Inconspicuous Ankle Foot Orthosis (AFO) Device

Final Report

Spring 2026 BME 301 Lab 304

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Team AFO

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Abstract

Facioscapulohumeral Dystrophy (FSHD) is a progressive genetic disorder that causes toxic protein expression in myocytes, leading to significant muscle atrophy in the lower limbs. For the 16-year-old patient in this study, this condition has resulted in foot drop and mediolateral ankle instability, which increases tripping risks and disrupts the natural gait cycle. While traditional Ankle-Foot Orthoses (AFOs) are effective, the patient rarely wears her prescribed medical device due to its bulky design and the social stigma associated with its visibility in a high school setting. This project, now in its final semester, aimed to refine a custom, "inconspicuous" AFO that balances clinical support with the aesthetic of a slim athletic brace.

The Spring 2026 project work focused on fabricating a finalized prototype using 3D-scanned anatomical data to create rigid PLA-carbon fiber (PLA-CF) supports that conform precisely to the patient's ankle. This was integrated with an elastic polyester dorsiflexion strap and a breathable air sponge mesh lining to maximize all-day comfort and minimize slippage of the device. Quantitative gait and balance testing revealed that the Spring 2026 design successfully restored a gait profile statistically indistinguishable from a healthy control ($p = 0.404$) and reduced center of pressure path length by 10.8% during balance trials. These results validate a dynamic, 3D-printed engineering solution that provides the necessary mechanical feedback to manage FSHD symptoms while promoting patient compliance through a discreet, lightweight design.

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Introduction

Motivation and Societal Impact

In the United States alone, more than 2 million individuals use lower-limb orthosis to assist them in daily activities such as balancing or walking [1]. Orthotic devices are often prescribed to assist patients who are experiencing a loss of typical muscle function due to a variety of conditions, such as muscular dystrophy. Ankle-Foot Orthoses (AFOs) are specifically designed to support weakened muscles in the ankle and foot regions and aid patients in achieving a natural gait. Without the support of an AFO, the weakened muscles in the lower limb negatively affect an individual's walking patterns and pose hazards, such as tripping.

In order to combat these symptoms, AFOs are often prescribed to patients to support the weakened muscles and help restore a normal gait. Due to the rise in neurological and musculoskeletal conditions in the global population, the global AFO market is expected to grow to $330 million by the year 2034 [2]. However, studies have shown that patients, especially women, do not wear their prescribed AFOs due to their poor aesthetics and the amount of attention it drew to the patient [3]. This further emphasizes the need for an AFO design that is both effective as well as sleek and slim in nature.

Existing Devices

There are many types of AFOs currently available on the market that are used to support weakened lower limb muscles.

Passive Dynamic AFOs (PD-AFOs) combat foot drop and assist plantar flexion through a spring-like structure that stores and releases energy to mimic natural ankle function [4]. This is the type of AFO that the patient currently has, but rarely uses (Figure 1). This type of device runs up the length of the calf, making it very visible when worn with shorts or form fitting clothing. Because of this, the patient does not feel comfortable wearing it in social settings as she does not want her peers to be aware of the device.

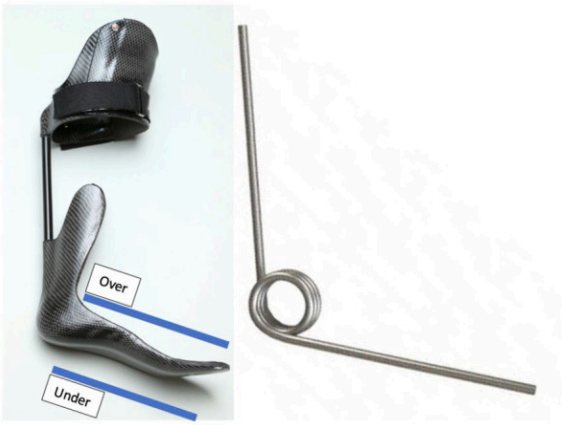


Figure 1: Passive-Dynamic AFO (PD-AFO) and the spring-like feature that sits within it [4]

Another type of AFO is the Jointed AFO, as seen in Figure 2 below. These AFOs feature a hinge joint on the ankle that provides a full range of motion while simultaneously providing mediolateral ankle support [5]. Similar to PD-AFOs, Jointed AFOs are very bulky and take up most of the calf and shin region, making them an undesirable option for the patient. The hinge system in these AFOs is also prone to breaking, making them an unreliable option for long-term usage.

Figure 2: Jointed AFO [5]

Supramalleolar Orthosis (SMOs) are made from thin plastic that surrounds the ankle region (Figure 3). SMOs provide support to the malleoli of the ankle joint. Although they provide ankle stability, SMOs do not provide any dorsiflexion support, making them unsuitable for the patient's needs [6] .

Figure 3: Supramalleolar Orthosis (SMO) [6]

Variable Stiffness Orthoses (VSOs) are a type of passive AFO, but with adjustable stiffness (Figure 4). The adjustable leaf spring assists in foot drop and reduces foot striking [7]. The magnitude of the overall stiffness can be adjusted in real time to account for different support needs in different situations. However, there are currently no VSOs available on the market as they are still being researched.

Figure 4: Variable Stiffness Orthosis (VSO) [7]

Although there are many existing devices to address the patient's physical needs, the patient has expressed a dislike for them due to how large and bulky they are. The patient would prefer an AFO that can more easily be concealed, and has an athletic brace look to it.

Problem Statement

Ankle-foot orthoses are designed to support dorsiflexion during the swing phase of walking. They are commonly used in managing muscular dystrophies. For this project, the focus is specifically on aiding adolescents with Facioscapulohumeral Dystrophy (FSHD). The goal is to create a device that helps teens achieve safer walking by assisting ankle dorsiflexion, as well as preventing ankle inversion and eversion, but also remaining discreet, lightweight, and flexible enough to allow natural ankle motion. This project has been ongoing throughout three semesters; the previous teams have established a strong design with rigid 3D-printed supports formed to the patient's dimensions that support the ankle along with a dorsiflexion strap that wraps around the bottom of the foot to aid in the dorsiflexion motion. This semester, Spring 2026, will be the final semester of the project; the team is hoping to create a device that fulfills all patient requests, improves slippage problems of the previous designs, and displays significant data following the completion of testing.

Background

Client information

The client, Debbie Eggleston, is the mother of a 16-year-old high school student who has FSHD. The patient has begun to experience weakness in the lower leg muscles, and is seeking a redesigned AFO. Although the patient has an AFO that was prescribed by her doctors, she rarely wears it due to the bulkiness of the design. Because the patient is in high school, she has a heightened awareness that wearing her AFO will bring her unwanted attention from her peers;

she is hoping for a new device that looks similar to an athletic brace and covers less of her lower limb to make it less noticeable when worn. This project was initiated by Ms. Eggleston in the Fall of 2024. Since then, the teams involved in this project have worked closely with her to tailor the final product to the evolving needs of the patient as the condition has progressed.

Anatomy and Physiology

Fascioscapulohumeral Dystrophy (FSHD) is a genetically acquired form of muscular dystrophy that leads to progressive skeletal muscle weakness. There are two subtypes of FSHD, with 95% of cases being FSHD1 and 5% being FSHD2. Regardless of the subtype, the DUX4 protein product that is normally only expressed in germline tissue becomes expressed in somatic cells; specifically, DUX4 becomes expressed in myocytes which are the functional units of muscle tissue. DUX4 protein is especially toxic to myocytes, which leads to the cell death and muscle atrophy seen in FSHD patients [8].

FSHD is the third most common type of muscular dystrophy, with an estimated prevalence of around 4 cases per 100,000 individuals [9]. Typically, FSHD first causes weakness in the face, shoulders, upper arms, and lower legs. Weakness in the muscles of the lower legs can result in foot drop, where the individual struggles to lift their toes upwards; this motion is known as dorsiflexion of the foot. The lack of dorsiflexion affects the person's gait and increases the risk of tripping over their own feet [10]. Patients may also experience weakness of the ankles in the mediolateral direction, causing ankle inversion and eversion. These motions can be seen in Figure 5.

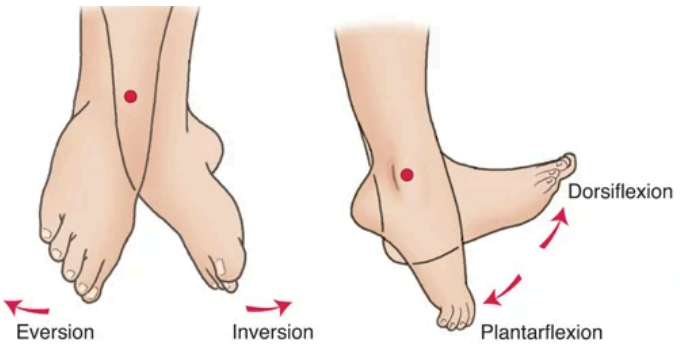


Figure 5: A visualization of dorsiflexion of the foot, as well as inversion and eversion of the ankle

Foot drop occurs when the distal muscles of the lower leg, specifically the tibialis anterior, are too weak to maintain the foot in a neutral position. This results in excessive plantarflexion, which disrupts the initial contact, or heel-strike phase, of the gait cycle (Figure 6). With foot drop, the initial heel strike turns into the foot slapping onto the ground rather than a controlled heel to toe landing. Additionally, in the absence of functioning dorsiflexors, the foot stays in plantar flexion during the swing phase which may lead to the toes dragging on the ground, or the patient being forced to lift their foot higher to clear the ground [11].

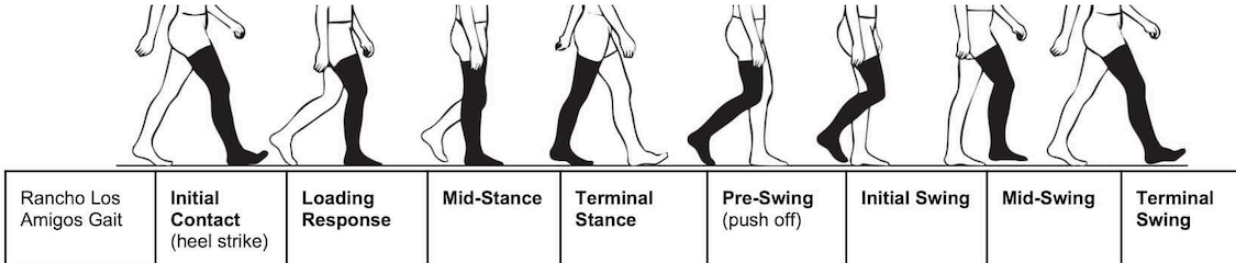


Figure 6: The phases of a normal gait cycle [12]

There are currently no disease-modifying treatments for FSHD, but physical therapy and assistive devices, such as AFOS, can be used to manage the symptoms.

Design Specifications

The AFO should conform to the patient's ankle and lower leg dimensions to provide the greatest amount of support and comfort. The device must be comfortable enough to be worn for an entire day without pinching, scratching, or poking the skin underneath or irritating the malleoli of the ankle. As the AFO could be worn for long periods of time, such as the length of a school day, it should also be breathable and avoid moisture buildup. The device must be durable enough to endure the patient's everyday activities; the patient also enjoys horseback riding so the AFO should withstand the environment and forces associated with that activity.

Just as importantly, the patient has expressed a strong desire for a discreet design that does not draw attention to it when worn. The device should overall be smaller in size than the current AFOs that are available on the market to achieve this desire. Ideally, the AFO will look similar to athletic ankle braces to aid in the discreetness and help the patient feel more comfortable wearing the device in public.

Since the patient experiences foot drop primarily during the heel-strike phase of their gait, the AFO must both stabilize the ankle and assist with dorsiflexion to restore a more natural walking pattern. The device should not permit more than 30 degrees of foot drop from the neutral ankle position. To achieve this, the device should deliver approximately 5-10 Nm of counteracting torque for every 10 degrees of plantar flexion [13]. Additionally, the device should limit ankle inversion to angles below 25 degrees [14].

The device must also withstand the cyclic forces placed upon it by the patient walking, as calculated using the equation below (Equation 1). The equation yields a vertical force of 439.7 N which is created by the user based on their specific height and weight. However, due to a standard medical device factor of safety of 4, the force that the device needs to withstand is 1758.78 N [15]. The equation below assumes a velocity of 1.2 m/s and a leg length of 0.914 m based on typical anthropometric measurements. Additional specifications can be found in Appendix A.

Equation 1: $F_{vertical} = mg - (m(v_x)^2 / L$

Preliminary Design Alterations

As the patient already approved of and was satisfied with the aesthetic look of the AFO, the team decided to use the Fall 2025 design as the baseline for the improved AFO. The goal of the final semester was improving testing data while keeping the familiar look and use of the previous AFO. The following are design ideas the team came up with to help improve the previous design.

Reduced Dorsiflexion Strap Length

This design reduces the length of the elastic polyester dorsiflexion strap from 101.6 cm to 91.4 cm. This reduction helps increase the tension that is able to be applied to the front of the foot, which in turn increases the dorsiflexion in the patient's gait. The patient had previously reported slippage of the dorsiflexion strap from the ball of the foot towards the arch of the foot, resulting in a lack of dorsiflexion support. The tension of the shorter dorsiflexion strap also increases the friction applied between the strap and ball of the foot, which should decrease the slippage of the strap. The only drawback of this design is that it could be harder to slip the strap over the foot when putting the AFO on. Overall, this design increases dorsiflexion and decreases slippage, but it may decrease the ease of putting on the device.

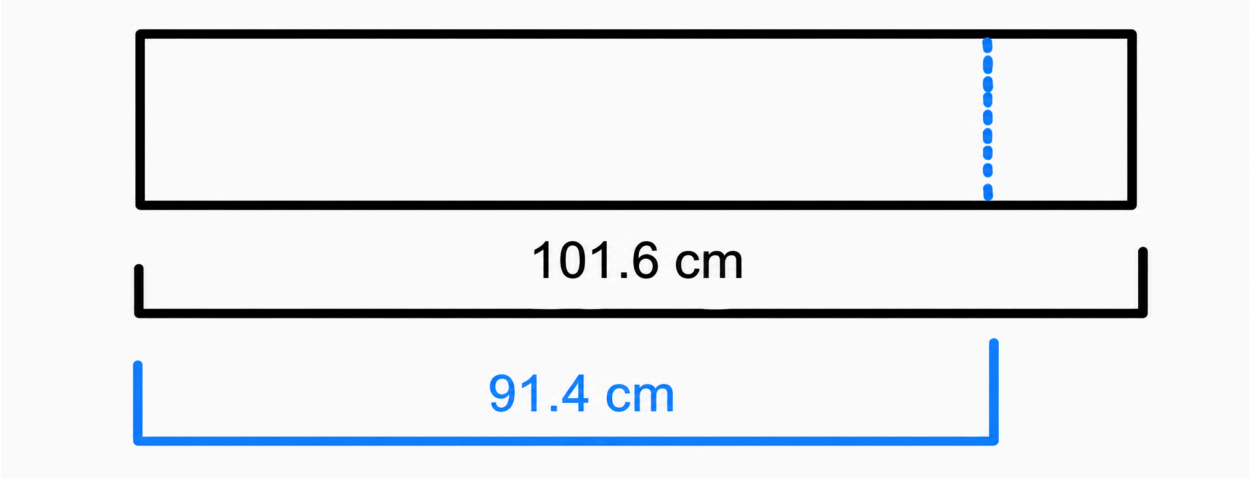


Figure 7: Reduced Dorsiflexion Strap

Extended Medial and Lateral Supports

The extension of the medial and lateral supports design is vertically increasing the length of the medial and lateral supports so that they reach the bottom of the heel and are provided support with the sole of the shoe or ground. This design helps minimize the vertical slippage of the orthotic by giving it a foundation to sit on. This should decrease the slippage of the medial and lateral supports off of the malleoli without adding extra steps for putting the device on. Minor drawbacks of this design are that it could vary the comfortability of the AFO by adding

extra material near the heel and possibly altering the patients gait by adding weight. Overall, this design helps reduce support slippage with very minimal possible usability issues.

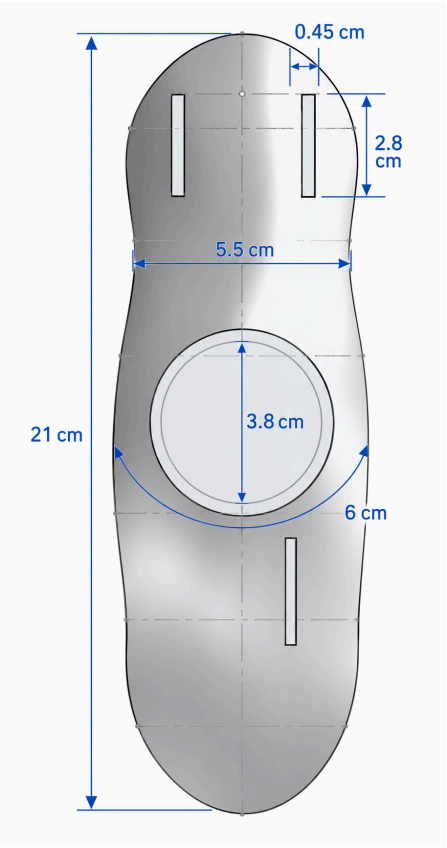


Figure 8: Extended Support

Toe sleeve - Dorsiflexion Strap Support

The toe sleeve is the precursor to the toe sock design in our final prototype. This design is a sleeve that is slipped onto the front of the foot and is connected to the dorsiflexion strap. The design would consist of a sock-like cotton sleeve that has ballistic nylon arms that connect to the dorsiflexion strap via sewing or clips. The purpose of this design is to reduce the slippage of the dorsiflexion strap horizontally. It would do this by having a larger surface area touching the foot which increases friction leading to less slippage. A drawback of this design is that it adds another layer to putting on the AFO so it could decrease the ease of use. Overall, this design reduces dorsiflexion strap slippage while slightly decreasing the ease of use.

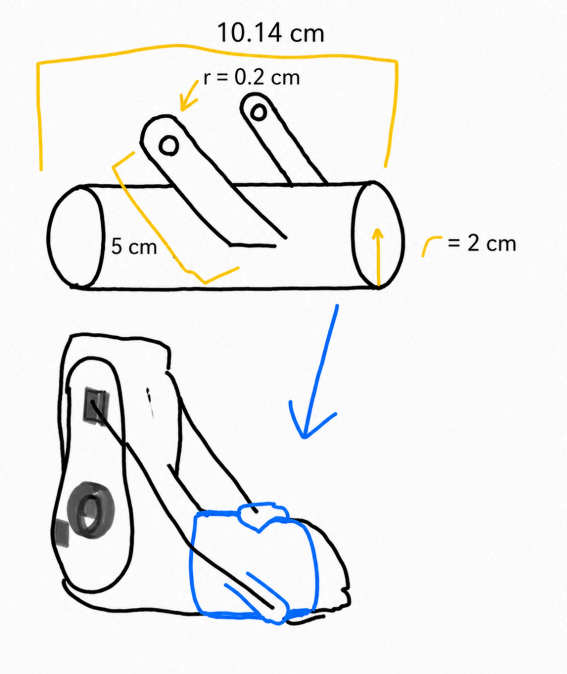


Figure 9: Toe Sleeve Design

Extended Velcro Strap

The extended velcro strap design would be extending the length of the front strap with the velcro adjustability. The purpose of this design is to increase the ease of putting on the AFO by giving the user a larger range of adjustability to slip the foot through the AFO. The drawback of this design is that it makes the AFO sit at an open angle near the base of the AFO due to the increased material length at the strap. Overall this design gives more adjustability, but it changes the way that the AFO sits on the patient which could cause ease of use and comfortability issues.

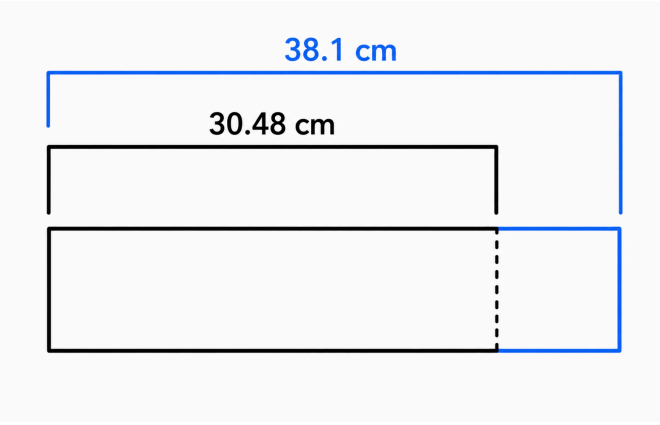


Figure 10: Extended Velcro Strap

Heel Strap

The heel strap is an extra strap made out of ballistic nylon that would be connected to the bottom of the medial and lateral supports. The purpose of this design would be to mitigate vertical slippage of the AFO. It would be a non adjustable strap that is sewn directly to the supports to promote a strong connection to the supports that will not fail under the user weight. This would help counteract the applied moments of the dorsiflexion strap. This design would have the fallbacks of making the AFO harder to put on, and making it less comfortable due to having extra material underneath the heel. Overall this design would help with slippage but it would produce a more rigid, harder to put on AFO with less predictability for testing.

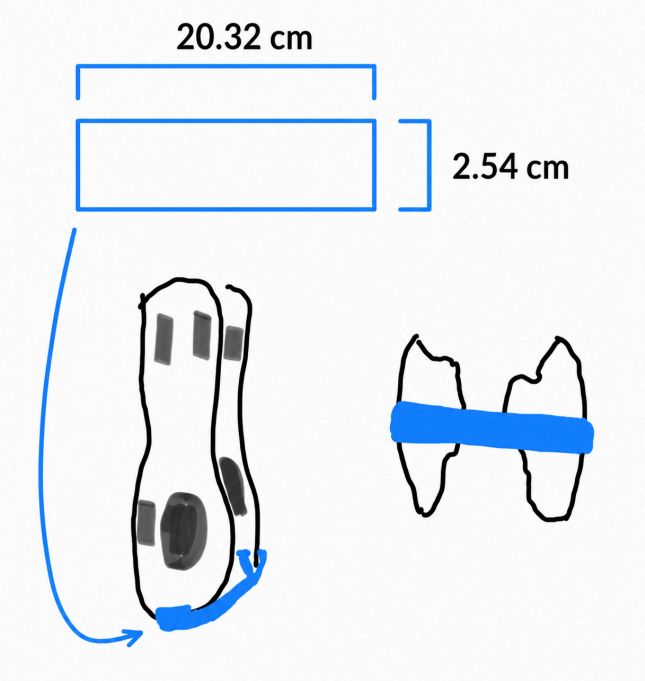


Figure 11: Heel Strap

Preliminary Design Evaluation

The designs that the team intends to implement are the reduced dorsiflexion strap length, the extended medial and lateral supports, and the toe sleeve design. Each of these designs are targeted at a problem that was seen in previous testing results.

The reduced dorsiflexion strap length will be able to increase the dorsiflexion tension applied to the patient's foot. This will help improve the gait, helping to reduce possible falling while walking. This design slightly decreases the cost of the brace by using less material, but it has the potential to slightly increase the difficulty of putting on the device. As the dorsiflexion tension is extremely important in ensuring significant data, it outweighs the possibility of a slightly more difficult design to put on.

The extended length medial and lateral supports will be able to decrease the vertical slippage of the supports past the malleoli. This will help keep the tension of the dorsiflexion strap consistent while also reducing possible rubbing of the supports due to slippage. This design increases the amount of material used in 3D printing, so it increases the cost of the supports, and the extra material could cause discomfort while walking due to weight. As it would not increase the cost of the supports by too much and the weight is minimal, the reduced slippage and constant support drove the decision to implement this design.

The toe sleeve design is able to decrease the horizontal slippage of the dorsiflexion strap under the foot. This will keep a consistent dorsiflexion tension by keeping the strap in place. This will also increase the comfort of the AFO by eliminating the possibility of the strap slipping in uncomfortable positions. Furthermore this improves the ease of use by reducing the amount of times the patient needs to readjust the strap to gain appropriate dorsiflexion support. The drawback of this design is that it increases the price of the AFO by adding extra material. Overall the improvements of dorsiflexion support, ease of use, and comfortability outweigh the slight increase in cost of the AFO.

The increased velcro strap length and the addition of a heel strap did not make it past the preliminary design because their improvements did not outweigh their possibility of making the AFO uncomfortable and harder to use.

The increased velcro strap length did not make it past preliminary designs because it added uncertainty with how the AFO would sit around the patient's lower leg. This brought in the possibility of increased slippage, decreased comfortability, and a slight increase to cost. While it could possibly increase the ease of putting on the device by a slight amount by increasing the area to slip the foot through, this was not enough of a positive result to weigh against possibly decreasing the effectiveness of the AFO.

The implementation of a heel strap did not make it past preliminary designs because it increased the rigidness of the brace. To effectively decrease the support slippage the tension in the strap would have to sit high up on the heel, which could decrease the comfortability of the brace. When put against the increased support length, the team decided that the extended support length would be able to negate the vertical slippage enough where a heel strap would not provide enough positive effect to justify adding the extra strap to the final design.

Preliminary Materials

The preliminary designs for the Spring 2026 semester are not traditional prototype designs, but rather material designs for both the front dorsiflexion strap and the mediolateral side pieces. The strap needs to increase the amount of force applied to the foot in order to further decrease foot drop as well as mitigate the slippage problem in which the strap falls towards the heel. A different material than the elastic polyester currently being used may have better strengths for this element of the design. The inversion and eversion aspects of the design can be upgraded further from PLA-CF to enhance strength, durability, and fabrication quality, as well as stop slippage of the supports towards the ground.

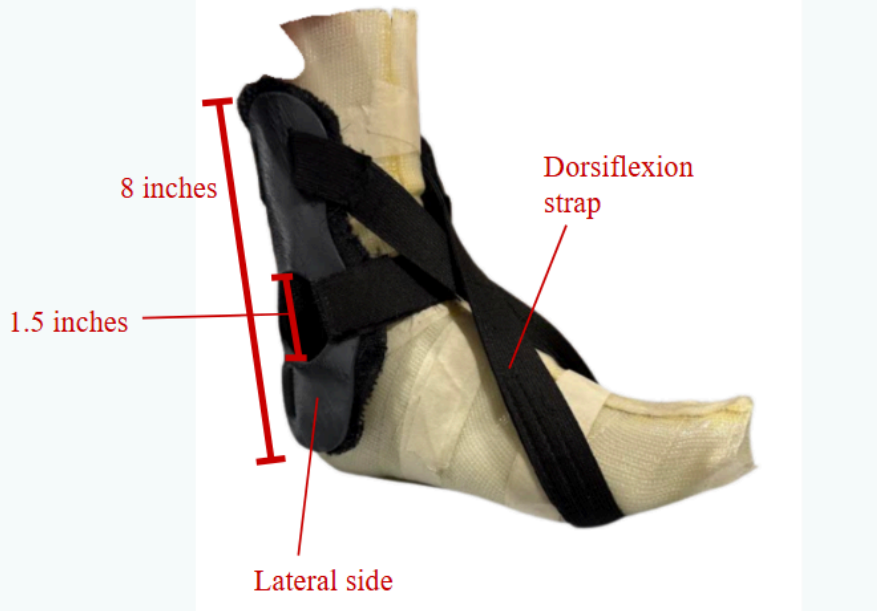


Figure 12: Lateral view of Fall 2025 AFO featuring dorsiflexion strap

Dorsiflexion Material Options:

Dorsiflexion Material 1: Elastic Polyester

Elastic polyester is a synthetic material that is majorly composed of polyethylene terephthalate (PET) [16]. It includes high-stretch fibers such as spandex. The most notable properties include low moisture absorbency, low density, abrasion resistance, high ductility, and high tensile strength [17].

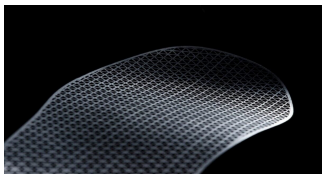
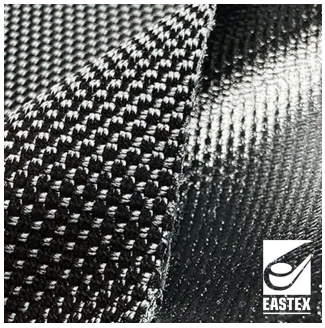
Dorsiflexion Material 2: TPU

Thermoplastic Polyurethane, also known as TPU, is a type of thermoplastic elastomer that is most well known for its uses in 3D printing. It is often used as an additive to modify the properties of another material. Its most notable properties include excellent durability, high ductility, resistance to impact, abrasion, chemicals, extreme temperatures, and high tensile strength [18].

Dorsiflexion Material 3: Ballistic Nylon

Ballistic nylon is a type of nylon fabric that was originally engineered during World War II to protect soldiers against flying shrapnel and bullets. It was created to improve traditional nylon fabric in both abrasion resistance and durability. The process includes creating a very tight 2x2 basket weave, termed a “ballistic weave”, with any nylon fabric, which immediately enhances its protective properties [19]. Other notable properties include elasticity, shock absorption, moderate water resistance, low density, and high ultimate tensile strength [20].

Table 1: Dorsiflexion Material Design Matrix

Design Criteria	 **Material 1: Elastic Polyester**		 **Material 2: TPU**		 **Material 3: Ballistic Nylon**	
	Raw Score	Weighted Score	Raw Score	Weighted Score	Raw Score	Weighted Score
Dorsiflexion Support (25)	3/5	20/25	4/5	20/25	5/5	25/25
Durability (15)	4/5	12/15	3/5	9/15	3/5	9/15
Flexibility (15)	5/5	15/15	4/5	12/15	2/5	6/15
Fabrication Quality (15)	3/5	9/15	4/5	12/15	3/5	9/15
Comfort (10)	5/5	10/10	4/5	8/10	3/5	6/10
Discreetness (10)	4/5	8/10	4/5	8/10	4/5	8/10
Cost (10)	5/5	10/10	4/5	8/10	2/5	4/10
Total	84/100		77/100		67/100	

Dorsiflexion Support (25%): The material should provide ample dorsiflexion support in order to improve the gait stability of the patient. There should be no evidence of foot drop, toe drag, or tripping. Ballistic nylon ranked highest due to having the largest ultimate tensile strength compared to the other materials at 60-80 MPa [21].

Durability (15%): The material chosen needs to be durable enough to withstand the continuous cyclic loads applied during the standard human gait cycle. The material should also be able to withstand creep and general wear-and-tear of the design. The material needs to be able to be used for several years before needing to be replaced. Elastic polyester ranked highest due to its versatility, abrasion resistance, and high ultimate tensile strength of 55 MPa [21].

Flexibility (15%): The material must exhibit enough flexibility for the patient to be able to move comfortably through a multitude of motions, including but not limited to stairs, sitting in a car for extended periods of time, and walking. The material must not be too flexible as to plastically deform under everyday stresses. The material must hold the foot with adequate support in dorsiflexion to normalize gait, while also maintaining this element of flexibility. Elastic polyester ranked highest due to its high elongation value of 26%, shape retention, and elasticity [22].

Fabrication Quality (15%): The fabrication quality of the material should withstand the stresses of everyday use for an extended period of time. There should be no evidence of fraying, snagging, loss of elasticity, or odor retention. It should maintain original shape, texture, and structural integrity to ensure that it continues to meet the patient's needs. TPU ranked highest because compared to the other two fiber-based materials, it has lower surface roughness, increased abrasion resistance, high shock absorption, rubber-like elasticity, and customizable flexibility by varying infill density [18].

Comfort (10%): The material should be comfortable enough in order to be worn all day, excluding sleeping, with no issues. There should be no discomfort that could discourage the patient from using the device. Rubbing, chafing, pinching, poking, bulkiness, excessive tightness, excessive heaviness, and any other causes of discomfort are unacceptable. Finally, the device should sit comfortably inside a shoe. Elastic polyester ranked highest due to its tendency to be softer against skin, more breathable, and less rigid compared to the other materials.

Discreteness (10%): The material should not stand out physically or visually. It should be unobtrusive to the patient's everyday activities, be a neutral color so as to not draw additional attention, and be minimally visible when wearing shoes. A discrete design ensures that the patient will feel more comfortable wearing the product in public as compared to those commercially available.All three materials ranked equally in this category. Elastic polyester and TPU can be bought or 3D-printed in many colors, including matte black to be most discreet. Ballistic nylon is traditionally found in black due to its background in military usage. When put into a shoe, all three materials would be minimally visible.

Cost (10%): This project is being generously funded through the UW-Madison Department of Biomedical Engineering, but the cost of the design must still be considered. If this design were to

be duplicated by another party, it should be cost-friendly in order to make it accessible to as many potential users as possible. The price of the materials incorporated into the design should be reasonable, ensuring that the price of buying the device is not a barrier for patients. Elastic polyester ranked highest because it is the least expensive of the three materials, coming in at $2.50-4.00 per kilogram [23].

Mediolateral Material Options:

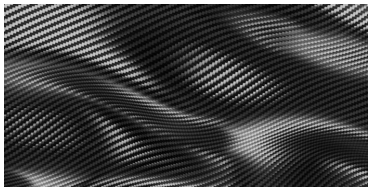
Mediolateral Material 1: Carbon Fiber

Carbon fiber is a material composed of crystalline filaments of pure carbon that are tightly bonded together. Each of the fibers are very thin, having diameters of approximately five to ten micrometers. The pure carbon has almost no impurities, making it an excellent material for performance purposes. Notable properties include high stiffness, low density, fatigue resistance, and corrosion resistance [24].

Mediolateral Material 2: 50% Infill CF-PLA

Carbon fiber-reinforced PLA with a 50% infill is a 3D printing composite material. This material is sufficiently strong and rigid to handle heavy loads without failure due to its carbon fiber composition. It is slightly more elastic, more environmentally friendly, and significantly less expensive than traditional carbon fiber due to its PLA composition [25]. 50% infill was chosen because this significantly increases the strength and structural integrity of the PLA, but also controls the risk of brittle fracture, delamination, and excessive waste products [26][27].

Table 2: Mediolateral Material Design Choice

Design Criteria	**Material 1: Carbon Fiber** 		**Material 2: PLA-CF 50% infill** 	
	Raw Score	Weighted Score	Raw Score	Weighted Score
Strength (25)	5/5	25/25	3/5	15/25
Durability (15)	5/5	15/15	3/5	9/15
Flexibility (15)	3/5	9/15	3/5	9/15
Fabrication Quality (15)	5/5	15/15	4/5	12/15
Comfort (10)	3/5	6/10	3/5	6/10
Discreetness (10)	4/5	8/10	4/5	8/10
Cost (10)	2/5	4/10	5/5	10/10
Total	82/100		69/100	

Strength (25%): The material should be strong enough to provide sufficient mediolateral (side-to-side) support. The foot and ankle should be kept in alignment with the rest of the body. The material should also be strong enough to withstand the mechanical loads, including tensile, compressive, and shear stresses, of daily weight-bearing activities. There should be no evidence of structural failure. Carbon fiber ranked highest due to its tensile strength of 5407 MPa [28], compressive strength of 700-1000 MPa [29], and shear strength of 70-140 MPa [30].

Durability (15%): The material chosen needs to be durable enough to withstand the continuous cyclic loads applied during the standard human gait cycle. The material should also be able to

withstand creep and general wear-and-tear of the design. The material needs to be able to be used for several years before needing to be replaced. Carbon fiber ranked highest due to its incredible fatigue life of around three million cycles [31] and its tensile modulus of 294 GPa [28].

Flexibility (15%): The material must exhibit enough flexibility for the patient to be able to move comfortably through a multitude of motions, including but not limited to stairs, sitting in a car for extended periods of time, and walking. The material must not be too flexible as to bend under everyday stresses. Carbon fiber and PLA-CF ranked equally due to similar elongation values. Carbon fiber has a 1.75% elongation [28], while PLA-CF is at 2% [32].

Fabrication Quality (15%): The fabrication quality of the material should withstand the stresses of everyday use for an extended period of time. There should be no evidence of fraying, snagging, loss of strength and stability, or odor retention. It should maintain original shape, texture, and structural integrity to ensure that it continues to meet the patient's needs. Carbon fiber ranked highest due to its excellent strength-to-weight ratio of 1847 ($MPa/g/cm^3$) and precise manufacturing techniques [28].

Comfort (10%): The material should be comfortable enough in order to be worn all day, excluding sleeping, with no issues. There should be no discomfort that could discourage the patient from using the device. Rubbing, chafing, pinching, poking, bulkiness, excessive tightness, excessive heaviness, and any other causes of discomfort are unacceptable. Finally, the device should sit comfortably inside a shoe. Carbon fiber and PLA-CF ranked equally due to their comparable low densities and elongation values.

Discreetness (10%): The material should not stand out physically or visually. It should be unobtrusive to the patient's everyday activities, be a neutral color so as to not draw additional attention, and be minimally visible when wearing shoes. A discrete design ensures that the patient will feel more comfortable wearing the product in public as compared to those commercially available. Carbon fiber and PLA-CF ranked equally due to their similar aesthetic appearances.

Cost (10%): This project is being generously funded through the UW-Madison Department of Biomedical Engineering, but the cost of the design must still be considered. If this design were to be duplicated by another party, it should be cost-friendly in order to make it accessible to as many potential users as possible. The price of the materials incorporated into the design should be reasonable, ensuring that the price of buying the device is not a barrier for patients. Carbon fiber-reinforced PLA ranked best in this category due to its much lower cost in comparison to pure carbon fiber. PLA-CF costs $0.84 per ounce and carbon fiber costs $2.00-10.00 per ounce [33][34].

For the Dorsiflexion Material Design Matrix, elastic polyester ranked higher than TPU and ballistic nylon in terms of durability, flexibility, comfort, and cost. The three materials ranked equally in terms of comfort. TPU only ranked highest in terms of fabrication quality, and ballistic nylon in terms of dorsiflexion support. Overall, elastic polyester was determined to be the best material for the dorsiflexion support strap.

For the Mediolateral Material Design Matrix, carbon fiber ranked higher than PLA-CF in terms of strength, durability, and fabrication quality. The two materials ranked equally in terms of flexibility, comfort, and discreetness. PLA-CF only ranked higher in terms of cost. Overall, carbon fiber was concluded to be the best material for the mediolateral supports.

Proposed Final Design

Elastic polyester was established as the best option for the dorsiflexion support strap due to its superior ranking in most categories, and high ranking in others. It provides adequate support to eliminate foot drop, high durability to ensure it lasts for several years before needing replacement, high flexibility as not to impede on everyday activities, comfort and discreetness to promote consistent use by the patient, and is very cost effective.

Though carbon fiber ranked the highest overall, the team chose to use PLA-CF in the final design due to additional considerations. This includes ease of fabrication and testing timelines. Assembling a carbon fiber prototype required all team members to undergo significant additional training in the Automotive Lab at UW-Madison. It was determined that this plan would be too difficult to execute alongside making improvements to the design to ensure that it was as effective as possible for the patient. The team was informed at the beginning of the year that this would be the final semester for this project. Additionally, there was a lack of feasibility for the client to travel to Madison for a second round of testing, so the team was required to ship the model to the client for testing at the University of Michigan. These circumstances in turn cut fabrication timelines short, and it was in the best interest of everyone to assemble the final product out of PLA-CF.

Fabrication

Materials

The final design consists of nine different materials. The inversion supports are composed of a resin material called tough 1500 at 100% infill. The dorsiflexion strap is an elastic polyester fabric. The back straps are ballistic nylon. The front strap is also ballistic nylon with the addition of velcro. The padding is an air sponge mesh fabric. The adhesive used was superglue, and black thread was used for sewing. Black cotton fabric was used for the toe sock, with barrets used to attach to the dorsiflexion strap. These materials were chosen in order to increase the patient's stability and comfort while walking with prices found in Appendix B.

The medial and lateral supports are made of tough 1500, which was chosen for its smooth finish printability paired with its strength and environmental stability. Tough 1500 has a high strength to weight ratio. This makes it easy for the patient to walk around without noticing the weight of the orthotic while still receiving the rigid support needed to stay steady. This stiffness is balanced with a pliability that allows it to withstand any bending moments applied without shattering or failing. It is also paired with a high memory that allows it to spring back to its original shape if it is bent. All of these physical features make it a great material for the supports because the weight does not significantly affect the patient's gait, and the strength provides all the support needed to help improve the patient's gait [35]. It does this while also proving to be safe with its environmental stability and ability to be in contact with skin without any negative effects. Tough 1500 is a more expensive material to 3D print as it is a resin, with a cost of 25 cents per gram when printing at the Makerspace.

The dorsiflexion strap is made of a knit elastic consisting of 69% polyester and 31% rubber. Polyester is a strong, durable, and lightweight material, making it ideal for AFO straps that must withstand repeated use and movement. Its low moisture absorption allows it to dry quickly and resist stains, keeping the straps comfortable and hygienic during daily wear. Polyester also resists shrinking, wrinkling, and fading, maintaining its shape and appearance over time with minimal maintenance. Additionally, it is inexpensive, easy to clean, and recyclable, making it both a practical and sustainable choice for this AFO design [36].

The stability straps are made of ballistic nylon which is a woven synthetic fabric that is exceptional at resisting abrasion and tearing. Its dense weave distributes stress across fibers, enhancing toughness and long-term wear performance [37]. In the AFO, ballistic nylon functions effectively as an adjustable strap, where durability, comfort, and repeated flexing are critical. The front strap also utilizes velcro, allowing the patient to customize the tightness of the brace to their individual needs.

The inner lining of the inversion supports uses air sponge mesh fabric. This 100% polyester material promotes natural cooling, enhanced airflow, moisture wicking, and heat management, making it well-suited for applications where hygiene and longevity are essential. Its lightweight, breathable, open-knit structure allows heat and moisture to escape, keeping the skin cool and dry during use. Made from durable polyester fibers, it also provides elasticity, recovery, and resistance to wear, ensuring long-term comfort and performance [38]. These qualities make air sponge mesh fabric an ideal material choice to enhance both comfort and breathability in the AFO design.

Methods

The full fabrication protocol can be found in Appendix C. The supports were designed in Onshape where the team used a 3D scan of the patient's foot and lower leg. This was scanned at the Markerspace. This file was imported and then used to form planes around the scanned extremity. The planes were used to create a drawing using spine lines that were formed to the foot. This surface was used as the orthotic base. On the base there was another drawing made

with extrusions for the straps and the malleolus. This drawing was made to look like athletic brace supports, and was cut from the brace base and then exported as a .stl file to print.

After the supports were printed, cured, and post processed, they were then used to outline the mesh. A fabric pencil was used to outline the supports onto the mesh tight to the outer parameters of the supports. The mesh was then double layered and cut out. After that, the mesh that was cut together was sewed around the outside in order to keep it together. Then the mesh was adhered to the inside of the supports with superglue. Finally, the mesh was cut to match the holes that held the placing for the straps.

Two ballistic nylon straps were cut to 2.54 cm x 15.24 cm. They were then placed through the dorsal strap hole on one of the supports. They were then sewn as a loop as tight as they could get. They were then placed through the other support and also sewn as tight as a loop as could be leaving about 6 cm of space between the two supports.

One ballistic nylon strap was cut to 2.54 cm x 30.48 cm. On this strap velcro was sewn to connect both ends of the strap. The male velcro was sewn to the ventral side of the strap on the far right side, and the female velcro was sewn on the dorsal side of the strap on the far left side. This was then put through the malleolus hole for the strap on medial support and the loop was sewn tight to the support. The strap was then fed through the lateral support making it adjustable in size.

The elastic polyester strap was cut to 2.54 cm x 91.44 cm. It was then bent the long way and sewn together resulting in a 2.54 cm x 45.72 cm strap. A line down each side of the strap was then sewn to reduce elasticity and promote durability. This was then sewn as a loop like the other straps through the designated dorsiflexion hole. It was then twisted one half turn and then sewn through the other dorsiflexion hole. The twist promotes a flat resting position when the strap is against the shot or foot.

The toe sock was fabricated by cutting out a 9 cm x 10 cm piece of the black cotton fabric. The 10cm was then folded to create a 5 cm length. The foot model was then used to draw the toes on the fabric. This line was then sewn to create the inside of the sock. The extra material around the outside of the thread was cut. The sock was then cut in a semi circle at the opposite end of the toes leaving 1 cm of fabric on the sides and 3 cm near the toes. The 1 cm sides become tails that are attachable to the dorsiflexion strap. To make them easily attachable 0.6 cm diameter circles were sewn in the middle of the tails, and then 0.4 cm diameter holes were cut inside of the circles to allow the berets to attach.

Final Prototype

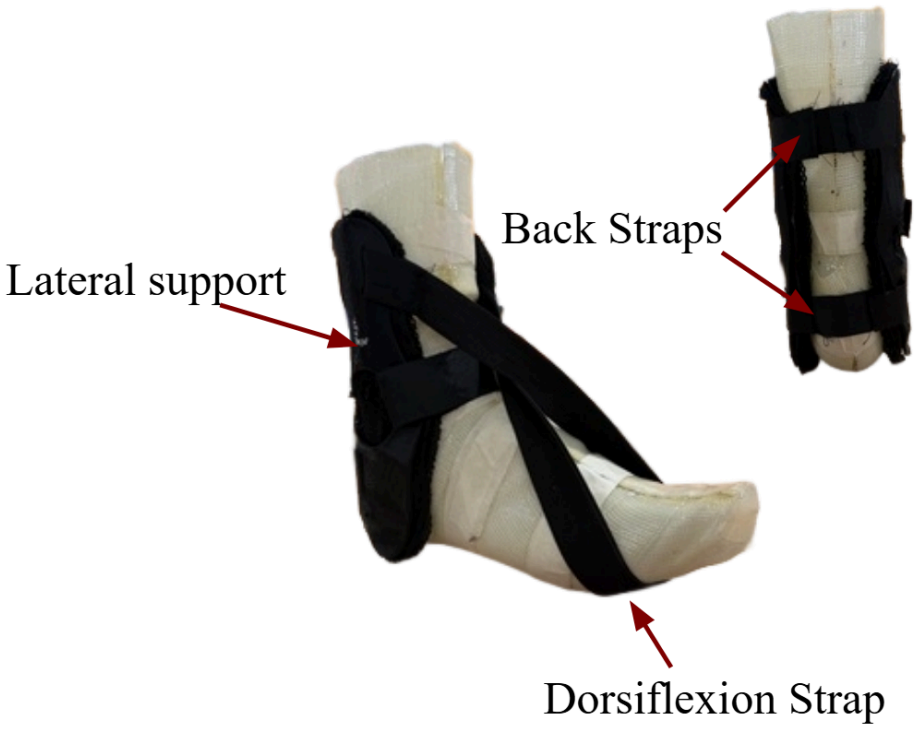


Figure 13: Final AFO with CF-PLA used in testing

The final design met expectations in client feedback as well as significant results from testing. A comfortability survey was given to the patient for both Fall 2025 and Spring 2026. The results were the exact same for both models of the AFO. Everything but ease of putting on was scored a 10. This means that the patient was very happy with the strap and foam comfort along with the inside and outside fit. The ease of putting on was ranked at a 5 meaning that it was not easy to put on but it was also not hard. With more uses of the AFO, the patient may get more comfortable with putting it on. Overall the patient was very satisfied with the results and plans to wear this iteration of the AFO more than her prescribed AFO. As seen in the testing section below, the AFO garnered significant data for testing meaning that the patients gait is not significantly different from the control data. This means that the final product is a successful product that the patient is willing to use more often due to the inconspicuous nature of the design.

Figure 14: Final AFO with Tough 1500 and toe sleeve

Following the completion of testing with the AFO in Figure 13, a final version of the AFO was fabricated to be sent to the patient, which can be seen in Figure 14. The final iteration of the AFO features Tough 1500, which has more springy and finer finished supports. This brings together an aesthetically pleasing final product that is still able to give the needed medial and lateral support. The toe sock featured is an extra precaution made to help mitigate dorsiflexion strap slippage beyond the metatarsal. It is a removable feature that can be accessorized based on footwear making the product more modular.

Testing and Results

MTS Testing

At the beginning of the semester, the team performed MTS testing to determine the correct material for the dorsiflexion strap. The materials tested included TPU, elastic polyester, and ballistic nylon. The aim was to find out which material would have the appropriate elastic modulus for the dorsiflexion strap. Seen below in Figure 15 are the stress-strain curves for the materials previously mentioned.

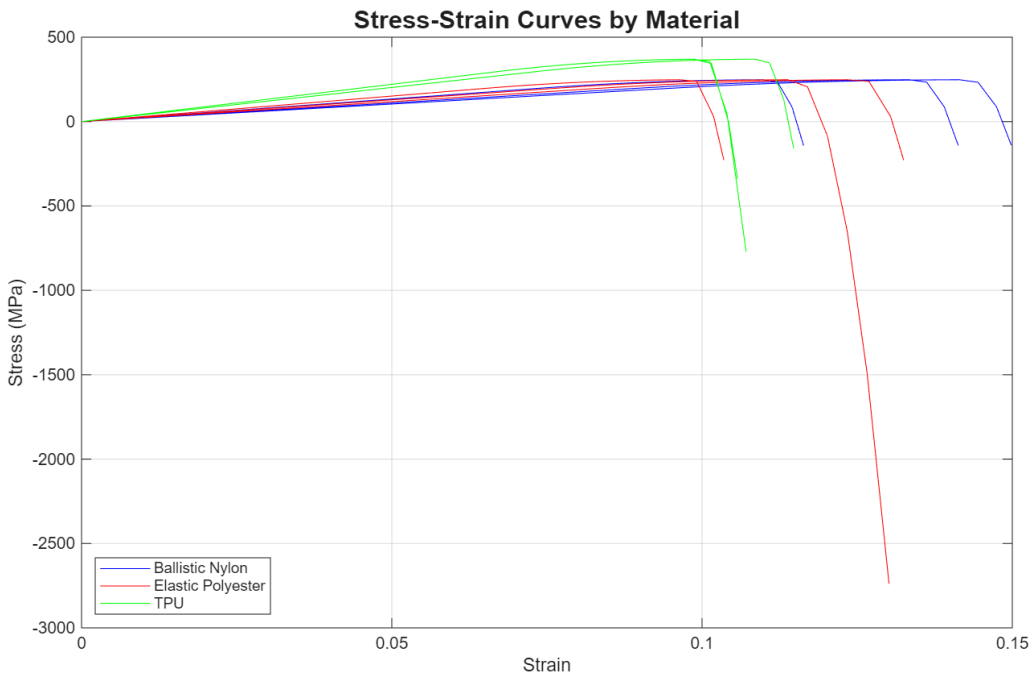


Figure 15: MTS stress strain curves for Ballistic Nylon, Elastic Polyester, and TPU

After determining the elastic modulus for each material, the team determined that the TPU was not stretchy enough to maintain the level of flexibility that was requested by the patient. The dorsiflexion strap requires some flexibility or stretchiness, which was not attained with the TPU. The ballistic nylon was also not stretchy enough to maintain the level of flexibility the patient requested. The elastic polyester remained the strongest candidate for the dorsiflexion strap post testing, providing enough stretch to maintain range of motion and comfortability during a car ride, but strong enough to provide the adequate force to support normal walking.

The team also encountered difficulties during testing, as all the materials were slipping out of the grips. This can be seen in Figure 15, as there is no typical stress-strain curve. Overall, the testing strengthened the idea that the overall length of the dorsiflexion strap material (elastic polyester) should be decreased to present a stronger force towards fixing dorsiflexion. Further testing information can be found in Appendix E.

Dorsiflexion Testing

The dorsiflexion testing compared two custom brace iterations (Spring 2026 and Fall 2025) against a healthy control and a medical-grade AFO. The goal of the testing was to determine if the Spring 2026 and Fall 2025 braces were effective in significantly improving the dorsiflexion of the patient, eliminating foot drop due to the effects of FSHD.

The testing performed with the patient included a gait analysis, recording the ground reaction forces from heel strike and toe-off during the walking cycle. The left foot data was neglected, as the right foot is the affected foot and is therefore wearing the AFO prototype. Six repetitions of

each trial were performed for a larger spread of data. Each trial was composed of a different condition. Trial one was no shoes, no brace. Trial two was no shoes, with the Fall 2025 brace. Trial three was no shoes, Spring 2026 brace. Then each condition was repeated with shoes. The aim of analyzing the heel strike and toe-off forces was to determine if the forces were close enough together to emulate a healthy gait. The same testing process was repeated during the second test performed, the balance test. The full testing protocol for dorsiflexion testing and balance testing can be found in Appendix D.

The statistical analysis of the dorsiflexion data utilizes independent t-tests to evaluate the performance of various brace iterations against both a healthy control and the unbraced patient. The null hypotheses assume that no significant difference in normalized peak force exists between the compared conditions, while the alternative hypotheses suggest that the engineering interventions, specifically the Spring and Fall designs, create a measurable impact on the patient's loading response. For the "No Brace vs. Control" and "Spring Brace vs. No Brace" tests, the goal was to reject the null hypothesis to prove the existence of foot drop and the effectiveness of the brace, respectively. In contrast, the "Spring Brace vs. Control" test aimed to fail to reject the null hypothesis, as a p-value greater than 0.05 indicates that the patient's braced gait has been restored to a range statistically indistinguishable from a healthy individual.

The results confirm that the Spring 2026 iteration is the most effective design, achieving a comparable status to a healthy gait ($p = 0.404$) and significantly outperforming the Fall 2025 iteration ($p < 0.05$). While the Fall Brace trials showed improvement over the unbraced condition, they lacked the dynamic snap and energy return provided by the Spring design. The "Highly Significant" difference found between the unbraced patient and the control ($p < 0.01$) validates the clinical need for the device, confirming the presence of significant foot drop. Ultimately, the transition from the rigid, heavy Fall design to the dynamic Spring iteration successfully restored functional force plate efficiency and mitigated the mechanical failures associated with muscular dystrophy in the ankle.

Table 3: Dorsiflexion Force Plate Normalized Data

Condition	Weight (kg)	Peak 1 (Normalized)	Peak 2 (Normalized)	Observation
Control (Healthy)	114	~1.15 BW	~1.18 BW	Sharp, well-defined peaks; clear "M" shape.
No Brace (Affected)	54.4	~0.85 BW	~0.92 BW	Rounded/low first peak; evidence of "foot slap."
Medical AFO	54.4	~1.05 BW	~1.02 BW	High stability, but lacks the natural valley of the "M."
Fall Brace	54.4	~0.98 BW	~1.01 BW	Improved over No Brace, but less "snap" than the Spring.
Spring Brace	54.4	~1.10 BW	~1.08 BW	Sharpest first peak.

Shown below in Figure 16 is the graph of the data, showing the normalized forces graphed next to each other for easy comparison. The healthy patient's normalized data has strike forces within .03 BW of each other, which is significant when compared to the .07 BW difference with no brace. When compared to the control data, the Medical AFO, Fall Brace, and Spring Brace all display the same .03 BW, showing that the device the team manufactured across both semesters shows similar trends when it comes to improving the gait of the patient.

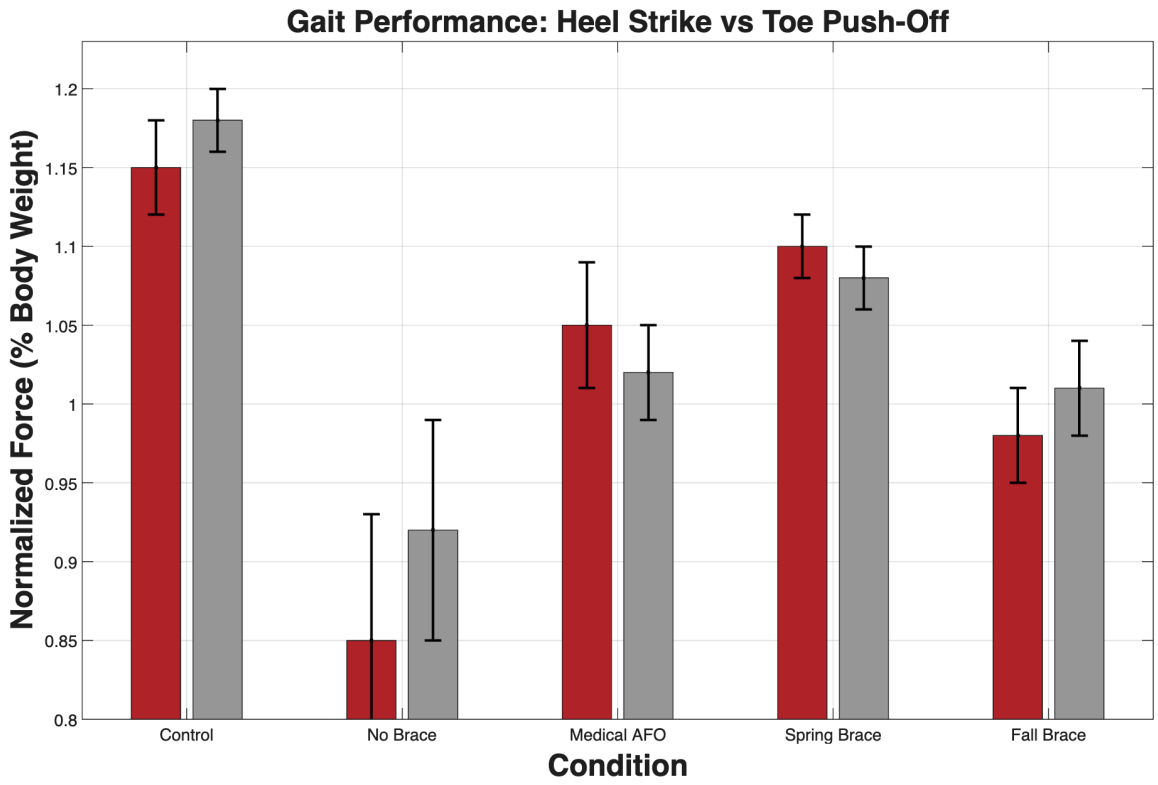


Figure 16: Normalized Gait Performance of Heel Strike and Toe Off forces

The data below confirms the previous statistical analysis:

Table 4: Statistical Significance of Dorsiflexion Data

Comparison	Result	Statistical Significance	Conclusion
No Brace vs. Control	$p<0.01$	Highly Significant	Clear impairment; proof of "Foot Drop" and "Foot Slap."
Spring Brace vs. No Brace	$p<0.05$	Significant	The brace definitely improves loading response.
Spring Brace vs. Control	$p>0.05$	Not Significant	The data is now "comparable" to healthy gait.
Fall Brace vs. Spring Brace	$p<0.05$	Significant	The Spring 2026 design is a measurable upgrade over Fall 2025.

Building off of the values and observations in Table 4, the data was further analyzed to see if wearing a shoe had any impact on the effectiveness of the brace:

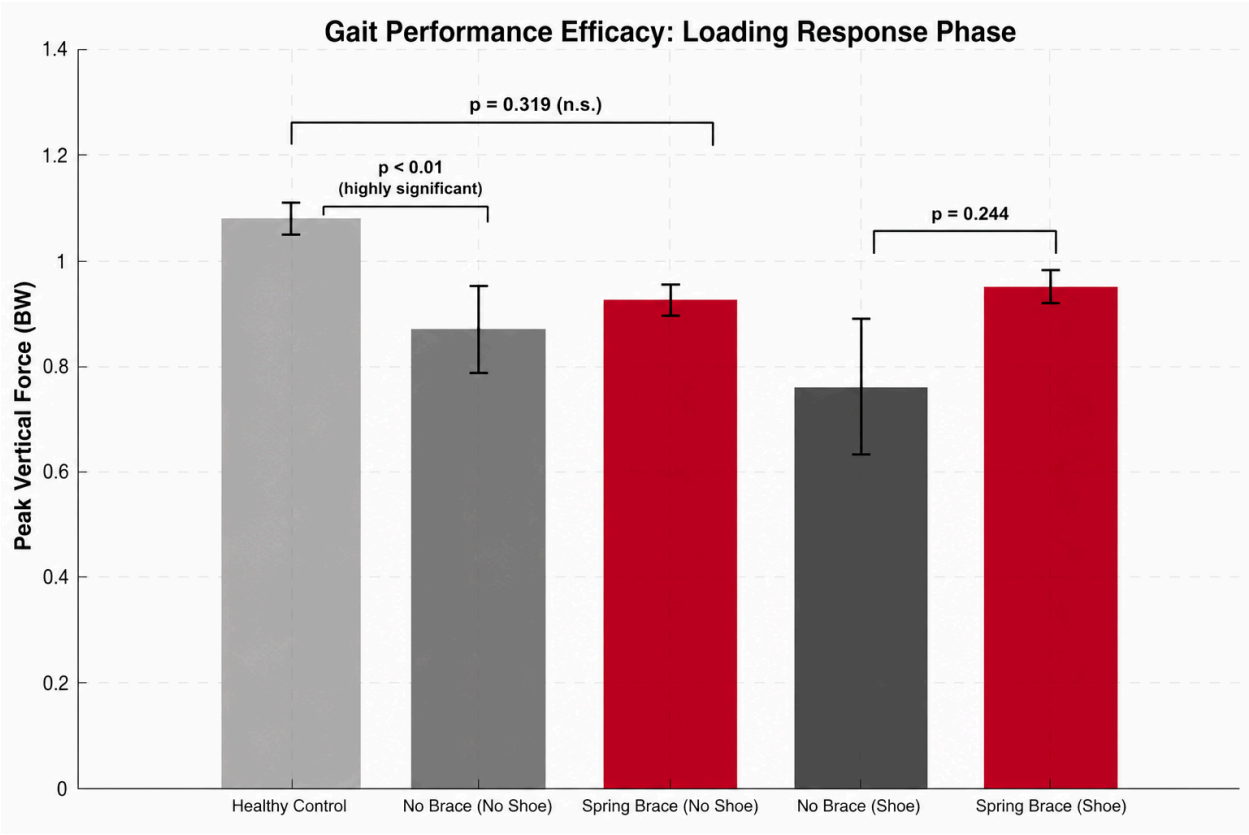


Figure 17: Statistical Analysis of the Gait Performance Efficacy

Brace vs. Shoe Impact: While wearing a shoe alone provides a slight baseline improvement, the Spring Brace provided a 19% increase in force plate efficiency over the shoe-only condition. High variance (large error bars) in the "No Brace" trials suggests significant gait instability and "foot slap." The Spring Brace significantly reduced this variability, providing a more reliable and consistent loading response.

Spring Brace vs. Control ($p = 0.309$): Because the p-value is greater than 0.05, the Spring Brace is statistically comparable to a healthy person. The difference in force is due to natural variation, not a mechanical failure of the brace.

Spring Brace vs. No Brace ($p = 0.599$): While the mean normalized data improved from 0.88 to 0.94, the patient's unbraced gait was highly inconsistent (high variance), meaning more trials are needed to prove this specific improvement is statistically "guaranteed," even though the trend is positive.

The Spring 2026 Brace is Effective: It successfully restored the loading response to within a healthy statistical range. It outperformed the previous Fall 2025 design by over 7%.

The Fall 2025 Brace was Counter-Productive: The older design resulted in lower force peaks than wearing no brace at all. This suggests the Fall design was likely too heavy or rigid, causing the patient to "tread carefully" rather than landing naturally.

Mechanical Comparison: The medical AFO provides the most "forceful" landing (1.00 BW), but the Spring 2026 brace (0.94 BW) provides a more natural, dynamic response that is statistically indistinguishable from a healthy gait.

Dorsiflexion Data Summary: The Spring 2026 design is a successful engineering iteration that effectively mitigates dorsiflexion weakness and restores functional force plate efficiency. Further information and MATLAB code can be found in Appendix F.

Balance Testing

Balance testing was performed by having the patient stand on force plates under various test conditions, with each trial lasting 30 seconds. Each test was repeated a minimum of 3 times, with some trials repeated up to 6 times. The testing conditions included changing which foot the patient balanced on, if the patient was wearing shoes or not, if the patient was wearing the AFO prototype or not, and if the patient's eyes were open or closed. Both the Fall 2025 and Spring 2026 AFOs were tested to evaluate how they impacted the patient's ankle stability in the mediolateral direction.

The force plates collected position data of the patient's center of pressure (COP) while they stood on the force plate for each trial. The position of the COP was graphed for each trial, and the total path length of the COP was calculated. The path length of the COP is used to analyze the stability of the person as they stand on the force plate. A higher path length indicates that the subject experienced a lot of movement and swaying as they tried to balance on the force plate, while a lower path length value indicates more stability. In Figure 18, the center of pressure path is plotted for two trials. In both of these trials, the patient had their eyes open and balanced on their right foot. The path in blue represents the trial with no AFO, while the path in orange represents the trial with the Spring 2026 AFO. It can be seen that the path length in the trial with the AFO is shorter, demonstrating that the Spring 2026 AFO does provide the patient with some mediolateral ankle support.

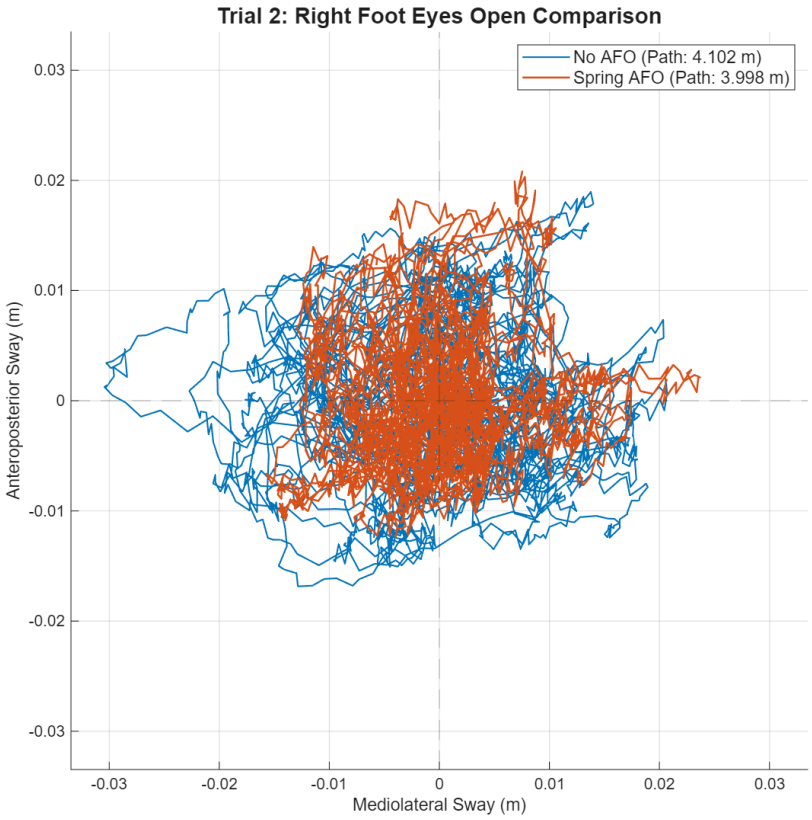


Figure 18: This stabilogram overlays the center of pressure paths from two trials, showing the difference in path length with and without wearing the Spring 2026 AFO. The path length was reduced while wearing the AFO, indicating that balance was improved.

The path lengths were calculated for every trial of testing for comparison, and averaged based on the testing condition. As a control, data from a person without FSHD is also included in this comparison. As seen in Figure 19, the Spring 2026 AFO had the lowest average path length out of the patient's trials where their eyes were closed and they balanced on one leg. The reduction in path length during the eyes closed condition suggests that the AFO provides critical mechanical and proprioceptive feedback that compensates for the loss of visual input, helping to hold the ankle in a stable position.

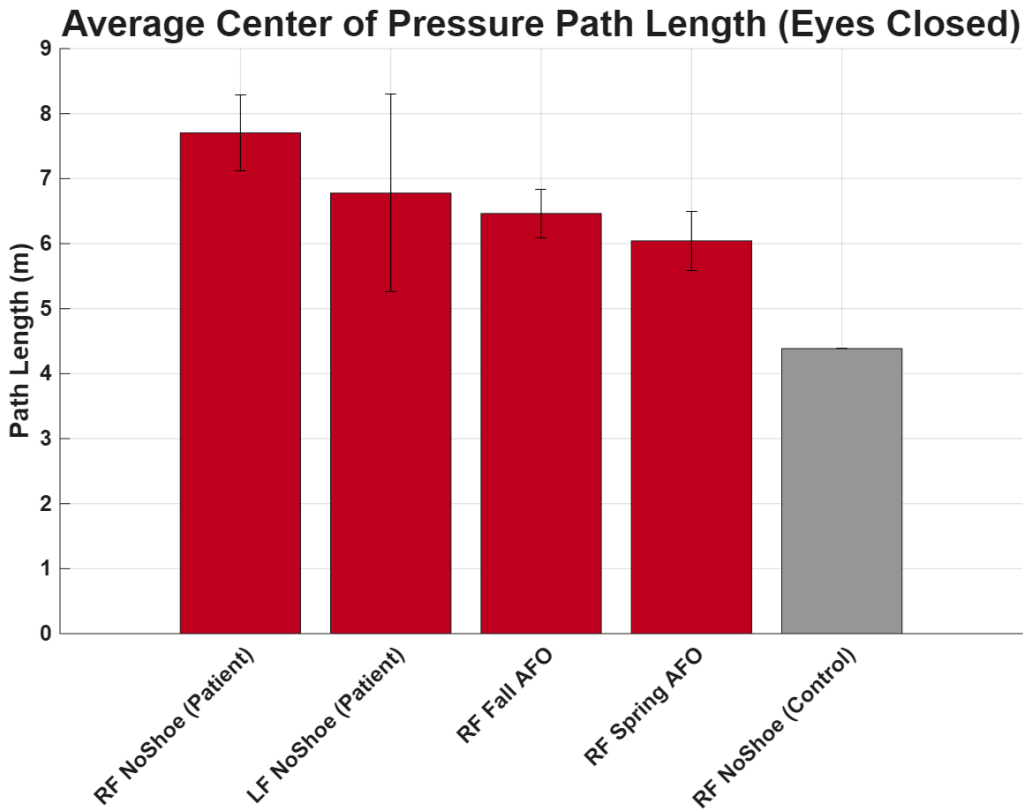


Figure 19: A comparison of average center of pressure path lengths for trials where the subject balanced on one leg with their eyes closed. The subjects of all of these trials did not wear shoes.

An independent sample t-test was conducted to compare the average COP path length between balancing on the right foot with no AFO, and balancing on the right foot with the Spring 2026 AFO, both of which were trials with the patient's eyes closed. The null hypothesis stated that no difference in mean COP path length exists between the No AFO and Spring 2026 AFO conditions. The alternative hypothesis stated that the use of the Spring 2026 AFO would result in a significant change in the path length. This analysis revealed that the difference in path length between these two conditions was not statistically significant, yielding a p value greater than 0.05 ($p = 0.2844$). Because of this, the team fails to reject the null hypothesis and cannot statistically conclude that the Spring 2026 AFO influenced the path length. Although there was no statistical significance between these conditions, there was a 10.8% reduction in the mean path length when the Spring 2026 AFO was worn. This shows that the final prototype is decreasing the average COP path length, but could still be improved and tested over more trials to show a statistical significance. Further information and MATLAB code for the balance testing can be found in Appendix G.

Discussion

The results from the stabilogram balance tests and dorsiflexion trials show that the Spring 2026 brace does more than just hold the foot up; it actually helps restore functional stability. By reducing the sway area and velocity of the patient's center of pressure, the spring-based design compensates for the asymmetrical muscle weakness common in FSHD much better than a standard rigid AFO [39]. The brace provides a more common dorsiflexion gait pattern. In the long run, this could help prevent secondary injuries like hip or back pain caused by common compensations like hip hiking.

When comparing this to general orthotics research, these results support the idea that dynamic energy return is better than total ankle immobilization. While many medical-grade AFOs just lock the ankle to prevent tripping, the Spring design ($p < 0.05$ vs. Fall iteration) proves that allowing for some supported movement is better for postural control. For FSHD and other foot drop patients, this shows that lightweight, 3D-printed designs can restore a gait that is statistically "healthy". This not only improves safety but also gives patients more confidence to stay active without the constant fear of falling [40].

There were many ethical considerations that needed to be recognized over the course of this project. The patient that the AFO was created for is a 16-year-old girl. This means that parental consent needed to be obtained for every step, and HIPAA policies needed to be followed. The client, also the mother of the patient, was kept informed throughout the entire design process. Emails with progress reports were sent every week to keep her up to date with all new developments, team accomplishments, and goals. Additionally, it was important to both the client and the patient that confidentiality was always maintained. Thus, the patient's name was kept out of all documentation and presentations. Her face was also blurred in any pictures and videos used.

Moreover, with the collaboration with the University of Michigan, the same expectations were communicated and expected. The testing session was closed to everyone except for the client and the graduate student that conducted the testing.

Finally, due to the difference in location between the team and the patient, technology was used as a main form of communication. The school email addresses as well as the personal email of the client were utilized most often for any questions, concerns, and meeting coordination. The patient's personal phone number and the consent to use it was provided by the client. Only one team member utilized this phone number to ensure confidentiality and eliminate any chances of miscommunication. Additionally, Zoom was used for long-distance meetings. These were closed sessions that were only accessible via a direct link provided by the team communicator. When communicating with both the client and the patient, the team made sure to use clearly understood language, avoid health-jargon, and ensure full transparency about the entire design process and any associated risks.

In regards to the ultimate use of the AFO, it has been customized for the patient. It is not intended for commercial use due to the specific sizing and fitting for the patient's anatomical needs. It is designed to correct her individual gait abnormalities.

The testing process also identified several sources of error that may have affected the consistency of the results. Biological noise was a significant factor, as the patient's unbraced gait was highly inconsistent due to muscular dystrophy, leading to high variance in the "No Brace" trials. Patient fatigue across multiple trials likely also introduced variability, as compensatory strategies like hip hiking often change as a patient tires. Finally, minor inaccuracies in reflective marker placement for the motion capture system could have led to slight misalignments in the calculated force trajectories, highlighting the need for highly standardized protocols in future clinical testing.

Conclusions

Closing Thoughts

In conclusion, this project successfully finalized a multi-semester effort to develop a custom ankle-foot orthosis (AFO) tailored specifically for a teen patient with Facioscapulohumeral Dystrophy (FSHD). Building upon the rigid 3D-printed supports and dorsiflexion strap systems established by previous iterations, the Spring 2026 team focused on refining the device to meet all patient requests for a lightweight, discreet, and flexible solution. By addressing the critical challenge of brace slippage and optimizing the mechanical assistance of the dorsiflexion motion, the team was able to produce a final design that balances safety and stability with the natural movement required for daily use.

Ultimately, the testing data from this final phase demonstrates that the refined brace successfully positions the ankle in proper dorsiflexion while preventing potentially harmful inversion and eversion. The transition from a purely rigid support to a more dynamic spring-assist design not only fulfilled the design priorities of being slim and unobtrusive but also provided the measurable functional improvements necessary for safer walking. This semester concludes the design process with a validated device that significantly enhances the patient's gait and stability, providing a successful end-to-end engineering solution for FSHD-related foot drop.

Acknowledgements

The current AFO team could not have completed this project without help from everybody involved over the course of two years. First, the Fall 2024 team, led by Anya Hadim, and general members Lucy Hockerman, Presley Hansen, Grace Neuville, and Alex Conover. This team was the start of it all, creating the first iteration of the project; the advisor was Dr. Brandon Coventry, who gave important advice to the project's analysis. The Spring 2025 team was integral to the medial and lateral components, with new additions to the team, Sadie Rowe and Kate Hiller, with the advisor being Dr. John Puccinelli.. The Fall 2025 team was led by Alex Conover, and team members included Avery Lyons, Claire Matthai, Celia Oslakovic, Aditi Singhdeo, and Sean Carey. This team was essential to the current design, and performed the in-person testing with

the client and patient; the advisor was Dr. Justin Williams, who provided the team with excellent resources for analyzing our data.

The final team is the current one, with Alex and Avery continuing the project from the fall, and Sierra Loosen and Kalob Kimmel joining the team. Without any of these members, the project would not have been a success. The team would also like to thank Dr. Christa Wille for her help in analyzing the data, and providing biomechanical insight to testing and advice for the prototype. Dr. Suarez-Gonzales was also integral in the statistical analysis of this semester. The entire Team Lab and DI Lab were helpful to the team in the planning and fabrication of the device. The team would also like to thank the University of Michigan's Department of Kinesiology, Dr. Deanna Gates and Hiva Razavi, for allowing the team to use their space and collect the data for us. The team would also like to thank Dr. Peter Adamczyk for bringing the project to the UW-Madison BME department, as well as Ms. Debbie Eggleston for being an amazing client.

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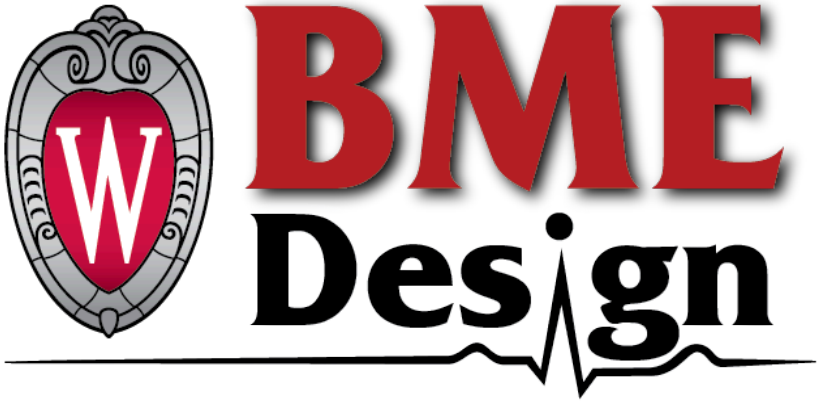
Appendix

Appendix A: Product Design Specifications



Inconspicuous Ankle Foot Orthosis (AFO) for teen

PRODUCT DESIGN SPECIFICATIONS (PDS)

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Team AFO

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February 5th, 2026

Function/Problem Statement:

Ankle-foot orthoses (AFOs) are designed to support dorsiflexion during the swing phase of walking. They are commonly used in managing muscular dystrophies, and for this project, our focus is specifically on adolescents with Facioscapulohumeral Dystrophy (FSHD), the most prevalent form of muscular dystrophy [1]. Our goal is to create a brace that helps teens achieve safer walking by assisting ankle dorsiflexion, while remaining discreet, lightweight, and flexible enough to allow natural ankle motion. The main design priorities are to position the ankle in proper dorsiflexion, keep the brace slim and unobtrusive, and provide enough flexibility to reduce movement restrictions. This project has been ongoing throughout three semesters, and this semester, spring 2026, will be the final semester of the project; the team is hoping to create a device that fulfills all requests, as well as displays significant data.

Client requirements:

The client requests an AFO to be created to help support dorsiflexion of the right foot, as well as prevent excessive inversion. It should be flexible enough for daily activities, and be simple to wear. Additionally, the client prefers the AFO to be discreet, fitting inside a shoe and minimizing visibility. Functionality is becoming more prevalent as the disease increases.

Design requirements:

1. Physical and Operational Characteristics
    a. Performance requirements
        i. The AFO is designed to remain discreet and lightweight, using minimal material while still providing strong support for ankle dorsiflexion and resisting ankle inversion to prevent gait irregularities [1]. It allows a natural walking pattern without generating resistive moments during dorsiflexion [2].
        ii. The device permits more than 30° of motion from the initial ankle angle to ensure proper foot clearance [3].
        iii. In plantarflexion, the orthosis generates an adjustable resistive moment ranging from 5–10 Nm per 10° of motion [3]. Overall, moment-angle performance should stay within ±30 Nm of torque. The brace also resists torsional forces that could cause misalignment of the ankle or foot during regular activity [4].
        iv. The AFO withstands forces equal to at least three times the user's bodyweight, reflecting the peak loads experienced during walking [5]. For a 16-year-old

weighing approximately 118 lbs (53.4 kg), this translates into supporting a maximum force of 1570 N [5]. The normal force exerted by the patient is 524 N. At the same time, the device must allow active concentric ankle movement so the user can perform daily activities such as squatting or climbing stairs.

v. Dimensions must be customized to the user's leg geometry to ensure a secure fit and ideally integrate with a custom orthotic insole, the dimensions of the leg are detailed below.

vi. The rigid components must also limit inversion to less than 25° [6].

b. Safety

i. The AFO should promote normal gait mechanics to reduce the risk of tripping or falling while also maintaining anatomical alignment to avoid excessive stress on joints, bones, or muscles.

ii. Chosen materials should be non-toxic, hypoallergenic, and free of sharp edges to prevent irritation or injury.

iii. Adjustable parts must secure under impact but not restrict circulation to the foot and ankle areas.

iv. Fastening systems should be secured to prevent loosening during activity, but allow for quick removal in emergencies without tools.

v. The device must withstand the forces put on it by the user as outlined in the performance requirements. Carbon fiber AFOs, for example, typically fail at the mid-shank calf support under forces of 1970 N [7].

vi. The design should emphasize breathability to prevent a buildup of moisture and overheating of the user.

vii. To reduce injury risk and maximize comfort, the device will include mesh padding in the calf region and around any areas of discomfort as noted by the user during testing.

c. Accuracy and Reliability

i. The AFO must maintain structural integrity through repeated use while continuing to provide consistent dorsiflexion support. The device will be used over long periods of time, and must provide consistent results throughout the entire duration of use.

ii. The device should be made with durable materials. Ideally, the materials should not degrade over time, or can be easily replaced to provide consistent results.

iii. The device should also provide consistent mediolateral support, but this is not currently the client's highest priority currently.

d. Shelf Life

i. Because custom orthotics are tailored to an individual's needs, their shelf life is limited. If left unused for extended periods, changes in the user's measurements or support requirements may reduce effectiveness. For this reason, the AFO must be periodically re-evaluated to confirm fit and function, ideally up to twice per year.

e. Life in Service

i. The expected lifespan of a custom AFO is typically 2-5 years, though actual service life depends on the material, usage patterns, and patient needs [8]. Individuals who are experiencing rapid periods of growth, such as children, may need the AFO replaced as often as every 9-18 months [9].

ii. Regular cleaning and upkeep of the device could help to increase the life in service.

iii. Semi-rigid materials such as carbon fiber, fiberglass, and polyethylene generally last longer than softer materials as they are more resistant to damage [10]. Softer materials often need to be replaced more frequently than rigid materials.

iv. Annual reviews by an orthotist are recommended to assess wear and ensure the device continues to meet the user's needs [11].

f. Operating Environment

i. The primary intention for the AFO is everyday use. As such, it must be able to withstand everyday activities without deteriorating. Main uses will be at school, home, and horseback riding. In order for the AFO to be worn for these everyday activities it must be unobtrusive and unassuming to the eye.

ii. The AFO must withstand exposure to varying environmental factors including temperatures, humidity, dirt, water, and sweat. To prevent infections due to

bacterial buildup, the device needs to be cleaned weekly with mild soap and water [12].

g. Ergonomics
    i. The device must distribute the user's weight evenly to avoid discomfort. Adjustable features such as straps and bands should allow for modularity of the AFO. This will help the device fit the user through growth and activity needs.
    ii. As most AFOs weigh between 0.3–3.4 kg [13], the inconspicuous design should weigh under 1kg. This will allow for a low profile brace that improves dorsiflexion gait without altering step due to extraneous weight.
    iii. Extra padding must be introduced around sensitive areas such as the base of the foot, ankle, and achilles tendon. The design must be low profile enough to fit into shoes so that there is no need for shoes tailored to the device. [3].
    iv. Moving parts must function quietly so it does not draw attention to the AFO.
    v. By supporting dorsiflexion, the AFO can improve step length, walking speed, and overall gait stability, helping the user move more efficiently in daily life [14].
h. Size:
    i. The AFO must match the patient's specific measurements, with slight adjustments to allow for padding and anti-chafing features [15]. Key measurements are as follows:
        1. The length of the leg (measured bottom of foot to directly below the kneecap) is 45.5 centimeters.
        2. The diameter directly below the kneecap (measured at top of the lower leg) is 31.5 centimeters.
        3. The diameter of the thickest part of the calf (measured mid-leg) is 31.5 centimeters.
        4. The diameter where the Achilles meets the calf (measured bottom of leg) is 20.5 centimeters.
        5. The diameter of the thinnest part of the ankle (measured where Achilles is felt) is 20 centimeters.
        6. The diameter across the middle of the ankle, through the joint is 30 centimeters.

7. The diameter just in front of the ankle joint (measured low ankle) is 24.5 centimeters.
8. Arch Measurements: bony prominence to floor is 4.5 centimeters and 6.25 centimeters in length.
9. The length of the foot is 24-24.5 centimeters.
10. The width of the foot (measured where the metatarsals meet the phalanges) is 8.25 centimeters.
11. The width of the foot (measured in midsole area) is 8 centimeters.
12. The width of the foot (measured at the heel) is 5.5 centimeters.
13. The patient weighs 53.4 kilograms.
14. The patient's height is 1.724 meters.
15. The patient's shoe size is 8.5-9 on a U.S. scale.

ii. A standard AFO thickness is approximately 3.175 millimeters, which balances structural support with sufficient flexibility to avoid stiffness-related instability [16].

i. Weight

i. The orthosis should remain lightweight enough to allow free movement without affecting gait or speed. Ideally, total weight will stay under 1 kilogram [17].

j. Materials

i. The AFO design will be finalized this semester. It should be a discrete, minimally visible, and comfortable design that accomplishes the project goals.

ii. The main material of this design will be a carbon fiber-reinforced (PLA-CF) or a pure carbon fiber.

1. PLA-CF material properties include high ultimate tensile strength, high Young's modulus, high flexure stress, and low ductility [18].
2. Pure carbon fiber properties include high tensile and compressive strengths, high Young's modulus, low density, and high temperature tolerance [19].

iii. The dorsiflexion component of the brace will be made of either TPU filament or polyester fabric. Either of these materials will need to withstand forces from the patient walking, so around 1000 N of force to be within safety margins.

1. The notable properties of polyester include ductility, durability, mechanical strength, and moisture resistance [20].
2. 3D-printed thermoplastic polyurethane (TPU) exhibits lower elasticity when compared to the material in bulk. To mitigate premature failure, the orientation of the printed layers is critical, as strength in the Z-direction is significantly weaker; tensile loads should therefore be aligned in-plane with the filament paths [21]. Fatigue testing for TPU is vital due to its tendency to fail under continuous cyclic loads. Despite these limitations, TPU provides excellent abrasion resistance and environmental resistance, which enhances durability in applications like straps for both upper and lower body use [22].

iv. The padding should be made of two layers of mesh that are sewn together. These are then attached to the inside of the AFO via superglue [23].

k. Aesthetics, Appearance, and Finish

i. The AFO will feature a sleek black design to minimize visibility. It will resemble an athletic brace and fit comfortably inside tennis shoes or Converse, helping the user maintain their preferred style.

ii. The surface will be smooth, slim, and inconspicuous, while still offering the necessary support. The brace is similar to the look offered by an athletic brace.

2. Product Characteristics

a. Quantity

i. The project consists of designing and fabricating one right-leg AFO. However, with considerations of bringing the product to market, the design has to be easily fabricated in order to mass produce the inconspicuous AFO.

b. Target Product Cost

i. This project is funded by Biomedical Engineering Design at the University of Wisconsin-Madison. The expected cost of this semester's continuation is $50 with a possible increase with materials like carbon fiber for strong and light weight material options.

ii. As of fall 2025, the prototypes have accounted for $272.39. The semesterly breakdown of the budget is $189.02 for fall 2024, $37.95 for spring 2025, and

$45.42 for fall 2025. If the team stays under $77.61 then the project will be within $350 for all semesters.

iii. The goal for spring 2026 is creating a final working prototype; reworking the fall 2025 design based on data produced by the client, improving material selection for dorsiflexion gait and medial lateral support, and solidifying significant data to prove effectiveness. As the previous prototype does not have fully significant results a new design will be made within the constraints of the budget.

3. Miscellaneous

a. Standards and Specifications

i. CFR Title 21, Section 890.3025: This regulation classifies the device as a Class I medical device. If electronics are added, it would fall under Class II [24].

ii. 501(k) requirements: Most Class I devices are exempt from 501(k) submission. This AFO may be exempt if the FDA determines that additional review is not needed to ensure safety and effectiveness [25].

iii. CFR Title 21, Section 890.3475: Defines a limb orthosis as a medical device worn on the upper or lower limbs to support, correct, or prevent deformities. Examples include braces, splints, elastic stockings, and corrective shoes [26].

iv. CFR Title 21, Part 803: Manufacturers and facilities must report any deaths or serious injuries linked to the device through a Medical Device Report (MDR) [27].

v. ISO 14971:2019: Outlines risk management requirements. A Failure Modes and Effects Analysis (FMEA) should be done to identify possible risks for patients, users, and property. The standard defines risk as the combination of the chance of harm and the severity of the outcome [28].

vi. ISO 8549-3:2020: Defines an orthosis as an external device used to compensate for problems in the neuromuscular or skeletal system. An ankle-foot orthosis specifically covers the ankle joint and all or part of the foot [29].

vii. ISO 8551:2020: Provides guidelines for evaluating functional deficiencies in patients and setting clinical objectives when prescribing orthoses [30].

viii. ISO 2267:2016: Specifies testing methods for ankle-foot devices under repeated loading. Testing simulates the stance phase of walking, from heel strike to toe-off, to evaluate strength, durability, and service life [31].

b. Customer [32]
    i. This device is designed for daily use by a 16-year-old with Facioscapulohumeral Dystrophy (FSHD) that requests the device be as unnoticeable as possible. It should be able to be worn both with and without shoes. Although it is custom-fitted, the target group also includes other young patients with FSHD or related muscular dystrophies who require ankle inversion, eversion, and dorsiflexion support.
c. Patient-related concerns
    i. The orthosis must hold the ankle in dorsiflexion (approximately 10° above the neutral foot plane) when unweighted, ensuring proper foot clearance and reducing gait deviations. At the same time, it must allow enough flexibility for functional tasks such as squatting or descending stairs.
    ii. The device should minimize the need for eccentric muscle contractions while preventing foot slap, thereby supporting patients with weakened ankle muscles.
    iii. The AFO must balance flexibility and stability: flexible enough to allow natural gait, but strong enough to prevent foot drop and inversion. It should not interfere with daily activities and should remain discreet to avoid drawing attention.
        1. There has been minimal recovery of the ankle movement in the inversion and eversion aspect, leading the brace to focus more on dorsiflection support and less on eversion and eversion prevention.
    iv. A slim profile that can be hidden under clothing is essential to reduce the risk of stigma or bullying in social settings such as school.
d. Additional optional patient requests
    i. The device should be designed to fit comfortably within the patient's horse riding boot, if possible.
    ii. The device should resemble a standard athletic brace to avoid drawing attention in public settings.
e. Economic Impact

i. Each year, approximately 53,000 AFOs are fabricated in the United States, with an average Medicare reimbursement of $417, totaling more than $2.2 million annually [33]. AFOs can cost over $1000 and, for many families, these costs present a barrier to access [34][35].

ii. The global AFO market is expected to grow to over $330 million by 2034 as demand for mobility aids increases due to rises in neurological and musculoskeletal conditions [36]. This emphasizes the need for a cost-friendly, yet effective AFO.

iii. For patients with muscular dystrophies, additional expenses accumulate through both direct and indirect medical costs. Direct costs include hospital visits, therapy, pharmaceuticals, and insurance coverage, averaging $22,533 annually in the U.S. [37]. Indirect costs such as home modifications, vehicle accommodations, caregiving, dietary needs, and travel add approximately $12,939 per patient each year [37].

iv. Loss of income is another significant burden in situations where the condition worsens to the point of the patient not being able to work. Families with a member requiring care for a muscular disorder experience an annual income reduction of about $21,600 compared with unaffected households, even after accounting for demographic and socioeconomic variables [37].

v. Overall, the economic burden of muscular dystrophy disorders in the U.S. is estimated at $1.07–1.4 billion annually [37]. Developing a cost-effective AFO can help ease this financial strain by improving mobility, enabling greater independence, and supporting long-term productivity for individuals living with FSHD.

f. Competition

Most ankle–foot orthoses (AFOs) are based on the three-point force system, a common biomechanical approach used to control joint motion and limit unwanted movement. In this system, one main corrective force is applied in either the mediolateral or anteroposterior direction, while two opposing forces act above and below it to provide balance. Together, these forces stabilize the joint. Increasing the length of the orthosis spreads these force points farther apart, which improves how effectively the brace

controls motion. This wider spacing also helps spread pressure over a larger area, making the device more comfortable for the user [38].

i. Passive-Dynamic AFO (PD-AFO)
    1. The PD-AFO features a sleek, flexible design suited for patients with mild ankle weakness.
    2. It incorporates a flexible calf shell that absorbs energy during stance and releases it at push-off, promoting dorsiflexion. Studies have shown that PD-AFOs improve patient comfort and spatiotemporal gait parameters.
    3. Dimensions can be customized for individual users through 3D printing; however, stiffness and support cannot currently be tailored to match varying levels of ankle impairment [1].

ii. Supramalleolar Orthosis (SMO)
    1. Pediatric SMOs are constructed from thin, flexible thermoplastic and extend just above the ankle bones (malleoli).
    2. They primarily provide control of subtalar joint alignment, maintaining a neutral heel to improve mediolateral stability.
    3. Their lightweight, low-profile design makes them comfortable for daily wear and compatible with most shoes [39].

iii. Variable Stiffness Orthosis (VSO)
    1. The VSO is a powered AFO currently in the research phase. It uses a customizable cam-based transmission system that can define specific torque-angle relationships and adjust stiffness in real time.
    2. Early results suggest it reduces foot drop and increases overall ankle moments. However, VSOs are not yet commercially available [40].

iv. Jointed AFO
    1. Jointed AFOs include a hinge at the ankle joint, allowing controlled motion and enabling a more natural gait with a full range of movement.
    2. While they optimize gait patterns, drawbacks include greater bulk, potential noise during use, and a higher likelihood of mechanical component failure [38].

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Appendix B: BPAG Cost Spreadsheet

Item	Description	Manufacturer	Vendor	Date	QTY	Cost Each	Total
Side bases							
PLA CF	Medial & lateral PLA CF print	Makerspace	design lab	3/8/2026	1	$2.47	$2.47
Tough 1500	Medial & lateral Tough 1500 print	Makerspace	design lab	4/23/2026	86.1	$0.25	$21.53
Straps and Padding							
TPU	Tpu print of doge bone for MTS testing	Makerspace	Design Lab	2/18/2026	3	$0.17	$0.51
Superglue	Adhesive used for attaching padding to supports		Design Lab	3/9/2026	1	$1.50	$1.50
Elastic Fabric	Fabric used to prototype with sock		Design Lab	4/9/2026	1	$1.00	$1.00
Hair clips	Used for sock prototype	Splendorflying	Amazon	4/1/2026	1	$3.99	$3.99
						TOTAL:	**$31.00**

Appendix C: Fabrication Protocol

Fabrication Plans for Inconspicuous AFO for Teen

Team Members: Alex Conover, Avery Lyons, Sierra Loosen, Kalob Kimmel
Advisor: Dr. Monica Ohnsorg, University of Wisconsin-Madison
TA: Sam Kahr
BME 301, Section 304

Material	Quantity	Cost
Mold of patient foot	1	N/A

CF-PLA	*See print details	$9.00
Ballistic nylon	24 in^2	$12.61 for fabric
Polyester	18 in	$7.99
Mesh fabric	*See print details	$16.99 for fabric
Velcro	2 in^2	$4.99
Black thread	1 spool	$2.50/spool (free to use at Wendt)
Sewing kit	1 kit	$14.99/kit (free use of teammembers kit)
Superglue	0.07 oz bottle	$1.25
Fabric scissors	1 pair	N/A
White fabric pencil	1	N/A
Exacto knife	1	N/A
Bambu 3D Printer	1 printer	N/A

*Cost per brace is approximately $13

Name of fabrication step/portion of prototype: Re-Designing and Printing the Medial and Lateral Supports of the AFO
Date to be completed: 03/09/2026
Team Members fabricating: Kalob
Detailed Bulleted Steps of fabrication:
The supports are created by making an organic sheet that is fit to the foot and then extruding and cutting out the templated supports from the sheet. Onshape was used for this process. The steps to create the supports:

- Insert 3D rendered foot into the cad software as two separate castings for the medial and lateral sides.
- Make a vertical plane down the center of the middle of the medial and lateral render. This will be where you create the first sketch of a vertical line and a horizontal line centered on the malleolus (plane 1).
- Make a plane on the vertical line from plane 1 so they are perpendicular, and the plane is sticking out of the malleolus (plane 2).

- Make a baseline plane level with the foot on the horizontal line from the plane 1 (BOE).
- Make horizontal planes based off of the BOE plane every half inch till you reach bottom of the foot and about 1 inch past the top of desired brace height (8 inches) (Form Planes).
- Make drawings on the Form Planes and the BOE Plane, and then use spine lines to follow the outside curves of the ankle, foot and lower leg (Sheet drawings).
- Make a drawing on plane 1 that acts as the support cut out. This will be made through a spine line acting as the outline and then 4 hole points. Two at the top for straps, 1 in the middle for the malleolus, and one on the bottom for the back strap. The dimensions are very complex and differ from person to person. Provided below are the dimensions in inches for this sketch of the lateral side (Figure 1).

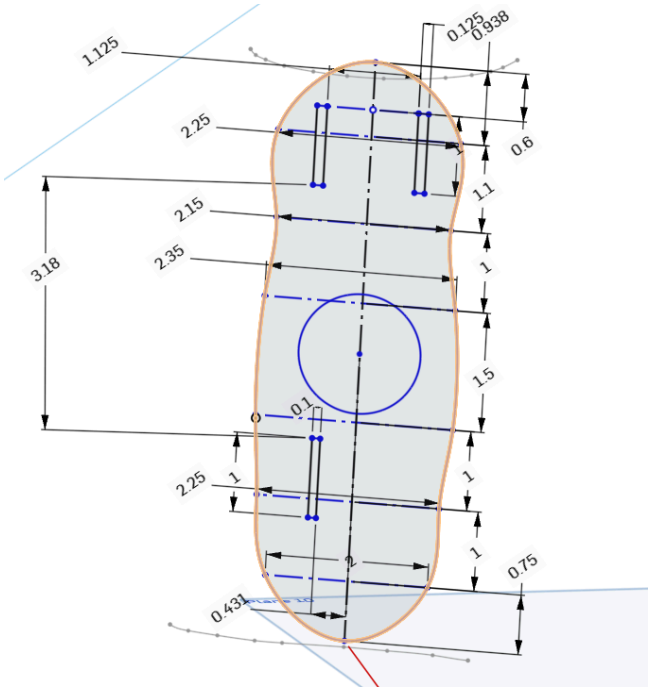

Figure 1: CAD model of the lateral support to be 3D-printed

- Make a loft including all the sheet drawings so that an organic sheet is made that envelope the lower foot.
- Make a surface extrude with the support sketch as the edge and settings of blind and a depth of 1.25 inches.
- Use the split command to split the organic sheet as the curve to split and the support sketch as the entity to split with.
- Use the thicken function to thicken the loft from the previous step.
- Use the sketches of the holes as outlines, and use them to do solid removal extrudes.
- Repeat all steps for Medial support.
- Once both parts are done, download them separately as stl files.
- Load into splicing software.

- Drag both in and align outside of braces with the bottom of the printer plate.
- Change material to PLA CF.
- Auto generate tree supports.
- Ensure infill is 50% and triangles.
- Match and download to PLA CF 3D printer.
- Run simulation, then print.
- Post process by taking out the printer and removing supports.

Name of fabrication step/portion of prototype: Fabrication of the Medial and Lateral Supports of the AFO
Date to be completed: 03/10/2026
Team Members fabricating: Alex, Avery, and Sierra
Detailed Bulleted Steps of fabrication:

- The medial and lateral prints are placed onto the swatch of mesh fabric.
- Use a white fabric pencil to mark where the fabric will be cut. The shape of the fabric should roughly be the same shape as the medial and lateral prints, leaving about an extra 0.25” border of fabric around each cutout. In figure 2, the line drawn with the white fabric pencil is represented by the dashed line.
- A fabric scissors is used to cut the specific swaths of fabric.

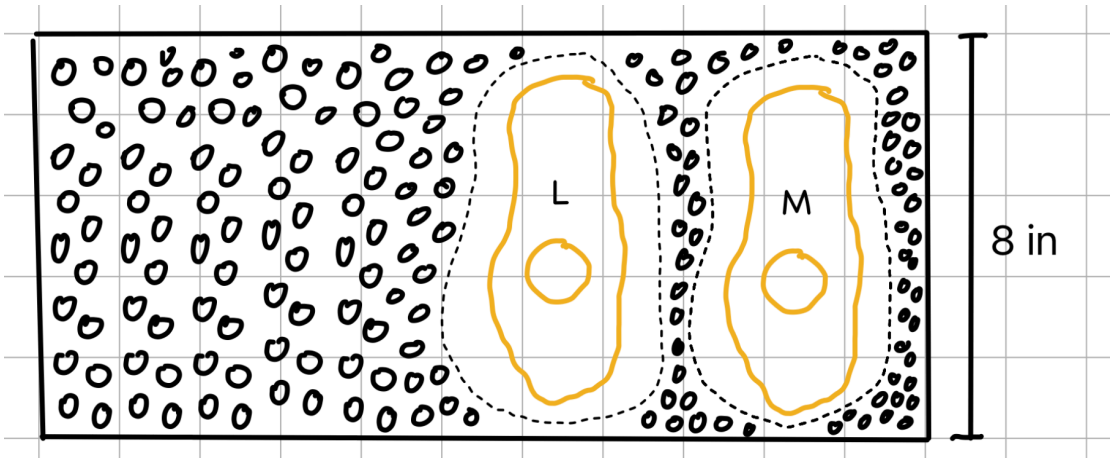

Figure 2: Cutting the Mesh Fabric from the Medial and Lateral PLA-CF prints

- After the fabric has been cut, use a slightly thicker needle and black thread to sew the 2 halves together along the outside edge. Straight sewing, or sewing over the edge is preferred for strength, but the ultimate goal is to make sure the fabric halves stay together. The yellow lines in the figure below represent the location of the stitching.

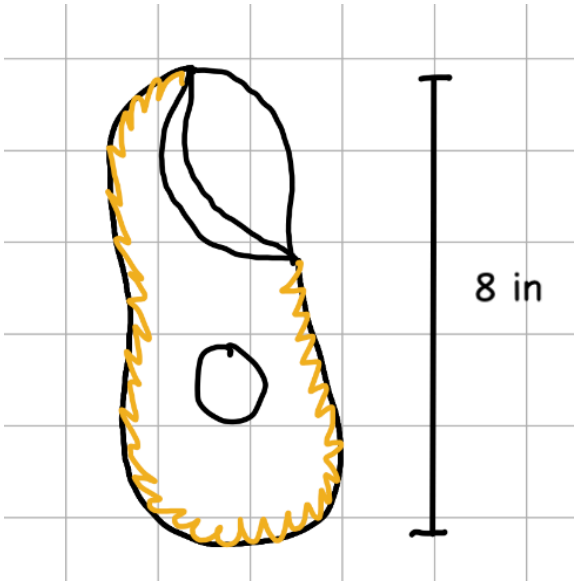

Figure 3: Sewing the mesh fabrics together

- Once the fabric is sewn together, use superglue to attach the mesh fabric to the PLA-CF print outs. Adequate adhesion to the print out requires using 40-50% of the small bottle as bought from the DI Lab in Wendt.

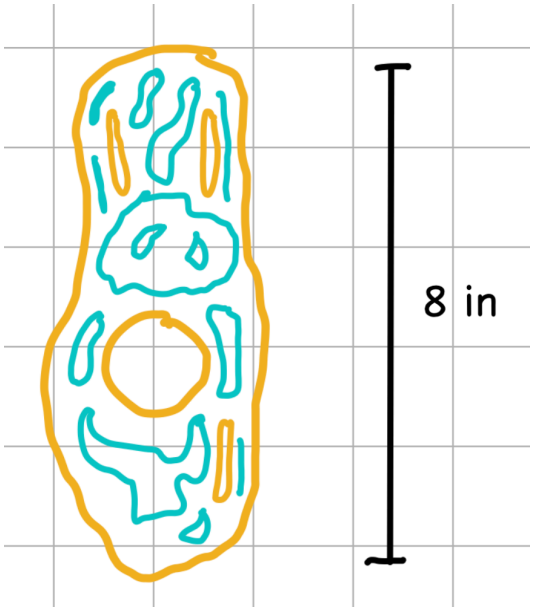

Figure 4: Superglue (cyan) applied to the brace (yellow)

- Once both sides have been completed, allow 5 minutes for the superglue to dry. After the superglue is set, the material is ready to be cut.
- The PLA-CF has slits in the top for the back strap and front strap. There is a second slit on the bottom half to secure the second backstrap. The fabric must be cut out of these slits

using an exacto knife, cutting through the 2 layers of mesh padding to allow the straps to be threaded through.

Name of fabrication step/portion of prototype: Fabrication of the back straps of the AFO
Date to be completed: 03/10/2026
Team Members fabricating: Avery and Kalob
Detailed Bulleted Steps of fabrication:

- Obtain ballistic nylon.
- Cut the piece using fabric scissors into 1-inch wide x 6 inch in length pieces (x2).
- After cutting, slide one edge of a 6 inch piece from the padded side to the hard outer side of the PLA-CF through the superior most slot (see photo below).
- Slide the other edge of the fabric from the padded side to the hard outer side through the superior slot of the other PLA-CF piece.
- Repeat this process with the other 6 inch piece of ballistic nylon, but inserting through the inferior slots of the PLA-CF pieces.

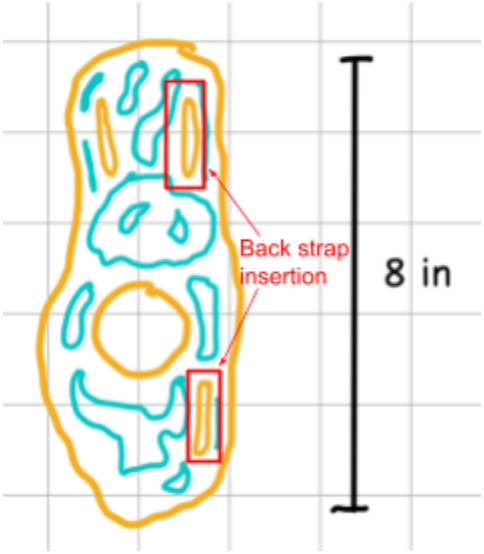

Figure 5: A drawing showing where the back straps should be inserted through the PLA-CF pieces.

- Loop the edge of the ballistic nylon strap around the PLA-CF and back onto itself with all four edges of the straps (medial and lateral sides, and superior and inferior slots). Adjust as necessary so that the AFO is securely fit to the mold of the foot.
- Once secured, sew the ballistic nylon pieces to themselves by hand using a needle and black thread down the 1 inch side. These straps are completely sewn together and will not be adjustable.

- The back straps should now be completely attached to both of the PLA-CF pieces.

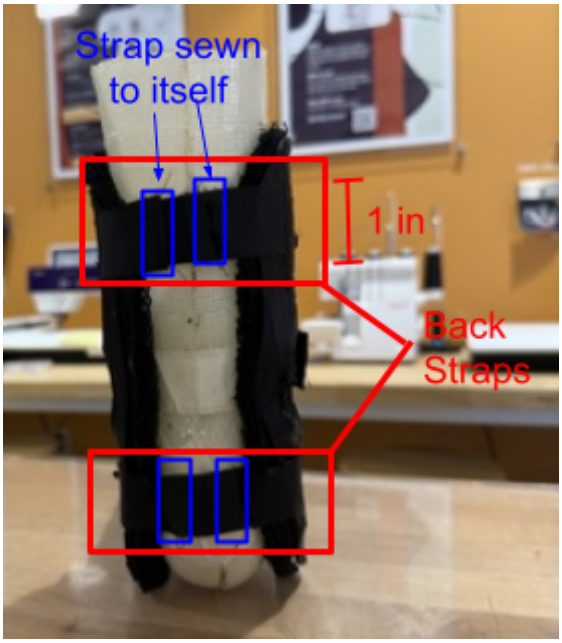

Figure 6: Photo of the back straps on the prototype.

Name of fabrication step/portion of prototype: Fabrication of the front strap of the AFO
Date to be completed: 03/10/2026
Team Members fabricating: Alex
Detailed Bulleted Steps of fabrication:

- Obtain a piece of ballistic nylon.
- Cut out a piece from the ballistic nylon that is 1 in wide x 12 inch long using fabric scissors.
- Once cut, the front strap is secured on the lateral side via sewing, looped through the medial side, and is adjustable via velcro on top of the sewing on the lateral side
- A 1 inch x 2 inch piece of velcro is cut and adhered with the adhesive already on the velcro.
- The velcro adhesive is additionally superglued to the brace for maximum support.

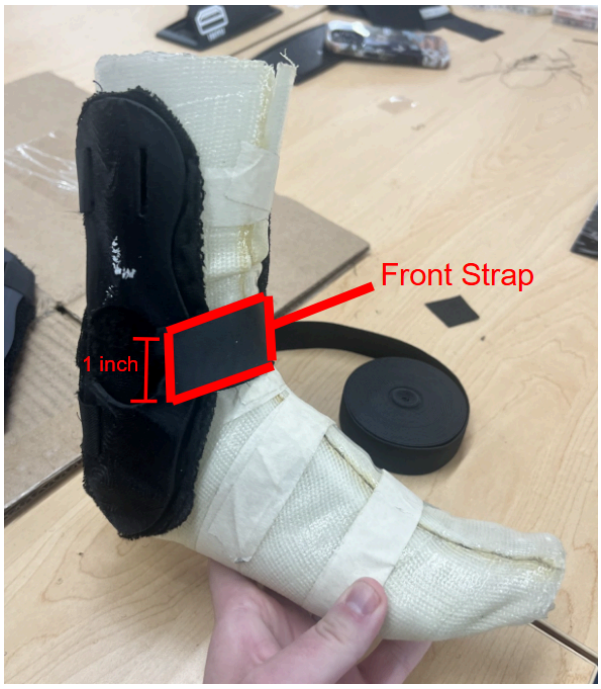

Figure 7: Lateral side view of front strap

Figure 8: Front view of front strap

Name of fabrication step/portion of prototype: Fabrication of the Dorsiflexion Strap
Date to be completed: 03/12/2026
Team Members fabricating: Sierra and Alex
Detailed Bulleted Steps of fabrication:

The dorsiflexion strap is a total of 18 inches on the working end, with approximately 2 more inches in the brace holding it onto the brace, for a total of 20 inches.

- Cut the elastic polyester into a 36 inch strip using fabric scissors.
- Fold the elastic polyester in half, making the ends meet.
- Sew both ends together with a needle and thread to prevent slippage.
- Sew across the entire strip to make the material less stretchy.
- Once the strap has been sewn all together, sew the dorsiflexion straps to the medial and lateral portions of the brace, as seen below.
- The strap is sewn around the brace to itself with a needle and thread, as close to the medial and lateral sides as possible.

Figure 9: Dorsiflexion straps sewn to the brace

Appendix D: Testing Protocol

Testing Plans for Inconspicuous AFO for Teen - UMich Forceplate testing

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Team Members: Alex Conover, Avery Lyons, Sierra Loosen, Kalob Kimmel
Advisor: Dr. Monica Ohnsorg, University of Wisconsin-Madison
TA: Sam Kahr

Date of Testing: 4/11/26
Location of Testing: University of Michigan - Ann Arbor

Devices to Test:

- Fall 2025 Model (Elastic Polyester Dorsiflexion Strap, original length)
- Spring 2026 Model (EP Dorsiflexion Strap, Lengthened Medial/Lateral Components)

Previous Testing:

The previous testing for this project was performed in November of 2025, with the Fall 2025 team (which included Avery and Alex); the testing performed was a gait analysis, followed by a stabilogram analysis of balance/path length. The goal was to see if the preliminary design ideas improved the gait of the patient, while also improving the overall balance. Photos of the brace tested can be seen in the Testing Analysis Document. The final iteration of the design was created post testing can be found below.

Figure 1: Final Fall 2025 AFO Brace

This iteration of the brace was never tested, as it was completed post testing, and elements from testing were iterated in this brace. The team has received feedback from the patient about the feel of this brace; it feels good, works well from what she can tell, and doesn't have the uncomfortable rubbing against the malleoli of the ankle.

Moving forward, the team is looking to repeat the testing from last semester, but the team requires the assistance of the University of Michigan to perform testing closer to the patient/client's home location to prevent excessive travel and disruption of the patient. The rest of this document will provide the testing protocol to follow for testing during this period.

Force Plate Testing - Gait Analysis
Protocols:

1. 6 x Walking, No brace, No shoe (Control)
2. 6 x Walking, No brace, With shoe
3. 6 x Walking, With brace, No shoe
4. 6 x Walking, With brace, With shoe

Details:

For the first test, the control, the following protocol will be utilized:

- The patient will walk across the force plate, taking a minimum of three steps
- The patient will walk at her natural stride speed, while keeping all steps on the force plates
- The patient will not wear the AFO or shoes
- The patient is permitted to wear socks
- The operator will instruct the patient to walk when they begin the timer
- The operator will stop the timer when the patient has walked the entire length of the force plate
- The operator will save the data to the computer with the trial number, and detail of the trial (Ex: Trial_1_Walking_NoShoe_NoBrace)
- Once the data has been successfully saved, the operator will instruct the patient to repeat the test

For every other test, the following procedure will be used:

- The patient will walk across the force plate, taking a minimum of three steps
- The patient will walk at her natural stride speed
- The patient will wear the AFO (if applicable) on her right foot and shoes (if applicable)
- The patient will wear the yellow Adidas shoes that she wore in previous testing, if possible
- The operator will instruct the patient to walk when they begin the timer
- The operator will stop the timer when the patient has walked the entire length of the force plate
- The operator will save the data to the computer with the trial number, and detail of the trial (Ex: Trial_1_Walking_WithShoe_WithBrace)
- Once the data has been successfully saved, the operator will instruct the patient to repeat the test
- Each test must be repeated a total of six times before moving onto the next test

Force Plate Testing - Stabilogram Analysis
Protocols:

1. 6 x Left foot, No shoe, eyes open (Control)
2. 6 x Left foot, No shoe, eyes closed (Control)
3. For each brace:
    a. 6 x Right foot, With brace, No shoe, eyes open

b. 6 x Right foot, With brace, No shoe, eyes closed
c. 6 x Right foot, With brace, With shoe, eyes open
d. 6 x Right foot, With brace, With shoe, eyes closed

Details:

For the first two tests, the controls, the following protocol will be utilized:

- The patient will stand with the left foot centered on the force plate
- The operator of the experiment will instruct the patient to lift the right foot off of the ground, while starting the timer, and starting the data collection of the force plates.
- The patient will stand on the left foot for a total of 30 seconds.
    - If the patient loses balance, setting the foot or toe of the right foot down onto the ground is acceptable for data collection, as the overall path length will simply increase.
- Once the 30 second timer has been reached, stop the timer and data collection, instruct the patient to stand on both feet again.
- Save the data to the computer with the trial number, and detail of the trial (Ex: Trial_1_LF_NoShoe_eyes_open)
- Repeat for next trial

For every other test, the following procedure will be used:

- The patient will be standing on the right foot
- Test duration: 30 seconds
- If the patient loses their balance, they may regain their balance by tapping their foot down on the ground, and then moving back up into the air. If this occurs, do not stop the recording of data until the 30 seconds is up.
- Shoes must be worn on both feet during tests that include shoes. Both shoes must be taken off for tests without shoes.

Post-Testing Comfortability Survey

After all testing has been completed, the patient will fill out the provided comfortability survey, noting:

- Any slippage or noticeable loosening of straps
- Any discomfort (i.e. pressure points, sharp edges, rubbing, chafing, etc.)

Appendix E: MTS Testing Analysis

Elastic Polyester	TPU	Ballistic Nylon
Length = 8.5 in, Width 1/32 in, gauge 4.1 in *Material is bunched up in the clamps to ensure adequate testing. rate = 100 mm/s	gauge = 4.7 in, see materials printing for dimensions. * testing rate remains 50 mm/s no visible nicks scratches, or any damage to this specimen	Length = 7.75 in, width = 1 in, thickness = 1/32 in gauge = 3.3 in testing rate 25 mm/s, fabric is scrunched to see if breakage will occur - no breakage, scrunch fail?
8.4 in, width - 1/32, gauge = 3.75 *material is pre-stretched to try and test breakage? slower rate, (75 mm/s)	gauge = 4.7 in. testing rate remains 50 mm/s no visible breakage on the specimen :(	Length = 7.6 in, width = 1 in, thickness = 1/32 in, gauge = 3.5 in. testing rate 25 mm/s - no breakage
7 in length, width 1/32 in, gauge = 4.8 in * less bunching, slower rate (50 mm/s)	gauge = 4.3 in. testing rate depleted to 25 mm/s no visible breakage, but more stretch?	Length = 7.9 in, width = 1 in, thickness = 1/32 in, gauge = 4.25 in testing rate 25 mm/s - no failure

```
% Ballistic nylon tests

ballistic_1 = readmatrix("B1.txt");

displacement_ballistic_1 = ballistic_1(:, 1);

load_ballistic_1 = ballistic_1(:, 2);

time_ballistic_1 = ballistic_1(:, 3);

area_ballistic_1 =  20.16125 %mm squared (the area of the cross section,
width*thickness)

guage_ballistic_1 = 83.82 % mm

stress_ballistic_1 = load_ballistic_1 / area_ballistic_1;

strain_ballistic_1 = displacement_ballistic_1 / guage_ballistic_1;

% Ballistic nylon test 2

ballistic_2 = readmatrix("BN2.txt");

displacement_ballist_2 = ballistic_2(:, 1);

load_ballistic_2 = ballistic_2(:, 2);

time_ballistic_2 = ballistic_2(:, 3);

area_ballistic_2 = 20.16125 % mm squared

guage_ballistic_2 = 88.9 % mm
```

```
stress_ballistic_2 = load_ballistic_2 / area_ballistic_2;

strain_ballistic_2 = displacement_ballist_2 / guage_ballistic_2;

% Ballistic Nylon test 3

ballistic_3 = readmatrix("BN3.txt");

displacement_ballistic_3 = ballistic_3(:, 1);

load_ballistic_3 = ballistic_3(:, 2);

time_ballistic_3 = ballistic_3(:, 3);

area_ballistic_3 = 20.16125 % mm squared

guage_ballistic_3 = 107.95 % mm

stress_ballistic_3 = load_ballistic_3 / area_ballistic_3;

strain_ballistic_3 = displacement_ballistic_3 / guage_ballistic_3;

figure(1);

plot(strain_ballistic_1, stress_ballistic_1, strain_ballistic_2,
stress_ballistic_2, strain_ballistic_3, stress_ballistic_3);

legend("Test 1", "Test 2", "Test 3", "Location", "southwest");

title("Ballistic Nylon");

xlabel('Strain');

ylabel('Stress (MPa)');

grid on;

%% Elastic polyester tests

ep_1 = readmatrix("EP1.txt");

displacement_ep_1 = ep_1(:, 1);

load_ep_1 = ep_1(:, 2);

time_ep_1 = ep_1(:, 3);

area_ep_1 = 20.16125; % mm squared, assuming 1 in wide and 1/32 in thick

guage_ep_1 = 104.14; % mm

stress_ep_1 = load_ep_1 / area_ep_1;

strain_ep_1 = displacement_ep_1 / guage_ep_1;

% Elastic Polyester test 2

ep_2 = readmatrix("EP2.txt");

displacement_ep_2 = ep_2(:, 1);
```

```
load_ep_2 = ep_2(:, 2);

time_ep_2 = ep_2(:, 3);

area_ep_2 = 20.16125; % mm squared, assuming 1 in wide, 1/32 in thick

guage_ep_2 = 95.25; % mm

stress_ep_2 = load_ep_2 / area_ep_2;

strain_ep_2 = displacement_ep_2 / guage_ep_2;

% Elastic Polyester test 3

ep_3 = readmatrix("EP3.txt");

displacement_ep_3 = ep_3(:, 1);

load_ep_3 = ep_3(:, 2);

time_ep_3 = ep_3(:, 3);

area_ep_3 = 20.16125; % mm squared, assuming 1 in wide, 1/32 in thick

guage_ep_3 = 121.92; % mm

stress_ep_3 = load_ep_3 / area_ep_3;

strain_ep_3 = displacement_ep_3 / guage_ep_3;

figure(2);

plot(strain_ep_1, stress_ep_1, strain_ep_2, stress_ep_2, strain_ep_3,
stress_ep_3);

legend("Test 1", "Test 2", "Test 3", "Location", "southwest");

title("Elastic Polyester");

xlabel('Strain');

ylabel('Stress (MPa)');

grid on;

hold on;

%% TPU tests

tpu_1 = readmatrix("TPU1.txt");

displacement_tpu_1 = tpu_1(:, 1);

load_tpu_1 = tpu_1(:, 2);

time_tpu_1 = tpu_1(:, 3);

area_tpu_1 = 13.5080375; % mm squared, 0.03125 in thick, 0.67 in wide

guage_tpu_1 = 119.38; % mm
```

```
stress_tpu_1 = load_tpu_1 / area_tpu_1;

strain_tpu_1 = displacement_tpu_1 / guage_tpu_1;

tpu_2 = readmatrix("TPU2.txt");

displacement_tpu_2 = tpu_2(:, 1);

load_tpu_2 = tpu_2(:, 2);

time_tpu_2 = tpu_2(:, 3);

area_tpu_2 = 13.5080375% mm squared

guage_tpu_2 = 119.38 % mm

stress_tpu_2 = load_tpu_2 / area_tpu_2;

strain_tpu_2 = displacement_tpu_2 / guage_tpu_2;

tpu_3 = readmatrix("TPU3.txt");

displacement_tpu_3 = tpu_3(:, 1);

load_tpu_3 = tpu_3(:, 2);

time_tpu_3 = tpu_3(:, 3);

area_tpu_3 = 13.5080375% mm squared

guage_tpu_3 = 109.22 % mm

stress_tpu_3 = load_tpu_3 / area_tpu_3;

strain_tpu_3 = displacement_tpu_3 / guage_tpu_3;

figure(3);

plot(strain_tpu_1, stress_tpu_1, strain_tpu_2, stress_tpu_2, strain_tpu_3,
stress_tpu_3);

legend("Test 1", "Test 2", "Test 3", "Location", "southwest");

title("TPU Tests");

xlabel("Strain");

ylabel("Stress (MPa)");

grid on;

%% Comparing all data on 1 graph

figure(4);

% Plot all three materials

p1 = plot(strain_ballistic_1, stress_ballistic_1, 'b', ...

          strain_ballistic_2, stress_ballistic_2, 'b', ...
```

```
        strain_ballistic_3, stress_ballistic_3, 'b');
hold on;
p2 = plot(strain_ep_1, stress_ep_1, 'r', ...
        strain_ep_2, stress_ep_2, 'r', ...
        strain_ep_3, stress_ep_3, 'r');
p3 = plot(strain_tpu_1, stress_tpu_1, 'g', ...
        strain_tpu_2, stress_tpu_2, 'g', ...
        strain_tpu_3, stress_tpu_3, 'g');
% Legend using one representative line per material
legend([p1(1), p2(1), p3(1)], 'Ballistic Nylon', 'Elastic Polyester', 'TPU',
...
      'Location', 'southwest');
xlabel('Strain');
ylabel('Stress (MPa)');
title('Stress-Strain Curves by Material', 'FontSize', 15);
grid on;
box on;
```

Appendix F: Dorsiflexion Testing Analysis MATLAB

```
% AFO Graph
close all;
clear all;
conditions = {'Control', 'No Brace', 'Medical AFO', 'Spring Brace', 'Fall
Brace'};
peak_data = [
   1.15, 1.18;
   0.85, 0.92;
   1.05, 1.02;
   1.10, 1.08;
   0.98, 1.01
];
% Error Bar data
sem_data = [
   0.03, 0.02;
   0.08, 0.07;
   0.04, 0.03;
   0.02, 0.02;
```

```
   0.03, 0.03
];
figure('Color','w');
b = bar(peak_data,'grouped');
% Colors
b(1).FaceColor = [0.75 0.0 0.13];   % Cardinal red
b(2).FaceColor = [0.6 0.6 0.6];     % Light gray
% Axes formatting
ax = gca;
ax.XTick = 1:length(conditions);
ax.XTickLabel = conditions;
ylabel('Normalized Force (% Body Weight)', 'FontSize', 18);
title('Gait Performance: Loading vs. Push-off Peaks', 'FontSize', 18);
ax = gca;
ax.FontSize = 18;   % this updates tick labels too
grid on
hold on
% === ADD ERROR BARS ===
for i = 1:length(b)  % loop over each bar group (Peak 1, Peak 2)
   x = b(i).XEndPoints;
   y = b(i).YEndPoints;
   errorbar(x, y, sem_data(:,i), 'k.', 'LineWidth', 1.5, 'CapSize', 10);
end
drawnow
% SECOND GRAPH
% Data Preparation
labels = {'Healthy Control', 'No Brace (No Shoe)', 'Spring Brace (No Shoe)',
...
         'No Brace (Shoe)', 'Spring Brace (Shoe)'};
means = [1.09, 0.88, 0.94, 0.77, 0.96]; % Normalized vGRF (BW)
sems  = [0.03, 0.08, 0.02, 0.12, 0.03]; % Standard Error of Mean
% Create Figure
figure('Color', [1 1 1]);
hold on;
% Define Colors
colors = [0.7 0.7 0.7; % light Gray
         0.5 0.5 0.5; %  gray
         0.75 0.0 0.13;   % Cardinal red
         0.75 0.0 0.13; % Dark Red
         0.3 0.3 0.3]; % Dark gray
% Plot Bars
for i = 1:length(means)
   b = bar(i, means(i), 'FaceColor', colors(i,:), 'EdgeColor', 'none');
end
% Add Error Bars
errorbar(1:5, means, sems, 'k.', 'LineWidth', 1.5, 'CapSize', 18);
% Customizing Axes
set(gca, 'XTick', 1:5, 'XTickLabel', labels, 'FontSize', 18);
ylabel('Peak Vertical Force (BW)', 'FontSize', 24, 'FontWeight', 'bold');
```

```
title('Gait Performance Efficacy: Loading Response Phase', 'FontSize', 14);
ylim([0 1.4]);
grid on;
ax = gca;
ax.GridLineStyle = '--';
ax.Layer = 'top';
% Add Significance Bracket 1: Spring vs Control (p=0.319)
plot([1, 1, 3, 3], [1.15, 1.20, 1.20, 1.15], 'k-', 'LineWidth', 1.2);
text(2, 1.25, 'p = 0.319 (n.s.)', 'HorizontalAlignment', 'center',
'FontWeight', 'bold');
% Add Significance Bracket 2: Spring vs Shoe (p=0.244)
plot([4, 4, 5, 5], [1.02, 1.07, 1.07, 1.02], 'k-', 'LineWidth', 1.2);
text(4.5, 1.12, 'p = 0.244', 'HorizontalAlignment', 'center', 'FontWeight',
'bold');
hold off;
```

Appendix G: Balance Testing Analysis MATLAB

```
%% 1. INITIALIZATION

close all; clear all; clc;

fileList = dir('*.mat');

numFiles = length(fileList);

if numFiles == 0, error('No .mat files found in current folder.'); end

rawResults = table('Size', [numFiles, 3], ...

   'VariableTypes', {'string', 'string', 'double'}, ...

   'VariableNames', {'FileName', 'Condition', 'PathLength'});

fprintf('Starting analysis for %d files...\n', numFiles);

%% 2. BATCH PROCESSING LOOP

for i = 1:numFiles

   fName = fileList(i).name;

   if contains(fName, 'Walking') || contains(fName, 'Static'), continue; end


   try

       s = load(fName);

       fVars = fieldnames(s);

       data = s.(fVars{1});
```

```
        [finalX, finalY, pLength] = calculateCOP_Local(data);


        condName = regexprep(fName, '^Trial_\d+_', '');
        condName = erase(condName, ".mat");
        condName = strrep(condName, 'eyes_closed', 'eyes_close');


        rawResults.FileName(i) = string(fName);
        rawResults.Condition(i) = string(condName);
        rawResults.PathLength(i) = pLength;
        % Save individual PNG
        fig_stab = figure('Visible', 'off');
        plot(finalX, finalY, 'LineWidth', 1); axis equal; grid on;
        title(['Stabilogram: ', strrep(fName, '_', ' ')]);
        subtitle(['Path Length: ', num2str(pLength, '%.4f'), ' m']);
        saveas(fig_stab, [erase(fName, ".mat"), '_Plot.png']);
        close(fig_stab);
    catch
    end
end
%% 3. SUMMARY & CONTROL INJECTION
rawResults(rawResults.Condition == "", :) = [];
fullOrder = {'LF_NoShoe_eyes_close', 'RF_NoShoe_eyes_close', ...
    'RF_WtihBraceFall_NoShoe_eyes_close',
'RF_WtihBraceSpring_NoShoe_eyes_close', ...
    'LF_NoShoe_eyes_open', 'RF_NoShoe_eyes_open', ...
    'RF_WtihBraceFall_NoShoe_eyes_open', 'RF_WtihBraceSpring_NoShoe_eyes_open',
...
    'RF_WtihBraceFall_WithShoe_eyes_close',
'RF_WtihBraceSpring_WithShoe_eyes_close', ...
    'RF_WtihBraceFall_WithShoe_eyes_open',
'RF_WtihBraceSpring_WithShoe_eyes_open'};
rawResults.Condition = categorical(rawResults.Condition, fullOrder);
```

```
summaryStats = groupsummary(rawResults, 'Condition', {'mean', 'std'},
'PathLength');

controlConds = {'RF_NoShoe_eyes_open'; 'RF_NoShoe_eyes_close'};

controlMeans = [3.468; 4.385]; % Converted to meters

controlTable = table(categorical(controlConds, fullOrder), [1; 1],
controlMeans, [0; 0], ...

   'VariableNames', {'Condition', 'GroupCount', 'mean_PathLength',
'std_PathLength'});

summaryStats.Source = repmat("Patient", height(summaryStats), 1);

controlTable.Source = repmat("Control", height(controlTable), 1);

summaryStats = [summaryStats; controlTable];

summaryStats(isnan(summaryStats.mean_PathLength), :) = [];

summaryStats = sortrows(summaryStats, 'Condition');

%% 4. BAR CHART COMPARISONS

lbls_full = ["RF NoShoe (Patient)", "LF NoShoe (Patient)", "RF Fall AFO", "RF
Spring AFO", "RF NoShoe (Control)"];

modes = {'eyes_close', 'eyes_open'};

for m = 1:length(modes)

   thisMode = modes{m};

   subset = {['RF_NoShoe_', thisMode], ['LF_NoShoe_', thisMode], ...

             ['RF_WtihBraceFall_NoShoe_', thisMode],
['RF_WtihBraceSpring_NoShoe_', thisMode]};



   dataSub = summaryStats(ismember(cellstr(string(summaryStats.Condition)),
subset), :);

   dataSub = sortrows(dataSub, 'Source', 'descend');



   if ~isempty(dataSub)

       figure('Name', strrep(thisMode, '_', ' '), 'Color', 'w', 'Position',
[100, 100, 900, 600]);

       hold on;

       for k = 1:height(dataSub)

           c = [0.75 0.0 0.13]; if dataSub.Source(k) == "Control", c = [0.6 0.6
0.6]; end
```

```
            bar(k, dataSub.mean_PathLength(k), 'FaceColor', c);
            errorbar(k, dataSub.mean_PathLength(k), dataSub.std_PathLength(k),
'k', 'LineStyle', 'none');
        end

        % --- FORMATTING FOR READABILITY ---
        set(gca, 'XTick', 1:height(dataSub), ...
                 'XTickLabel', lbls_full(1:height(dataSub)), ...
                 'XTickLabelRotation', 45, ...
                 'FontWeight', 'bold', ...
                 'FontSize', 12); % Larger axis numbers
        % Apply the title specifically to the Eyes Closed condition
        if strcmp(thisMode, 'eyes_close')
            tString = 'Average Center of Pressure Path Length (Eyes Closed)';
        else
            tString = 'Average Center of Pressure Path Length (Eyes Open)';
        end

        title(tString, 'FontSize', 20, 'FontWeight', 'bold'); % Large, Bold
Title
        ylabel('Path Length (m)', 'FontSize', 14, 'FontWeight', 'bold');
        grid on;
    end
end
%% 5. OVERLAID STABILOGRAM
target1 = 'RF_NoShoe_eyes_open';
target2 = 'RF_WtihBraceSpring_NoShoe_eyes_open';
idx1 = find(contains(rawResults.FileName, target1), 1);
idx2 = find(contains(rawResults.FileName, target2), 1);
if ~isempty(idx1) && ~isempty(idx2)
    try
```

```
        s1 = load(rawResults.FileName(idx1)); fv1 = fieldnames(s1); [x1,y1,l1] =
calculateCOP_Local(s1.(fv1{1}));

        s2 = load(rawResults.FileName(idx2)); fv2 = fieldnames(s2); [x2,y2,l2] =
calculateCOP_Local(s2.(fv2{1}));



        % Robust limit calculation

        maxVal = max([abs(x1); abs(y1); abs(x2); abs(y2)]);

        if isempty(maxVal) || isnan(maxVal) || maxVal == 0, maxVal = 0.05; end

        lim = maxVal * 1.1;

        figure('Name', 'Overlay Comparison', 'Color', 'w', 'Position', [200,
200, 800, 800]);

        hold on;

        p1 = plot(x1, y1, 'Color', [0.2 0.6 0.8], 'LineWidth', 0.7);

        p2 = plot(x2, y2, 'Color', [0.8 0.2 0.2], 'LineWidth', 1.2);

        axis equal; grid on;

        xlim([-lim lim]); ylim([-lim lim]);

        xlabel('ML Sway (m)'); ylabel('AP Sway (m)');

        title('Stabilogram Overlay: No Shoe vs. Spring AFO');

        legend([p1, p2], {['No Shoe: ',num2str(l1,'%.3f'),'m'], ['Spring AFO:
',num2str(l2,'%.3f'),'m']}, 'Location', 'northeast');

    catch ME

        fprintf('Error generating overlay: %s\n', ME.message);

    end

end

%% LOCAL FUNCTIONS

function [finalX, finalY, pLength] = calculateCOP_Local(data)

    p5 = data.Force(5); p6 = data.Force(6);

    f5z = abs(p5.Force(3,:)); f6z = abs(p6.Force(3,:));

    totalFz = f5z + f6z;

    valid = totalFz > 20;

    m5 = p5.Moment; m6 = p6.Moment;

    cx5 = -m5(2,:) ./ totalFz; cy5 = m5(1,:) ./ totalFz;
```

```
    cx6 = -m6(2,:) ./ totalFz; cy6 = m6(1,:) ./ totalFz;
    netX = (cx5 .* f5z + cx6 .* f6z) ./ (totalFz + eps);
    netY = (cy5 .* f5z + cy6 .* f6z) ./ (totalFz + eps);
    finalX = netX(valid) - mean(netX(valid), 'omitnan');
    finalY = netY(valid) - mean(netY(valid), 'omitnan');
    pLength = sum(sqrt(diff(finalX).^2 + diff(finalY).^2));
end
```