Inconspicuous Ankle Foot Orthosis (AFO) for Teen

FINAL REPORT

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Abstract

Facioscapulohumeral Dystrophy (FSHD) is one of the most common types of muscular dystrophy, affecting approximately 4-10 in 100,000 individuals. This genetic disorder leads to progressive muscle weakness, and while there is currently no cure, physical therapy, orthotics, and surgery can help manage symptoms. [1] There is limited clinical research focusing on children with FSHD, leaving gaps in understanding and addressing their specific needs. This project aims to raise awareness of FSHD and explore the benefits of discrete ankle-foot orthoses (AFOs) for individuals with progressive muscle weakness.

Current devices that provide the flexibility the patient seeks include flexible-dynamic, jointed, and passive-dynamic AFOs. However, these designs fail to meet key patient needs, such as discreteness, sufficient flexibility, and ankle inversion prevention. To address this, a discrete AFO for the right foot will be fabricated for a teenager diagnosed with FSHD. The design aims to support dorsiflexion to prevent foot drop, provide flexibility for daily activities, and prevent ankle inversion. The current rigid support design balances appearance and functionality. It will be made from carbon fiber reinforced PLA (CF-PLA), include a gel pad to minimize comfort, and be attached to the foot with elastic straps. The bungee cord system from the Fall 2024 design team will be incorporated with the new rigid support design to allow for adjustable tension, facilitating dorsiflexion. Current plans for testing this design include Finite Element Analysis, Comfortability Testing, and IMU and MoCap Testing.

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Introduction

Motivation & Global Impact

Currently, the majority of AFOs are rigid, bulky, and unappealing. The patient is a teenager in high school who has FSHD, and is in need of an AFO for their right foot. Current devices are not aesthetically appealing and the patient does not want to draw attention from unwanted peers. The goal is to create a comfortable and flexible AFO that the patient will wear without feeling self-conscious. This will allow the patient to go about their normal day-to-day activities without worrying about their safety or about peers' opinions.

Additionally, there are currently limited clinical trials for FSHD in young individuals. The device will raise awareness for FSHD affecting young individuals, and increase the amount of research on how FSHD affects children.

On a global scale, the device can be made customizable to other young individuals who need an AFO but do not want to draw attention to their condition. The device could help other children with FSHD who are feeling self-conscious about their condition. This device can also potentially meet other markets, such as adults who also want an inconspicuous AFO. Additionally, it could be worn by individuals who have a different type of muscular dystrophy, or simply need extra support in their ankle.

Existing Devices & Current Methods

Current orthotics use the Three-Point Pressure system. This is where the force on the corrected joint is countered by forces above and below the joint so the sum of all forces is zero. This relieves discomfort for the patient [2].

Pediatric Supra Malleolar Orthosis (SMOs), depicted in Figure 1, are made from thin and flexible thermoplastic that provides support just above the ankle bones (malleoli). SMOs are primarily used to control subtalar joint alignment to maintain a vertical or neutral heel to help improve mediolateral movement. They are comfortable to wear in shoes due to their thin and minimally restrictive design [3].

Figure 1: Supra Malleolar Orthosis (SMO) [4].

Jointed AFOs, as seen in Figure 2, are designed with a hinged joint that allows for a full range of motion; however, some limitations to this device are that it is bulky and difficult to fit in standard footwear. This device can be noisy due to the hinge mechanism and may break more easily. The patient is looking for an orthosis that allows for some range of motion; however, due to the bulkiness of this device, it does not meet the criterion the patient is looking for [5].

Figure 2: Jointed AFO [6].

The Passive-Dynamic AFO (PD-AFO), as seen in Figure 3, consists of a calf shell and a foot plate. This AFO provides flexibility by allowing both dorsiflexion and plantarflexion. As the calf shell bends in the stance phase, elastic potential energy is stored. The energy is later released during the push-off phase, supporting the user throughout the gait cycle. More commonly, PD-AFOs are made from 3D-printed materials; however, due to the flexibility of this device, it should be worn by individuals with less severe foot drop [7]. This device meets the flexibility requirements of the patient, but may not support the patient medially, so it is not a viable option.

Figure 3: Front View of Passive-Dynamic AFO [8].

Variable Stiffness Orthoses (VSOs), as shown in Figure 4, are powered and feature a customizable cam-based transmission able to specify any torque-angle and change the magnitude of its overall stiffness in real time. VSOs are found to lead to reduced foot drop and increased total ankle moments. VSOs are currently in the research stage and not on the market [9].

Figure 4: Variable Stiffness Orthosis [10].

Problem Statement

AFOs are designed to provide dorsiflexion support during the swing phase of walking. These devices are primarily used to treat muscular dystrophies. This project aims to raise awareness of FSHD and explore the benefits of discrete AFOs for individuals with progressive muscle weakness. The team aims to design a discreet, flexible brace that supports natural gait, enhances walking safety, and allows for a functional range of ankle motion. The brace will be tailored specifically for the patient. The key objectives of this device include positioning the ankle in adequate dorsiflexion, restricting medio-lateral ankle motion, maintaining a narrow, thin, and discrete design, and ensuring sufficient flexibility to minimize any restriction of movement.

Background

The team was tasked with creating an AFO for the client, Debbie Eggleston, and her patient. The patient has FSHD, and the disease has progressed to the point of needing an AFO. The client explored traditional AFOs but found them too bulky and restrictive for the patient's needs. The patient is a sophomore in high school, and is concerned about the questions and judgment they may receive; therefore, they are looking for a more inconspicuous AFO. An AFO will be created that is similar to an ankle brace, limiting questions the patient may receive, while still providing the necessary support for the right ankle.

Anatomy & Physiology

FSHD is a rare neuromuscular condition that causes progressive weakness of muscles in the shoulder girdle, hip girdle, and lower limbs. Therefore, FSHD patients tend to suffer from foot drop because of their weakened muscles which affects their gait cycle and increases their risk of falling. FSHD is the third most common type of muscle dystrophy, and 1 in every 15,000 people suffer from FSHD. [11] It most commonly affects females in their late twenties to early thirties. There are two types, FSHD1 and FSHD2, in which 95% of patients have FSHD1. [11]

The client specifically has FSHD1, which is an autosomal dominant muscle disorder, mapped to 4q35 on chromosome 4. The EcoRI fragment is partially deleted, less than 35 kb [kilobase] in length, instead of a normal 35-300 kb or with repeated copies. [13] Additionally, it has been found that mutations in epigenetic regulators lead to FSHD. [14] The overexpression of the DUX4 gene on chromosome 4 in the D4Z4 region is another cause of FSHD. The DUX4 protein is active at low levels during fetal development and is silenced as development progresses and in most adult cells. Hypermethylation, a large number of methyl groups attached to the DNA of the D4Z4 region on chromosome 4, silences the DUX4 protein. [15] However, if the DUX4 protein is activated again, it damages muscle cells leading to FSHD diagnosis.

The AFO's primary purpose is to support the foot in dorsiflexion, as shown below in Figure 5, in order to fix foot drop.

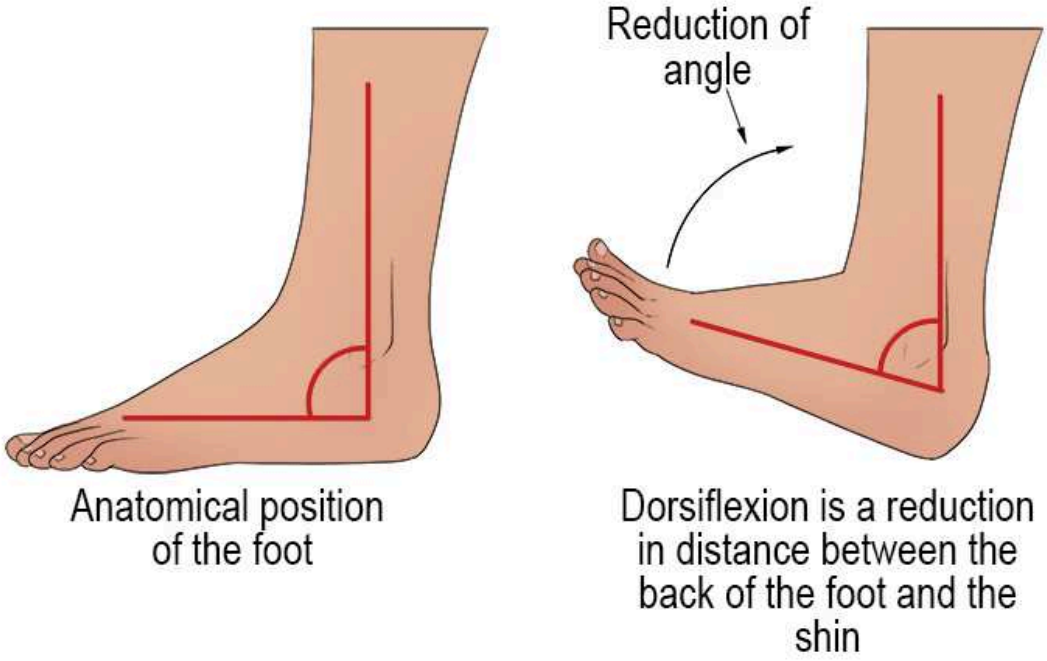


Figure 5: Foot dorsiflexion [16].

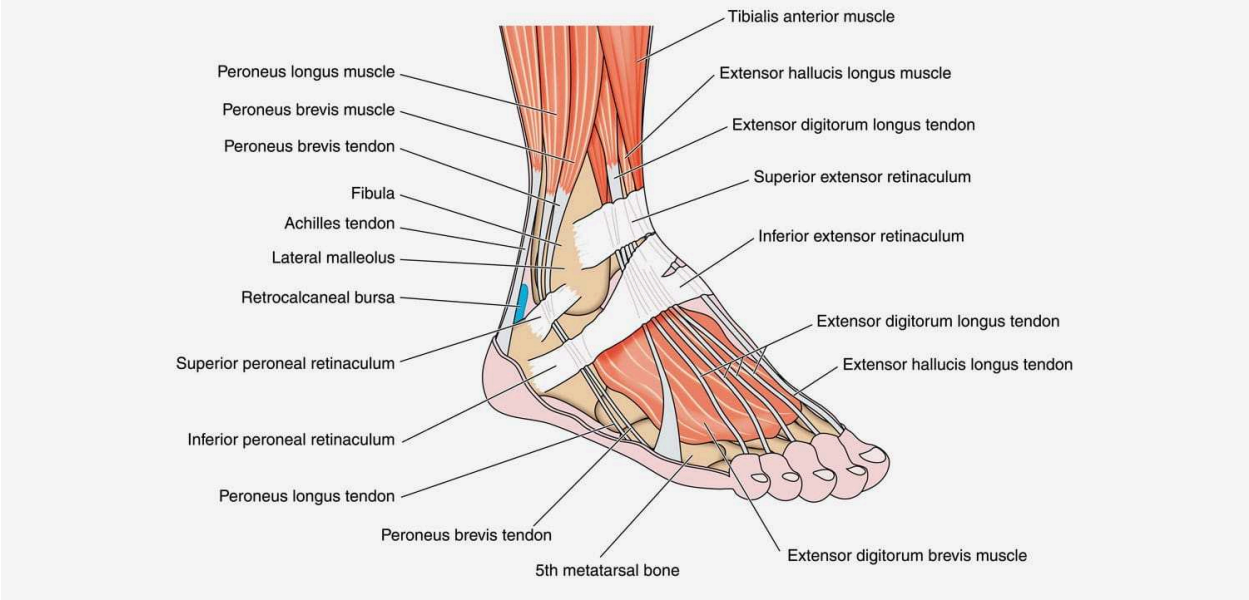


Figure 6: Ankle Muscle Anatomy [17].

Foot drop occurs when the muscles extending distally to the ankle, specifically the tibialis anterior shown above in Figure 6, are too weak to support the foot in a normal position, leading to an excess drop of the foot. This affects the initial contact or heel strike phase of the gait cycle shown in Figure 7. The patient's foot can catch on the ground, increasing the risk of fall while walking.

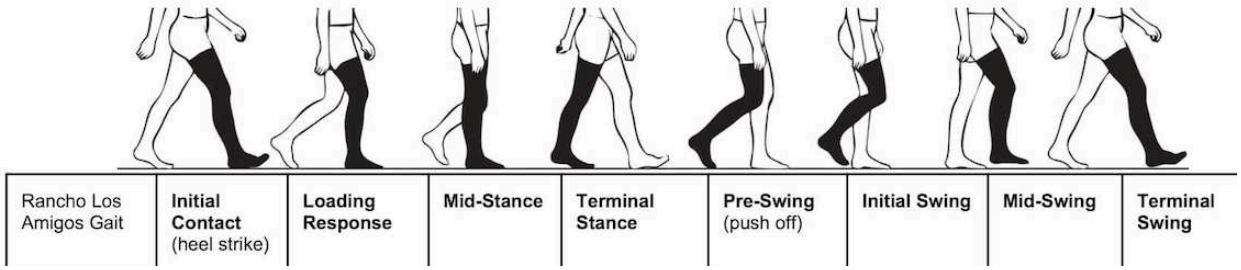
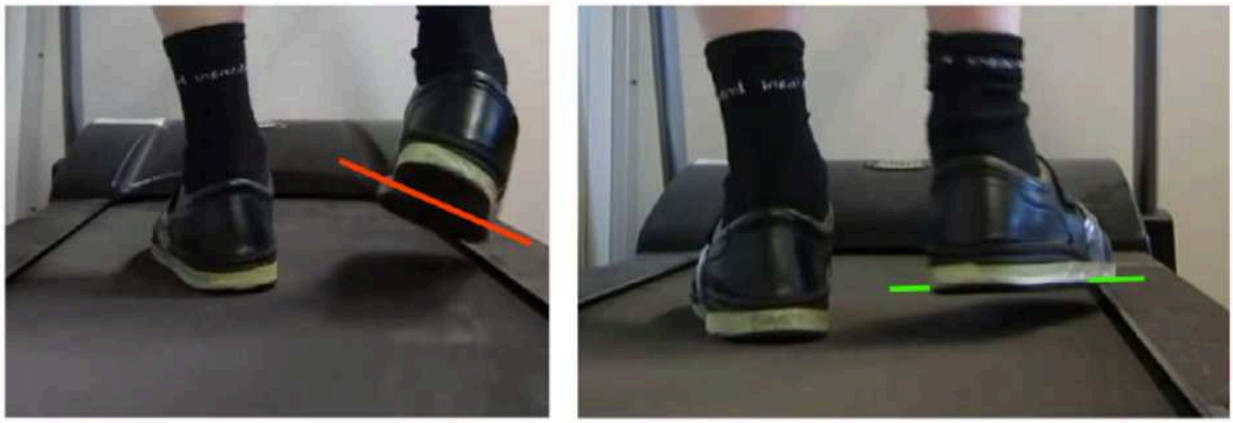


Figure 7: Normal Gait Cycle [18].

The patient also experiences ankle inversion shown in Figure 8. The medial side of the foot is rotated inward in compression, while the lateral side of the foot is in tension. This poses a risk of straining the tendons of the ankle. The patient also requires arch support in the foot, which is already implemented in the form of inserts in athletic shoes.



(d) excessive foot inversion (negative roll) (e) neutral inversion/eversion (zero roll)

Figure 8: Ankle Inversion During Gait vs. Normal Gait [19].

Client Information

The client, Debbie Eggleston, is a physical therapist and activist for FSHD. She introduced the team to the patient, who is receiving the AFO. After weeks of a lack of progress with the client, Ms. Eggleston worked closely with the University of Michigan to diagnose the patient's disorder, FSHD1, in December 2022. Ms. Eggleston has worked closely with other FSHD specialists to advocate and bring awareness to the disease. She has also joined several Facebook groups to fundraise and bring awareness to the condition. Ms. Eggleston has been advocating for FSHD patients for more than five years now, and she continues to be a driving force in advocating for FSHD patients.

This project started in the Fall 2024 semester; therefore, the team has met with Ms. Eggleston at various points over the course of the semester to provide updates regarding manufacturing of the AFO, as well as her providing updates with the patient. The patient's condition has progressed to the point of needing a professional AFO, and Ms. Eggleston provided the team with the information of the orthotist so that future groups can continue to work in tandem with the doctor and herself.

Product Design Specifications

The AFO will be custom designed to fit the patient's specific dimensions and comfort preferences. The device must withstand day-to-day usage, including horseback riding, which the patient enjoys. Additionally, the patient requests that the device has a discrete design, as they would not like to draw attention to the AFO. The device will be 31 cm in length, extending proximally from the distal end of the foot. The device will have a rigid aspect and an adjustable bungee mechanism to provide customizable mechanisms of support for dorsiflexion and ankle inversion. The client experiences foot drop during the heel strike phase of the gait cycle. Therefore, the AFO will need to support normal gait and allow for more than 30° of motion from

neutral ankle position. To support dorsiflexion, the device will provide 5-10 Nm of counteracting force for every 10° of plantar flexion [20]. The device must also prevent inversion angles greater than 25°[21] . In regards to the rigid support, it must resist 30 Nm of torsional forces and withstand a load of 260 N [22]. Equation 1 depicts the mathematical analysis used to identify the peak transverse load applied by the patient during ankle inversion.

$$F_I = W \cdot tan(\theta)$$

- F_I is inversion force
- W is patient weight in Newtons
- θ is the angle of ankle inversion in degrees

$$F_I = 556.03 \cdot tan(25) = 259.28\,N$$

Equation 1: Calculation of Ankle Inversion Peak Force.

The budget of this project is $100, but this value could change as it is funded by the UW-Madison's Department of Biomedical Engineering. Additional details on the Product Design Specifications can be found in Appendix A.

Preliminary Designs

Design One: Pivot Pro

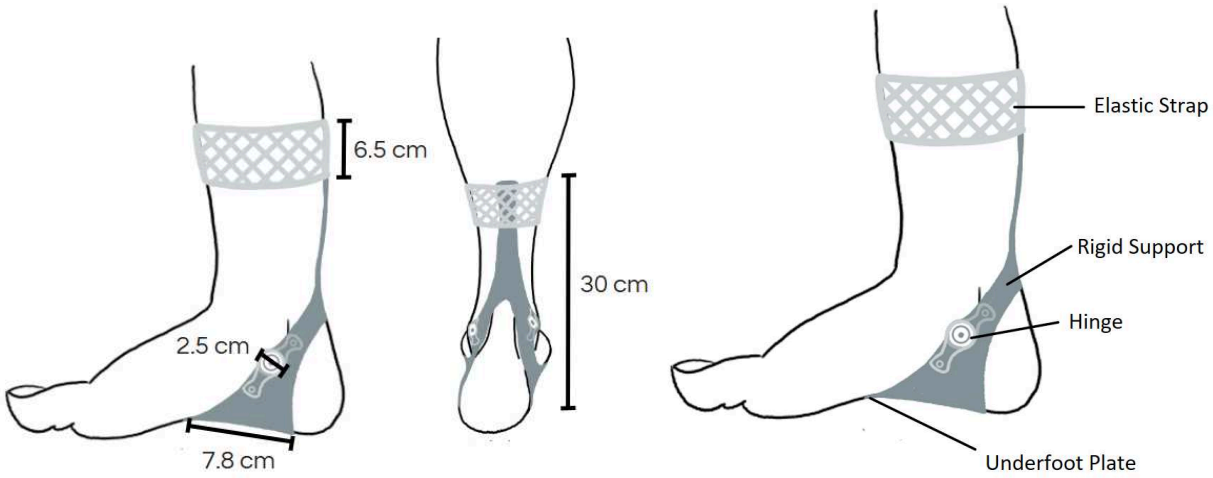


Figure 9: Pivot Pro Dimensions and Components.

The design shown above is called the Pivot Pro, named after the hinge located on the lateral and medial sides of the ankle. The design incorporates a rigid support modeled to the patient's ankle anatomy. It wraps around the back, upper portion of the heel and underneath the posterior arch of the foot. The rigid support travels up the back of the calf and connects to an

elastic strap holding the design in place. The hinge allows the underfoot plate to rotate to allow for dorsiflexion and plantarflexion. The hinge will be compact to reduce bulk in the shoe and discomfort for the patient. The entirety of this design will slide underneath the bungee cord apparatus that was constructed last semester.

Design Two: Calf Hugger

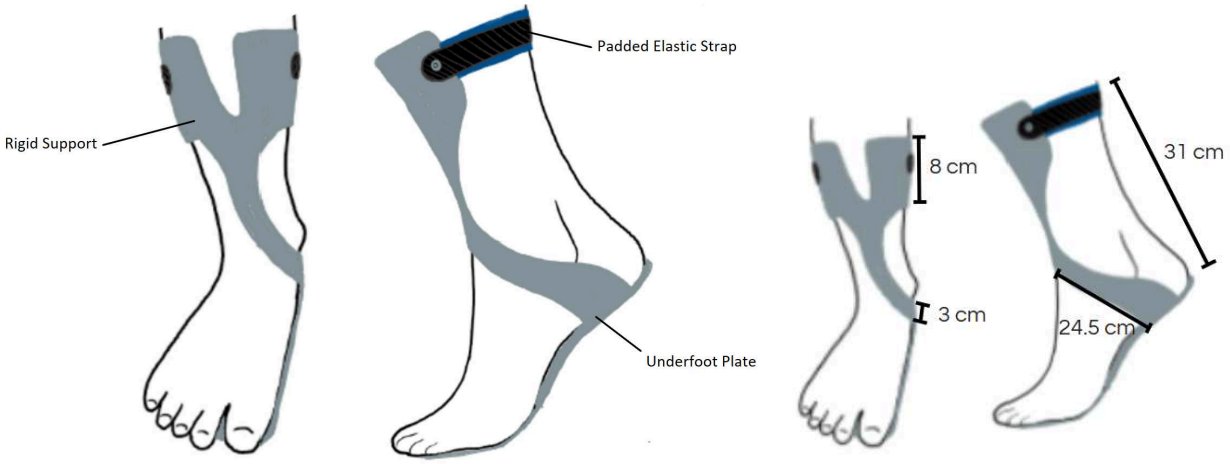


Figure 10: Calf Hugger Dimensions and Components.

The calf hugger design is modeled to wrap around the patient's calf and foot with extreme specificity. This will allow for maximum comfort and discreteness. The front piece follows the shin of the patient and wraps around the side of the foot, passing just below the ankle bone. As ankle inversion occurs on the subtalar joint below the bulk of the ankle protrusion, the design will limit rubbing on the ankle joint and prevent ankle inversion at the subtalar joint. The long rigid support is connected to a rigid plate underneath the foot that provides greater stability of the design within a shoe. Lastly, there are elastic straps connected to the top of the brace via velcro or buttons to hold the design upright. The lack of rigidity along the back of the ankle allows for comfortable dorsiflexion and plantarflexion and reduces rubbing during the gait pattern.

Design Three: We Support U

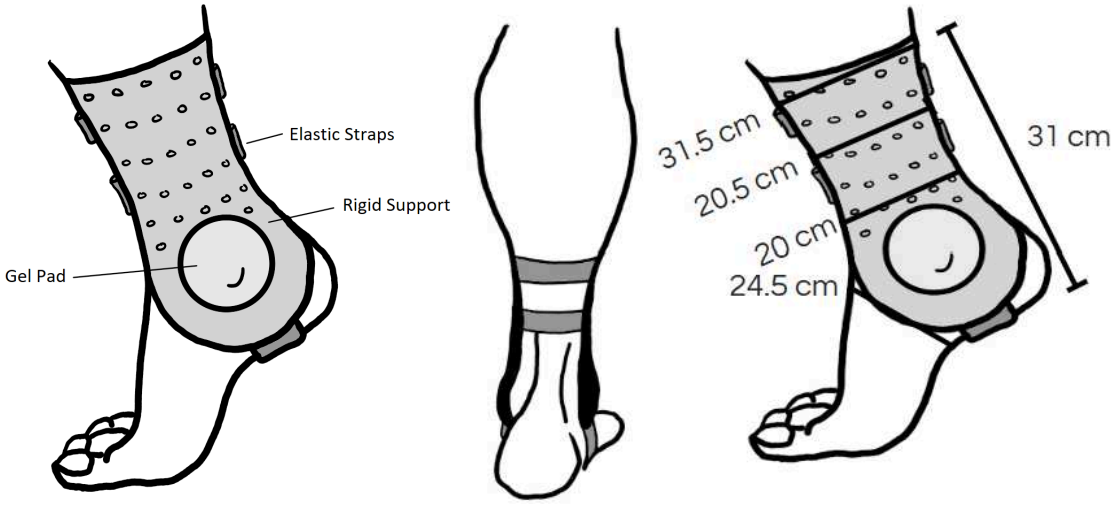


Figure 11: We Support U Dimensions and Components.

We Support U is the final design, as it resembles a “U” shape along the ankle and offers support while walking. The design consists of two rigid plates placed medially and laterally along the side of the shin to the subtalar joint. The plates may incorporate perforations to allow for breathability and airflow. There are two adjustable elastic straps along the front and back of the calf to allow for customizability and adjustability for the patient. These straps will travel through slots along the side of the brace and wrap back onto themselves. A gel pad is inserted into a cavity on the lateral sides of the brace to ensure the ankle does not rub on the rigid support. Lastly, another elastic band dives under the posterior arch of the foot to fasten the bottoms of the slabs together.

Preliminary Design Evaluation

Rigid Support Design Matrix

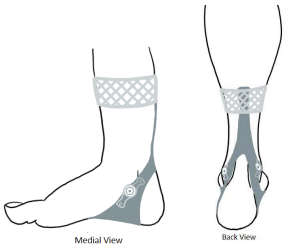
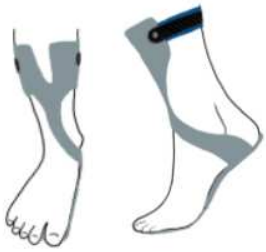
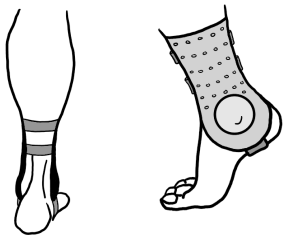
Criteria	 Design 1: Pivot Pro		 Design 2: Calf Hug		 Design 3: We Support U	
	Raw Score	Weighted Score	Raw Score	Weighted Score	Raw Score	Weighted Score
Dorsiflexion range of motion (20)	3/5	12/20	2/5	8/20	5/5	20/20
Mediolateral support (20)	3/5	12/20	3/5	12/20	4/5	16/20
Ease of user assembly (15)	4/5	12/15	4/5	12/15	3/5	9/15
Comfort (15)	2/5	6/15	3/5	9/15	4/5	12/15
Discreteness (10)	3/5	6/10	4/5	8/10	2/5	4/10
Ease of Fabrication (10)	1/5	2/10	2/5	4/10	4/5	8/10
Cost (5)	3/5	3/5	4/5	4/5	3/5	3/5
Safety (5)	4/5	4/5	5/5	5/5	5/5	5/5
Total	57/100		62/100		77/100	

Table 1: Design Matrix for Preliminary Designs.

Summary of Design Matrix

To properly evaluate the three designs, a multitude of criteria were considered. The criteria were chosen based on client-expressed desires, product design specifications, and fabrication concerns. The following criteria and constraints are outlined below:

1. **Dorsiflexion range of motion:** Dorsiflexion range of motion refers to the degree of dorsiflexion and plantarflexion motion that each design allows. This category was given a weight of 20 because one criteria of the brace is to provide the patient additional dorsiflexion support. If this motion is inhibited by the rigid support, then the overall functionality of the brace decreases. A higher score represents a design that allows for more dorsiflexion range of motion. Design 3 was ranked the highest in this category because it has nearly zero inhibition of dorsiflexion motion. The first two designs, however, both include rigid components that might inhibit the range of motion of the foot by limiting ankle motion or general foot motion.
2. **Mediolateral support:** Mediolateral support refers to how effectively each design prevents inversion (and, ideally, eversion) of the ankle. This category is also weighted 20 because the second most imported criteria provided by the client was protection against ankle inversion. A higher score represents a design that allows for more mediolateral support and prevents ankle inversion best. We Support U won this category because it includes the most material along the ankle, on both sides, providing the greatest amount of support. The Calf Hugger and the Pivot Pro both provide similar support in that they only inhibit motion at the subtalar joint. While this is location ankle inversion transpires, it does not provide nearly as much stability during the gait cycle.
3. **Discreetness:** Discreteness refers to inconspicuousness of the brace. The brace should fit underneath the patient's pants and shoes while sporting an athletic-like appearance. This category is weighted a 10 because the client is a high schooler who does not want to draw attention to her ankle. She is often ridiculed for her appearance and gait pattern at school, so the goal is to reduce the noticeability of the brace. The highest score represents the most discreet design. The Calf Hugger design was ranked the highest in discreteness because it uses the least amount of material and has the most sleek and narrow design. The Pivot Pro rigid support is designed to fit within the shoe, and the strap will fit under the pants. It resembles a little less like an athletic brace due to the strap rising above the brace. Lastly, the We Support U design ranked the lowest because it is the bulkiest and uses the most material to provide support.
4. **Ease of user assembly:** Because the client intends to use this device everyday, it must be easy and simple to don and take off. Ease of user assembly is weighted a 15 because it is a downfall of most braces on the market currently. The highest score represents the most user-friendly brace. Pivot Pro and Cald hug won this category because both designs require the patient to simply slide her foot into the brace. The We Support U brace,

however, requires the user to strap the design together upon wearing it to allow for customizability and stability.

5. **Comfort:** The comfort category considers how comfortable the brace is for the client to wear. Comfort is weighted 15 because the client intends to wear this brace during all of her daily activities. If the brace causes discomfort, the client will not wear it. A higher score represents a design that is more comfortable for the user. We Support U won this category because it has padding around the ankle and no material on the front or back of the heel, which is usually a pain point for brace wearers. Pivot Pro placed last in this category because of the hinge directly below the ankle joint and the material pressing near the achilles tendon. Lastly, the Calf Hugger placed second because it has less material on the ankle joint, but does have a piece running down the front of the shin that may cause discomfort during dorsiflexion.
6. **Ease of fabrication:** Ease of fabrication considers how easy each design is to fabricate, including the accessibility of materials needed and the time required for fabrication. Ease of fabrication is weighted a 10 because, although the design needs to be practical to produce, there is considerable flexibility in the time and fabrication processes that can be used. The We Support U design will be the easiest to fabricate because it requires the least amount of specific patient anatomy and specificity. The Calf Hugger's success relies heavily on the team's ability to accurately model the patient's foot and ankle anatomy. This adds additional fabrication challenges because the patient is currently not available in person. The Pivot Pro design is more simple in its molding, but requires a hinge to be effective. Fabricating a small, flat, thin hinge that allows for dorsiflexion and plantarflexion motion will be extremely challenging.
7. **Cost:** The cost category considers the fiscal burden of each design. Cost was ranked a 5 because each design should have a similar total cost, as they all use the same materials. Low scores indicate higher costs. The Calf Hugger will be the cheapest design, as it has the least amount of material to produce. The Pivot Pro and We Support U designs are tied in this category because the Pivot Pro requires purchasing or fabricating a hinge joint and the We Support U brace requires gel padding, elastic straps, and more material than the other two designs.
8. **Safety:** Safety refers to how safe the rigid support is when worn. This category is ranked a 5 because none of the designs have considerable safety risks. The Pivot Pro ranked slightly lower, however, because the hinge might add a pain point or pinch point when in motion. The Calf Hugger and We Support U designs do not include any components that will lead to injury.

Proposed Final Design

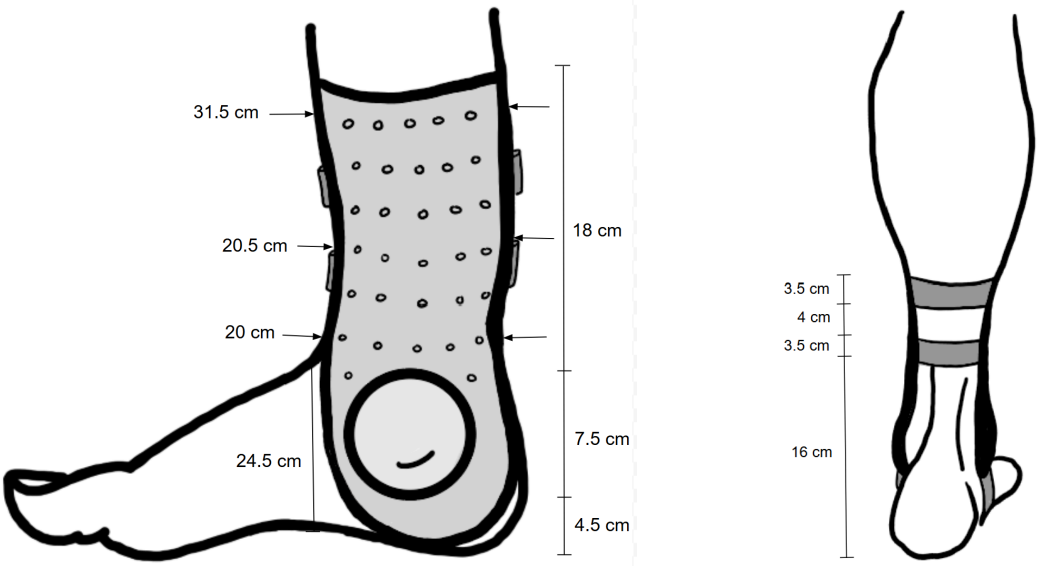


Figure 12: Proposed Final Design Dimensions.

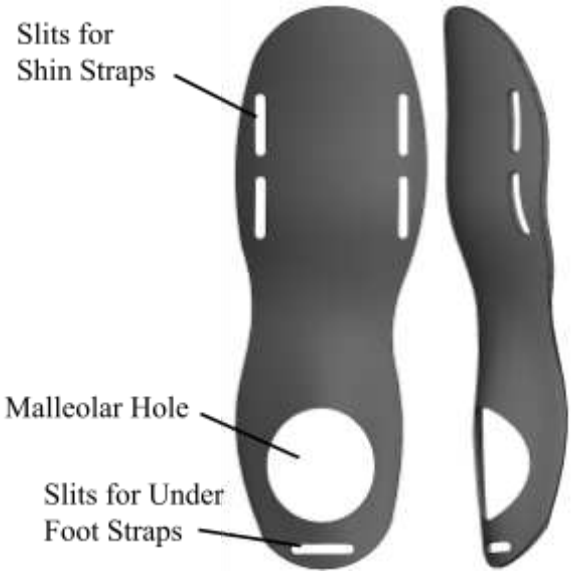


Figure 13: We Support U Preliminary SolidWorks Model, Anterior and Lateral Views.

After evaluation of the three rigid support designs, the We Support U design was chosen. This design balances appearance with functionality, as the client will likely grow in her condition, needing more support over time than the other two braces can offer. The CF-PLA supports will prevent ankle inversion and eversion at the subtalar joint. The gel pad will reduce discomfort and rubbing on the ankle joint, and the elastic straps will allow for adjustability and customizability for the patient. Strap attachment methods are being considered, but will likely be

composed of slots travelling along the exterior edge of the brace and one under the gel pad for the underfoot strap, as shown in Figures 12 and 13.

Fabrication and Development Process

Materials

The final design will incorporate six materials across three components of the device: the foot sleeve, bungee cord mechanism, and inversion support. Specifically for the inversion support, the team considered different materials in attempts to select the most appropriate material. The design matrix and criteria for the inversion support gives insight on our decision making process below.

Materials Design Matrix

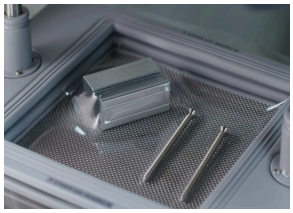
Criteria	 Carbon Fiber reinforced PLA composite (CF-PLA)		 Fiberglass Plaster		 Thermoplastics	
	Raw Score	Weighted Score	Raw Score	Weighted Score	Raw Score	Weighted Score
Strength/rigidity (30)	5/5	30/30	4/5	24/30	4/5	24/30
Ease of Fabrication (20)	4/5	16/20	5/5	20/20	1/5	4/20
Cost (20)	5/5	20/20	3/5	12/20	4/5	16/20
Safety (20)	5/5	20/20	3/5	12/20	5/5	20/20
Environmental Impacts (10)	5/5	10/10	4/5	8/10	2/5	4/10
Total	96/100		76/100		68/100	

Table 2: Design Matrix for Inversion Support Material

Summary of Material Design Matrix

To evaluate the three materials effectively, criteria was selected to assess the mechanical properties, fabrication process, cost, safety, and environmental impacts of each considered material. The following criteria and scoring decisions are outlined below:

1. **Strength and Rigidity:** This criteria is the highest priority as it is the main determining factor of the support's functionality. Rigidity is assessed based on flexural strength because it will be subjected to bending forces that the material must effectively support the ankle and resist inversion during daily activities. CF-PLA ranks the highest, with a flexural strength of 470 MPa according to the Makerspace material reference sheet. Fiberglass and thermoplastics, while strong, have lower flexural strengths of 50 Mpa [23] and 10-50 Mpa [24].
2. **Ease of Fabrication:** This criteria evaluates the complexity and time required for fabrication, weighted at 20% because it is important to ensure practical material selection and allow for appropriate time for testing and revisions. Fiberglass plaster ranks highest due to its water-based application process, which eliminates the need for precise foot dimensions or modeling. CF-PLA has the next highest rating, as 3D printing is relatively simple but requires a 3D scan for customization. Thermoplastics rank lowest due to their complex fabrication process, which involves a heat gun and vacuum sealing.
3. **Cost:** Cost is weighted as 20% due to the $100 budget and funding through BME design. CF-PLA received the highest rating because the Makerspace offers minimal 3D printing cost compared to fiberglass plaster, which requires bulk purchasing, and thermoplastics, which are inexpensive but still more expensive than 3D printing.
4. **Safety:** The primary safety concern is skin irritation as the material will be in direct contact with the skin for an extended period of time. CF-PLA and thermoplastics scored highest due to their smooth surfaces, while fiberglass plaster ranked lower due to potential skin irritation from fiberglass dust or fragments.
5. **Environmental Impacts:** Environmental impact considers the material's effect on the Earth, particularly its recyclability. While this is an important factor, our design is customized for an individual patient and is unlikely to be mass-produced, so this criterion is weighted at 10%. CF-PLA and fiberglass plaster have similar environmental impacts, but CF-PLA scores highest due to its high recycling rate and improved strength after remanufacturing [25]. Thermoplastics rank the lowest because non-degradable plastics can release methane and harm wildlife [26].

Inversion Support: Carbon Fiber Reinforced PLA Composite (CF-PLA)

As decided from the materials design matrix, the rigid support pieces around the ankle will be made from CF-PLA, chosen for its lightweight, high flexural strength, and sleek, low-profile design.

CF-PLA's lightweight nature will allow for ease of use, enabling better movement while reducing fatigue and pain for the user. Its sturdiness ensures resistance to everyday wear and tear, providing long-term support. With a flexural strength of 470 MPa, CF-PLA maintains its integrity under high bending loads. The ankle experiences an average force of 266 N generated mediolaterally for an individual with typical gait patterns. CF-PLA well exceeds the strength required to prevent inversion. The extra strength helps withstand higher force caused by increased inversion due to FSHD symptoms during dynamic movements and potential falls, ensuring effectiveness in real-world conditions. These combined properties optimize ankle stabilization for overall gait improvement [27].

Although this device is custom-made to fit the patient's dimensions and not intended for mass production, CF-PLA has a high recycling rate, and its mechanical properties improve after remanufacturing. Recycling CF-PLA involves reversing the 3D printing process by using a hot air gun to melt the composite and recover the carbon fiber to be used in the next printing process. Through this recycling approach, 100% of the carbon fiber and 73% of the PLA matrix are recovered and reused, requiring only 67.7 MJ/kg - significantly less energy than original CF/PLA production [25].

Additionally, CF-PLA is low in cost at $0.05 per gram of material [28]. Granted access to University of Wisconsin-Madison's Design Innovation Lab allows for fabrication processes including 3D scanning, 3D printing, and additional CF-PLA manual refinement with minimal costs.

Foot Sleeve: **Nylon, Polyester, and Latex**

The foot sleeve of the brace will be composed of a blend of nylon, polyester, and latex. These materials were chosen for their specific properties that enhance both functionality and comfort. The sleeve's breathability will absorb sweat and keep the foot dry, providing comfort during extended use. The material will also be tight and strong, ensuring that the sleeve stays securely in place without sliding. Additionally, the fabric is smooth and soft, adding comfort, while its graduated compression promotes circulation, providing support and pain relief to the user [29].

Nylon is specifically selected for its low elongation, strength, high-temperature resistance, and ability to make the brace visually appealing and lightweight [30]. Polyester, known for its durability and strength, is ideal as it retains its shape and resists wrinkles, shrinking, and environmental elements like water and wind, which is crucial since the device will frequently be exposed to outdoor conditions [31]. Latex contributes flexibility, durability, and excellent resistance to liquids, making it an effective barrier against moisture while maintaining overall strength [32]. Since this device will be worn on the foot during activities that involve sweating, these properties are essential to ensuring both the functionality of the design and the comfort of the user.

Bungee Cord Mechanism: Lock Lace, Bungee Cord, and Casing

A thin black bungee cord that is ⅛ inch in diameter and has 100 lb max tensile strength will be used. This specific cord was chosen because it is less bulky and requires less cord displacement, but still offers the patient the support needed for dorsiflexion. The bungee cord will apply adequate tension, strength, recoil, and flexibility needed for gait support.

The bungee cord is securely sewn at the base of the foot and anchored by a 3D-printed black PLA casing, which houses a spring-loaded cord lock from Lock Lace, positioned on the anterior side of the shin. This mechanism ensures consistent tension while the brace is worn.

Last Semester's Prototype

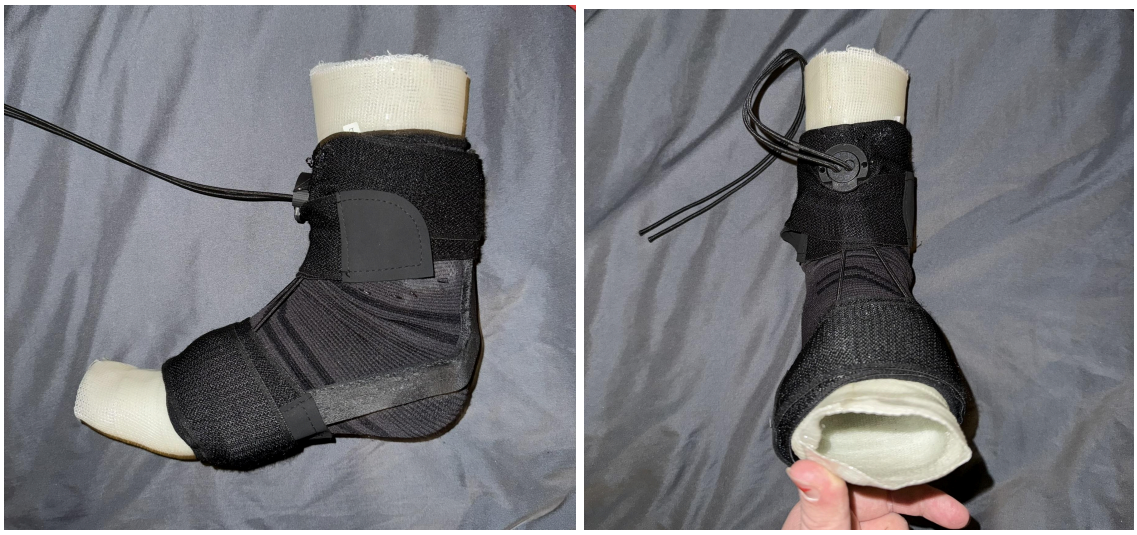


Figure 14: Bungee Brace Assembly, Lateral and Anterior Views.

Last semester's final design consists of three main components: the inner and outer compression socks, the bungee cord mechanism, and the rigid inversion support. Each element was designed to address the patient's difficulties with maintaining dorsiflexion and preventing inversion during daily activities.

The inner compression sock features strategically placed gel pads at key pressure points, including the medial malleolus and the posterior leg. These gel pads serve to cushion areas susceptible to pressure from the rigid inversion support, enhancing comfort and reducing irritation. The gel pads are intended to accommodate individual needs by adjusting the gel placement based on personal preference. The outer compression sock secures the rigid support and anchors the bungee cord mechanism while maintaining a sleek and black "athletic brace" look. Two adjustable Velcro straps provide a customizable fit, ensuring that the brace remains secured throughout movement.

The bungee cord mechanism incorporates a Lock Lace system housed within a custom 3D-printed holder, positioned on the anterior leg. The bungee cord can be adjusted to match

specific support needs. The more tension in the bungee cord will provide an increased dorsiflexion support.

The rigid inversion support extends from the posterior side of the leg and runs along the medial side of the right foot, offering stabilization against excessive inversion. The inversion component is 3D-printed from a CF-PLA composite. The design was tailored using x, y, and z coordinate measurements from a team member.

Overall, the Bungee Brace is ideally built to effectively prevent falls, minimize foot inversion, and support dorsiflexion, providing a practical and user-friendly solution for everyday wear. For additional information on previous design fabrication methods and expenses, refer to Appendices B and C.

Strengths and Limitations of the Current Design

The current prototype incorporates good fundamental components, which were evaluated through gait analysis on a healthy individual (see the testing section for more details). Among the different components, the bungee cord mechanism proved to be the most promising as it maintained an increased dorsiflexion angle throughout gait. Additionally, the brace featured a sleek, black aesthetic, designed to resemble an athletic brace to remain discrete.

Despite these strengths, the prototype has areas that require further updates and refinements. One of the primary concerns is the rigid inversion support, which the team aims to improve this semester. Gait analysis results indicated minimal to no mediolateral change, suggesting that the current inversion support may not be effective. This could be attributed to the inaccuracies in the customized dimensions caused by the limitations in SolidWorks to build specified dimensions in three different planes. The current prototype has visible gaps along the leg where the rigid support should be flesh with the skin, causing bulkness and functional concerns. Another key challenge is the brace's usability. With multiple components and two tight compression sleeves, the current design can be difficult and time consuming to put on and remove. Moving toward, the team aims to create a more cohesive design with improved usability that incorporates the bungee cord mechanism and updated inversion support.

Methods

A large focus of this semester's design was patient dimensional specificity. A plaster cast, shown in Figure 15, was provided last semester from the client. An attempt to create a mold last semester left the cast with a smooth, epoxy-coated interior, limiting the dimensional accuracy of the cast. This likely caused errors in the dimensioning aspect of fabrication. A dremel was used to cut the cast in two parts, exposing the medial and lateral dimensions of the patient's right leg. This process is shown in Figure 16. The cast was then 3D scanned and modeled using the Creality RaptorX gun in the Makerspace, shown in Figure 17. This device was selected because it was readily available, did not require high definition parts, and can depict crevices in the item. This wireless blue laser scanner does not require the use of sprays, can model objects ranging from five millimeters to 157 inches long, and can fully replicate the model in color [33]. Using

the device's software, both sides of the cast were scanned into a mesh file, smoothed, simplified, and refined. After being exported as an "obj" file, the mesh file was imported into OnShape to build splines off of. The splines were lofted and the solid extruded into the desired shape and dimensions before being 3D printed with carbon fiber reinforced PLA.

Figure 15: Fiberglass Plaster Cast.

Figure 16: Dremel Sawing of the Plaster Cast.

Figure 17: 3D Scanning of Cast using Creality Raptor X.

Two pieces of mesh foam padding were layered on top of one another and carefully sewn around the edges to create a double layered cushioning. They were dimensioned to the shape and configurations of the supports they were adhered to. The foam was then attached to the inside of the inversion support using a liquid adhesive, Weld-On step four.

Using 1-inch-wide strips of thick scrap fabric found at the makerspace, velcro was sewn onto the ends and threaded through the slits in the inversion supports located around the shin. Additional velcro straps were sewn onto the bottom slit on one inversion support and secured to the other support with a piece of velcro attached to the outer surface of the support. The full outline of this fabrication process can be found in Appendix E.

To engineer a more cohesive and user-friendly design, additional fabrication methods are needed to incorporate the bungee cord mechanism. The goal is to enhance ease of use, making the brace simpler to put on and remove while maintaining both functionality and aesthetics.

Final Design

Figures 18 and 19: Red inversion support with dimensions (left) and fully assembled brace including inversion support underneath compression sock with bungee cord mechanism (right).

The final inversion support design features two rigid supports designed to avoid potential pressure points around the ankles. Each support includes a hole on the lower portion to prevent any discomfort on or around the malleoli. For testing purposes, the lengths and dimensions were altered to create two different versions of the supports. The red brace features a longer lateral side support and shorter medial side support, whereas the black brace had a longer medial side and a shorter lateral side. The specific dimensions of the two versions are provided below in Table 3 with the diagram of dimensions shown in Figure 20.

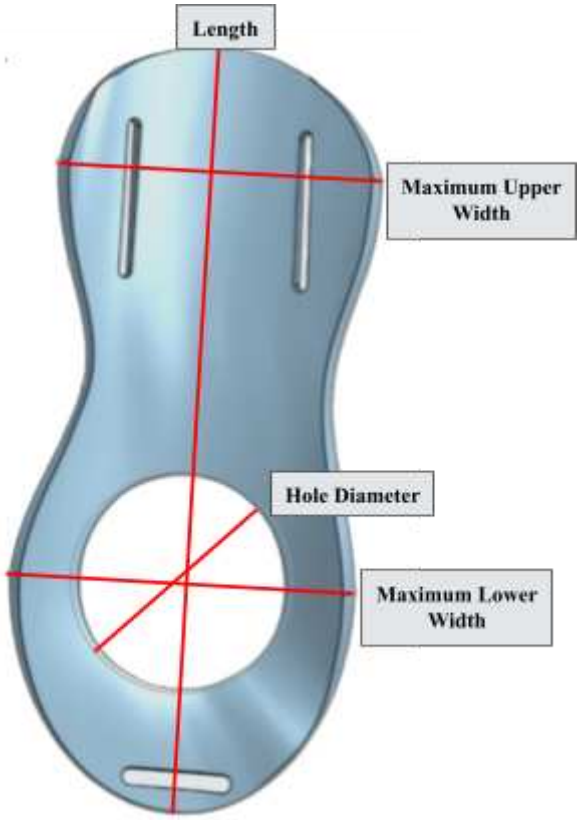


Figure 20: Dimension Locations of Final Design.

Version	Side	Height (cm)	Hole Diameter (cm)	Maximum Upper Width (cm)	Maximum Lower Width (cm)
Red Brace	Lateral	15.87	4.32	5.94	6.08
	Medial	16.74	3.84	5.62	6.00
Black Brace	Lateral	19.32	4.32	5.93	6.08
	Medial	13.32	3.84	5.62	6.00

Table 3: Dimensions of Inversion Supports.

The brace features two velcro straps located around the shin and underneath the heel, allowing for minor adjustments to create a more comfortable and secure fit around the ankle. For additional comfort, the brace includes double-layered mesh foam on the inside of both supports.

The full assembly of the brace involves first placing the inversion support and securing it on the correct location on the ankle using the velcro straps. Next, the compression sock with the bungee cord mechanism is slid over the inversion supports and the bungee cord is tightened until the foot is held in a dorsiflexed position. Finally, to secure the whole brace, additional velcro straps are wrapped around the bungee cord and compression sock at the base of the foot and around the shin. The final design did not exceed the $100 design budget. A detailed expense breakdown can be referenced in Appendix D.

Testing

Previous Semester Testing

During the previous semester, the team conducted testing to evaluate the effect of the AFO on gait in a healthy individual using Runeasi, a device equipped with an internal measurement unit (IMU) located on the lower back to measure biomechanical data [34]. Using the acceleration of the body center of gravity, Runeasi was capable of calculating measuring four main metrics:

- Dynamic instability (%): the mediolateral movement during walking.
- Ground time contact (ms): the duration of foot to ground contact.
- Impact magnitude (G): the vertical force transmitted to the pelvis at initial contact.
- Cadence: refers to the number of steps a person takes per minute during walking or running.

Testing was performed under three conditions: walking with the brace, without the brace, and with the brace minus the rigid support. Results demonstrated that the previous semester's prototype brace had no adverse impact on gait in a healthy individual and maintained high comfort levels throughout use. However, testing revealed no improvement of dynamic instability indicating the device's limitations in mediolateral support. Additionally, despite slight slippage in the bungee cord lock mechanism, the brace successfully supported dorsiflexion by raising the resting foot angle by 38° compared to the natural resting position.

Although Runeasi provided useful insights, it is no longer available for use, necessitating alternative testing procedures to evaluate device performance requirements.

Testing Limitations

Debbie Eggleston and the patient reside in Michigan, making in-person testing challenging. To identify necessary initial adjustments, the team first evaluated the device's mechanical properties and performed force plate testing on an unaffected individual before shipping the final design and testing materials to the client for at-home comfortability and functionality testing.

MTS Testing

MTS testing was performed to validate the strength of the rigid support component. When in ankle inversion, the loading pattern of the brace undergoes a three point bend. Consequently, a three point bending test was used to replicate these conditions, with the positioning shown in Figure 21. A load of 260 N must be withstood by each brace, as this is the force applied to the support during a 25 degree ankle inversion by the client, as specified in the design specifications. Three 3D printing infills were tested. By increasing the infill, you increase the print's strength, rigidity, print time, cost, and weight. 15%, 35%, and 50% infills were tested using the bending test protocol and code outlined in Appendix F.

Figure 21: MTS Testing Setup.

After testing, the three prints were evaluated for crack patterns and location of break. As shown in Figure 22, all three rigid supports cracked in the same location in the same orientation. This could be a result of the design of the support or the layering of the 3D printer. Because each sample cracked in the thinnest portion of the support, the designs could be remodeled to have a thicker rim around the holes for the malleoli. This, however, would require greater specificity in the patient's dimensions and could cause more discomfort around the ankle. Another solution would be including a heterogenous infill throughout the rigid piece, adding greater infills in the regions most susceptible to cracking. Additionally, the printing pattern of the 3D printer utilized could have created fault lines that were more prone to break. If the layers in the printing process are not well bonded, they will separate when placed under stress. Other factors, including printing speed, nozzle clogging, and incorrect flow settings can also affect the breaking patterns.

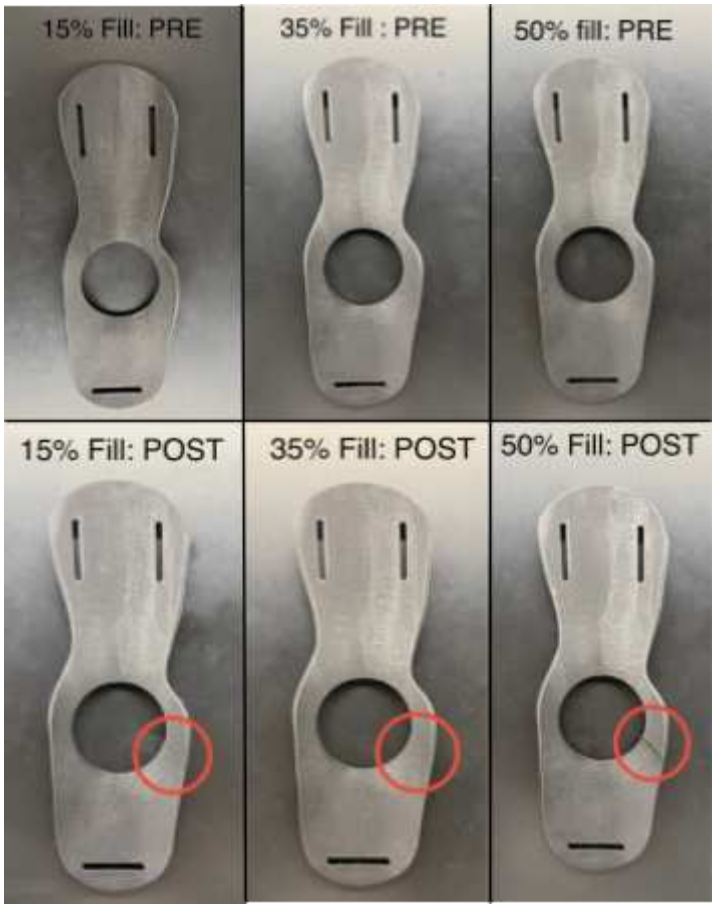


Figure 22: Pre- and Post-MTS Testing Samples.

The stress-strain curve below depicts the mechanical performance of each infill under three-point bending. As shown in Figure 23, the 35% infill failed first under the 260 N load, exhibiting a more brittle failure with minimal deformation. The 15% infill showed increased resistance, with a slight plastic region before fracture. The 50% infill sustained the highest load, with the broadest curve, indicating the best overall performance in terms of both stiffness and strength.

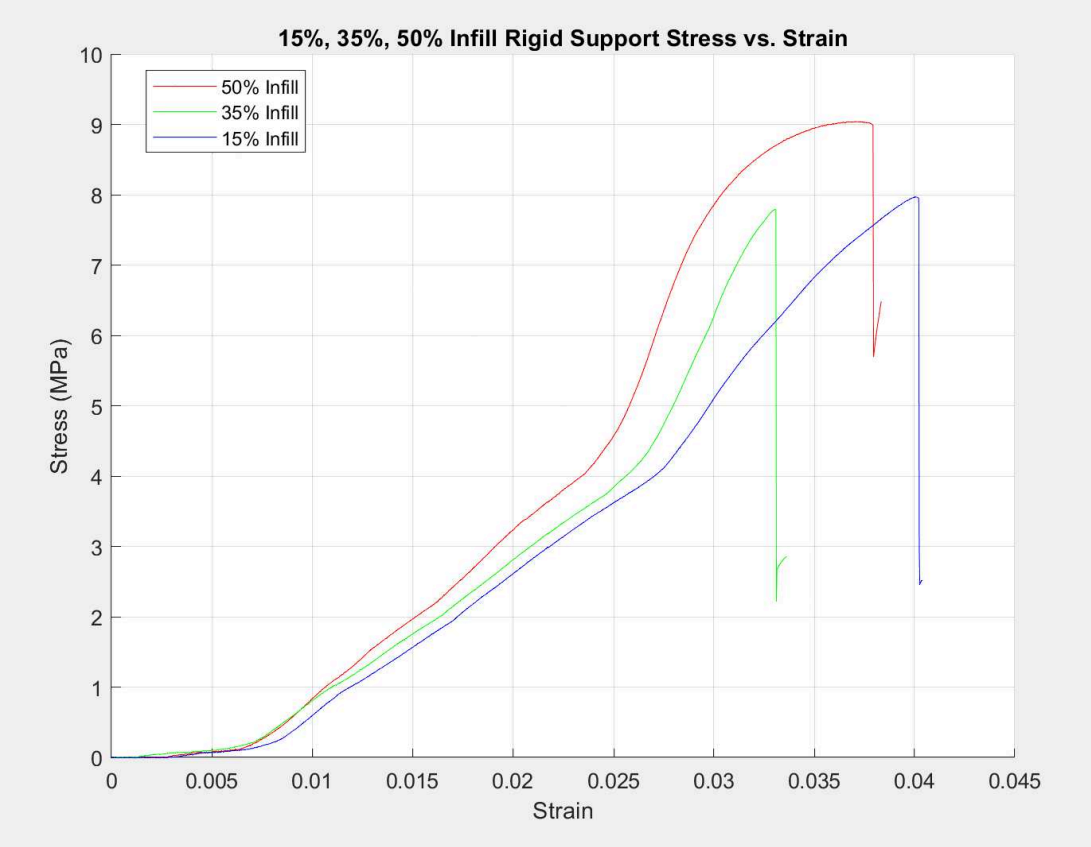


Figure 23: MTS Testing Stress Strain Curve.

These results suggest that a 50% infill provides the most structurally sound option for the rigid support, though trade-offs in weight and comfort must be considered. All three infills, however, met or exceeded the 260 N specification, meaning they are all viable options for our rigid component. In clinical applications, an optimal balance between strength and comfort should be sought, potentially through selective reinforcement of high-stress regions rather than uniformly increasing infill density. Future testing could include varying layer orientation, implementing post-processing treatments, or exploring alternative materials to further enhance performance. 15% infill was chosen for the final prototype design. Figure 24 depicts the maximum three-point bending force experienced by each sample before fracture.

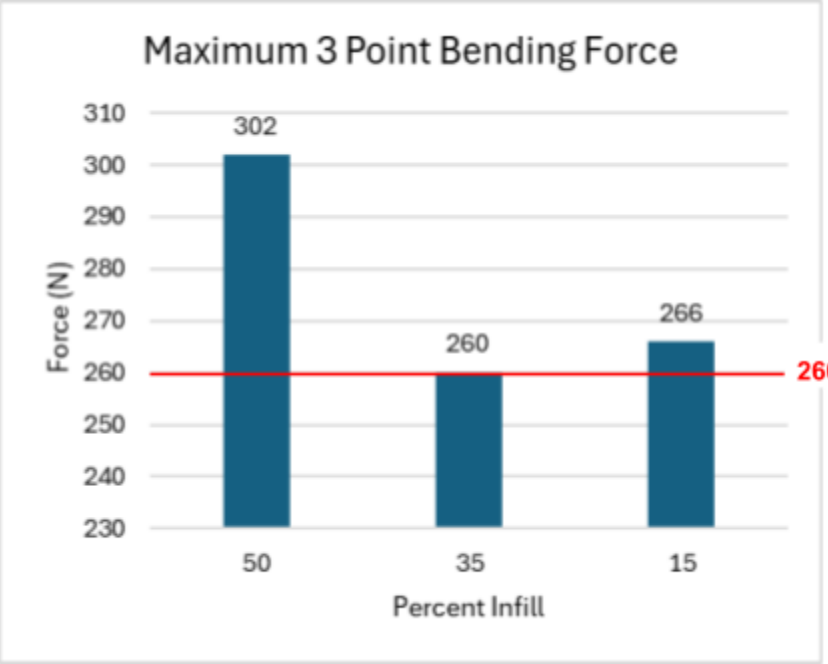


Figure 24: Maximum Three-Point Bending Force Bar Graph.

Force Plate Testing

Once the optimal printing infill was selected for the brace design, force plate testing was conducted to determine the impact of each brace on stability. Due to client unavailability, force plate testing was performed on healthy subjects. According to the protocol in Appendix G, each participant completed three trial conditions: eyes open, eyes closed, and wedge-induced stance. For each condition, trials were conducted using the black brace, red brace, and no brace, with participants maintaining a 30-second hold per trial. Trial success was assessed by tracking the center of pressure (COP), which was then visualized using stabilograms. Representative stabilograms are shown in Figure 25 and the comprehensive set of stabilograms is found in Appendix H.

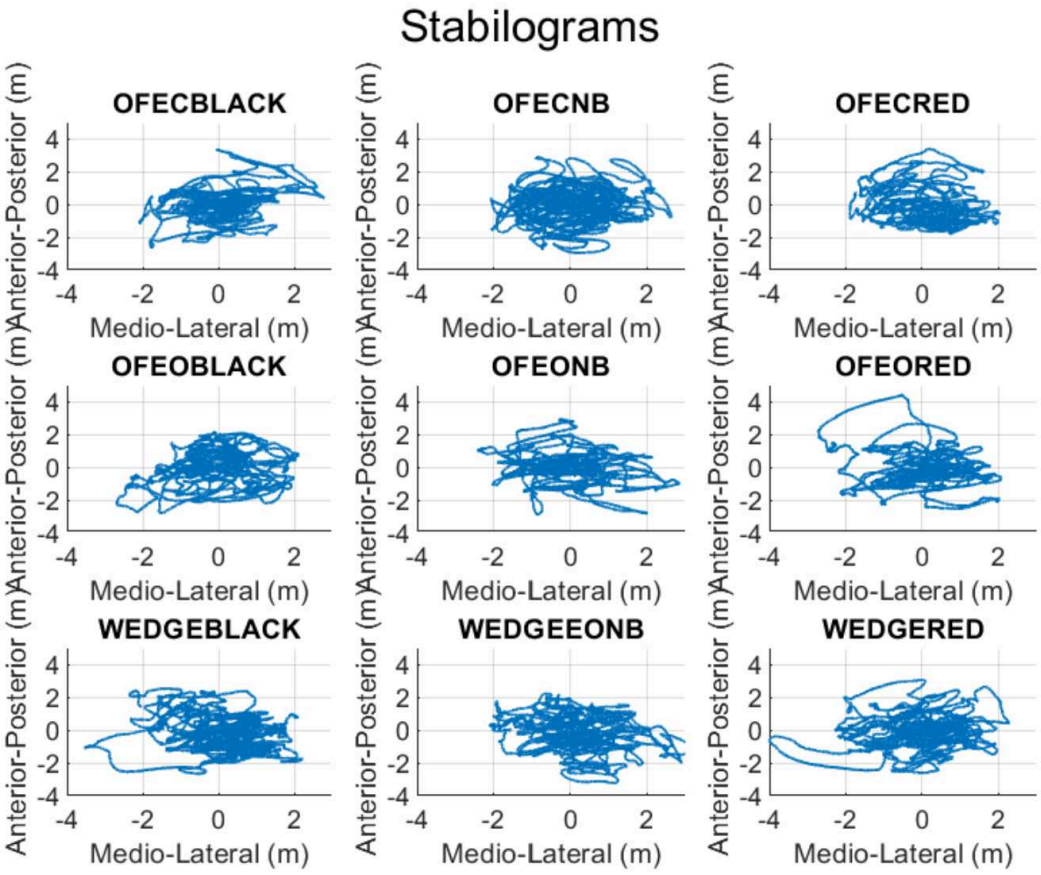


Figure 25: Stabilograms: One Foot & Eyes Closed & Black Brace (top left), One Foot & Eyes Closed & No Brace (top middle), One Foot & Eyes Closed & Red Brace (top right), One Foot & Eyes Open & Black Brace (middle left), One Foot & Eyes Open & No Brace (middle middle), One Foot & Eyes Open & Red Brace (middle right), Wedge & Black Brace (bottom left), Wedge & No Brace (bottom middle), Wedge & Red Brace (bottom right).

Additionally, the total distance the COP travelled in each trial was calculated and averaged to obtain the change in COP path length with and without an AFO. Each trial was repeated five times with five different subjects, meaning there were 45 trials total. Eliminating inconclusive data, 37 quality trials were obtained. Figure 26 depicts the average path lengths for each condition, along with their respective standard deviations. In all trials, the black brace demonstrated the greatest improvement in stability, exhibiting the shortest average COP path length. The red brace consistently outperformed the no AFO condition across all trials. Although statistical analysis using a one-sided T-Test revealed that the differences in path length were not significant, as seen in Table 4, the data still reflected the expected trend. Further force plate testing on a single subject may help reduce standard deviation caused by inter-individual variability. Expanding the force plate analysis could also strengthen the observed trend in COP path length.

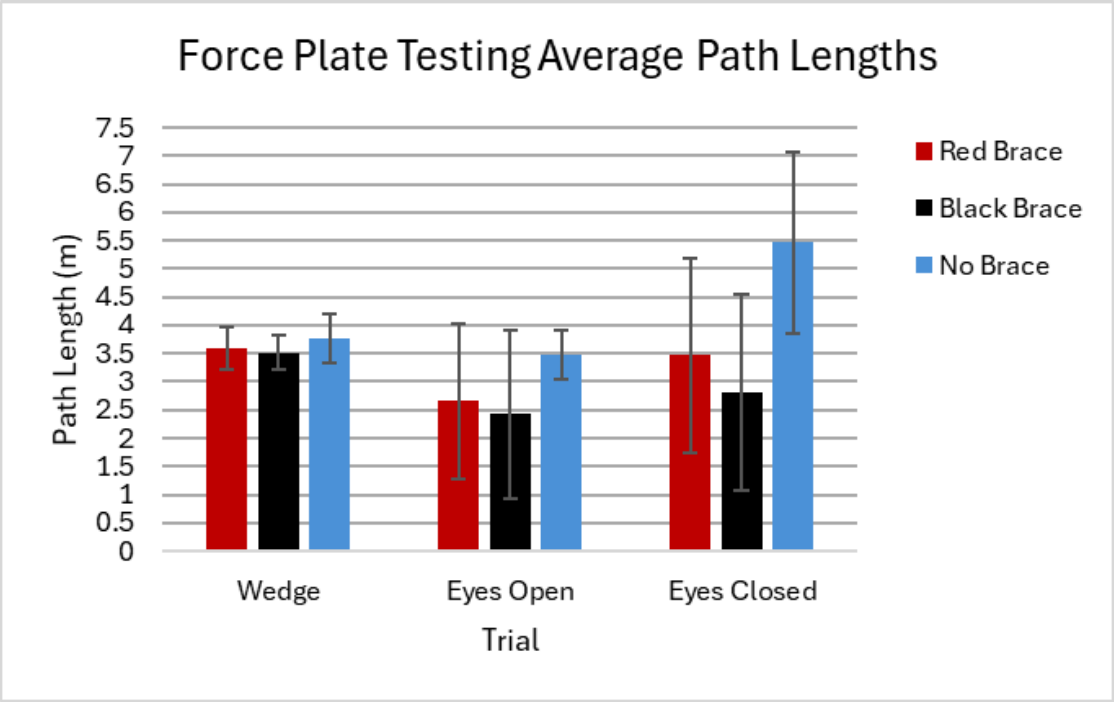


Figure 26: Force Plate Testing Average Path Length Bar Graph.

Trial	Significance Against no AFO	Samples (n)	Significant
Eyes Closed, Black Brace	0.0536	3	No
Eyes Closed, Red Brace	0.0785	4	No
Eyes Open, Black Brace	0.155	3	No
Eyes Open, Red Brace	0.1797	4	No
Wedge, Black Brace	0.1792	5	No
Wedge, Red Brace	0.2825	5	No
* Alpha value of 0.05, Confidence Level of 95%			

Table 4: Path Length T Test Significance Levels.

Comfortability Testing

To meet both design specifications and client expectations, the AFO must maintain a low profile and fit comfortably within standard footwear without the need for specialized shoes. Due to the client's inability to conduct testing in-person, prototype materials for both prototype iterations were shipped to facilitate at-home testing. To assess the comfort level of the device during ambulation, the client underwent a structured comfort assessment protocol. During this process, the client wore each brace and completed a comfort evaluation form that rated several key areas: inside support comfort, outside support comfort, strap comfort, and foam padding comfort, each on a scale from one to ten. Each section also included a notes area for the

participant to provide feedback and suggest potential design modifications. Additionally, the evaluation form featured diagrams of the medial and lateral aspects of the foot, where the participant was prompted to circle any points of discomfort or pain. Completed and original forms are available in Appendices I and J.

A consistent pain point was identified on the medial side of the foot for both braces, likely caused by friction from the rigid supports. The participant rated the red and black braces similarly in most categories, with the main difference being the fit of the medial support. Discomfort in the red brace was attributed to slippage of the rigid supports, which occurred when the compression sock was applied, leading to misalignment and subsequent pain.

The most challenging aspect of the design for the user was donning the brace. The compression sleeve and strap system were found to be difficult to adjust and confusing to use, particularly when trying to fit the brace into a shoe.

This subjective assessment has highlighted critical areas for improvement to ensure the final device meets both functional and comfort-related requirements. Based on the findings, the design will be modified to include a lace-up closure system. This adjustment will simplify the donning process, provide enhanced dorsiflexion support, and ensure the rigid components remain securely positioned, minimizing discomfort and maximizing user satisfaction.

Motion Capture Testing

To effectively assess the AFO design against physical performance requirements outlined in the PDS, the team conducted testing using an optical motion capture (MoCap) technology through OpenCap. This testing was utilized to quantify specific gait parameters while wearing the orthotic, including: ankle inversion, hip flexion, and knee angle motion. OpenCap is a high-precision optical, markerless motion capture software that utilizes video input from multiple smartphone cameras to estimate human joint positions. By applying computer vision algorithms and musculoskeletal modeling, OpenCap maps 2D joint positions to generate 3D reconstruction of body motion [35]. The platform enables the extraction of joint angles, kinematics, and other biomechanical data without the need for traditional markers or specialized laboratory equipment. OpenCap was selected for its simple setup, allowing the client to perform motion capture independently at home with provided materials and detailed instructions.

Motion Capture Testing was first conducted by the team on a healthy individual to obtain baseline data and edit testing procedures prior to client-use. To begin, two iOS devices were connected to the OpenCap web application and calibrated with a 4x5 checkerboard calibration target, which was positioned on a blank wall within the designated data collection space. Cameras were positioned approximately 15 feet away from the wall and 10 feet apart from one another. Following successful calibration, subjects entered basic anthropometric data including height and weight, into the web application. For each trial, the subject performed walking movements under three orthotic conditions. After initiating each video recording via the online interface, the subject waited five seconds before walking five consecutive steps either toward the cameras or along the wall. Each movement was recorded separately and trials were kept as brief

as possible to minimize video processing time. OpenCap's cloud-based system automatically processed each video, generating 3D joint reconstructions through triangulation and musculoskeletal modeling [35].

A total of eight testing conditions were outlined, including walking with no AFO, with the client's personal AFO, and with each of the two prototype designs (black and red AFOs). Each condition was performed in two directions, toward and along the wall, and repeated three times, resulting in 24 total trials. However, due to mechanical failure of the black AFO prior to motion capture testing, the client only completed testing with the red prototype, reducing the total number of trials to 18. Detailed testing procedures can be found in Appendix K.

Upon successful processing, data was downloaded for subsequent biomechanical analysis. Raw data was sorted to remove inconclusive trials, as represented in Appendix L. Data from both team testing and client testing were then analyzed using MATLAB to generate plots displaying the fluctuations of knee flexion angle, hip flexion angle, and subtalar inversion/eversion angle over a portion of the participant's gait cycle. Comparative figures of the team and client testing are presented in Figures 27, 28, and 29. A sample of the MATLAB code used to produce the following plots can be accessed in Appendix M.

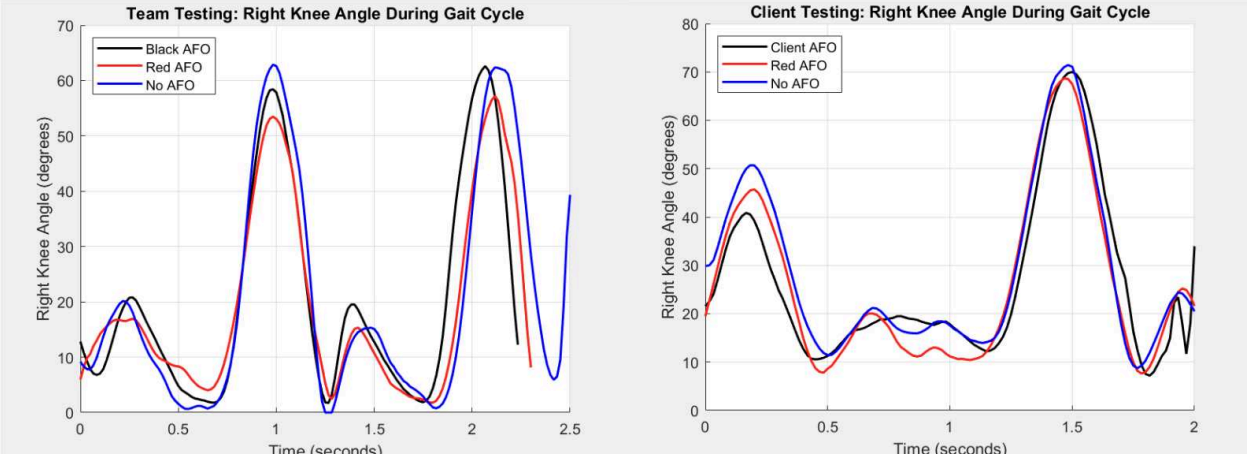


Figure 27: Comparison of right knee flexion angles during the gait cycle for team testing (left) and client testing (right). Angular fluctuations are plotted over time to assess gait dynamics under each condition.

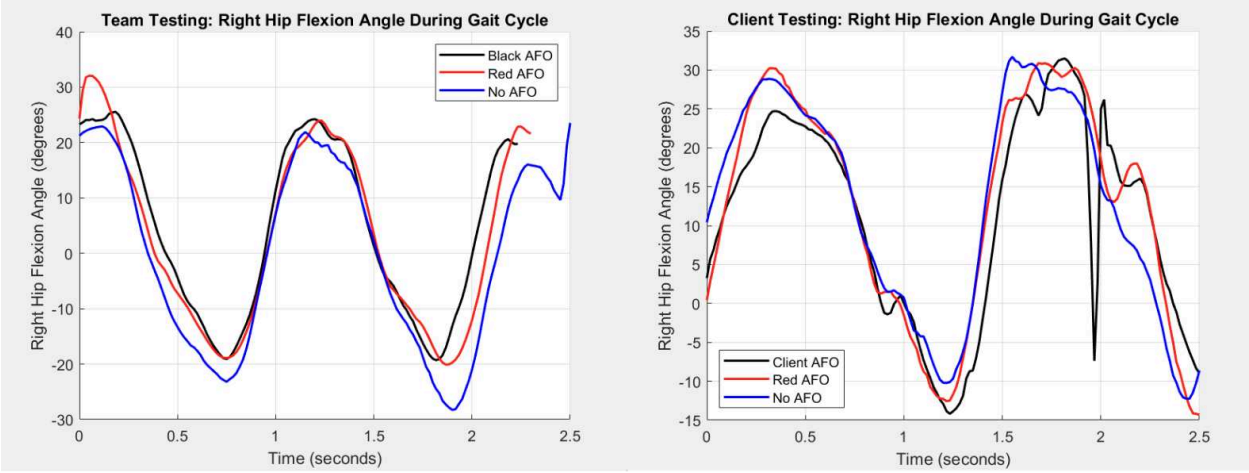


Figure 28: Comparison of right hip flexion angles during the gait cycle for team testing (left) and client testing (right). Angular fluctuations are plotted over time to assess gait dynamics under each condition.

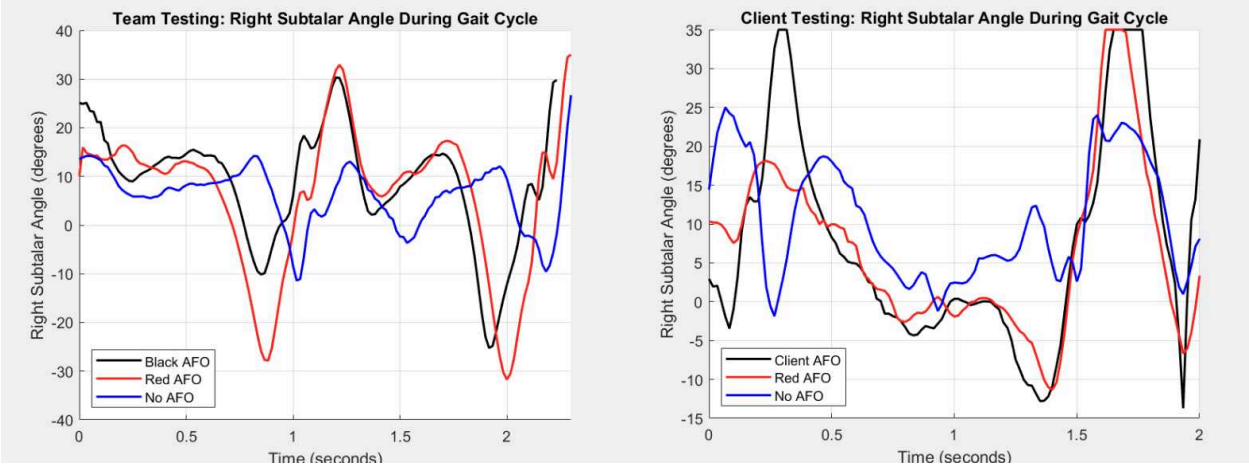


Figure 29: Comparison of right subtalar angles during the gait cycle for team testing (left) and client testing (right). Angular fluctuations are plotted over time to assess gait dynamics under each condition.

As shown in Figure 27, the right knee flexion angle does not substantially differ between testing conditions in either the team or client testing. The similarity between plots, with the no AFO condition represented in blue, the red rigid supports in red, and either the black rigid supports (team testing) or the client's personal AFO (client testing) in black, supports the conclusion that the designed rigid supports and combined AFO structure do not adversely affect unintended knee joint motion during gait.

A similar relationship can be obtained from Figure 28, where hip flexion angles do not significantly differ between the testing conditions for either team or client testing. A noticeable dip in knee flexion angle is present in the client testing plot for the client's existing AFO condition (black line) around the two-second mark. This discrepancy is likely due to a processing

error within OpenCap or a momentary occlusion of the body segment from the camera views, as the curve realigns to the expected pattern by approximately 2.3 seconds. This anomaly is not considered representative of true motion and was excluded from interpretation.

Unlike Figures 27 and 28, Figure 29 displays significant variation in the right subtalar angle between each of the three testing conditions. This angle represents ankle inversion and eversion, where eversion–tilting of the sole outward away from the midline–is indicated by a positive subtalar angle, and inversion–tilting inward toward the midline–is indicated by a negative subtalar angle.

In both the team and client testing plots, the no AFO condition demonstrated the least fluctuation in subtalar angles. In contrast, across both testing groups, the use of the client's existing AFO, the red prototype, or the black prototype was associated with increased inversion and eversion angles. This trend was unexpected, particularly for the client's existing AFO (black line, client testing), where improved gait performance was visually observed and subtalar angles would typically be expected to remain stable or decrease relative to the no AFO condition.

The unexpected increase in subtalar angle fluctuations may be attributed to displacement errors during motion capture. Specifically, the additional material from the compression sleeve of the orthotics may have been interpreted by the OpenCap software as part of the limb, artificially altering the calculated joint positions. This artifact could explain the consistent increase in inversion and eversion angles observed across all conditions involving an AFO. Due to the inaccuracies observed in Figure 29, inversion angle data obtained from motion capture testing are considered inconclusive. Further testing using non-optical analysis methods is required to draw more meaningful conclusions regarding the relationship between the designed rigid supports and their effectiveness in preventing ankle inversion. This adjustment is discussed further in the future work portion of the report.

Discussion

Ethical and Safety Concerns

To address ethical concerns, the patient must be fully informed about the potential risks associated with using the prototype, as well as the various testing methods involved. This ensures the patient can make an informed decision regarding their voluntary participation. Informed consent must be obtained and is able to be withdrawn at any point during the project. To prioritize patient safety and well-being, pain levels will be regularly monitored throughout the testing. If pain levels exceed a low to moderate threshold, testing must immediately stop, and the prototype will be refined to further enhance comfort.

It is essential to recognize that not everyone has equal access to orthotics. Socioeconomic differences significantly influence the availability of healthcare resources, limiting many individuals' access to necessary orthotics. The treatment of rare diseases, although individually uncommon, is associated with substantial health and financial burdens globally. Custom orthotics, in particular, require a high level of expertise and resources that are not readily

available. On average, the total societal burden of muscular dystrophy ranges between $80,120 and $120,910 per patient annually, increasing significantly with disease progression. The corresponding household burden is estimated to be between $58,440 and $71,900. Given the substantial costs associated with such conditions and treatment options, it is vital to consider the socioeconomic impacts of engineering design [36].

The use of advanced materials, such as carbon fiber composites, enhances the durability and performance of orthotic devices; however, these advancements come with significant expense. While the durability of the device may improve with the use of more advanced materials, the increased cost raises concerns about affordability and poses important questions about accessibility among diverse populations. To satisfy ethical considerations, it is essential to consider reducing the cost of replication to ensure broader accessibility in the case of a reproducible design.

Considering safety factors, the device must be able to withstand the cyclic loading during walking and provide proper anatomical alignment. Alignment must be maintained to avoid excessive tension, compression, or shear forces on joints, bones, and muscles to ensure long-term musculoskeletal health. The device's use of new materials raise allergy concerns that must be considered during design analysis. There must be a protective layer between the user and the CF-PLA composite to avoid skin irritation and maximize comfort while protecting the user. In case of emergency, there must be a protocol for easy and quick removal of the designed orthotic

Given the task of manufacturing a single AFO for use, it is crucial to evaluate the environmental impact of its production, including material sourcing, energy consumption, and waste generation. While the impact of a single device may be minimal, scaling up to mass production introduces additional sustainability concerns. In analysis of such design factors, it is especially important to consider manufacturing processes. Among industry standards, additive manufacturing (AM), particularly 3D printing, has emerged as a promising alternative to traditional fabrication methods. AM enables customization to minimize material waste and reduce mass and energy consumption that contribute significantly to greenhouse gasses and global warming [37]. Consideration of these factors early in the design process ensures a more responsible approach to both individual and large-scale production.

Design Evaluation

The previous semester's brace design demonstrated the design effectively supported dorsiflexion without interfering with normal gait. However, the current rigid support did not provide adequate support and difficulty donning of the brace. Over the course of the project, the patient's condition had progressed and the team focused on creating a rigid support to prevent ankle inversion. Therefore the We Support U design was selected for its relatively simple donning and fabrication method. The brace aimed to provide additional ankle support through the rigid component while preserving free range of dorsiflexion and plantar flexion. This rigid support is to be integrated with the current adjustable bungee mechanism from the previous

semester's design. The brace will remain easily concealable, offer sufficient ankle stability, and accommodate the client' progressive condition while also meeting her aesthetic preferences.

The rigid support proved to be mechanically effective, but for future iterations of the device the team would like to test stronger materials like carbon fiber. Material selection for the brace was limited to accessible 3D printable materials as 3D scanning a cast of the client's foot was used to create the rigid support. The client was unable to travel to Madison during the fabrication process and therefore supplied the team with the cast for reference. Overall, the comfort of the entire brace was sufficient with the client particularly liking the comfortability of the foam and strap selection. However, ergonomic refinements of the rigid support must be made to minimize the localized discomfort around the medial malleolus. During at-home testing this semester, the client concluded that the bungee cord mechanism was not sufficient to provide dorsiflexion support as the client's condition has progressed. Therefore, improvements must be made to the compression sleeve in future device iterations to address this issue.

Potential Sources of Error

In the design and fabrication of the device, potential sources of error may arise from measurement inaccuracies and modeling discrepancies. Slight errors in capturing the client's foot dimensions or inconsistencies in the translation of the 3D-scanned mesh into SolidWorks could lead to an imperfect fit of the rigid support. To mitigate this, the team provided the client with different models of the prototype and consulted with the client to address necessary custom adjustments.

Many methods of testing were completed this semester, and with each form of testing came with limitations. Since the client resides in another state, initial testing last semester was conducted on a healthy subject, whose biomechanics will not perfectly match the client's. While this may introduce discrepancies in testing results, it provided a baseline for PDS evaluation. This semester, arrangements were made for the client to complete in-person motion capture and force plate testing to evaluate dorsiflexion angles, inversion angles, and the brace's influence on the client's gait cycle. However, due to unforeseen circumstances, the client was unable to travel to UW-Madison for in-person testing. Therefore, the team created a protocol and sent the client two versions of the brace to complete at-home comfortability and motion capture testing. The team opted to use OpenCap, an optical markerless motion capture software, to collect data about the client's gait cycle and joint angles. However, by using this markerless software there are potential position data discrepancies from the brace thickness which would affect collected data.

Force plate testing protocol was altered to induce instability as healthy subjects had to be used for this testing method. The data collected from this testing could be skewed from a physiological bias the subject has as subjects knew if they were wearing a brace and what brace they were wearing. For more accurate data, the team needs to conduct testing directly on the client to conclude if the brace prevents ankle inversion and induces ankle stability for the client.

While the data collected from MTS testing was conclusive and supported the max force the brace must withstand in the PDS, there is a source for potential error. A make-shift fixture

was created for the 3-point bend test inorder to apply force where the rigid support would experience it during ankle inversion. It is important to recognize potential sources of error to ensure correct conclusions can be made from the collected data.

Additionally, material properties and fabrication inconsistencies may contribute to variation in the rigid support's performance. Differences incurred by 3D printing precision, carbon-fiber reinforcement distribution within the PLA composite, and bungee cord tensioning could lead to slight variations between prototypes. All of these factors have the potential to affect overall rigidity, support, and ease of use.

By addressing these potential sources of error through iterative prototyping, controlled testing conditions, and direct client feedback, the team aims to strategically improve the design's accuracy, functionality, and overall effectiveness.

Conclusions

The development of this custom AFO aimed to provide necessary functional support for a high school student with FSHD, while maintaining a discreet and aesthetically appealing design. Current designs fall short of the clients needs and fail to avoid drawing unwanted attention from peers. Given the client's concern regarding an inconspicuous AFO, the final design integrates a CF-PLA composite rigid support along with a bungee lock lace mechanism, previously validated from last semester's design. This combination aims to provide sufficient dorsiflexion assistance and ankle inversion stability while still resembling a standard athletic brace, thus minimizing visibility and social discomfort during use. By achieving a balance between functionality and subtlety, this AFO seeks to enhance mobility, confidence, and overall quality of life for the patient.

Future Work

Looking forward, the team desires to integrate the bungee mechanism compression sleeve and the rigid support into one cohesive device. Additionally, since the client's condition has progressed, improvements should be made to the bungee cord mechanism. The team learned that the client's ankle needs to be at an even greater dorsiflexed position at a resting ankle position to prevent foot drop during the swing phase of the gait cycle. The team would like to explore a lace-up mechanism with the bungee embedded into the brace. By creating one cohesive device and integrating a lace-up mechanism, this should allow for easier donning of the brace and motivate the client to wear the brace more often. The combined brace would also ideally be even sleeker than the current design to enhance the device's inconspicuousness.

Additionally, the team would like the client to travel from Michigan to Madison to complete testing and brace adjustments. The team desires to conduct ergonomic refinements of the rigid support by directly 3D scanning the patient's foot to improve the rigid support dimensions and mitigate localized discomfort. The rigid support fabrication process is relatively straightforward, allowing for easy remanufacturing to accommodate the patient's support needs and growth.

While the client is at UW-Madison, the team would like to conduct non-optical marker motion capture testing. The testing would ideally take place on a treadmill to enforce a constant cadence to easily analyze collected data. The team would consider using electromagnetic field motion capture, inertia measurement units (IMUs), or strain-gauges for measurement of joint angles and gait analysis. Lastly, the team would like to conduct force plate testing with the client, focusing on how the brace affects the client's ankle stability and inversion. By completing in-person testing, the team aims to collect statistically significant data and prove the device's effectiveness.

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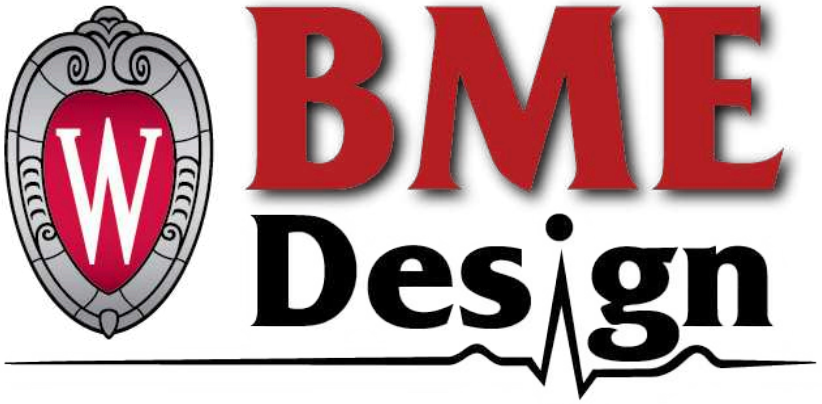
Appendices

Appendix A: Product Design Specification (PDS)



Inconspicuous Ankle Foot Orthosis (AFO) for teen

PRODUCT DESIGN SPECIFICATIONS (PDS)

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Rise and Stride

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February 6th, 2025

Function/Problem Statement:

Facioscapulohumeral Dystrophy (FSHD) is one of the most common types of muscular dystrophy, affecting approximately 4-10 in 100,000 individuals. This genetic disorder leads to progressive muscle weakness, and while there is currently no cure, physical therapy, orthotics, and surgery can help manage symptoms [1]. However, there is limited clinical research focusing on children with FSHD, leaving gaps in understanding and addressing their specific needs. This project aims to raise awareness of FSHD and explore the benefits of discrete ankle-foot orthoses (AFOs) for individuals with progressive muscle weakness. The team aims to design a brace for the patient to aid in natural gait for safer walking while being easily concealable and flexible enough to allow for a functional ankle range of motion. The key objectives of this device include positioning the ankle in adequate dorsiflexion, restricting medio-lateral ankle motion, maintaining a narrow, thin, and discrete design, and ensuring sufficient flexibility to minimize any restriction of movement.

Client requirements:

The client requests that the AFO supports dorsiflexion and prevents foot inversion while remaining flexible enough for the patient to carry out their daily activities. Additionally, the client prefers the AFO to be discreet, fitting inside a shoe and minimizing visibility. Functionality, however, is starting to outweigh discreteness as the patient's disease progresses.

Design requirements:

1. Physical and Operational Characteristics
    a. Performance requirements
        i. The AFO must be designed to be discrete and involve minimal material while providing strong support for ankle dorsiflexion and resist ankle inversion to prevent irregular gait [1].
        ii. The device should mimic normal gait without providing any resistive moment during dorsiflexion [2]. Additionally, it should allow a range of motion of more than 30° from the initial ankle angle to facilitate foot clearance [3].
        iii. The device should generate a resistive moment in the plantarflexion direction and should be adjustable in the range of 5-10 Nm per 10° of plantarflexion [3].

iv. Moment-angle characteristics should be maintained within a torque range of ±30 Nm. Additionally, the design must resist torsional forces that could lead to misalignment of the ankle or foot during typical activities [4].

v. The design must withstand the maximum bodyweight forces exerted by a teen. The average weight of a 16-year-old teenager in the United States is 136.2 lbs [5] and during walking, forces exerted on the AFO are estimated to be three times the body weight [6].

vi. Design must possess sufficient flexibility to allow for active concentric ankle movement, enabling the user to perform daily activities effectively, such as squatting and ascending/descending stairs.

vii. The AFO dimensions must be tailored to the client's leg geometry and customizable to ensure a secure fit. Ideally, the design should incorporate the patient's custom-made orthotic insole.

viii. The rigid inversion must support a minimum flexural strength of 260N in order to restrict unwanted mediolateral movement.

ix. AFO design must prevent inversion angles greater than 25° [8].

b. Safety

i. To prevent tripping and falling, the brace must facilitate normal gait patterns and enhance balance. Proper anatomical alignment must be maintained to avoid excessive tension, compression, or shear forces on joints, bones, and muscles to ensure long-term musculoskeletal health.

ii. In a manufactured and marketed design, the chosen material must be non-toxic and hypoallergenic to minimize the risk of skin irritation or allergic reactions. All surfaces must be smooth and free of sharp or ridged edges to prevent risk of surface wounds or abrasions.

iii. The outer surface of the AFO material must offer sufficient traction to prevent slipping when worn without shoes.

iv. Adjustable components of the design must remain secure under strong impacts while avoiding restriction of blood flow.

v. Any fastening mechanisms should prevent loosening or dislodgement during regular physical activity.

vi. The AFO must feature mechanisms for quick and easy removal in case of emergency without the need for specialized tools.

vii. The design should allow breathability to prevent overheating and moisture buildup.

c. Accuracy and Reliability

i. The AFO design must maintain structural integrity with repetitive use while consistently providing support to ensure proper anatomical alignment of the ankle and foot. Carbon fiber AFOs typically fail at the mid-shank region of the calf support under forces of 1970 N [9]. To limit the possibility of injury, the calf support should include a padding layer to protect the user in case of material failure. Additionally, the soft padding material must be easily replaceable after extended use to prevent user discomfort from padding degradation.

d. Shelf Life

i. Custom orthotics are tailored to the patient's specific measurements and support needs. They are designed for immediate and continuous use. If left unused for an extended period, the patient's measurement or support requirements could change, making the AFO ineffective. Therefore, the shelf-life should be limited, and the AFO should be regularly assessed to ensure it continues to meet the patient's evolving needs.

e. Life in Service

i. The lifespan of an AFO depends on several factors, including material composition, frequency or intensity of use, and changes to the patient's needs. Generally, it should last around 5 years [10].

ii. AFOs fabricated using semi-rigid materials like carbon fiber, fiberglass, or polyethylene may last longer than less-stiff materials [11].

iii. An orthotist should review the AFO at least once a year to ensure it continues to meet the user's needs and to check for any signs of deterioration [12].

f. Operating Environment

i. This AFO is designed for day-to-day use and must withstand transportation and frequent use. It will mainly be used around the house, during the school day, and

for horseback riding. The bulkiness of the device should be considered so that it can still be inserted into horseback riding shoes.

ii. It will be used both indoors and outdoors, exposed to varying temperatures, humidity, dirt, water, and sweat. The AFO should be cleaned with mild soap and water at least once a week to prevent bacterial build-up [13].

g. Ergonomics

i. The AFO must be capable of withstanding the maximum downward force exerted by the user's weight while distributing this force in a way that avoids excessive pressure points.

ii. The design should allow for adjustments to accommodate the growth, such as adjustable straps to ensure a secure and personalized fit for the intended product life in service.

iii. Current AFOs commonly weigh between 0.3 and 3.4 kg depending on the material and bulkiness of the device. The AFO should be as lightweight as possible while maintaining proper function to ensure normal gait patterns and reduce fatigue [14].

iv. Padding should be provided around sensitive areas, such as the Achilles tendon, ankle, and foot base, to prevent discomfort and skin irritation.

v. The orthosis should maintain a low profile and be able to fit comfortably within a standard shoe, without requiring the user to wear specialized footwear [3].

vi. Any moveable components of the design should function quietly.

vii. AFOs can increase step length and step velocity of patients which results in more fluid body movement and less energy excursion. AFOs can provide more stability during gait which improves the patient's daily life [15].

h. Size:

i. The size of the AFO must be tailored to the patient's dimensions. Measurements have been taken to closely match the leg. The orthotic should match these measurements, with minor adjustments for padding or other anti-chafing mechanisms in the design [16].

1. Length of the leg (measured bottom of foot to directly below kneecap) is 45.5cm.

2. Diameter directly below the kneecap (measured at top of the lower leg) is 31.5cm.
3. The diameter of the thickest part of the calf (measured mid-leg) is 31.5cm.
4. Diameter where the Achilles meets the calf (measured bottom of leg) is 20.5cm.
5. The diameter of the thinnest part of the ankle (measured where Achilles is felt) is 20cm.
6. Diameter across the middle of the ankle, through the joint is 30cm.
7. Diameter just in front of the ankle joint (measured low ankle) is 24.5cm
8. Arch Measurements: bony prominence to floor is 4.5cm and 6.25cm in length.
9. Length of the foot is 24-24.5cm.
10. Width of the foot (measured where the metatarsals meet the phalanges) is 8.25 cm.
11. Width of the foot (measured in midsole area) is 8cm.
12. Width of the foot (measured at the heel) is 5.5cm.

ii. Typically, an AFO's thickness will be 3.175 mm to adequately support the foot [17]. The device should deform only minimally during use while maintaining enough flexibility to avoid excessive stiffness that could cause instability [3].

iii. AFO must be small enough to fit comfortably inside of a shoe.

i. Weight

i. The orthosis will be light enough to allow a full range of motion without hindrance. The weight will not impair the patient's walking gait or velocity. It should be minimized as much as possible, ideally weighing less than 1 kg [18].

j. Materials

i. The foot sleeve of the brace as well as the bungee cord will be composed of a blend of nylon, polyester, and latex. These materials were chosen for their specific properties that enhance both functionality and comfort [19].

1. The sleeve's breathability will absorb sweat and keep the foot dry, providing comfort during extended use. The material will also be tight and

strong, ensuring that the sleeve stays securely in place without sliding [20].

2. The fabric is smooth and soft, adding comfort, while its graduated compression promotes circulation, providing support and pain relief to the user [20].
    a. Nylon is specifically selected for its low elongation, strength, high-temperature resistance, and ability to make the brace visually appealing and lightweight [20].
    b. Polyester, known for its durability and strength, is ideal as it retains its shape and resists wrinkles, shrinking, and environmental elements like water and wind, which is crucial since the device will frequently be exposed to outdoor conditions [21].
    c. Latex contributes flexibility, durability, and excellent resistance to liquids, making it an effective barrier against moisture while maintaining overall strength [22].

ii. The effectiveness of preventing ankle inversion depends highly on the rigid strength of the cast. Fiberglass substrates impregnated with polyurethane resin offer a strength proportional to the square of their thickness. By wrapping the fiberglass twice, the rigid support can withstand a bending deflection of 50 N minimum. With an increase in thickness, the piece can provide exponential strength [23].

iii. The rigid supporting piece along the ankle will be constructed using fiberglass polymer tape. This material was selected for the following characteristics: lightweight, moldable, radiolucent, resistant to degradation by water, inexpensive, high strength-to-weight ratio, and thin profile [23].

iv. There is potential to include a 3D printed component within the fiberglass tape that is modeled after the patient's own anatomy and acts as a reinforcement against ankle inversion. This piece would be fabricated using PLA.

v. Fiberglass substrate's lightweight nature will allow for ease of use, enabling better movement while reducing fatigue and pain for the user. Its sturdiness ensures resistance to everyday wear and tear, providing long-term support. Additionally,

fiberglass substrate's porous nature will allow the patient comfort and breathability. These combined properties maximize the aid needed for foot-dragging prevention, ankle stabilization, and overall gait improvement.

k. Aesthetics, Appearance, and Finish
    i. The AFO will have a black, sleek design to reduce its visibility. It will resemble an athletic brace, promoting a more natural appearance when worn in public. The brace is designed to fit comfortably in tennis shoes and Converse, giving the patient the freedom to maintain their personal style.
    ii. The brace will have a smooth finish and a slim appearance, making it as inconspicuous as possible while still providing the necessary support.

2. Production Characteristics
    a. Quantity
        i. This project consists of making one right-leg AFO. However, considering mass production, the quantity would meet market demands among teens needing right-leg and/or left-leg AFOs.
    b. Target Product Cost
        i. This project is funded by Biomedical Engineering (BME) Design at the University of Wisconsin-Madison. The monetary supplementation is $100 with room for expansion where needed.
        ii. The initial prototype accounted for $189.02 of last semester's budget of $300. $8.71 was covered by the BME department, so the total spent through BME Design funding was $180.30.
        iii. The remaining budget is to be spent on fabricating a rigid support to inhibit the patient's ankle inversion. Because fiberglass substrates are relatively inexpensive, this support should cost under $50 to implement in an initial prototype [24].

3. Miscellaneous
    a. Standards and Specifications
        i. Code of Federal Regulations Title 21, Section 890.3025
            1. This device is classified as a Class I Medical Device. The device will be considered a Class II Medical Device if an electronic component is incorporated [25].
        ii. 501(k) requirements, premarket submission

1. Most class I medical devices are exempt from 501(k) requirements. The device may be exempt if the FDA determines that a 501(k) is not required to provide reasonable assurance of the safety and effectiveness of the device [26].

iii. Code of Federal Regulations Title 21, Section 890.3475

1. This defines a limb orthosis as a medical device worn on upper or lower limbs to support, correct, prevent deformities, or align body structures to improve bodily function. Examples of limb orthoses are as follows: a whole limb and joint brace, a hand splint, an elastic stocking, a knee cage, and a corrective shoe [27].

iv. Code of Federal Regulations, Title 21, Chapter 1 Part 803

1. Manufacturers and facilities that use the device must report deaths and serious injuries that the device has caused or contributed to through a Medical Device Report (MDR) [28].

v. ISO Standard 14971:2019

1. Risk analysis through Failure Modes and Effects Analysis (FMEA) should be completed to identify potential risks for the patient, operator, and property. This includes gathering data and reviewing literature about the risks of similar medical devices. This standard states the concept of risk involves the probability of occurrence of harm and the severity of its consequences [29].

vi. ISO Standard 8549-3:2020

1. Defines orthosis as an externally applied device utilized to compensate for impairments in the structure and function of the neuromuscular and skeletal system; ankle-foot orthosis is defined as an orthosis that encompasses the ankle joint and the whole or part of the foot [30].

vii. ISO Standard 8551:2020

1. Covers functional deficiencies in prosthetics and orthotics. The standard provides guidelines for the person to be treated with an orthosis, the clinical objectives of treatment, and the functional requirements of the orthosis [31].

viii. ISO Standard 2267:2016

1. This standard outlines a specific testing procedure for ankle-foot devices and foot units used in external lower-limb prostheses.
2. Testing should be completed on how the prosthetic device performs under repeated, cyclical loading conditions that simulate the forces and motions experienced during the complete stance phase of walking. This includes the moment the heel strikes the ground to the moment the toe leaves the ground (toe-off). The testing will provide performance characteristics of the prosthetic device such as its strength, durability, and service life, ensuring the prosthesis meets quality and safety standards [32].

b. Customer [33]

i. The device is intended for everyday use by a 16-year-old teenager, who has been diagnosed with Facioscapulohumeral Dystrophy. While the orthosis will be custom-fitted to the patient's ankle, the primary target audience includes all young individuals diagnosed with Facioscapulohumeral Dystrophy or similar muscular dystrophies that require an ankle orthosis.

ii. The device must be discreet, featuring a slim and narrow design that allows it to be easily hidden under pants or remain minimally noticeable with any type of clothing, ensuring it doesn't draw attention to the individual's physical limitation.

iii. The device must be capable of holding the ankle in dorsiflexion (angle 10 degrees upwards from straight foot plane) when unweighted to ensure foot clearance and prevent gait deviations.

iv. The device must have enough flexibility to ensure that other functional activities, such as squatting or descending stairs, are minimally affected.

v. The device must minimize the need for eccentric muscle contractions while preventing foot slap to support individuals with ankle weakness.

c. Patient-related concerns

i. The device must be flexible enough to allow for natural gait movement while being sturdy enough to support the patient's ankle weakness and prevent foot drop as well as foot collapse (foot inversion specifically).

ii. The device must not interfere with daily activities or draw attention to itself or the patient.
iii. The device must be discreet to prevent drawing unwanted attention and reduce the risk of bullying at school and in other public settings.

d. Additional optional patient requests
   i. The device should be designed to fit comfortably within the patient's horse riding boot.
   ii. The device should resemble a standard athletic brace to avoid drawing attention in public settings.

e. Economic Impact
   i. Given that approximately 53,000 AFOs are fabricated each year in the United States at an average Medicare reimbursement of $417, more than $2.2 million per year are spent on them [34]. These costs may make AFOs inaccessible to low-income families.
   ii. For patients with muscular dystrophies, expenses incur in the form of direct and indirect medical costs.
      1. Direct medical costs include hospital visits, therapy, pharmaceutical treatments, and insurance coverages. The sum of these factors amounts to about $22,533 per year in the United States [35].
      2. Indirect costs include home renovations, vehicle accommodations, home relocations, professional caregiving, dietary supplements, travel expenses, and more. In total, indirect costs in the United States cost approximately $12,939 per patient per year [35].
   iii. One of the largest contributors to the loss of income in families with a member who suffers from a muscular disorder is loss of productivity. While race, age, gender, duration of disease, level of education of the primary income member, and number of adults in the family are considered, the annual loss of income for a family with a patient requiring care is $21,600 less than those who do not need care [35].
   iv. Overall, the total cost of muscular dystrophy disorders in the United States ranges from $1.07 to $1.4 billion per year [35].

v. Providing a cost-effective AFO will alleviate the long-term economic burden of FSHD and increase the productivity of individuals who have the condition, raising their income levels and allocating more funds to treating the condition.

f. Competition

When constructing AFOs, the Three-Point Force system is essential for creating an orthosis that stabilizes a joint or segment to reduce angular rotation. The force is applied either medio-laterally or anteroposteriorly, with counter forces applied above and below the primary force, all summing to zero. The longer the lever of the orthosis, the farther apart the points of force are, resulting in greater correction. This technique can also help reduce pressure and discomfort when wearing the orthosis. This system is incorporated in the majority of existing AFO designs [36].

i. Passive-Dynamic AFO (PD-AFO)
    1. Sleek and flexible design, ideal for patients with less severe ankle weakness.
    2. Have a flexible calf shell, which can absorb energy to promote dorsiflexion by releasing additional energy during the push-off phase of walking.
    3. Shown to provide better comfort and improve spatio-temporal parameters.
    4. Able to customize dimensions for the user through 3D printing but unable to adjust stiffness and support to match the patient's level of ankle impairment [1].

ii. Supramalleolar Orthosis (SMO)
    1. Pediatric SMO's are made from thin and flexible thermoplastic that provides support just above the ankle bones (malleoli).
    2. Primarily used to control subtalar joint alignment to maintain a vertical or neutral heel to help improve mediolateral movement.
    3. Comfortable to wear in shoes due to their thin and minimally restrictive design [36].

iii. Variable Stiffness Orthosis (VSO)

1. This powered AFO features a customizable cam-based transmission able to specify any torque-angle and change the magnitude of its overall stiffness in real time.
2. Found to lead to reduced foot drop and increased total ankle moments.
3. In the research stages, not currently on the market [38].

iv. Jointed AFO

1. Features a hinge at the ankle joint, allowing for motion
2. Optimizes gait patterns and allows for a full range of motion.
3. Drawbacks include being bulkier, potentially noisy, and prone to parts breaking more easily [36].

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Appendix B: Previous Fabrication Methods

The carbon fiber attachment was designed in SolidWorks and subsequently 3D printed at the UW-Makerspace using the Bambu Labs printer [38]. The material will undergo an initial testing evaluation on Solidworks prior to being printed (see testing section for more details). This preliminary testing will assess the strength, flexibility, and overall functionality of the carbon fiber component in the device.

The ankle brace and bungee cord will be purchased (see BPAG cost sheet for pricing details), but the bungee cord will be customized to meet the specific dimensions and support requirements of the patient. The cord will be cut and modified to optimize the level of tension needed to assist with walking. These modifications will be made based on assumptions and initial bungee cord testing and then fine-tuned after an in-person testing session with the patient (see the Testing and Results section for more detailed procedures). To ensure ease of adjustability, the bungee cord will be threaded through a “lock lace” plastic cord lock, which will also be purchased and integrated into the design.

The attachment for the Locklace will also be designed in SolidWorks and 3D printed at the UW-Makerspace using the Ultimaker printer [38]. It will be printed using PLA material also on the Bambu Labs printer, and the Locklace will be assembled by fitting snugly and being glued to the inside the printed piece. Both the Locklace and the 3D-printed piece, when assembled, will be sewn onto the foot brace through two holes on either side of the printed component. This design increases the surface area for improved grip, ensures the Locklace is securely positioned, and facilitates ease of use and adjustability on the brace.

To assemble all components, the gel-padded compression sock will remain separate as an additional layer of comfort and support for the user. The gel pads will be strategically sewn onto the sock at three key locations—behind the calf, around the ankle bone, and near the second attachment point of the carbon fiber support, around the ball of the foot. These placements were determined based on the pressure points identified by team members during and after testing. The carbon fiber attachment will be securely sewn onto the foot sleeve brace using purchased sheets of nylon fabric. This will hold the carbon fiber in place without adding unnecessary bulk or restricting movement. This assembly will be completed by hand using basic black nylon thread and sewing needle. The plastic cord lock and its attachment will be sewn onto the top portion of the foot sleeve, while the bungee cord, once placed under tension, will be threaded through the cord lock, ensuring adjustability. The bungee will then be covered and secured using diagonal Velcro straps, which wrap across the front of the ankle to stabilize the brace. The bottom of the bungee cord will be sewn to the front of the brace, approximately 15.24 centimeters from the top, using additional nylon fabric that will be glued down with strong fabric glue for extra support and reinforcement.

Once fully assembled, the user will be able to put on the brace by first slipping on the compressive sock, followed by sliding the brace onto their foot, both processes like a regular sock. The bungee cord can then be tightened to the user's preference using the cord lock, and the Velcro straps will be fastened as the final step. The design prioritizes simplicity, speed, and ease

of use, as the AFO will be worn daily and taken on and off frequently. This streamlined assembly and adjustment process ensures that the device will be comfortable, user-friendly, and highly functional for everyday use.

Appendix C: Fall 2025 Expenses

Item	Description	Manufacturer	Mft Pt#	Vendor	Vendor Cat#	Date	QTY	Cost Each	Total	Link
Ankle Brace - Component 1										
Ankle Brace	Cloth brace	**Abiram**		Amazon		10/10/2024	1	$14.88	$14.88	Link
Gel padding	medical grade padding	**Shechekin**		Amazon		10/10/2024	1	$15.81	$15.81	Link
Gel sock	Compressive sock to support the carbon fiber	**KEMFORD**		Amazon		10/10/2024	1	$15.95	$15.95	Link
Plastic cord locks	End of the bungee	**HeadoUS**		Amazon		10/10/2024	1	$3.98	$4.20	Link
Nylon Fabric	fabric/cloth to sew carbon fiber	**MYUREN**		Amazon		11/6/2024	1	$12.61	$12.61	Link
Bungee pt 2	stronger bungee to support better dorsiflexion	**LuckyStraps**		Amazon		10/23/2024	1	18.99	$20.03	Link
Bungee	thinner bungee	**Huouoo**		Amazon		10/25/2024	1	$6.32	$6.32	Link
Mini caribener	small sized caribener to hold bungee	**REI**		REI		11/4/2024	1	$6.00	$6.00	**In-store**
Shock cord	thinner and stronger bungee	**REI**		REI		11/4/2024	1	$5.95	$6.61	**In-store**
Lock laces	lock laces to fix the slipping problem of the plastic cord lock	**Lock Laces**		Amazon		11/4/2024	1	$12.65	$12.65	Link
Fabric Glue	glue to attach the cord locks to the fabric	**E6000**		Amazon		11/08/2024	**1**	$8.14	$8.14	Link
Needles and Thread	Stronger needles and thread to attatch various fabrics	**Basic Home**		Amazon		12/03/2024	**1**	$8.43	$8.43	Link
Carbon Fiber piece - Component 2										
3D printing prototype	3D printing of back support	**Bambu printer**		Makerspace		11/8/2024	1	1.4	$1.40	***covered by our**

										given $50 per team
3D printing prototype - 3 variants	3D printing of back support	**Bambu printer**		Makerspace		11/12/2024	1	3.8	$3.80	***covered by our given $50 per team**
3D printing prototype	3D printing of back support	**Bambu printer**		Makerspace		11/13/2024	1	1.71	$1.71	***covered by our given $50 per team**
Lock lace piece	3D printing the lock lace piece	**Bambu printer**		Makerspace		11/18/2024	1	0.23	$0.23	***covered by our given $50 per team**
3D Printing Final Prototype	3D printing of back support	**Shen Printer**		Makerspace		12/3/2024	1	1.57	$1.57	***covered by our given $50 per team**
Epoxy Mold - Component 3										
Epoxy	Take cast of the leg	**Easy Pour Epoxy**		Amazon		11/14/2024	1	$39.97	$39.97	Link
Mold release	PVA release agent - Prevent bonding to	**Mrealeazy**		Amazon		11/14/2024	1	0	$0.00	***Used the**

Agent	the cast									**provided materials in ECB**
								TOTAL:	**$189.02**	

Appendix D: Spring 2025 Expenses

Item	Description	Manufacturer	Mft Pt#	Vendor	Vendor Cat#	Date	QTY	Cost Each	Total		Total Budget Spent	Link
Category 1 - Rigid Support												
CF-PLA	Carbon Fiber PLA 3D Print	Shen Printer		MakerSpace		2/28/2025	1	$0.86	$0.86			
CF-PLA	Carbon Fiber PLA 3D Print	Shen Printer		MakerSpace		3/5/2025	1	$2.42	$2.42			
CF-PLA	Carbon Fiber PLA 3D Print	Shen Printer		MakerSpace		3/14/2025	1	$3.66	$3.66			
CF-PLA (red)	Carbon Fiber PLA 3D Print	Shen Printer		MakerSpace		4/4/2025	1	$3.92	$3.92			
CF-PLA	Carbon Fiber PLA 3D Print	Shen Printer		MakerSpace		4/4/2025	1	$1.94	$1.94			
Category 2 - Straps and Padding												
Carpet Tape		Capitol	705-1560	Menards	7051560	4/2/2025	1	$7.36	$7.36		$7.36	link
Mesh Padding	3D Air Sponge Mesh Fabric	Tong Gu		Amazon		3/7/2025	1	$16.99	$16.99		$16.99	link
Velcro	Velcro pieces			MakerSpace		2/28/2025	2	$0.40	$0.80			
								TOTAL:	**$37.95**	**Budget Spent:**	$24.35	

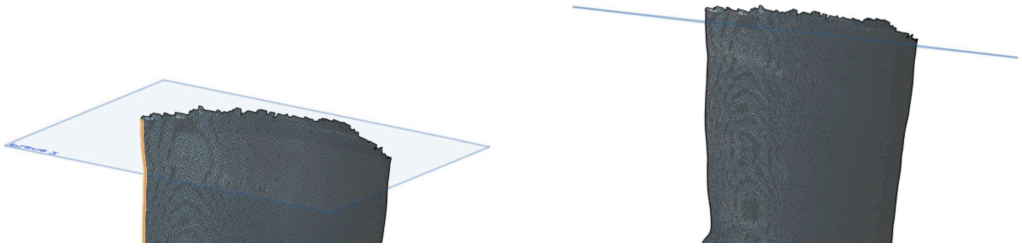
Appendix E: Fabrication of Rigid Support

3D Scanning:

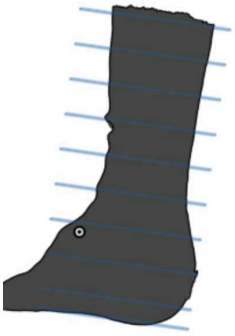
1. Enter the Creality software on the Makerspace computer.
2. Select "New Design"
3. Place your part on the rotating disc
4. Set up dotted blocks around your part to allow for spatial awareness
5. Click the green arrow to begin scanning your part.
6. Hold the wand about one foot away from your part, carefully watching the light on the handle.
    1. Green = acceptable distance from the part
    2. Orange = too close to the part
    3. Red = Extremely close to the part
    4. Light blue = too far away from the part
    5. Dark blue = Extremely far away from the part
7. Continue scanning the part until the part is fully defined and green on the screen
8. Once done scanning (about 15 minutes), click the all in one edit tool to smooth, refine, minimalize your mesh.
9. Delete the components of the mesh that do not include your design/part.
10. Select "export" and select "obj" file.

3D Modeling of Rigid Support:

1. Open OnShape
2. Import stl or obj file by clicking "create" --> "import files"
3. If prompted, import the part in mm
4. Click on the document that displays your part.
5. Using the plane tool, create a 3 point plane that intersects the top of the cast.



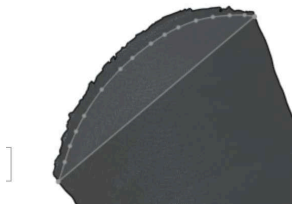
6. Using the plane feature again, create offset planes every 1 inch down the cast until you reach the bottom.



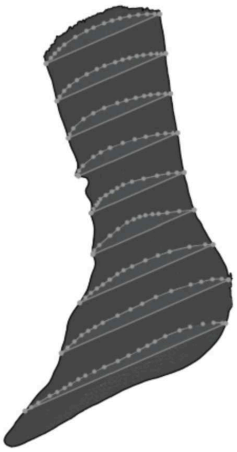
7. Create another plane using 3 point selection that splits the cast to create a smooth edge along the backside.
8. Using the split tool, select the cast mesh as the surface and the plane you created in step 7. Unselect "keep both sides" and ensure the deleted side is the back half of the cast to produce the part shown below.

9. On the face of the plane created in step 5, create a sketch.

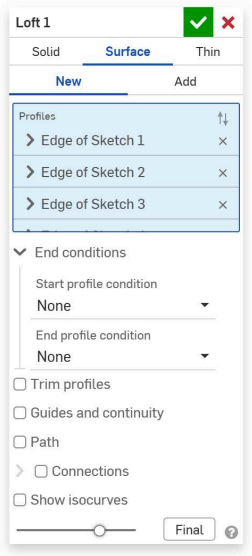
10. Use the spline tool to draw a curved line that matches the curvature of the cast at that intersection.
11. Use the line tool to connect the ends of the spline.



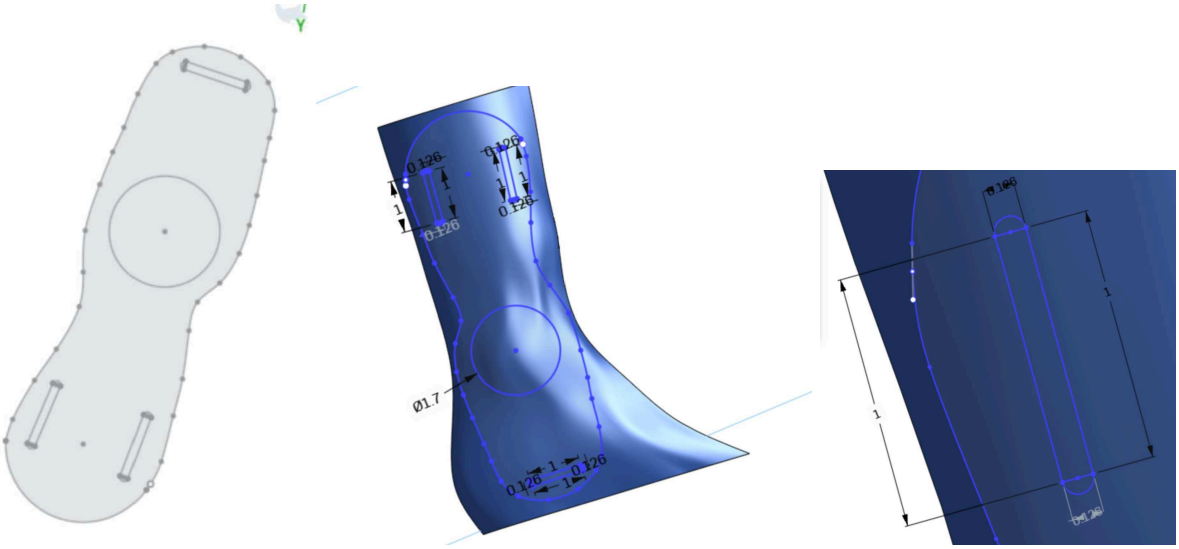
12. Repeat steps 9-11 on every plane down the cast mesh.



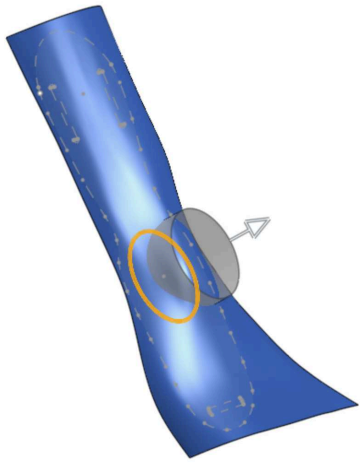
13. Using the "loft" tool, connect the sketches in the order they appear on the cast.



14. On the resulting surface, create a sketch that includes 3 one inch rounded slots and a 1.7 inch diameter circle using the plane created in step 6 and 7.



15. Click on the extrude tool
16. Select the lofted surface
17. Select the circle in sketch created in step 14
18. Extrude the circle fully through the part



19. Repeat steps 15 - 18 for every line/component in the same sketch.
20. Using the cut tool, select the surface to cut and select the extruded pieces to cut with
21. Uncheck both sides and ensure the holes are removed, not the surface itself.
22. Use the thicken tool to thickent the final design to 0.105 inches.

Foam Attachment:

1. Cut two pieces of the mesh off of the sheet
    a. Ensure they are big enough to cover the full rigid support dimensions
2. Trace one outline of a rigid support onto the smooth side of the mesh padding sheet, including the slits for the straps
3. Lay a second mesh sheet on top of the first, ensuring the smooth side you traced onto is visible on one side and the other has mesh
4. Sew the two pieces together along the traced lines, excluding the strap slits.
5. Trim the pieces down to about 1 mm outside of the sewn lines
6. Place the piece of mesh onto the rigid support (smooth side facing the support) and ensure the shapes are consistent with one another
7. Using the Weld-On 4 (shown below), apply the liquid to the concave side of the rigid support until fully covered

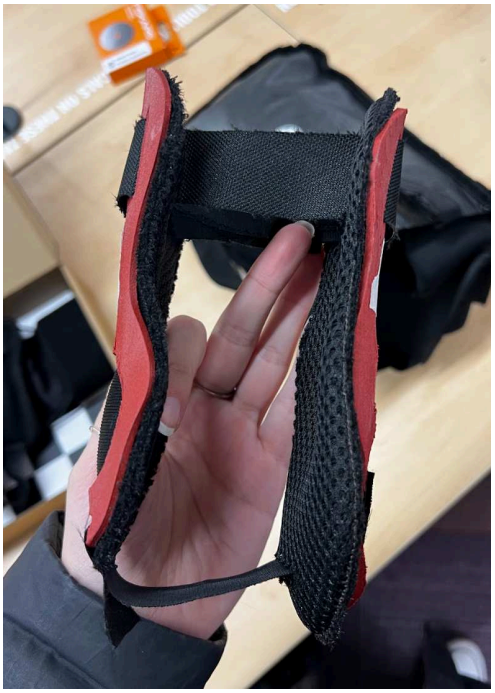
8. Quickly place the smooth foam side onto the adhesive and press until firmly adhered
9. Let the full part sit for about 15 minutes before moving
10. Check the edges for loose pieces of foam and apply more adhesive if needed
11. Using scissors, cut the slits for the straps out of the mesh padding
12. Leave the circle over the rigid support hole intact!

Straps:

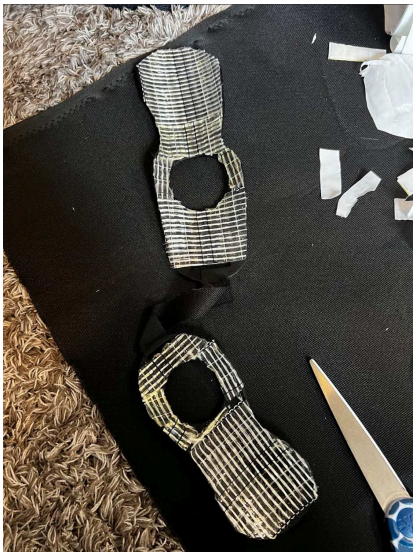
1. Cut 1 foot-long strips of robust black material about one inch thick (cut 4)
2. Cut four more straps, one inch thick, to the patient's specific foot dimensions
3. Cut 4 velcro squares of the adhesive velcro about one inch wide
4. Peel the paper off of the back of these squares and adhere to the outside of the 3D printed rigid support component
    a. You can see the velcro piece adhered to the outside of the brace just above the strap slit
    b. Repeat for all four braces

5. Sew four velcro strips (length can be modified for patient dimensions) to one end of the smaller underfoot straps
    a. Do for all four braces
6. Loop the other end of the underfoot strap through the bottom of the brace ad sew it to itself
    a. Ensure the underfoot strap is an appropriate length for your client with minimal adjustments required
    b. Repeat for all four braces
    c. Note: the underfoot strap lengths may change based on support lengths
7. Cut four pieces of non-adhesive velcro and sew them into the longer, one foot long straps
8. Sew one velcro piece onto each end and on opposite sides of the strip
    a. Repeat for all four straps
9. Loop the straps through the foam slits and into the holes in the rigid support to secure






** FAILED ATTEMPT USING CARPET TAPE ADHESIVE SHOWN BELOW **



Appendix F: MTS Testing Protocol and Code

Protocol:

1. Obtain three 3D printed rigid supports with infills of 15%, 35%, and 50%
2. Record the mass of the sample with scale and dimensions of the samples. Examine the sample for defects that could potentially lead to failure or errors.
3. Know the maximum possible load for the sample. Choose a proper load cell and fixture that are correct for the sample.
4. Insert the three point bending attachment pieces on to the MTS Criterion Model 43 Machine
5. Attached steel bars via rubber bands to the two end pieces
6. Open Test suite on the adjacent computer
7. Set values for testing rate 0.05 mm/mm, 10 hz, and 191 mm gauge length
8. Set the Safety Stops.
9. Lock the machine.
10. Place the sample into the fixture.
11. Rotate the emergency switch to the right, and it should pop outwards.
12. Double click the lock button, and the light should turn green.
13. Zero the load by right-clicking on the load cell on the software and hitting “zero signal”.
14. Use the wheel to scroll down until the top clamp is barely touching the sample. Carefully watch the load values and wait for a small spike in load as the fixture makes contact with the sample.
15. Record the test rate.
16. Lock the MTS machine by pressing the lock button on the hand controls.
17. Zero the crosshead and the load using the same process as before (Step 13).
18. Press the play button on the computer screen to start the test trial.
19. Enter the sample’s measured diameter.
20. Allow the MTS machine to run until sample failure, the curve on the graph flattens, or until the load almost (but does not) reach load cell capacity or fixture capacity.
21. Press Stop. Kill the switch or press the stop button if something goes wrong.

22. Repeat testing trails using the same steps above for each rigid support with a different infill.
23. Export the data by right-clicking on the trial and selecting the proper file location.
24. Power off the machine, return components, and clean up the sample and surrounding area.

Code:

```
%% RS Dimensions

L = 191; % mm

length = 12.557; % mm

width = 2.7; % mm

surface_A = 9200.346; % mm
```

```
x_sectional_A = 33.38; % mm
%% Data Imports
[if_50, if_50_path] = uigetfile('*','TestRun3_50infill');
if_50 = importdata([if_50_path,filesep,if_50]);
[if_35, if_35_path] = uigetfile('*','TestRun2_35infill');
if_35 = importdata([if_35_path,filesep,if_35]);
[if_15, if_15_path] = uigetfile('*','TestRun4_15infill');
if_15 = importdata([if_15_path,filesep,if_15]);
%% Variable Assignment
disp_if_50=if_50.data(:,1); % mm
force_if_50=if_50.data(:,2); % N
time_if_50=if_50.data(:,3); % sec
disp_if_35=if_35.data(:,1); % mm
force_if_35=if_35.data(:,2); % N
time_if_35=if_35.data(:,3); % sec
disp_if_15=if_15.data(:,1); % mm
force_if_15=if_15.data(:,2); % N
time_if_15=if_15.data(:,3); % sec
%% Force Plot
figure(1);
hold on;
title("15%, 35%, 50% Infill Rigid Support Forces");
plot(force_if_50, 'r');
```

```
plot(force_if_35, 'b');
plot(force_if_15, 'g');
xlabel("Time (ms)");
ylabel("Force (N)");
grid on;
hold off;
%% Max Displacements
max_disp_50 = max(disp_if_50);
max_disp_35 = max(disp_if_35);
max_disp_15 = max(disp_if_15);
%% Load vs. Deflection Plots
figure(2);
hold on;
title("15%, 35%, 50% Infill Rigid Support Load vs. Displacement")
plot(disp_if_50,force_if_50, 'r');
plot(disp_if_35,force_if_35, 'b');
plot(disp_if_15,force_if_15, 'g');
grid on;
xlabel("Displacement (mm)");
ylabel("Force (N)");
hold off;
%% Peak Load
if_50_pl = max(force_if_50); % N
```

```
if_35_pl = max(force_if_35); % N

if_15_pl = max(force_if_15); % N

%% Stress Data

if_50_stress = force_if_50/x_sectional_A;

if_35_stress = force_if_35/x_sectional_A;

if_15_stress = force_if_15/x_sectional_A;

%% Strain Data

if_50_strain = disp_if_50/L;

if_35_strain = disp_if_35/L;

if_15_strain = disp_if_15/L;

%% Stress vs Strain Plot

figure(3);

hold on;

title("15%, 35%, 50% Infill Rigid Support Stress vs. Strain")

plot(if_50_strain,if_50_stress, 'r');

plot(if_35_strain,if_35_stress, 'g');

plot(if_15_strain,if_15_stress, 'b');

grid on;

xlabel("Strain");

ylabel("Stress (MPa)");

hold off;

%% Max Stress From Data

if_50_maxstress2 = max(if_50_stress); %MPa
```

```
if_35_maxstress2 = max(if_35_stress); %MPa

if_15_maxstress2 = max(if_15_stress); %MPa
```

Appendix G: Force Plate Testing Protocol and Code

Protocol:

Recording Data with the Force Plates:

a. **Pre-Test Measurements**

   i. Measure the subject's foot length in centimeters.

b. Power on the amplifier boxes for each force plate you intend to use (**Figure 1a**)

Figure 1. A.) Amplifier boxes for the in-ground force plates. The blue arrow indicates where the power switch is. **B.)** Bertec Acquire 4 software for data collection of force plate data.

c. Log in to the motion capture control computer using your CAE login

d. Open the Bertec Acquire 4 Software (**Figure 1b**)

e. Allow the software several minutes to identify the force plates. Ensure the force plates and all channels are selected (figure 8)

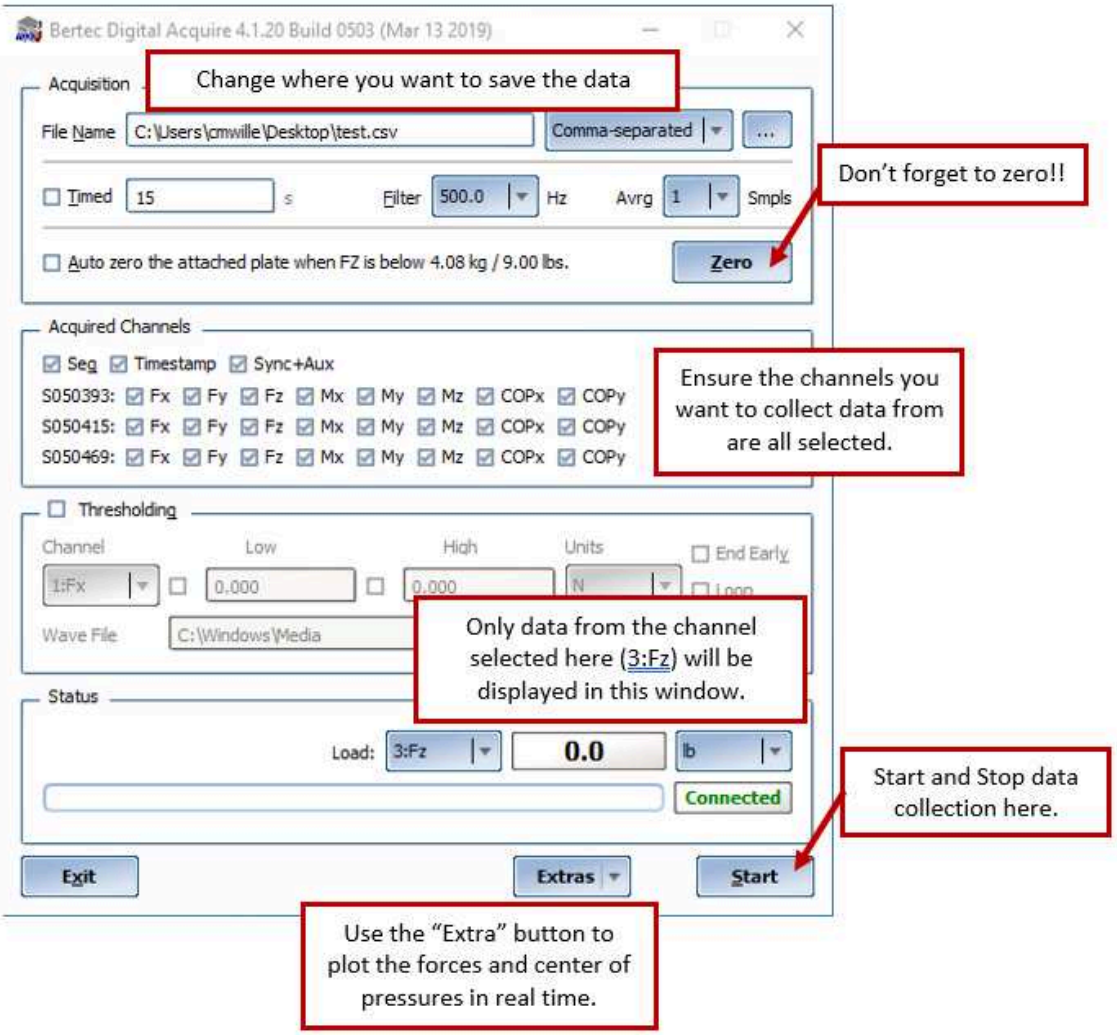


Figure 8. Annotated user interface for Bertec Acquire 4.

f. In the Acquisition box in **Figure 8**, select the time box and input 30 seconds. This ensures each text runs for exactly 30 seconds.

g. Zero the force plates (**Figure 8**)

h. Ensure the data is saved in the desired location:

    i. Under acquisition, rename each test with a descriptive title for each trial (see below), making sure it is a .csv file

i. Follow the instructions for trial 1

j. Press start to begin recording
k. The test will automatically end after 30 seconds
l. Data will be saved in a .csv file format, and output data will be in Newtons and meters
m. Repeat steps 1, j, k, and l for each trial listed below

TRIALS

1. One footed stance – eyes open – no brace (OFEONB)
    a. Zero the force plates
    b. Place your right foot in the center of the force plate **without the brace** on and keep your **eyes open**
    c. Stand on the force plate for thirty seconds without placing their left foot on the plate
2. One footed stance – eyes open – with RED brace (OFEORED)
    a. Zero the force plates
    b. Place your right foot in the center of the force plate **with the brace** on and keep your **eyes open**
    c. Stand on the force plate for thirty seconds without placing their left foot on the plate
3. One footed stance – eyes open – with **BLACK** brace (OFEOBLACK)
    a. Zero the force plates
    b. Place your right foot in the center of the force plate **with the brace** on and keep your **eyes open**
    c. Stand on the force plate for thirty seconds without placing their left foot on the plate
4. One footed stance – eyes closed – no brace (OFECNB)
    a. Zero the force plates
    b. Place your right foot in the center of the force plate **without the brace** on and keep your **eyes closed**

    c. Stand on the force plate for thirty seconds without placing their left foot on the plate
5. One footed stance – eyes closed – with RED brace (OFECRED)
    a. Zero the force plates
    b. Place your right foot in the center of the force plate **with the brace** on and keep your **eyes closed**
    c. Stand on the force plate for thirty seconds without placing their left foot on the plate
6. One footed stance – eyes closed – with **BLACK** brace (OFECBLACK)
    a. Zero the force plates
    b. Place your right foot in the center of the force plate **with the brace** on and keep your **eyes closed**
    c. Stand on the force plate for thirty seconds without placing their left foot on the plate
7. One footed stance – eyes open – with wedge – no brace (WEDGEEONB)
    a. Place your 10 degree wedge on the marked tape section of the force plate
    b. Zero the force plates
    c. Place your right foot **without the brace** onto the wedge
    d. Place your left foot on the force plate adjacent to the wedge in the marked foot zone
    e. Stand in this position for 30 seconds
8. One footed stance – eyes open – with wedge – with red brace (WEDGERED)
    a. Place your 10 degree wedge on the marked tape section of the force plate
    b. Zero the force plates
    c. Place your right foot **with the brace** onto the wedge
    d. Place your left foot on the force plate adjacent to the wedge in the marked foot zone
    e. Stand in this position for 30 seconds
9. One footed stance – eyes open – with wedge – with **BLACK** brace (WEDGEBLACK)
    a. Place your 10 degree wedge on the marked tape section of the force plate
    b. Zero the force plates

c. Place your right foot **with the brace** onto the wedge
d. Place your left foot on the force plate adjacent to the wedge in the marked foot zone
e. Stand in this position for 30 seconds

Code:

```
% Close figures and clear out other variables that have been assigned
close all;
clear all;
% Data is saved from the force plates in Newtons and meters.
stabilogram_folder = "C:\Users\madis\Downloads\BME 301\Stabilograms";
%% OFECBLACK
[file1, path1] = uigetfile('*','OFECBLACK');
data1 = importdata([path1,filesep,file1]);
time1 = data1.data(:,1);
fx1_2 = data1.data(:,4);
fy1_1 = data1.data(:,5);
fz1_1 = data1.data(:,6);
cop1_x_1 = data1.data(:,10);
cop1_y_1 = data1.data(:,11);
figure(1);
plot(cop1_x_1,cop1_y_1);
title('Eyes Closed | One Foot | Black Brace - Stabilogram');
xlabel('Medio-Lateral (m)');
```

```
ylabel('Anterior-Posterior (m)');
grid on;
axis equal;
saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_OFECBLACK.png'));

%% OFECNB
[file2, path2] = uigetfile('*','OFECNB');
data2 = importdata([path2,filesep,file2]);
fx1_2 = data2.data(:,4);
fy1_1 = data2.data(:,5);
fz1_1 = data2.data(:,6);
cop1_x_2 = data2.data(:,10);
cop1_y_2 = data2.data(:,11);
figure(2);
plot(cop1_x_2,cop1_y_2);
title('Eyes Closed | One Foot | No Brace - Stabilogram');
xlabel('Medio-Lateral (m)');
ylabel('Anterior-Posterior (m)');
grid on;
axis equal;
saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_OFECNB.png'));
%% OFECRED
[file3, path3] = uigetfile('*','OFECRED');
```

```
data3 = importdata([path3,filesep,file3]);
fx1_2 = data3.data(:,4);
fy1_1 = data3.data(:,5);
fz1_1 = data3.data(:,6);
cop1_x_3 = data3.data(:,10);
cop1_y_3 = data3.data(:,11);
figure(3);
plot(cop1_x_3,cop1_y_3);
title('Eyes Closed | One Foot | Red Brace - Stabilogram');
xlabel('Medio-Lateral (m)');
ylabel('Anterior-Posterior (m)');
grid on;
axis equal;
saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_OFECRED.png'));
%% OFEOBLACK
[file4 path4] = uigetfile('*','OFEOBLACK');
data4 = importdata([path4,filesep,file4]);
fx1_2 = data4.data(:,4);
fy1_1 = data4.data(:,5);
fz1_1 = data4.data(:,6);
cop1_x_4 = data4.data(:,10);
cop1_y_4 = data4.data(:,11);
figure(4);
```

```
plot(cop1_x_4,cop1_y_4);
title('Eyes Open | One Foot | Black Brace - Stabilogram');
xlabel('Medio-Lateral (m)');
ylabel('Anterior-Posterior (m)');
grid on;
axis equal;
saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_OFEOBLACK.png'));
%% OFEONB
[file5, path5] = uigetfile('*','OFEONB');
data5 = importdata([path5,filesep,file5]);
fx1_2 = data5.data(:,4);
fy1_1 = data5.data(:,5);
fz1_1 = data5.data(:,6);
cop1_x_5 = data5.data(:,10);
cop1_y_5 = data5.data(:,11);
figure(5);
plot(cop1_x_5,cop1_y_5);
title('Eyes Open | One Foot | No Brace - Stabilogram');
xlabel('Medio-Lateral (m)');
ylabel('Anterior-Posterior (m)');
grid on;
axis equal;
saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_OFEONB.png'));
```

```
%% OFEORED
[file6, path6] = uigetfile('*','OFEORED');
data6 = importdata([path6,filesep,file6]);
fx1_2 = data6.data(:,4);
fy1_1 = data6.data(:,5);
fz1_1 = data6.data(:,6);
cop1_x_6 = data6.data(:,10);
cop1_y_6 = data6.data(:,11);
figure(6);
plot(cop1_x_6,cop1_y_6);
title('Eyes Open | One Foot | Red Brace - Stabilogram');
xlabel('Medio-Lateral (m)');
ylabel('Anterior-Posterior (m)');
grid on;
axis equal;
saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_OFEORED.png'));
%% WEDGEBLACK
[file7, path7] = uigetfile('*','WEDGEBLACK');
data7 = importdata([path7,filesep,file7]);
fx1_2 = data7.data(:,4);
fy1_1 = data7.data(:,5);
fz1_1 = data7.data(:,6);
cop1_x_7 = data7.data(:,10);
```

```
cop1_y_7 = data7.data(:,11);

figure(7);

plot(cop1_x_7,cop1_y_7);

title('Wedge | Black Brace - Stabilogram');

xlabel('Medio-Lateral (m)');

ylabel('Anterior-Posterior (m)');

grid on;

axis equal;

saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_WEDGEBLACK.png'));

%% WEDGEEONB

[file8, path8] = uigetfile('*','WEDGEEONB');

data8 = importdata([path8,filesep,file8]);

fx1_2 = data8.data(:,4);

fy1_1 = data8.data(:,5);

fz1_1 = data8.data(:,6);

cop1_x_8 = data8.data(:,10);

cop1_y_8 = data8.data(:,11);

figure(8);

plot(cop1_x_8,cop1_y_8);

title('Wedge | No Brace - Stabilogram');

xlabel('Medio-Lateral (m)');

ylabel('Anterior-Posterior (m)');

grid on;
```

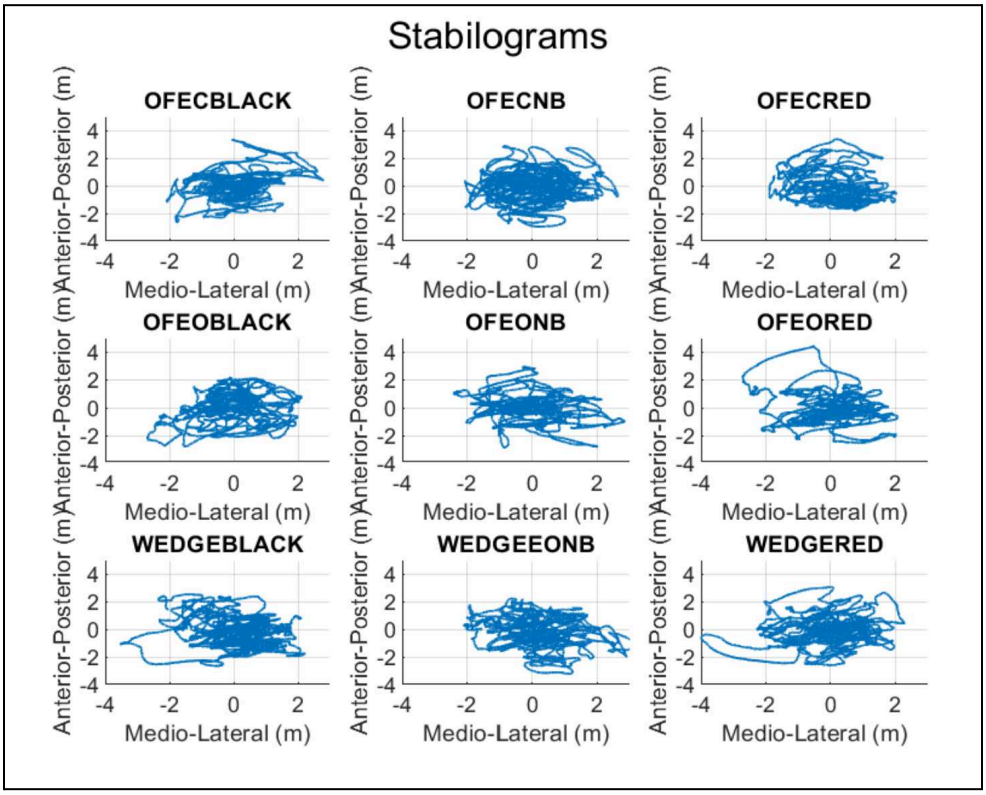
```
axis equal;

saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_WEDGEEONB.png'));

%% WEDGERED

[file9, path9] = uigetfile('*','WEDGERED');

data9 = importdata([path9,filesep,file9]);

fx1_2 = data9.data(:,26);

fy1_1 = data9.data(:,27);

fz1_1 = data9.data(:,28);

cop1_x_9 = data9.data(:,32);

cop1_y_9 = data9.data(:,33);

figure(9);

plot(cop1_x_9,cop1_y_9);

title('Wedge | Red Brace - Stabilogram');

xlabel('Medio-Lateral (m)');

ylabel('Anterior-Posterior (m)');

grid on;

axis equal;

saveas(gcf, fullfile(stabilogram_folder, 'stabilogram_WEDGERED.png'));

%% All 9

x_data = {cop1_x_1, cop1_x_2, cop1_x_3, cop1_x_4, cop1_x_5, cop1_x_6, cop1_x_7,
cop1_x_8, cop1_x_9};

y_data = {cop1_y_1, cop1_y_2, cop1_y_3, cop1_y_4, cop1_y_5, cop1_y_6, cop1_y_7,
cop1_y_8, cop1_y_9};

N1 =  normalize(cop1_x_1);
```

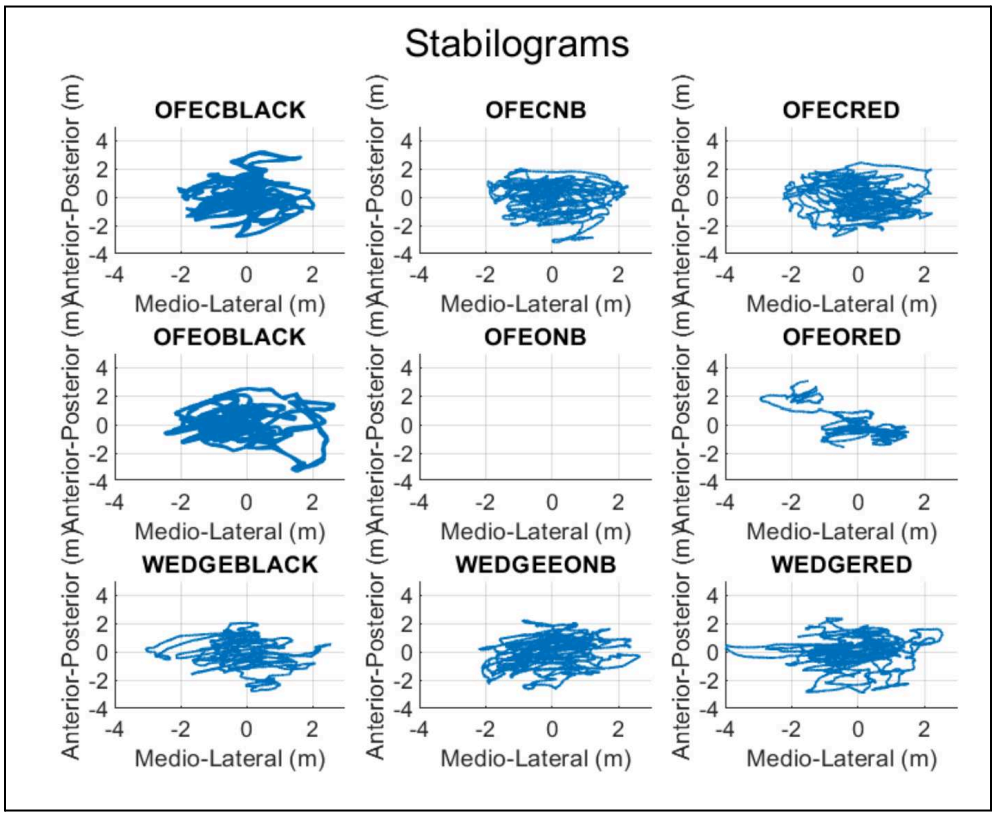
```
N2 =  normalize(cop1_x_2);
N3 =  normalize(cop1_x_3);
N4 =  normalize(cop1_x_4);
N5 =  normalize(cop1_x_5);
N6 =  normalize(cop1_x_6);
N7 =  normalize(cop1_x_7);
N8 =  normalize(cop1_x_8);
N9 =  normalize(cop1_x_9);
M1 =  normalize(cop1_y_1);
M2 =  normalize(cop1_y_2);
M3 =  normalize(cop1_y_3);
M4 =  normalize(cop1_y_4);
M5 =  normalize(cop1_y_5);
M6 =  normalize(cop1_y_6);
M7 =  normalize(cop1_y_7);
M8 =  normalize(cop1_y_8);
M9 =  normalize(cop1_y_9);
x_stand = {N1, N2, N3, N4, N5, N6, N7, N8, N9};
y_stand = {M1, M2, M3, M4, M5, M6, M7, M8, M9};
labels = {
    'OFECBLACK',
    'OFECNB',
    'OFECRED',
```

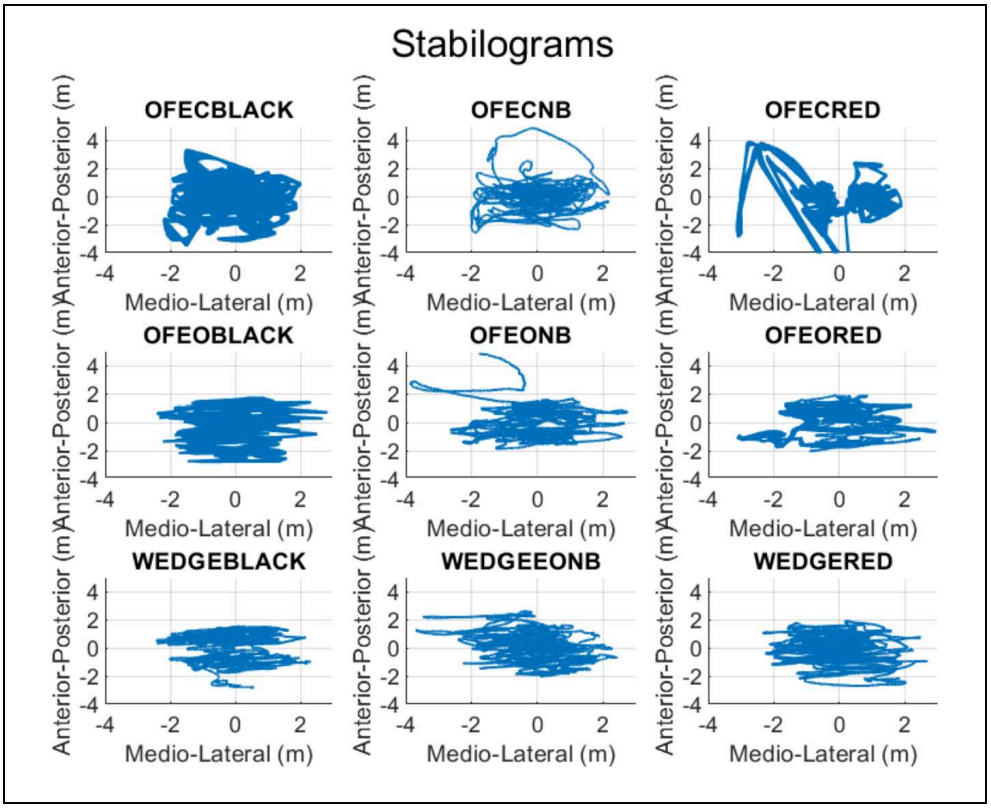
```
    'OFEOBLACK',
    'OFEONB',
    'OFEORED',
    'WEDGEBLACK',
    'WEDGEEONB',
    'WEDGERED',
};
figure(10);
for i = 1:9
    subplot(3, 3, i);
    hold on;
    plot(x_stand{i}, y_stand{i}, 'LineWidth', 1);
    xlabel('Medio-Lateral (m)');
    ylabel('Anterior-Posterior (m)');
    grid on;
    title(labels{i});
    xlim([-4,3]);
    ylim([-4,5]);
end
sgtitle('Stabilograms');
saveas(gcf, fullfile(stabilogram_folder, 'stabilograms_all.png'));
%% Path Lengths
path_lengths = zeros(1, 9);
```

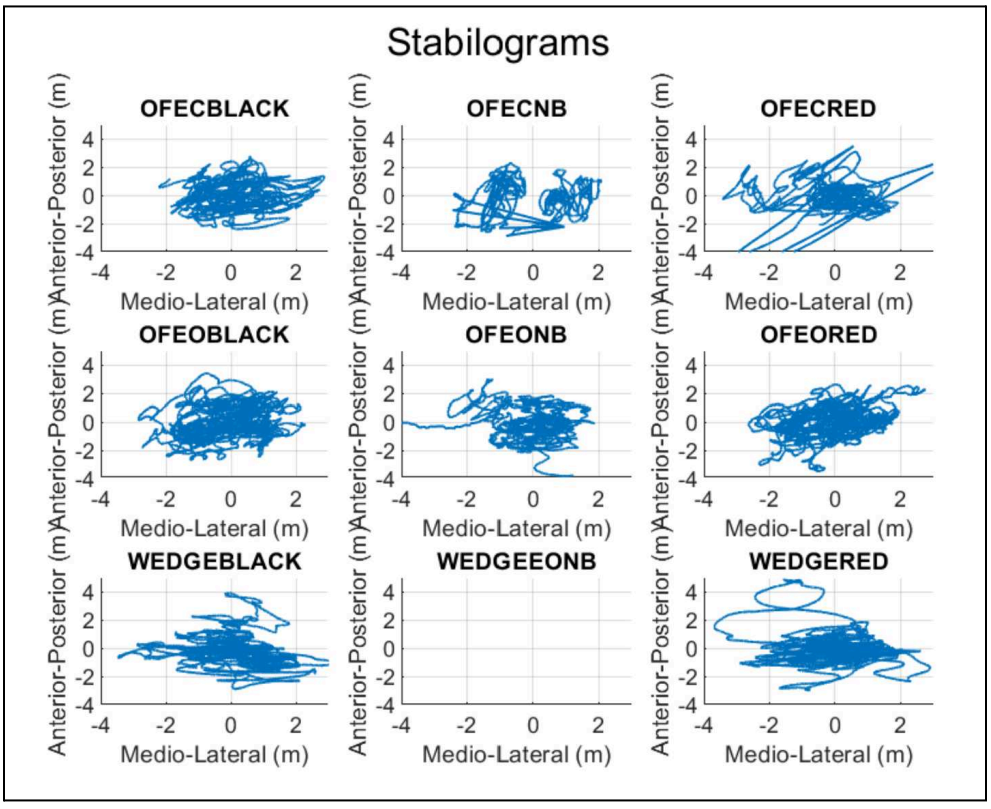
```
x_data = {cop1_x_1, cop1_x_2, cop1_x_3, cop1_x_4, cop1_x_5, cop1_x_6, cop1_x_7,
cop1_x_8, cop1_x_9};

y_data = {cop1_y_1, cop1_y_2, cop1_y_3, cop1_y_4, cop1_y_5, cop1_y_6, cop1_y_7,
cop1_y_8, cop1_y_9};

for i = 1:9

    dx = diff(x_data{i});

    dy = diff(y_data{i});

    path_lengths(i) = sum(sqrt(dx.^2 + dy.^2));

end

path_lengths = path_lengths(:);

T = table(labels, path_lengths, 'VariableNames', {'Condition', 'PathLength (m)'});

disp(T);
```

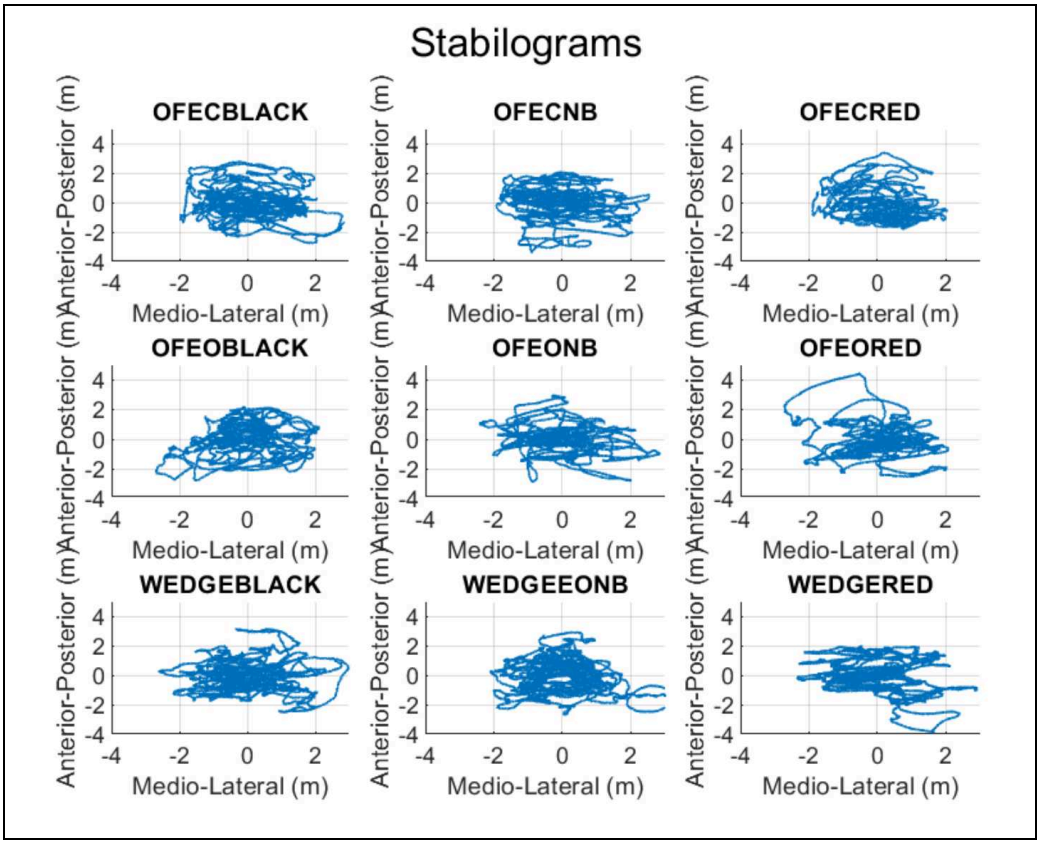
Appendix H: Force Plate Testing Stabilograms






Stabilograms
OFECBLACK
OFECNB
OFECRED
OFEOBLACK
OFEONB
OFEORED
WEDGEBLACK
WEDGEEONB
WEDGERED
Anterior-Posterior (m)
Medio-Lateral (m)


Stabilograms
OFECBLACK
OFECNB
OFECRED
OFEOBLACK
OFEONB
OFEORED
WEDGEBLACK
WEDGEEONB
WEDGERED
Anterior-Posterior (m)
Medio-Lateral (m)


Stabilograms
OFECBLACK
OFECNB
OFECRED
OFEOBLACK
OFEONB
OFEORED
WEDGEBLACK
WEDGEEONB
WEDGERED
Anterior-Posterior (m)
Medio-Lateral (m)
4
2
0
-2
-4

Appendix I: Blank Comfort Testing Form

Short Outside and Long Inside (Red Brace)

Ease of Putting On:

1 2 3 4 5 6 7 8 9 10

Notes: -

Strap Comfort:

1 2 3 4 5 6 7 8 9 10

Notes:

Foam Comfort:

1 2 3 4 5 6 7 8 9 10

Notes:

Outside Support Fit:

1 2 3 4 5 6 7 8 9 10

Notes:

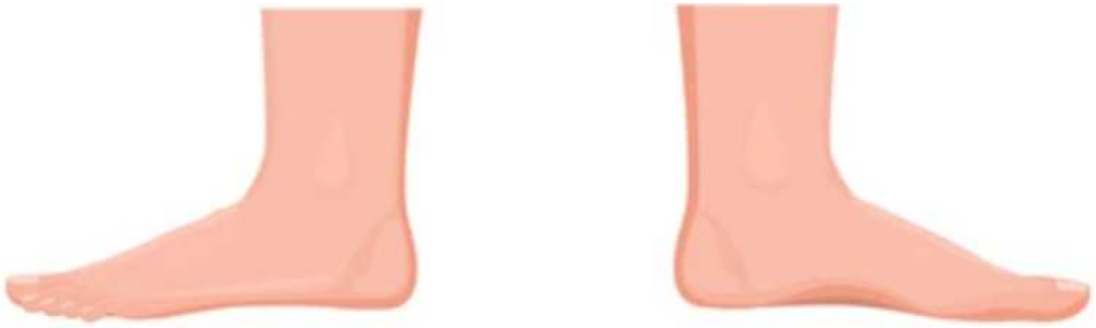
Inside Support Fit:

1 2 3 4 5 6 7 8 9 10

Notes:

Please circle on the foot where you feel discomfort:

Outside Inside



Anterior to medial malleolus

Long Outside and Short Inside (Black Brace)

Ease of Putting On:

(hard)1 2 3 4 5 6 7 8 9 10 (easy)

Notes:

Strap Comfort:

1 2 3 4 5 6 7 8 9 10 (most comfort)

Notes:

Foam Comfort:

1 2 3 4 5 6 7 8 9 10

Notes:

Outside Support Fit:

1 2 3 4 5 6 7 8 9 10

Notes:

Inside Support Fit:

1 2 3 4 5 6 7 8 9 10

Notes:

Please circle on the foot where you feel discomfort:

Outside

Inside

Anterior to medial malleolus

Appendix J: Completed Comfort Testing Form

Short Outside and Long Inside (Red Brace)

Ease of Putting On:

1 2 3 4 5 6 7 8 9 10

Notes: - see notes for black. same

Strap Comfort:

1 2 3 4 5 6 7 8 9 10

Notes:

Foam Comfort:

1 2 3 4 5 6 7 8 9 10

Notes:

Outside Support Fit:

1 2 3 4 5 6 7 8 9 10

Notes:

Inside Support Fit:

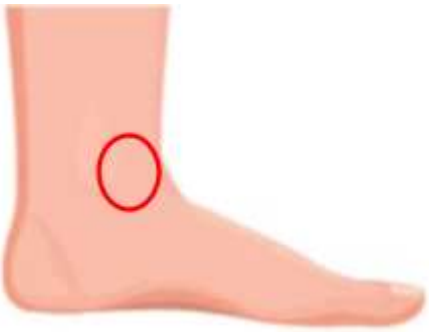
1 2 3 4 5 6 7 8 9 10

Notes: Plates are pretty easy. Compression sleeve is pretty hard. (In comparison, her SMO is 9 and AFO is 9.5 - much easier.) Hard to deal with 2 straps and to get shoe on over 2 straps.

Please circle on the foot where you feel discomfort:

Outside

Inside



Anterior to medial malleolus

Long Outside and Short Inside (Black Brace)

Ease of Putting On:

(hard)1 2 3 4 5 6 7 8 9 10 (easy)

Notes: Plates are pretty easy. Compression sleeves are pretty hard. (In comparison, her SMO is 9 and AFO is 9.5 - much easier.) Hard to deal with 2 straps and to get shoes on over 2 straps.

Strap Comfort:

1 2 3 4 5 6 7 8 9 10 (most comfort)

Notes:

Foam Comfort:

1 2 3 4 5 6 7 8 9 10

Notes:

Outside Support Fit:

1 2 3 4 5 6 7 8 9 10

Notes:

Inside Support Fit:

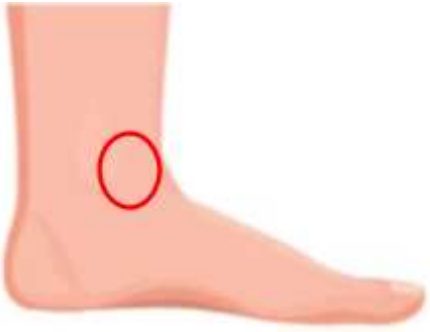
1 2 3 4 5 6 7 8 9 10

Notes: Slips which makes it uncomfortable.

Please circle on the foot where you feel discomfort:

Outside

Inside



Anterior to medial malleolus

Appendix K: Optical Markerless Motion Capture Testing Procedure via OpenCap

OpenCap is a smartphone application and a web application that enables cloud computing. To collect data, users open an application on two or more iOS devices and pair them with the OpenCap web application (on a computer). The web application enables users to record videos simultaneously on the iOS devices and to visualize the resulting 3-dimensional (3D) kinematics. In the cloud, 2D keypoints are extracted from multi-view videos using open-source pose estimation algorithms. The videos are time synchronized using cross-correlations of keypoint velocities, and 3D keypoints are computed by triangulating these synchronized 2D keypoints. These 3D keypoints are converted into a more comprehensive 3D anatomical marker set using a recurrent neural network (LSTM) trained on a large motion capture dataset. 3D kinematics are then computed from marker trajectories using inverse kinematics and a musculoskeletal model with biomechanical constraints. Finally, dynamic measures are estimated using muscle-driven dynamic simulations that track 3D kinematics.

Equipment:

- Computer
- 2 Apple IOS devices (iPad and/or iPhone)
- 2 Tripods (or chairs to set up the cameras)
- Printed checkerboard calibration target

I. Data Acquisition

For recommended best practices, additional tips, and set up videos please visit this link

Tips for smooth data collection:

1) Have a stable internet connection.
2) All devices should be connected to the same WIFI network.
3) Do not move the two cameras once the calibration is completed and the reference frame is set.
4) Note that the checkerboard calibration target is only used for calibrating the cameras. The checkerboard can remain on the wall of

II. Equipment set up

a. Take the checkerboard calibration target and hang it in the area where you expect to do the focused data collection.

b. On the **iOs device** you intend to do the data collection on, install the OpenCap App from the App store or use this link.

c. Position tripods and cameras (phones/iPads) in the orientation shown in Figure 1.

d. Go to the OpenCap's web app on a **laptop** (any operating system), create a login and follow the instructions on the web app for the calibration and data collection.

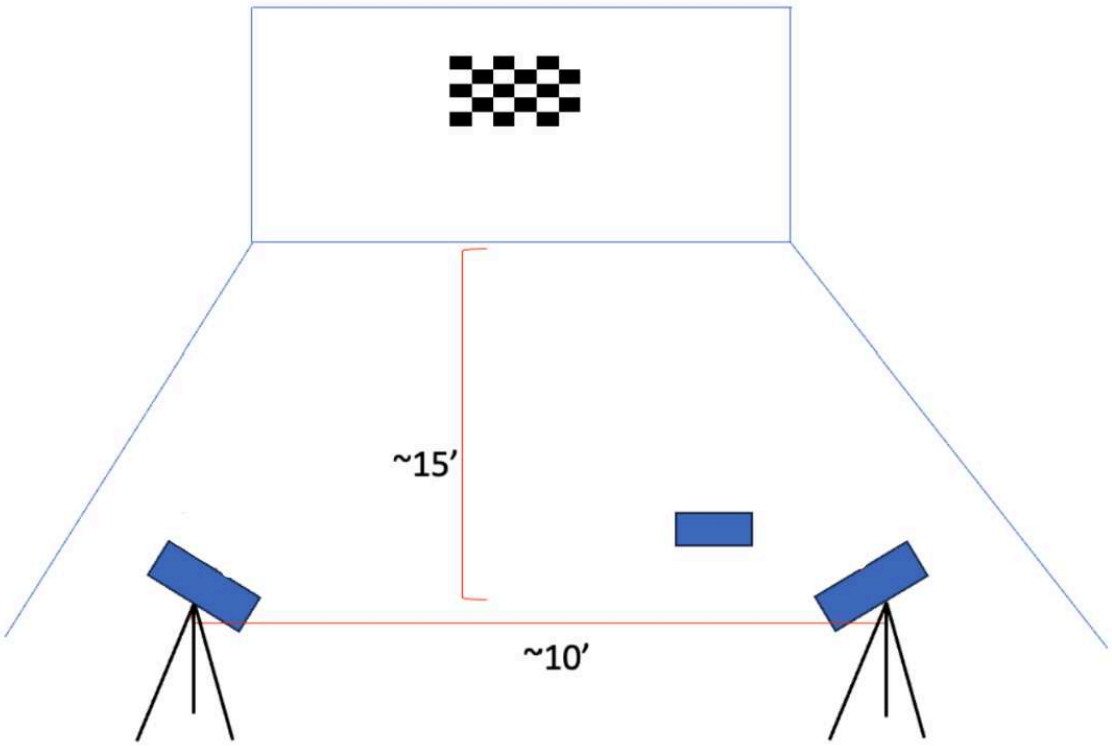


Figure 1. Set up recommended for the positioning of the cameras relative to the area of data collection (rectangle containing the checkered calibration square).

III. Data Collection

a. Log in to OpenCap on your computer. It should provide a QR code that needs to be scanned by both of your data collection devices (in the OpenCap mobile app). Scan the code with both, then follow the instructions to lead you to the point of data collection.
    i. NOTE: A red error message will appear if the devices are not properly connected or the checkered calibration sheet is not in frame. If this occurs, rescan the QR codes and adjust camera angles. Both devices should be pointed in the same direction. If the issue persists, attempt using a different IOS device for data collection.
    ii. Do not touch devices once calibrated.
b. After calibration, you will see this screen below. You'll need to enter a new subject, including weight in kg and height in meters.

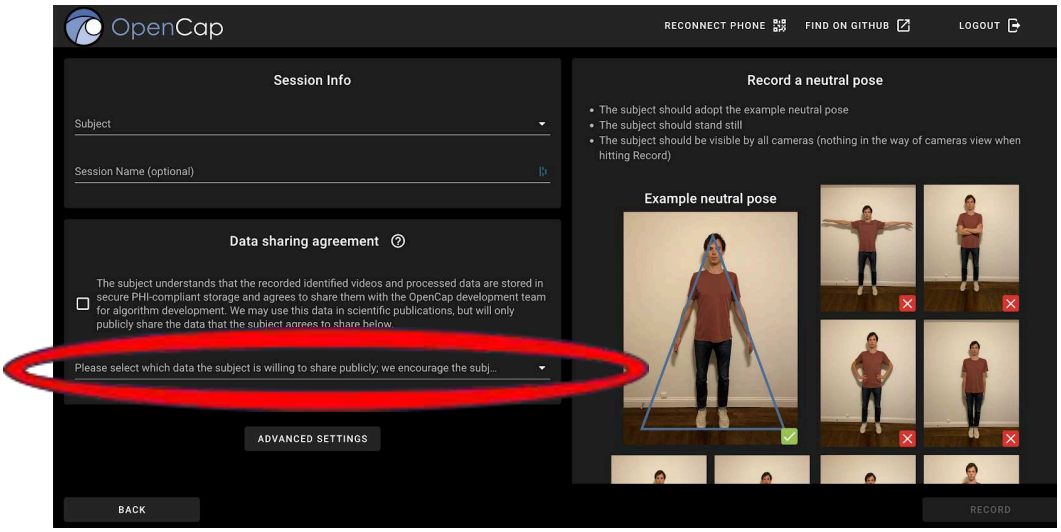
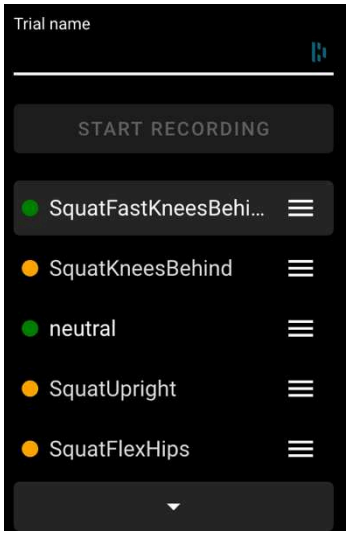


Figure 2. Initial screen prompting user for subject information and providing prompts for the data calibration.

c. After entering subject information, read and check the box under Data Sharing Agreement. Then select in the dropdown below it (circled) “Share No Data”. Name your session something descriptive- include your subject’s name, for example “Maggie_No_AFO” Once all of the fields are filled out on this page, you can move on to the recording screen. Click the button in the bottom right corner- **RECORD** (this actually will not start the recording yet).
d. On the next screen, name the file before starting to record (File Names Listed Below). You will record one file for each movement.
e. Press record.
f. Wait 5 seconds before walking.
g. Walk 5 steps.
h. Stop recording.
    i. NOTE: Keep the videos as short as possible- they can take quite a while to process.
    ii. Processing time: approximately 2 minutes
i. Your recording will show up under the START RECORDING button. A green dot means it has been processed and is ready to download for analysis in



OpenSim. Orange means it is still processing. Red means it has failed and the video should be re-taken.

i. Once green, click on the title of the trial and ensure the skeleton model resembles recorded movement.

j. Repeat action for all conditions listed in the table below for 3 trials. There should be **24 total tests**.

k. Download your data once all movement files show a green dot (this can sometimes take several minutes). Click on **DOWNLOAD DATA** and follow the prompts to save your files on your computer.

i. Do not unzip files

ii. Drag and drop all files into the Data folder within the OpenCap folder in the Google Drive

Perform the following 8 Testing Conditions, repeat each condition 3 times (stop data collection between each trial):

Table 1: Test Motion, Conditions, and corresponding file names

Test Motion	Condition	File Name
Walk 5 steps toward cameras	No AFO, With Shoes	No_AFO_Toward
Walk 5 steps along wall	No AFO, With Shoes	No_AFO_Along
Walk 5 steps toward cameras	Maggie's AFO, With Shoes	Maggie_AFO_Toward
Walk 5 steps along wall	Maggie's AFO, With Shoes	Maggie_AFO_Along
Walk 5 steps toward cameras	Black Prototype, With Shoes	Black_AFO_Toward
Walk 5 steps along wall	Black Prototype, With Shoes	Black_AFP_Along
Walk 5 steps toward cameras	Red Prototype, With Shoes	Red_AFO_Toward
Walk 5 steps along wall	Red Prototype, With Shoes	Red_AFO_Along

Appendix L: Optical Markerless Motion Capture Data

Extracted data from MATLAB processing including maximum knee angle, knee angle standard deviation, maximum and minimum hip angles, and hip angle standard deviations. The number of consistent data points and additional notes regarding testing inconsistencies were recorded. Specific trials are marked as good (green) or bad (red) data collections. The most consistent data collection trials were used in MATLAB graph data representation.

Client Data:

Trial	Number of (consistent) KNEE Peaks	Average Max Knee Angle (Degrees)	Knee Standard Deviation	Number of (consistent) HIP Peaks	Average Max Hip Angle (Degrees)	Average Min Hip Angle (Degrees)	Hip Standard Deviation Max	Hip Standard Deviation Min	Notes
NB_foward 1	1 (averaged 2)	53.7	11.71	(2 max, 1 min)	32.49	-19.73	5.67	0	
NB_foward 2	1 (averaged 2)	54.62	14.45	(2 max, 1 min)	30.85	-18.03	3.96	0	
NB_foward 3	1 (averaged 2)	53.29	18.06	(2 max, 1 min)	32.42	-17.68	5.45	0	Labeled as No_AFO_Toward
NB_side 1	2 (averaged 3)	66	10.25	(2 max, 2 min)	33.07	-9.42	0.41	0.83	
NB_side 2	2	74.1	3.3	(2 max, 2 min)	30.41	-11.02	1.86	1.49	
NB_side 3	2	76.01	6.32	(3 max, 2 min)	34.7	-10.85	4.47	0.74	Labeled as No_AFO_Away
AFO_foward 1	1	68.31	0	(2 max, 1 min)	27.07	-21.16	7.51	0	
AFO_foward 2	1	67.09	0	(2 max, 1 min)	29.27	-16.45	6.44	0	
AFO_foward 3	1	70.6	0	(2 max, 1 min)	32.16	-16.28	4.09	0	
AFO_side 1	2	71.14	0.67	(2 max, 2 min)	28.25	-11.53	4.75	3.41	
AFO_side 2									Bad data
AFO_side 3	2	69.78	6.44	(2 max, 2 min)	28.39	-10.58	3.51	1.86	
RB_foward 1	1 (averaged 2)	50.02	15.46	(2 max, 1 min)	32.86	-18.2	2.11	0	Number labeled 3
RB_foward 2	1 (averaged 2)	51.55	13.3	(2 max, 1 min)	32.7	-14.74	3.2	0	Number labeled 4
RB_foward 3	3	41.93	3.87	(1 max, 2 min)	48.03	0.96	0	2.48	Number labeled 5
RB_side 1	2	72.12	4.13	(2 max, 2 min)	30.66	-13.14	0.52	1.34	Number labeled 3
RB_side 2	2	68.73	5.95	(2 max, 2 min)	34.18	-11.37	5.2	0	Number labeled 4
RB_side 3	2	72	5.62	(3 max, 2 min)	32.89	-12.77	2.23		Number labeled 5

Healthy Subject: Team Member Data:

Trial	Number of (consistent) KNEE Peaks	Average Max Knee Angle (Degrees)	Knee Standard Deviation	Number of (consistent) HIP Peaks	Average Max Hip Angle (Degrees)	Average Min Hip Angle (Degrees)	Hip Standard Deviation Max	Hip Standard Deviation Min	Notes
NB_foward 1	1	53.7	16.62	1	25.58	-19.8	2.23	0	
NB_foward 2	1	54.52	9.1	1	21.55	-20.85	10.16	0	
NB_foward 3	1	55.34	7.08	1	25.23	-21.9	6.94	0	
NB_side 1	2	62.8	0.29	2	22.6	-25.4	0.74	3.72	
NB_side 2	2	75.04	3.3	2	27.28	-26.58	0.29	3.32	
NB_side 3	2	58.32	1.31	1	24.96	-22.43	0	1.85	Odd dip in first hip peak
BB_foward 1	1	53.8	12.39	1	26.63	-19.8	0	0	
BB_foward 2	1	56.05	14.16	2	18.75	-19.4	3.57	0	
BB_foward 3	1	62.8	0	1	23.69	-19.38	7.08	0	
BB_side 1	2	60.4	3.27	3	23.47	-19.03	2.61	0.07	
BB_side 2	2	64.01	0.29	3	19.67	-25.16	3.13	0.5	
BB_side 3	2	61.76	2.31	2	24	-22.03	0.58	0.39	
RB_foward 1	2	53.39	9.34	1	23.56	-18.92	0	0	
RB_foward 2	2	48.14	18.1	1	27.27	-19.97	0	0	
RB_foward 3	2	52.04	14.35	1	26.79	-20.96	0	0	
RB_side 1	2	59.58	2.31	2	22.86	-19.93	1.72	6.53	
RB_side 2	2	62.71	1.54	2	24.23	-23.12	10.43	1.73	
RB_side 3	2	55.56	3.05	3	26.72	-19.58	5.3	0.58	

Appendix M: Sample MATLAB Code for Motion Capture Plots

```
clc,clear;
%% Code for each
% BLACK BRACE:
data_anglesBB = readtable('BBSidewalk1mot.xlsx', MissingRule="omitrow");
data_anglesBB.Properties.VariableNames = ["time", "pelvis_tilt", "pelvis_list",
"pelvis_rotation", "pelvis_tx", "pelvis_ty", "pelvis_tz", "hip_flexion_r",
"hip_adduction_r", "hip_rotation_r", "knee_angle_r", "ankle_angle_r",
"subtalar_angle_r", "mtp_angle_r", "hip_flexion_l", "hip_adduction_l",
"hip_rotation_l", "knee_angle_l", "ankle_angle_l", "subtalar_angle_l",
"mtp_angle_l", "lumbar_extension", "lumbar_bending","lumbar_rotation",
"arm_flex_r", "arm_add_r", "arm_rot_r", "elbow_flex_r", "pro_sup_r",
"arm_flex_l", "arm_add_l", "arm_rot_l", "elbow_flex_l", "pro_sup_l"];
% figure;
% plot(data_anglesBB.time, data_anglesBB.knee_angle_r, 'b-', 'LineWidth', 1.5);
% xlabel('Time (s)');
% ylabel('Knee Flexion Angle (degrees)');
% grid on;

% RED BRACE (shortest file - everything has to line up to this one)
data_anglesRB = readtable('RBSidewalk3.mot_EXCEL.xlsx', MissingRule="omitrow");
data_anglesRB.Properties.VariableNames = ["time", "pelvis_tilt", "pelvis_list",
"pelvis_rotation", "pelvis_tx", "pelvis_ty", "pelvis_tz", "hip_flexion_r",
"hip_adduction_r", "hip_rotation_r", "knee_angle_r", "ankle_angle_r",
"subtalar_angle_r", "mtp_angle_r", "hip_flexion_l", "hip_adduction_l",
"hip_rotation_l", "knee_angle_l", "ankle_angle_l", "subtalar_angle_l",
"mtp_angle_l", "lumbar_extension", "lumbar_bending", "lumbar_rotation",
"arm_flex_r", "arm_add_r", "arm_rot_r", "elbow_flex_r", "pro_sup_r",
"arm_flex_l", "arm_add_l", "arm_rot_l", "elbow_flex_l", "pro_sup_l"];
%figure;
% plot(data_anglesRB.time, data_anglesRB.knee_angle_r, 'b-', 'LineWidth', 1.5);
% xlabel('Time (s)');
% ylabel('Knee Flexion Angle (degrees)');
% grid on;

% NO BRACE
data_anglesNB = readtable('NBSidewalk1mot.xlsx', MissingRule="omitrow");
data_anglesNB.Properties.VariableNames = ["time", "pelvis_tilt", "pelvis_list",
"pelvis_rotation", "pelvis_tx", "pelvis_ty", "pelvis_tz", "hip_flexion_r",
"hip_adduction_r", "hip_rotation_r", "knee_angle_r", "ankle_angle_r",
"subtalar_angle_r", "mtp_angle_r", "hip_flexion_l", "hip_adduction_l",
"hip_rotation_l", "knee_angle_l", "ankle_angle_l", "subtalar_angle_l",
"mtp_angle_l", "lumbar_extension", "lumbar_bending", "lumbar_rotation",
"arm_flex_r", "arm_add_r", "arm_rot_r", "elbow_flex_r", "pro_sup_r",
"arm_flex_l", "arm_add_l", "arm_rot_l", "elbow_flex_l", "pro_sup_l"];
% figure;
% plot(data_anglesNB.time, data_anglesNB.knee_angle_r, 'b-', 'LineWidth', 1.5);
% xlabel('Time (s)');
```

```
% ylabel('Knee Flexion Angle (degrees)');
% grid on;
%% Offset - match up all 3 data sets to the same point
% Shortest: set offset to 1
% Others: (set offset to the length of its signal - shortest signal length) + 1
BBoffset = 127;
RBoffset = 1;   % set offset for shortest signal to 1, adjust the rest roughly
by its signal length - the shortest signal length + 1
NBoffset = 187;
time = data_anglesNB.time; % taking the longest sample and temporarily storing
time?
% Going from offset to the height (bottom of columns) & taking this for all
columns:
data_anglesBB = data_anglesBB(BBoffset:height(data_anglesBB), :);
data_anglesRB = data_anglesRB(RBoffset:height(data_anglesRB), :);
data_anglesNB = data_anglesNB(NBoffset:height(data_anglesNB), :);

%% Plotting:
figure(1);
hold on;
% plot(time, knee angle data)
% dimensions need to match
% this won't run unless all the dimensions match
% need to see how long arrays are  and plot the time indices from (1:number of
rows)
plot(time(1:135), data_anglesBB.knee_angle_r, 'k-', 'LineWidth', 1.5);
plot(time(1:139), data_anglesRB.knee_angle_r, 'r-', 'LineWidth', 1.5);
plot(time(1:156), data_anglesNB.knee_angle_r, 'b-', 'LineWidth', 1.5);
xlabel('Time (seconds)');
ylabel('Right Knee Angle (degrees)');
title('Team Testing: Right Knee Angle During Gait Cycle');
xlim([0 2.5]);
legend('Black AFO','Red AFO', 'No AFO', 'Location', 'northwest')
grid on;
hold off;
```