The Design of a Soft, Passive Ankle Device for Providing Assistance During Walking

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Abstract— **This paper introduces a novel wearable ankle device that provides assistance to a human user while walking. The device has two main components: a pneumatic artificial muscle (PAM) that provides assistive torque in the sagittal plane, and an air reservoir under the foot that, when stepped on, causes the PAM to actuate. Therefore, this device is passive and does not require any air compressor or power source to actuate, and the actuation timing is determined by the placement of the reservoir under the foot. A prototype of this device was modeled, fabricated, and then tested in an experiment that aimed to determine if the device could provide assistance while walking. The preliminary study on one subject demonstrated strong potential for the device, since it was shown that the actuator successfully provided dorsiflexion assistance. There was also an indication of a slight reduction in muscle effort required for dorsiflexion. The results also indicated design changes, such as the reservoir placement and design, for future iterations of this device. Overall, the device demonstrated potential for being a lightweight and low-cost device that could be more accessible than current state-of-the-art robotic devices, but more functional than commonly used ankle foot orthoses.**

I. Introduction

In robotics, the term "soft" typically refers to highly compliant materials, closely resembling those found in living organisms. The field of soft wearable robotics is a new and rapidly growing area of research. This field offers the opportunity to wear robots like garments or accessories to assist the movement of specific parts of the body. The use of soft materials provides an advantage with human motion by minimizing restrictions to the wearer and eliminating the need to carefully align a rigid robot with biological joints, making soft materials important to the development of robotic systems that are more comfortable and lower cost alternatives compared to their rigid counterparts [1].

The nature of these materials allows these soft robots to be mechanically biocompatible and capable of lifelike functionalities. From biologically-inspired field robots for exploration to soft, lightweight cooperative robots that safely interact with people, the applications are countless. Increasingly common applications include those involved with lower limb assistance.

II. Background

Moving the human foot requires flexing the muscles in the calf, ankle, and the foot itself. These common actions are crucial for tasks like walking, but can be difficult for people who develop muscle weakness or those with chronic ankle instability. To aid in recovery for those with disabilities, wearable robotics are becoming a viable option to provide assistance to people while walking [2]. Such robotic devices have also found non-rehabilitative applications, such as in industrial and military settings to augment the strength and/or endurance of healthy individuals [3]. Many of these wearable robotic devices are focused on providing assistance to the ankle joint, since this joint directly interfaces with the environment and significant muscle effort is exerted to control the ankle while walking.

Before considering state-of-the-art wearable robots for walking, it is important to recognize that most people who require assistance for walking are not currently using wearable robots due to their high cost and limited commercial availability. Most people requiring assistance for walking are patients experiencing abnormal gait patterns that can arise from a multitude of ankle injuries commonly derived from neuromuscular disorders, sports injuries, and ageing. For patients with such issues, the ankle's range of motion is heavily influenced, restricting patients' dorsiflexion and plantarflexion when walking. An example that is caused by neuromuscular disorders or injury is drop foot. Weakening in the anterior muscles in the shank causes difficulty in dorsiflexion in patients, hindering their ability to walk unassisted [4]. Attending physical therapy or utilizing an assistive device are the main treatments for drop foot and similar disorders. Physical therapy provides an effective treatment, but restricts a patient to attend in-person while also being cumbersome financially for many. The other solution would be assistive devices: passive and active ankle orthoses.

The main passive devices for walking assistance are ankle-foot orthoses (AFOs). They restrict the ankle to a neutral position of about 90° to correct the ankle position and limit the ankle's range of motion. AFO's are rigid wearables that usually require physical therapy with its use. Overuse of an AFO causes dependence in day-to-day life. Such AFOs have also been shown to cause atrophy and muscle weakening [1,5].

Modern devices have been developed in recent years to support lower limb motion as they work to emulate muscles in parallel with the human user. These devices often use a form of pneumatic actuator (e.g., McKibben muscles or flexible pneumatic muscles) since they are typically lightweight, easy to fabricate, and are self-limiting, or have a maximum contraction, making them similar to human skeletal muscles.

Many current active solutions combine rigid AFO frames with an pneumatic artificial muscle (PAM). Such devices have been shown to assist in plantarflexion. Although these devices have been shown to reduce metabolic cost, they may not be practical in day to day life. One such device is not intended for use outside of research and another is heavy, but both are restricted by their air compressors [6, 7].

A less rigid approach uses three air muscles to "mimic" human muscles. It is capable of multiple motion assistance including dorsiflexion and eversion, and has a feedforward feedback system, inertial measurement units (IMUs), sensors, that all assist in accurate device usage. However, these components add weight and restrictions as they have to be housed and supported. Although the system can be battery powered, it is restricted to an air compressor [4] as the more rigid devices are.

To avoid these bulky and heavy exoskeletons, an approach using light, soft, and shapely wearable devices is introduced. One such design is a soft, sock-like AFO exosuit with fabric-based, thermally-bonded nylon actuators that is meant to be worn over the user's shoes. The system uses a portable pump and battery to actuate the exosuit, and it is shown to assist with dorsiflexion and aid in natural gait restoration [8].

More complex, state-of-the-art solutions include a full lower body, pneumatically-actuated system with its components stored in a backpack: using sensors and air pressure, specific actuators expand or compress to assist lower body movement [9]. Additionally, a commercially available product called the *Restore Exosuit* by Rewalk Robotics uses motors and cable tension instead of pneumatic actuators, but achieves similar walking assistance [10]. It is capable of three unique modes: assist to provide active assistance, slack to effectively allow normal walking with no interference, and brace, to provide constant stiff cables to act as an AFO. These state-of-the-art solutions tend to be expensive, not practical, and are oftentimes restricted by air compressors.

To balance the trade-off between inexpensive AFOs (which do not provide active assistance) and expensive state-of-the-art robotic devices (which are currently cost prohibitive for widespread use), we propose a soft wearable ankle device that is as inexpensive as simple AFOs, but still provides the dynamic assistance that parallels walking gait that is seen in the more active and expensive solutions. This paper presents the preliminary study of the first prototype of such a device that can provide walking assistance.

III. Methods

A. *Device Concept*

Our device is designed as a closed pneumatic system that is actuated using the human user's weight during the stance phase of walking. By placing a reservoir of air under the user's foot, we can convert the energy of the user pushing their foot downwards on the ground into pneumatic energy that can be used by a PAM. Therefore, the reservoir serves as both the device's power source and sole sensor for determining the gait phase. By placing the reservoir at different locations under the foot, we can tune at what phase of the gait cycle we want the PAM to actuate. The PAM in our device is connected between the user's toes and their shank just below the knee so that actuating the PAM generates torque about the ankle joint.

The PAM used by our device is a low-cost actuator that consists of a latex twisting balloon (like those used to make balloon animals) and the fabric outer lining of an expandable garden hose. When the balloon is put inside the fabric lining and constrained so that ends of the balloon must move with the ends of the fabric lining, the balloon can be filled with air so that the actuator expands in length. The fabric lining surrounding the balloon constrains the expansion of the balloon to cause the PAM to expand in a single degree-of-freedom (DOF). Therefore, the PAM used in this paper is an extensional device, unlike common PAMs (such as the McKibben muscle) which is a contractile device. Another important feature of our PAM is that it will quickly contract if air is not being forced into it. Therefore, only while the reservoir of our device is being stepped on will the PAM extend in length.

While the specifics of the reservoir and actuator design will be described in the next subsection, it is helpful to first have a high-level understanding of the device's design and how it interfaces with the human user during the walking gait. The central goal of the device is to reduce the human user's effort while walking by using the PAM to actuate in parallel with the human user's muscles. More specifically, our device is focused on the muscles used to control the human ankle in the sagittal plane, corresponding to the dorsiflexion (pointing toes upwards) and plantarflexion (pointing toes downwards) movements. While muscles other than those specific to controlling ankle motion in the sagittal plane are important for human locomotion, the ankle muscles controlling motion in the sagittal plane play a critical role in propulsion [11].

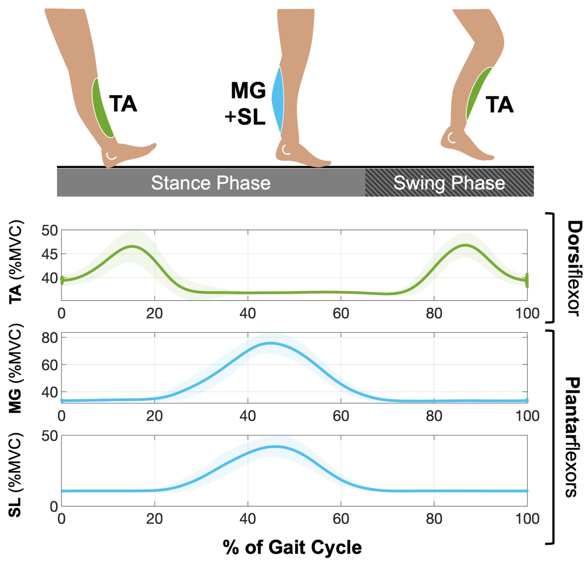


Fig. 1. A visual representation of the human muscles of interest which control motion in the sagittal plane throughout the human walking gait cycle. The three plots show the average, filtered EMG response as a percent of MVC for a single human subject walking barefoot on a treadmill.

The strongest muscles controlling movement of the ankle in the sagittal plane are the tibialis anterior (TA) which allows the ankle to move in the dorsiflexion direction and the soleus (SL) and medial gastrocnemius (MG) which allow the ankle to move in the plantarflexion direction. Our assistive device aims to actuate at the correct times in the gait cycle to provide assistance in these directions of motion. Therefore, the information of when these muscles are contracting, and therefore injecting energy into the system during walking, is critical to the design of the device which aims to use its actuator in parallel with the human muscles.

Previous research has demonstrated that providing assistance at the wrong times will cause the user to expend additional effort compensating for what the body feels is a disturbance [12], so actuating the device at the correct times during the gait cycle is important. The aforementioned muscles and plots of their respective muscle activations throughout the gait cycle can be seen in Fig. 1. Coupling this information with the previous description of the PAM connected between the toes and shank of the human user, the actuator should be inflated during the later half of the stance phase so that it promotes plantarflexion at the appropriate time. Furthermore, when the actuator is not inflated, its length and placement can be selected so that the actuator pulls the ankle upwards in dorsiflexion during the swing phase and at the beginning of the stance phase. To achieve this timing of the actuator, the reservoir can be placed at the ball of the foot to effectively "sense" when the user is in the later half of the stance phase. A diagram of the device's concept is shown in Fig. 2.

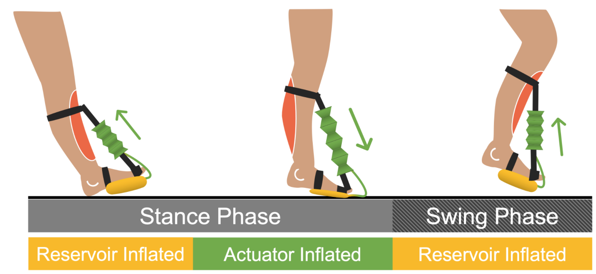


Fig. 2. Visual representation of the concept of the device. The reservoir is indicated in yellow, the actuator (PAM) in green, and the human muscle in red. Arrows are used to indicate the direction in which the actuator is pulling/pushing the toes.

B. *Modeling and Fabrication*

For the concept described above to work in practice, there are a number of important design questions that we considered. First, the length of the actuator needed to be determined so that it could pull the ankle up in dorsiflexion when deflated and push the ankle downwards in plantarflexion when inflated. Second, the design of an air filled reservoir to actuate the actuator needed to be determined.

First, the methodology for selecting the actuator length will be described. The stretchable, latex balloon within the actuator allows the actuator to generate a significant contractile/pulling force when the actuator is pulled from its equilibrium position. When the balloon is filled with air and the actuator extends, the equilibrium position (determined by the length of the actuator) and pulling force change. Therefore, the PAM can be modeled as a variable stiffness actuator in that the stiffness magnitude and stiffness equilibrium position change as a function of the air inside of the actuator. Since our device is designed so that the actuator quickly transitions between the inflated and deflated states, we will model the length and stiffness at these two states of the actuator only.

The length of the actuator was determined by the natural range of motion of the human ankle in the sagittal plane. Considering the shank and foot to be 90 degrees from one another in the neutral position, the ankles range of motion has a maximum dorsiflexion of 20° upwards, and a maximum plantarflexion of 50° downwards [13]. The ankle, shank, and actuator form a triangle with known foot and shank lengths, as well as a known angle between them. From the shank to the foot, maximum dorsiflexion would have a 70° angle, and maximum plantarflexion would have a 140° angle. Using the law of cosines, the ideal minimum and maximum lengths of the actuator can be determined based on the actuator's attachment point on the leg and the foot length. Equation (1) determines the ideal minimum length of the actuator, l_{min}, which corresponds to the unstretched neutral length, and (2) calculates the ideal maximum actuator length, l_{max}, when filled with air and stretched.

$$l_{min} = \sqrt{h^2 + l_f - 2h(l_f)cos(70°)}$$

$$l_{max} = \sqrt{h^2 + l_f - 2h(l_f)cos(140°)}$$

where h is the actuator's attachment point and lf is the foot length. With the minimum length found in (1), and the maximum length found in (2), an actuator of the ideal length could be fabricated for any size human user. Since these are ideal values, it may not be feasible to design an actuator that perfectly transitions between l_{min} and l_{max} when it is filled with air, but these equations describe the geometric relationship between the size of the user and the ideal length of the actuator which can be used to inform the fabrication of the actuator.

Next, the stiffness of the actuator can be characterized experimentally. At both l_{min} (when the actuator is deflated) and at l_{max} (when the actuator is inflated), the actuator will have different stiffness properties caused by the change in tension of the latex balloon when it is unfilled versus filled with air.

An experiment was performed to model the stiffness of the actuator. A simple linear stiffness model, Hooke's Law, (3) was used to determine the spring constant of the actuator while filled and unfilled with air [14].

$$F = - kx \tag{3}$$

where x represents the spring's displacement and k is the spring constant. This equation can be converted into the equilibrium expression specific to the experiment (4):

$$W = kx \tag{4}$$

where W is the weight of the applied mass at the end of the actuator. The spring constant can be determined by plotting the applied weight and the spring's displacement measurements against each other with the spring constant, k, being the slope of the plot. Fig. 3 shows the setup for measuring the spring's displacement when a range of weights is applied. Weight was applied to the actuator in 100 gram increments, then the change in length was measured in inches and compared to the initial actuator length.

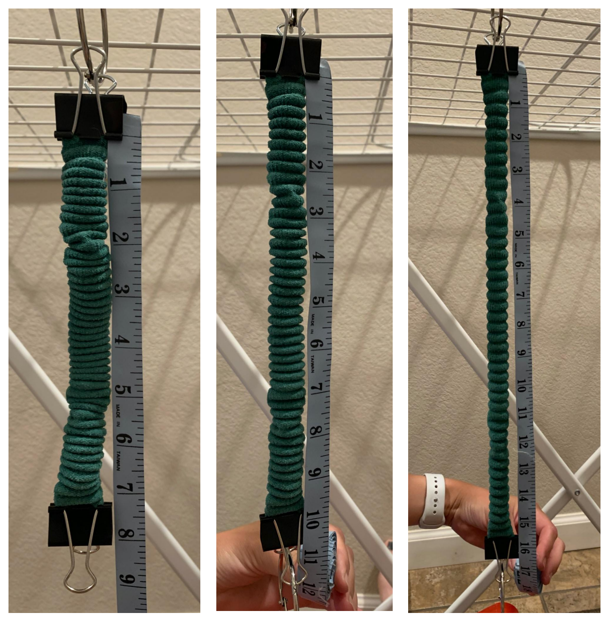


Fig. 3. Experimental setup to model the stiffness properties of the actuator. Three different conditions are shown demonstrating the change in the length of the actuator.

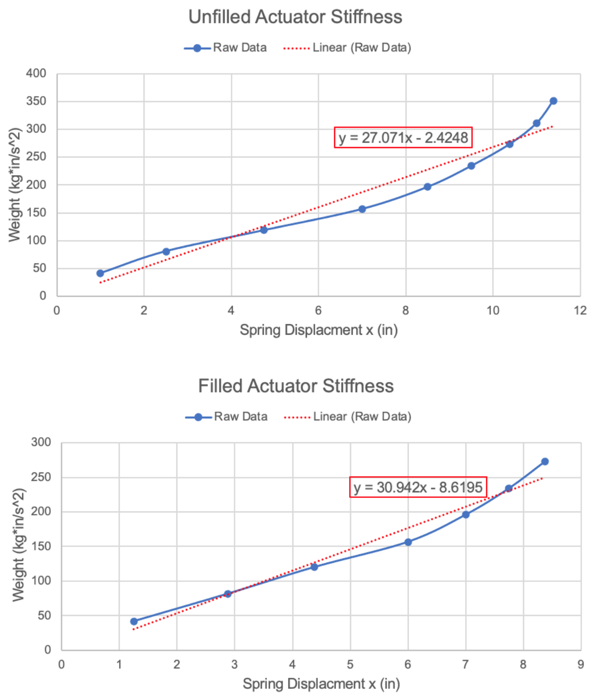


Fig. 4. Plots showing the experimentally collected data used to calculate the stiffness of the actuator when filled and unfilled with air. Blue points are the raw data collected in the experiments, and the red line shows the linear fit.

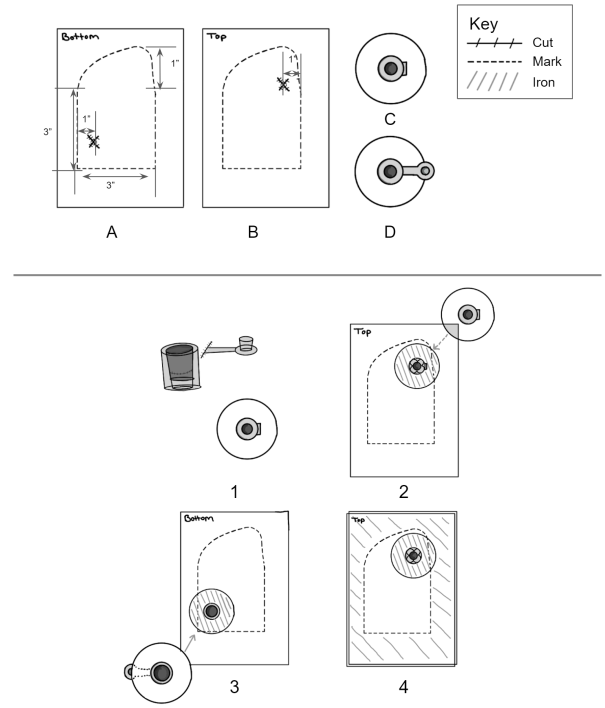


Fig. 5. Reservoir parts for assembly and manufacturing steps. Two vinyl marked cut outs, A and B, and two inflatable valves, C and D, are needed to manufacture the reservoir. The four manufacturing steps are listed and shown (1-4). The dimensions shown in the figure are specific to the size of the subject's foot.

Plots of the weight and spring displacement are displayed in Fig 4 where a linear equation has been fit to the data. The slope of this line is an estimate of the spring constant, k, of the actuator. The k value of the actuator when unfilled with air was ~27 N/m and the k value of the actuator when filled with air was about ~31 N/m. These values show the expected results that the actuator stiffens as it is filled with air, but is still considerably stiff in its deflated state.

With the actuator parameters described and modeled, the next step was to consider the design of the reservoir. The amount of air required to fill the actuator was found experimentally by fully inflating the actuator and collecting the air into a measuring cup underwater. The final reservoir design, as seen in Fig 5, was crafted with rectangular vinyl cut outs, taken from inexpensive rafts, and circular valves cut from cheap inflatables.

To manufacture the reservoir, two square vinyl cut outs (A and B) were marked with the appropriate actuator size, and an x shaped slit is cut through it to easily adhere the valves (C and D) to it. To combine parts A-D, four simplified manufacturing steps are shown in Fig 5. Step 1 is to cut the plug arm from valve C as well as cutting out the inner plastic piece to allow for continuous air flow (note, valve D is unaltered). Step 2 is to push the top of valve C through the bottom of B using the x shaped slit, and then to iron the materials together. Step 3 takes valve D and puts it through the x-shaped slit on top of B so the valve faces downward. It is then ironed together. Step 4 then irons A and B together forming a reservoir with a valve coming out the bottom to add air in, and a valve coming out the top to actuate the actuator. Following these steps ensure that the reservoir stays airtight, can withstand the weight of the user, and contains intuitive valve placement to fit the final device design.

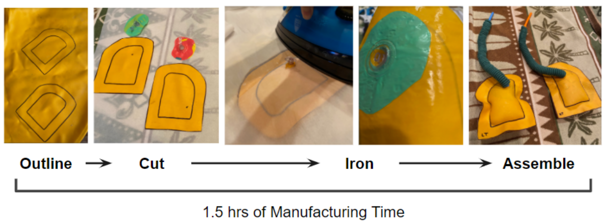
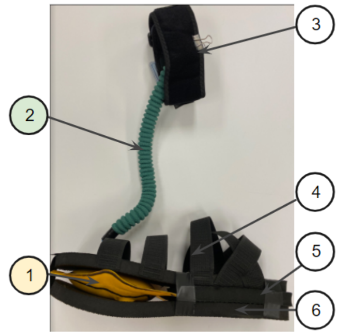


Fig. 6. Depiction of reservoir manufacturing process. The images shown are from the refined manufacturing process, with outlining, cutting, ironing and assembling two working reservoirs taking around 1.5 hours.



Bill of Materials	
#	**Part Name**
1	Reservoir
2	Actuator
3	Velcro Strap
4	Modified flip flop with elastic straps
5	Wedge spacer
6	Modified flip flop base

Fig. 7. Full device assembly with bill of materials. Six major components of the device are listed. The reservoir, the actuator, a velcro strap to secure the actuator to the leg, a flip flop to attach the device to the foot, a cut flip flop to help space out the reservoir, and a flip flop to act as the device's sole.

Photographs of the reservoir manufacturing process can be seen in Fig. 6, showing the main steps and time it takes to manufacture two working reservoirs. With around 1.5 hours of time to create them, it allows for rapid prototyping for future iterations of the device.

With the reservoir and actuator constructed, the system was designed using altered flip flops, elastic straps, and a velcro strap. As seen in Fig 7, the device consists of 6 main components. *Part 1* and *Part 2*, the reservoir and actuator respectively, have been previously described. They act as a closed loop system that transfers air at specific points in the gait cycle to provide the necessary dorsiflexion and plantarflexion assistance. *Part 3*, the velcro strap, is a common tennis elbow orthotic used to attach the actuator to the shank just below the knee. *Part 4* is a custom modified flip flop with the strap removed from the sole. Elastic straps were added instead to anchor the foot to the sole and prevent the heel from lifting during walking. *Part 5* is a flip flop sole cut to act as a wedge spacer so the reservoir does not hinder walking, encouraging reservoir compression towards the end of the stance phase. Lastly, *Part 6* is a flip flop sole to create a steady base to walk on, protecting the reservoir, and the user from harmful inversion and eversion of the foot.

Important device features are highlighted in Fig 8, labeled A-C. *Feature A* is a square cut out of the base flip flop to allow for the bottom reservoir valve to be easily accessible. A small hand pump could then be used to inflate the reservoir if needed. The valve then sits flush with the bottom of the flip flop base ensuring it does not inhibit walking. *Feature B* is an elastic strap to curve the base flip flop upwards to prevent it from dragging or tripping the user. Lastly, *Feature C* takes the top reservoir valve through the hole left behind from the flip flop strap to actuate the actuator. A small piece of tubing to connect the valve and the actuator acts as the flip flop strap and rests comfortably between the user's toes.

Both a left and a right shoe were created to run experiments and test how effective the device is in providing assistance during walking.

C. *Experimental Protocol*

We performed a preliminary study that aimed to determine whether or not the device could reduce the effort of the human user while walking, and to quantify the effect of the device on the user's gait. Since the device could only be worn comfortably by a single person available to participate in our study, one subject's data from the experimental protocol is presented in this paper.

One, healthy female subject (age: 24) was selected as the preliminary subject for this study. The device used in the experiment was designed to fit this subject. From (1) and (2) and measurements of the subject's leg and foot length, an actuator of uninflated length (l_{min}) was selected to be 0.23m. The volume of air was found to be ~100mL for the given actuator length. Therefore the reservoir was designed to hold at least 100mL of air when fully inflated. The width and length of the reservoir, shown on A and B, in Fig. 5 are meant to fit under the subject's foot (US women's size 9 shoe), and hold approximately 100 mL of air to actuate the actuator.

The study required the subject to walk on a split-belt treadmill (Bertec Treadmill, Columbus, OH, USA) (Fig. 9A) in three different experimental conditions: baseline, passive, and active (Fig. 9B). The baseline condition was before the device was put on the subject. The passive behavior was when the device was placed on the subject but without the actuator connected below the knee. This condition meant that while the air in the reservoir was still able to move into the actuator, the device was not providing any torque about the ankle. Finally, the active condition was when the device was placed on the subject with the actuator connected and provided assistive torque about the ankle joint. As the subject walked in the active condition, the actuator would pull the foot upwards in dorsiflexion when it was not inflated, then as the reservoir was stepped down on at the end of the stance phase, the actuator filled with air and allowed the foot to plantarflect without having to overcome the stiffness of the actuator in its deflated state.

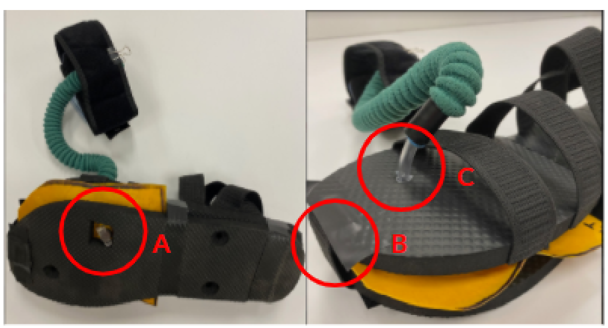


Fig. 8. Important device features. Features A-C display necessary device design points through easy access to the reservoir valve (A), an elastic strap to prevent tripping (B), and the actuator attachment to reservoir (C).

During all three experimental conditions, ankle angle data in the sagittal plane was collected by an electro-goniometer (Biometrics Ltd., Newport, UK). The goniometer was placed along the bottom of the ankle and captured data for dorsiflexion and plantarflexion. Electromyography (EMG) data was collected by surface EMG sensors (Delsys Inc., Natick, MA, USA) (Fig. 9C). The EMG sensors were placed on four muscles important for ankle motion: the tibialis anterior (TA), soleus (SL), medial gastrocnemius (MG), and peroneus longus (PL). As mentioned previously, the TA allows for ankle movement in the dorsiflexion direction, the SL and MG are for the plantarflexion direction, and the PL is for the eversion direction. While assisting motion in the frontal plane (eversion and inversion directions) is not a goal of our device, the PL EMG data was useful for understanding how the design of the device affects user's stability in the frontal plane. The PL data is not presented in this paper since it did not show any significant differences between the three experimental conditions.

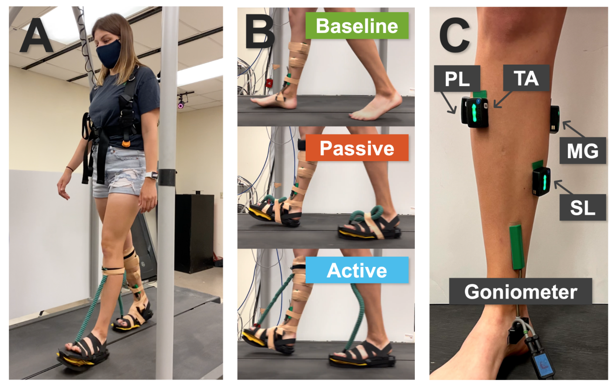


Fig. 9. The experimental set-up and conditions. **A**: A full view of the subject walking on the split-belt treadmill while wearing the device on both feet and sensors on the right leg. The subject wore a safety harness which did not bear any weight. **B**: Photos taken during the experiment in each of the experimental conditions. **C**: A close-up view of the sensors on the right leg. The four EMG sensors are indicated by the muscle that they correspond to, and the goniometer can be seen connected to the ankle.

Prior to the experiment, the subject was required to perform a maximum voluntary contraction (MVC) test whose data was used to scale the collected EMG data. The MVC test was performed based on typical muscle testing guidelines [15, 16]. With this collection of data, we were able to find the maximum activation specific to the subject so we could calculate the results in terms of a percent muscle activation compared to the maximum the subject was able to perform. Ankle position data was also collected with the ankle at known angles in the sagittal plane, which was used to convert the raw goniometer voltage signal to position data.

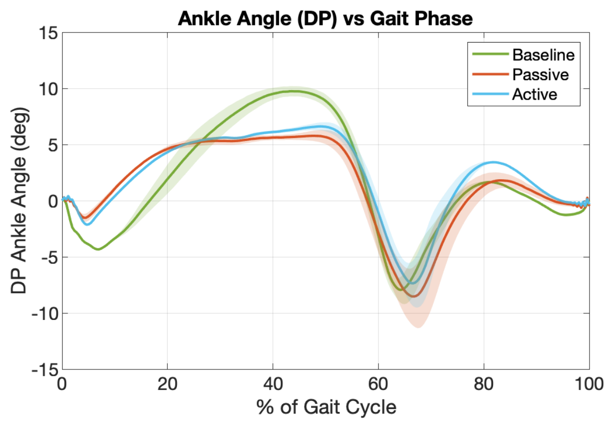


Fig. 10. The filtered angle position data in degrees over the gait cycle, with 0% corresponding to the heel strike. The darker, solid lines represent the mean of the sensor data for each stride and the lighter color around the line is the mean ±1 standard deviation (STD).

D. *Data Analysis*

The data collected from the experiment needed to be processed in order to make any conclusions about how the device impacted the subject's gait. Since only a single subject's data was collected for this preliminary study, the results were analyzed by looking directly at plots of the subject's ankle position (collected by the goniometer) and muscle activation (collected by the EMG sensors) for the three experimental conditions. From these plots, we could understand how the device impacted the subject's position and muscle activation responses throughout the gait cycle.

Before the plots could be generated, an important consideration was how to filter and scale the data collected from the goniometer and EMG sensors. The filters for both the goniometer and EMG sensor data were selected based on previous work on designing active AFOs for providing assistance while walking [17]. The goniometer data was filtered using a 2nd order Butterworth low pass filter with a cutoff frequency of 10 Hz and fit to calibration data. The EMG data was demeaned, rectified, and filtered using a 2nd order Butterworth low pass filter with a cutoff frequency of 5 Hz, and scaled by MVC. By scaling by MVC, the voltage information collected from the EMG sensors could be considered as a percentage of the subject's highest possible effort for each muscle.

The force data collected from the split-belt treadmill was used to identify the heel strike of each stride for the right leg. With this information, the plots of the average ankle position and muscle activation could be created by overlapping the data corresponding to each stride of the gait cycle. These plots are in the following section, and are used to understand the effect of our device on the subject's gait.

IV. Results and Discussion

The effect of the device on the subject's gait will be considered from the average position results for the subject, shown in Fig. 10. The baseline results show the expected curve for the subject's ankle angle over time. When this curve is compared with both the passive and active conditions, it is clear that there are some differences in the subject's ankle angle throughout the gait cycle. However, some of the effects can be attributed to the design of the reservoir and shoe components of the device, while other effects can be attributed to the actuator/PAM.

First considering the position results during the stance phase (between 0 and 60% of the gait cycle) there is a strong indication that the design of the reservoir and surrounding shoe caused a difference in how the subject's ankle was able to move. Compared with the baseline condition, both the passive and active conditions show similar results during the stance phase, with the subject moving their ankle less in both the dorsiflexion and plantarflexion directions. This result can be attributed to the wedge shape of the shoe which likely caused the subject to rely on the shape of the shoe rather than the bending of their ankle to transition between heel strike and toe off. The addition of a shoe to a person's foot will inevitably change their ankle movement, but this result gives some indication that a goal of future designs should be to make the shoe and reservoir thinner and flatter.

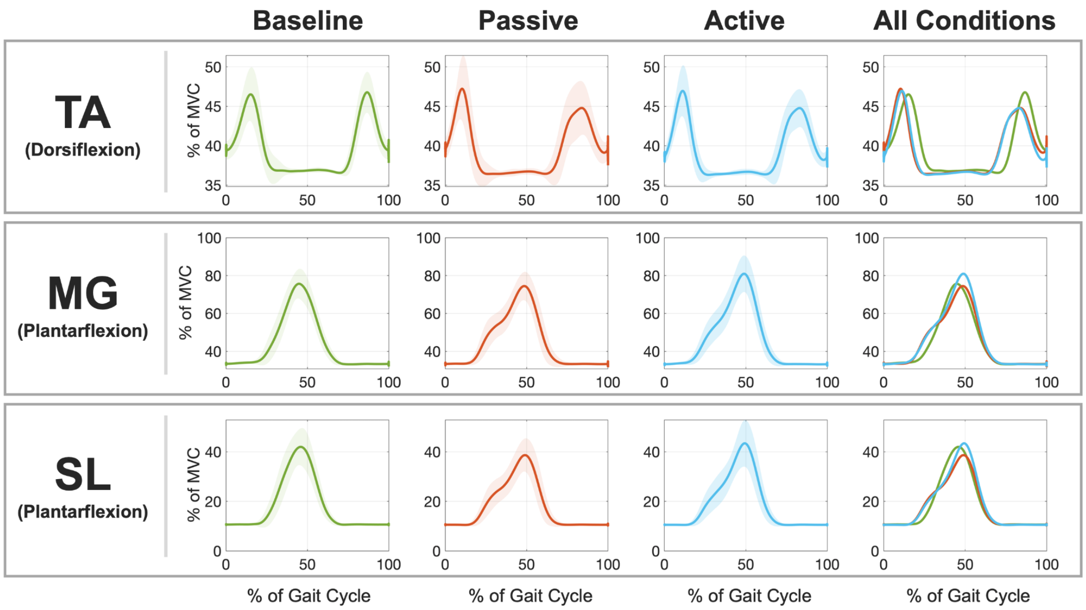


Fig. 11. The filtered EMG data in % MVC over the gait cycle, with 0% corresponding to the heel strike. The darker, solid lines represent the mean of the sensor data for each stride and the lighter color around the line is the mean ±1 standard deviation (STD).

Next, the position results during the swing phase (between 60 and 100% of the gait cycle) indicate that the subject's ankle movement during swing was affected by the actuator. While the baseline and passive results follow similar curves during the swing phase, the active condition results demonstrate that the actuator did provide torque about the ankle joint in the positive direction during the swing phase. For applications like helping those with foot drop, this result is promising as it demonstrates that the stiffness of the actuator is substantial enough to pull the toes up in dorsiflexion during the swing phase.

Now, the muscle activation results collected by the EMG data will be discussed. The plot in Fig. 11 shows these results separately and overlapped for each condition and muscle of interest. Due to the inherent noisiness of EMG data and high variability, it is difficult to make any strong, quantitatively-backed conclusions, but there are some important differences seen from the muscle activation results that will inform future designs.

First, the muscle activation results for the TA muscle indicate an effect of the shoe design on when the subject was activating their TA muscle and a possible reduction of effort caused by the actuator. Comparing the baseline results with those of the passive and active conditions, there is a noticeable time shifting of when the TA muscle is activating that was likely caused by the shoe design. Most likely, the thickness of the shoe changed the time when the subject's foot was in the air versus on the ground, therefore shifting the times when the subject would dorsiflect. The other result of the reduction in effort caused by the actuator can be seen near the end of the swing phase where the active condition curve is lower than the curves of the other conditions. While this reduction in effort is low, this information, coupled with the position results previously discussed, is a positive result showing that the device allowed for an increase in dorsiflexion motion but at a reduced effort. It is also possible that if the subject were given more time to acclimate to the device during before the active condition experiment, there would be a greater reduction in TA effort and a smaller increase in motion in the dorsiflexion direction.

Finally, the muscle activation results for both the MG and SL muscles indicate that there may be a slight increase in effort caused by the actuator. As mentioned previously, the goal of our device is to provide dorsiflexion assistance during the swing phase, but then to allow the user to plantarflect properly during the end of the stance phase when the actuator fills with air and becomes longer. The slight increase in muscle activation seen in the active condition results indicates that the current placement of the reservoir under the foot is not ideal. More specifically, it appears that while the actuator was long enough when inflated to allow for uninhibited plantarflexion motion, the actuator may not have been inflating soon enough in the gait cycle, and therefore the subject had to overcome the stiffness of the deflated actuator for a short period of time, causing an increase in muscle activation.

Taken together, the results demonstrate the device being effective in providing dorsiflexion assistance during the swing phase, but also indicate some changes to the design to make the system more transparent during the stance phase and actuate at the correct time. These preliminary results indicate that designing a thinner, flatter shoe with the reservoir further back on the foot would be a good candidate for the next design iteration of our device.

Some comparisons can be made between these results and the results of other devices that have been designed for helping people walk. Previous research has shown that non-robotic AFOs (like those for foot-drop) can often provide assistance for one direction of motion, but at the expense of requiring an increase in effort in the opposite direction [18]. By improving the reservoir placement of our device, we could alleviate this trade-off by changing the actuator's length and stiffness during the correct time of the stance phase.

Like state-of-the-art, robotics solutions for helping people walk [4, 8-9], our device was able to use a PAM in parallel with the human muscles to actuate at certain times by identifying different gait events. However, our device did not require any expensive sensors, power sources, or heavy equipment for the user to carry, so our device is much more lightweight, practical, and accessible than current state-of-the-art robotic devices.

V. Conclusion and Future Direction

Our inexpensive and accessible design of a pneumatic system that is actuated by the user's weight worked to assist the user's dorsiflexion while walking. The device parallels the robotic and dynamic devices seen on the market and in recent research, but at a lower cost to users. This design can be applicable to a wide range of markets whether it is utilized by people who hope to reduce the efforts needed while walking, people who need extra assistance with dorsiflexion because of an injury, or even space exploration applications. The design presents a solid baseline with many different applications within different industries.

The testing and results highlighted some design flaws that should be improved upon for future design iterations. The footbed did not provide proper support throughout the gait cycle as the elastic straps stretched and allowed the wearer's foot to slide. The strap material and layout can be adjusted to reduce slipping and improve the effectiveness of the system in future iterations.

The preliminary results also gave insight into some adjustments that should be made for the reservoir design and placement. Creating a thinner, flatter reservoir that is placed further back under the foot could aid in reducing walking efforts throughout the whole gait cycle compared to the current design that is optimized to reduce dorsiflexion efforts. It would also be beneficial to create a device that can fit a wider range of foot sizes in the future opposed to the current design which was optimized for a single foot size. The current design and any future iterations should be tested and directly compared against other devices on the market.

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