibrate FSM force thresholds every time the shoe is worn.

1. Connect all the sensors in the shoe to the sensing PCB, and allow the user to wear it while sitting.

2. While the foot is in the air, record the resting value for the sum of the FSR outputs (by taking average of 20 consecutive readings) as $tFSR\_Sw$.

3. Allow the user to stand with both feet on the ground but bearing more weight on the unaffected leg. Record the sum of the FSR values as $tFSR\_St$.

4. Ensure that on loading different parts of the leg, the corresponding FSR values are able to rise sufficiently. According to the observations, the fractions $tFSR\_HS$ and $tFSR\_TO$ can be adjusted if deemed necessary.

Angle Thresholds: The angle thresholds are fixed either based on data collected from user trials for specific tasks, or based on the human knee angle required for similar tasks. This need not be calibrated every time - if it has been correctly obtained once, the same thresholds may be used as long as IMUs are calibrated.

4.3.4 Motor Position auto-calibration

`void cable_taut(float vRef)` is used to detect when a cable becomes taut by driving a motor at a given reference velocity and monitoring the actuator's velocity and torque. The function stops once the cable is confirmed taut, an interrupt is received, or a timeout occurs.

Inputs

- `vRef`: Reference velocity for the actuator to begin pulling the cable.

Key Conditions for Tautness

- The actuator was previously moving (`v_out` exceeds threshold).
- Velocity has dropped below a small threshold (`ZERO_VELOCITY_BAND`).
- Torque feedback is sustained above a threshold.

Failsafes

- 8-second timeout.
- Serial interrupt via `Serial.available()`.

Algorithm 1 `cable_taut(vRef)`

```
Initialize taut = false, moving = false, counter = 0
Declare filtered_t_out[3]
Set timer = millis()
while !taut and !Serial.available() and millis() - timer < 8000 do
    if abs(v_out) > 3 then
        moving = true
    end if
    Set kp_in = 0, kd_in = 5, v_in = vRef
    if abs(t_out) < 15 then
        Store in filtered_t_out[counter]
    end if
    Increment counter; wrap to 0 after 3
    Compute average torque from filtered_t_out
    do_each_loop('f')
    if abs(v_out) < ZERO_VELOCITY_BAND and avg_t_out > 0.35 and moving then
        taut = true; break
    end if
end while
Set v_in = 0, t_in = 0, kp_in = 0, kd_in = 0
do_each_loop('f')
```

4.4 Controls

4.4.1 Motor Controls and Algorithms

The lower-level control was performed by the KV80 driver provided with the AK80-64 motor. It can be configured (through CAN) for different control modes such as position, velocity, torque or combinations of these. We used the MIT Mode, which used the following control block diagram:

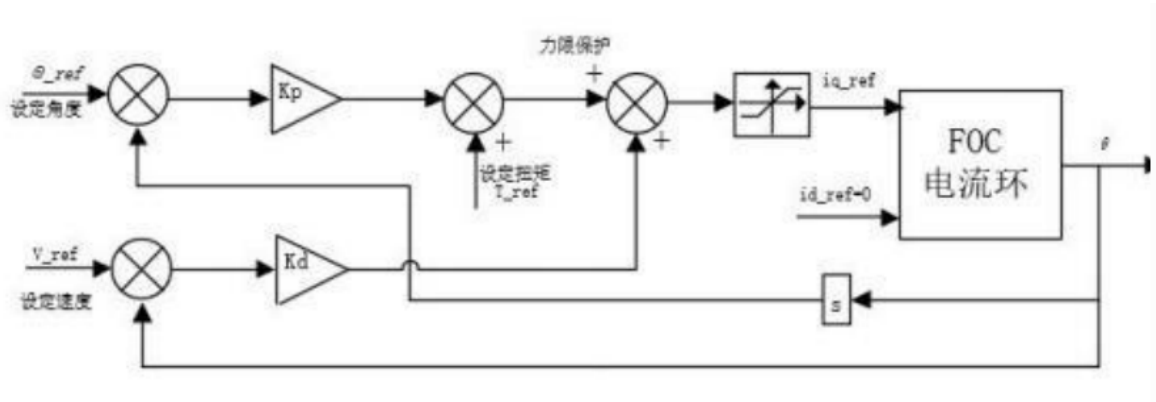


Figure 30: MIT Mode Control Loop CubeMars

Communication to the driver board was through special CAN codes, and we wrote functions that send these CAN messages to provide a high-level interface for motor control that had the following utilities:

- Enter motor control mode
- Exit motor control mode
- Set current motor position as zero
- Reset all control inputs to zero
- Set Kp, Kd
- Set desired position θ_{ref}
- Set desired velocity v_{ref}
- Set desired torque T_{ref}
- Cable taut (for motor position auto-calibration)

The functions in this interface can be used real-time by passing serial inputs to the ESP32 directly through USB or over a web-server. Using these functions, we further developed various task-specific controllers along with relevant safety measures.

4.4.2 Stair Ascent

`void Stair_Ascent_Loading()` applies a staged torque profile to assist in lifting the limb during stair ascent. It monitors thigh and shank angles to determine appropriate torque output, ramps the torque over time, and modifies it further based on knee angle conditions. The function exits when the pose no longer matches a stair-ascent loading condition.

Key Inputs

- Sensor data:
  - `sensors.thighAngle`
  - `sensors.shankAngle`
  - `sensors.kneeAngle`
- Control parameters:
  - `iRef1`, `iRef2`, `rampTime1`, `rampTime2`

Torque Phases

- **Phase 1 (Initial ramp)**: Feedforward torque profile increases to `iRef1`.
- **Phase 2 (Ramp to peak)**: Torque linearly increases from `iRef1` to `iRef`.
- **Phase 3 (Steady torque)**: Torque remains constant at `iRef`.
- **Knee-based reduction**: If knee angle is below threshold, torque is attenuated proportionally.

Algorithm 2 `Stair_Ascent_Loading()`

```
CAN_receive(), GRF_FSRs()
Initialize constants: iRef1 = -2.0, iRef = -4.5, rampTime1 = 75, rampTime2 = 175
iRef = constrain(iRef, -10, 10)
Initialize: counter_loc = 0, enter_highAngle = false, startTime = millis()
while thighAngle > 10 && abs(shankAngle) < 20 && kneeAngle > 10 do
    Increment counter_loc
    if counter_loc < rampTime1 then
        t_in = (iRef1 / rampTime1) * counter_loc
    else if counter_loc < rampTime2 then
        t_in = iRef1 + (iRef - iRef1)/(rampTime2 - rampTime1) * (counter_loc - rampTime1)
    else
        t_in = iRef
    end if
    if kneeAngle < 50 then
        if enter_highAngle == false then
            iRef2 = t_in, enter_highAngle = true
        end if
        if kneeAngle > 25 then
            t_in = iRef2 * (kneeAngle / 25)
        else
            t_in = 0
        end if
    end if
    do_each_loop('s')
end while
```

4.4.3 Stair Descent

`void descentController()` is a control function designed to apply a variable torque profile for safe and stable descent. It modulates torque based on the relative displacement between an input and output position signal (`p_in`, `p_out`) and the output velocity (`v_out`). The control operates in multiple regions defined by positional thresholds.

Key Variables

- `::p_in`, `::p_out`: Input and output positions.
- `::v_out`: Measured output velocity.
- `::t_in`: Torque command.
- `k1`, `k2`, `k3`, `d1`, `d2`, `d3`: Gains Scheduled for different bands in the descent gait (scheduled as a function of θ_k)
- `region1`, `region2`, `region3`: Position difference thresholds dividing the control strategy.

Behavior

The function computes the relative position error:

$$e = -\mathtt{p\_in} + \mathtt{p\_out}$$

and applies a scheduled impedence controller, based on different regions that e falls into.

Algorithm 3 `descentController()`

```
Compute e = -p_in + p_out
if 0.05 < e < region1 + 0.1 then
    t_in = -k1 * e - d1 * v_out
    Print "1"
else if region1 < e < region2 then
    t_in = -0.3 - k2 * e - d2 * v_out
    Print "2"
else if region2 < e < region3 then
    t_in = -0.3 - k3 * e - d3 * v_out
    Print "3"
else if e > region3 then
    t_in = -2
    if v_out > 5 then
        t_in = -2 - 0.08 * v_out
    end if
    Print "4"
end if
do_each_loop('d')
```

4.4.4 Monitoring and Safety Checks

This section summarises the key safety checks, bailout mechanisms, and monitoring strategies implemented across the three critical control functions

- `cable_taut()`
- `descentController()`
- `Stair_Ascent_Loading()`

1. *Velocity Monitoring*
    - `cable_taut()`: Uses `v_out > 3` to check if the joint is moving, and later uses a velocity deadband (`ZERO_VELOCITY_BAND`) to identify when motion ceases.
    - `Stair_Ascent_Loading()`: Includes velocity clamps for safe operation (e.g., applying damping torque if `v_out < -7`).
2. *Angle Constraints*
    - `Stair_Ascent_Loading()`: Main loop runs only while:
        - `thighAngle > 10`
        - `abs(shankAngle) < 20`

- `kneeAngle > 10`

These thresholds restrict torque application to kinematically stable regions.

3. *Torque Clamping*
   - `Stair_Ascent_Loading()`: Torque reference `iRef` is clamped using `constrain(iRef, -10, 10)` to ensure actuator safety.
   - `descentController()`: Region 4 sets an upper bound on braking torque (`-2 - 0.08 * v_out`).
4. *Temporal Limits*
   - `cable_taut()`: Includes an 8-second timeout to prevent indefinite looping.
   - `Stair_Ascent_Loading()`: Optional timeout after 8 seconds triggers an error condition:
     - `ERROR_STATE = 1`, `ERROR_CODE = 3`
5. *Mode Monitoring*
   - `Stair_Ascent_Loading()`: Contains a bailout if `currentMode == Passive`, allowing external FSM to abort early.
6. *Bailout Conditions*
   - Cable timeout: If cable does not reach taut condition in 8 seconds, `cable_taut()` exits cleanly with zero torque.
   - Angle thresholds: In `Stair_Ascent_Loading()`, torque is reduced to 0 if knee angle drops below 25°, avoiding unsafe loading.
   - Manual override: Several places a runtime exit based on velocity overshoot, passive mode, or timeout.

5 Trials

5.1 User Selections

In order to select a user, we had a networking event at NCAHT, IITM Research Park, where we invited prosthetic leg users from all over Chennai and had them participate in some fun games and activities which helped the clinical team analyze their activities. Besides this, the clinical team also diagnosed the users one-on-one. At the end, there was a talk covering useful topics such as how to maintain hygiene inside the socket and how to fall safely.

The selection of our user was based on the following criteria:

- Unilateral transfemoral amputation
- Aged between 18 to 40
- Weight between 40 to 100 kgs
- Amputation level K2-4
- No residual knee function

- Muscle strength (hip flexors, hip extensors, hip abductors, hip internal rotators, hip exteral rotators)
- Endurance and balance assessment
- Ability to stabilize neck and head
- Ability to stabilize trunk
- Sensory ability in the residual limb
- Use of any other assistive devices (pacemakers, implanted medication pump, etc.)

Besides these, personal and logistical criteria were taken into account, such as the availability of users in the IITM campus and their willingness.

Based on all of these points, two candidates with differing levels of abilities were shortlisted:

- User 1 (Krishna): Primary pilot, experienced transfemoral prosthesis user
- User 2 (Tamilalagan): Secondary pilot, first-time prosthesis user

5.2 Prosthetist Training Exercises

A new socket was manufactured for both users, and the passive leg with Kadam knee was assembled. There were 4-5 training sessions with each user just to get them accustomed to the Kadam knee dynamics, where some basic exercises were performed to improve balance and gait:

- **Static**: These exercises aimed to improve the balance of the user and ensure proper weight-bearing.
  - Static Standing - wide base of support
  - Static Standing - narrow base of support
  - Standing on unaffected limb >10 seconds
  - Standing on affected limb >10 seconds
  - Tandem standing with unaffected leg forward
  - Tandem standing with affected leg forward
- **Dynamic**: There exercises help improve endurance, balance and agility.
  - Normal Walking
  - Walking laterally and forward across obstacles
  - Walking up and down a ramp
  - Walking on uneven surfaces
  - Turning 360°
  - Tandem walking
  - Climbing down stairs with unaffected leg
  - Climbing up stairs with affected leg

5.3 Biofeedback Trials

5.3.1 Positioning of Tactors

First, to decide where to place the tactors, we had to test the sensitivity of our user to the vibrations at different points on his stump. We did this by placing one tactor at different points on the residual limb, and giving the highest amplitude of vibration possible for our chosen coin vibration motors. We made the following observations:

- The sensitivity of the skin reduces in more distal areas, i.e. closer to the site of amputation.
- The sensitivity is highest along the medial aspect of the limb, when assessed circumferentially.

Next, we placed four tactors on small velcro patches and stuck them at different lengths along the velcro band which was worn by the user around his thigh. Placing tactors less than 2cm apart results in perceptual overlap, making it difficult for users to distinguish which tactor is vibrating. After some trials, we reached an arrangement wherein the tactors were spaced apart equally, so that the user could easily differentiate between the vibrations from different tactors.

5.3.2 Vibration signal patterns

To investigate which vibration signals and variations users could interpret, we placed two tactors on the thigh and two FSR sensors on the feet. Each tactor was controlled by the measurement of one FSR, with the aim of conveying which part of the foot was bearing load. During the test, we manually activated the tactors and asked the user to indicate which tactor they believed was being loaded.

- Ramp up/down frequency: When the vibration frequency was varied from zero to maximum, users could feel the difference only when the change was big. Small changes in frequency were hard to notice, so only a small amount of information could be communicated this way. Hence, the frequency was fixed at 250 Hz, which is the vibration frequency to which the skin has maximum sensitivity.
- Ramp up/down amplitude: Small steps in amplitude variation could be noticed more easily, so this was a viable option.
- Although we experimented with vibration patterns such as single pulses, double pulses, and bursts, our user indicated that it was harder to interpret their significance. Modulating the amplitude of vibration emerged as the most intuitive method for conveying force information from the foot.

5.3.3 Feedback Modes

Having understood the best method to convey information to the user, now we had to decide what information was to be conveyed and when. Some of the feedback strategies we attempted were:

1. **Corrective Feedback:** Used for tasks like walking on a balance beam. Whenever the knee angle deviated beyond a predefined threshold indicating an imbalance, vibrotactile feedback was delivered through the tactors to prompt the user to

adjust their posture or gait. This feedback was intended to train the user to maintain correct alignment during dynamic movements.

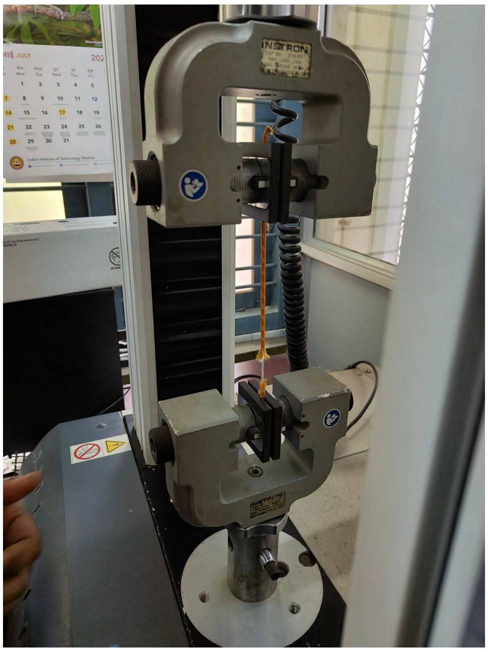
2. **Event-Based Feedback:** This mode provided haptic cues when specific events occurred, such as heel strike or toe-off, detected using the FSRs on the foot. These cues helped improve gait timing and foot placement awareness.
3. **Proportional Feedback:** The intensity of vibration from the tactors varied proportionally with the knee flexion angle, allowing the user to perceive how much the knee was bending in real time. This continuous feedback helped in fine-tuning limb control during various activities.
4. **Threshold-Based Alerts:** In this mode, feedback was delivered only when the measured force or angle exceeded a certain limit (e.g., overloading during stance phase). This helped in preventing compensatory movements and improving safety.

To measure the effectiveness of the vibrotactile feedback, we followed a structured testing protocol. We recorded quantitative metrics such as the time taken to complete specific tasks and the number of times participants looked at their prosthetic leg during the activity, as visual attention can indicate a lack of confidence or proprioceptive awareness. However, due to time constraints and limited access to users, we were unable to conduct long-term trials. Extended usage would be necessary to evaluate learning effects and adaptation to the feedback system, which are crucial for determining its long-term usability and effectiveness.

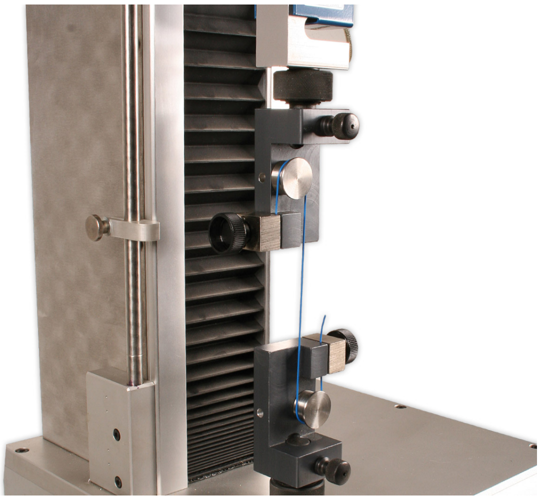
5.4 Mechanical Validation

5.4.1 UTM Test

To check the maximum tensile strength of our rope we conducted the test on a UTM machine in the Composite technology centre opposite MDS. A picture of the setup is given below.



This is however not the right way to measure the tensile strength for ropes but instead this is the recommended setup.



We tried two different rope materials:

1. Polyamide 3mm diameter
2. Dyneema 2mm diameter

The polyamide rope was able to withstand a maximum tension of 0.8kN before breaking. However, it broke at the points where it was clamped by the fixtures, indicating that it could have been due to shear and not tensile loads. The dyneema cable was able to withstand upto 1.4kN before breaking.

The result of the UTM test is shown below:

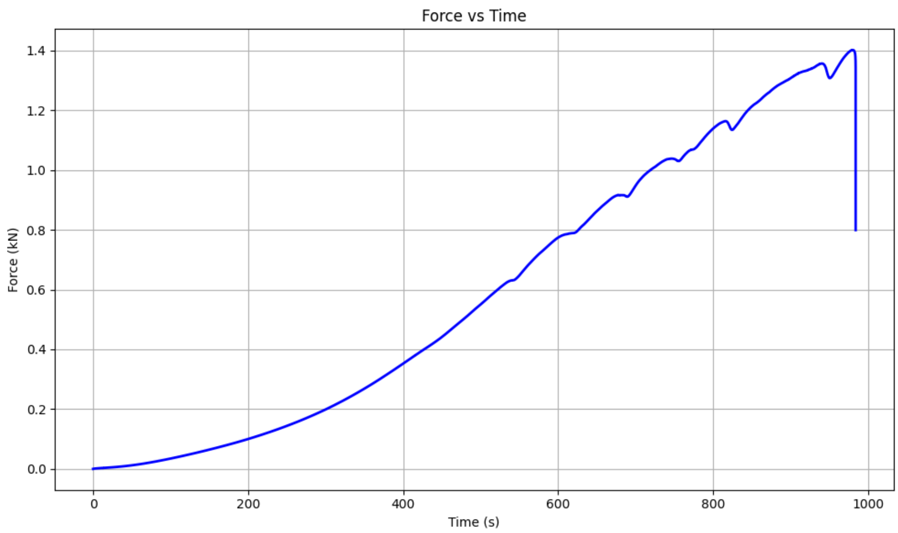


Figure 31: Force (kN) vs time (s)

5.5 Data Collection

For setting the threshold values in various logics used in our controllers and finite state machine, we needed user data from Kadam knee trials. We equipped our user with the sensorized passive Kadam knee, containing the two IMUs (shank and thigh) and four FSRs on the foot. These sensors were connected to two ESP32s, which sent data using WiFi to a laptop running a Python logging code. Besides, we recorded videos of the user and matched the time-stamps to the collected time-series data. We collected the data for a variety of tasks performed at NCAHT:

- Normal walking
- Walking up / down a ramp
- Stair ascent / descent
- Walking on uneven terrain (stones)
- Tandem walking

Based on the ground reaction force, shank angle, thigh angle and knee angle data collected from these passive tasks, we established the values for various thresholds used in our active setup as described in Sections 4.2 - 4.4.

5.6 Actuation Trials

Write about different approaches we tried and then our results

5.6.1 Stair-Ascent

To enable powered stair ascent using the Kadam knee prosthesis, we explored and evaluated multiple control strategies. Each strategy was developed with the goal of ensuring smooth, reliable motion across a range of users, while accounting for the non-linear characteristics of the knee joint.

1. Velocity-Based Trajectory Control

   The initial approach involved tracking a predefined joint velocity trajectory derived from literature. This trajectory was mapped using data from a survey paper that aggregated average stair ascent velocities across different age groups and body dimensions.

   Limitations:

   - The Kadam knee exhibits strongly non-linear dynamics.
   - Applying a linear velocity trajectory resulted in poor tracking performance, as the prosthetic could not respond accurately to the commanded velocities.
   - The mismatch between the control input and the system's response made this method unsuitable for consistent stair ascent.

2. Feedforward Torque-Based Control with State Machine

   To better accommodate the prosthetic's non-linear behavior, we transitioned to a **feedforward torque trajectory** approach. This method leveraged a predefined torque profile designed to account for the Kadam knee's specific dynamic characteristics.

   This control strategy was integrated into a finite **state machine**, with three key phases:

(a) **Foot Loading Phase:** Represented the initial contact and weight transfer onto the stair. This phase remained passive to allow the user to stabilize and transfer weight safely.
(b) **Active Torque Phase:** During this phase, the feedforward torque controller was activated to provide the necessary knee extension torque to propel the user upward onto the stair.
(c) **Unloading and Leg Swing Preparation:** After completing the extension, the system gradually ramped down the torque and executed controlled cable unwinding. This introduced slack in the mechanism, allowing the leg to be lifted and positioned onto the next stair with minimal resistance.

This strategy significantly improved stair ascent performance by aligning control inputs with the prosthesis's mechanical characteristics and the dynamic requirements of stair climbing.

5.6.2 Stair-Descent

Stair descent presents unique challenges for prosthetic knee control, particularly due to the need for active energy dissipation and controlled support throughout the motion. Unlike level-ground walking, stair descent requires the prosthesis to absorb impact forces and allow smooth knee flexion while bearing the user's body weight. Our objective was to enable confident, safe, and natural stair descent using the Kadam knee prosthesis.

1. Initial Impedance Control Approach

   Our first strategy employed a fixed **impedance controller** that applied a combination of stiffness and damping to the knee joint. This controller mimicked the behavior of a passive mechanical spring-damper system, allowing the user to descend stairs with controlled resistance that prevented sudden free-fall or buckling of the prosthetic knee.

   To realize this, we implemented a position-based control loop on the motor, effectively using motor position tracking to regulate the apparent impedance felt by the user. The stiffness ensured the knee resisted undesired flexion under load, while the damping component helped absorb shocks and smooth out the descent.

   $$\tau = -K(\theta - \theta_0) - B\dot{\theta} \tag{1}$$

   where:

   - τ is the output torque,
   - K is the stiffness gain,
   - B is the damping gain,
   - θ is the measured knee angle,
   - θ_0 is the desired equilibrium angle, and
   - $\dot{\theta}$ is the angular velocity of the knee joint.

   This approach proved effective in providing basic descent capability and support. However, it lacked adaptability across the different stages of stair descent and among different users. Some phases of the descent required high support, while others demanded increased flexibility. A single fixed impedance profile could not optimally serve all needs.

2. Gain-Scheduled Impedance Control with Joint-Angle Dependency

   To address the limitations of the fixed impedance controller, we developed a more responsive control scheme using **gain-scheduled impedance control**. In this framework, the prosthetic dynamically adjusted the virtual stiffness and damping values based on real-time estimates of the knee and hip joint angles.

   The rationale for this approach was rooted in biomechanical observations:

   - In the **early phase of stair descent**, when the user is initiating the step down and transferring weight onto the prosthesis, high support is critical. Increased stiffness and damping in this phase provide resistance to knee buckling and instill user confidence.
   - In the **middle phase**, as the user transitions weight and the knee flexes under load, moderate support is needed to manage impact and allow smooth progression.
   - In the **late phase**, just before swing, lower impedance is beneficial. It reduces the effort required to lift the prosthetic leg and allows natural transition to the next step.

   To implement this, we defined discrete **bands** based on the combination of knee and hip angles. Each band was associated with a specific set of impedance gains. As the user moved through these bands, the prosthesis adaptively transitioned its behavior using a finite state machine or a continuous scheduling function.

6 Cybathlon 2024

CYBATHLON is a global competition hosted by ETH Zurich where research laboratories collaborate with assistive technology users to compete in a series of real-world, challenge-based tasks. The event pushes the boundaries of both academic research and clinical training, fostering innovation in fields such as brain-computer interfaces (BCIs), advanced prosthetics, functional electrical stimulation (FES), and more.

Team ASAI from the R2D2 Lab at IIT Madras proudly represented India in the Leg Prosthesis Race, competing with the Kadam Knee, a powered above-knee prosthesis developed in-house. The team scored 50 out of 100 points, securing 8th place globally among top institutions and research groups.

The experience at CYBATHLON provided invaluable exposure to cutting-edge research, as the team interacted with international labs, each bringing specialized innovations in assistive mobility. The event not only served as a technical benchmark but also reinforced the importance of designing for real-world functionality, user comfort, and robustness.

Below are some pictures of the team and pilot from the event.

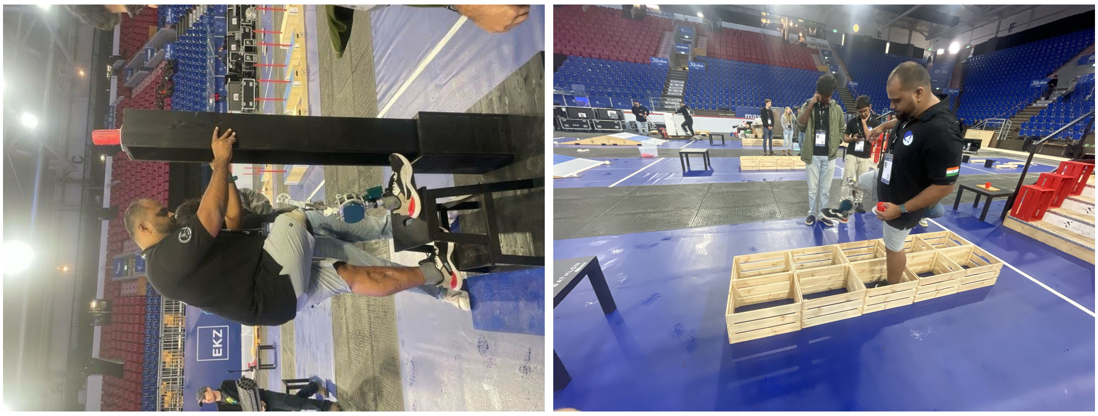


Figure 32: Our Pilot Krishna practicing at Cybathlon, Switzerland

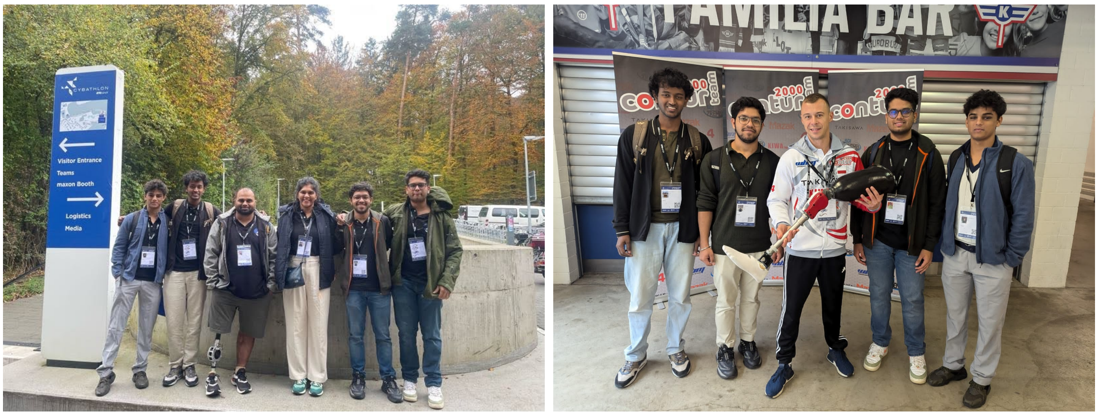


Figure 33: Team ASAI representatives at Cybathlon

Figure 34: More photos of our pilot Krishna at the competition