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DEVELOPMENT OF WEARABLE SOFT PNEUMATIC ACTUATORS DESIGNED FOR KNEE SUPPORT

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DEVELOPMENT OF WEARABLE SOFT PNEUMATIC ACTUATORS DESIGNED FOR KNEE SUPPORT

By

Jungwon Lee

A capstone project submitted for Graduation with University Honors

May 7, 2026

University Honors

University of California, Riverside

APPROVED

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University Honors

ABSTRACT

The ability to perform athletic activities is often limited for individuals with lower limb mobility issues due to the sustained physical demands required to maintain bent knee postures for extended periods of time. This research explores the development of a lightweight and wearable soft pneumatic actuator which aims to address this issue. The actuator design consists of an inflatable bladder encased within a fabric sleeve, with the sleeve opening secured by 3D-printed end caps. When inflated, the actuator resists bending to provide supportive assistance, while deflation allows for unrestricted movement. Positioned behind the knee, the actuator aims to deliver adaptive support that enables users to comfortably maintain physically demanding postures for longer durations. The performance of the actuator will be characterized using a custom test setup. This setup includes a force sensor that will accurately measure the force generated by the actuator at a range of different pressures and angles. Testing various sizes, shapes, and materials will allow us to choose ideal designs for certain desired characteristics. Future works will include further evaluation which will be done by analyzing the body position changes of a user during a sustained squatting task in order to compare the exerted effort of a user with and without the actuator assistance. The ultimate goal of this research is to enhance accessibility and participation in physically demanding activities for individuals with lower limb mobility impairments.

ACKNOWLEDGEMENTS

I would like to express my deepest gratitude to my faculty-mentor, Dr. Jun Sheng, for his mentorship and support throughout this project. Doing research in the RaMS lab has given me valuable experience as I begin to pursue a career through internship and job opportunities as a mechanical engineer. Thank you for guiding me throughout this capstone project. Your expertise and feedback were invaluable for my research experience.

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INTRODUCTION

Soft pneumatic actuators are a type of actuator that is flexible and compliant in nature. This makes them ideal for wearable technologies as they provide safer and less intrusive methods of adaptive body mobility support. Due to these desirable characteristics, the development of soft actuators in wearable medical devices has seen a rapid increase in recent years. Considering most of the previous wearable devices used for research or medical applications have been based on more rigid mechanical framed designs, the growing research related to wearable soft actuators is not surprising. Previous works in medical mobility assisting soft pneumatic actuator devices by Pan et al. have already proven to provide superior flexibility and weight making them much preferred over their competition [1]. Other notable works in the wearable soft pneumatic actuator space include works done by O'Neill et al. and their textile based straightening wearable pneumatic actuator specifically engineered for shoulder mobility rehabilitation purposes [2]. The research outlined in this paper also explores the development of a similar straightening fabric based pneumatic actuator designed for lower limb mobility support around the knee joint. In this work, the basic actuator design will be characterized across multiple design iterations to determine important parameters and ideal configurations to maximize performance.

DESIGN METHODOLOGY

This project begins with the fabrication of a soft pneumatic actuator and the development of a lower limb integration method to make the design wearable. However, in order to start designing soft pneumatic actuators for wearable applications, it is necessary to first collect background knowledge of preexisting wearable devices to understand and be aware of the potential strengths and limitations these technologies have. Because this research is concerned with lower limb mobility assistance around the knee, actuators with the ability to adapt to the

user's body proportions and desired actions, while also providing adequate resistance against bending are ideal.

Most soft wearable robots (SWR) focused on lower limb mobility assistance and rehabilitation, fit into the category of soft pneumatic actuating devices. This is because of the performance and light weight properties that air pressure can provide for wearable applications. The two main actuator types for joint mobility devices that this project is concerned with are bending and straightening actuators. There are a variety of soft bending actuator designs. These designs are often constructed with a combination of elastic and rigid type materials to initiate singular or multi axial bending as seen in the actuator design from Capello et al. [3]. Their soft actuator design uses a two layered fabric sleeve design with one side of the sleeve being knit fabric and the other side being woven fabric. These two fabric types react differently to stretching which causes the actuator to bend to one side. The knit type fabric expands while the woven fabric retains its form. This combination of stretch and strain properties within the fabric of the actuator generates a bending force when air pressure is applied to expand the cavity. Other bending type actuators rely on the compression properties of certain geometrical structures when creating negative pressure inside of the actuator cavity, as seen in the works of Zheng et al. [4]. Both of these works related to the development of bending type soft actuators are either meant for only bending, or a combination of bending/straightening sequences for body joint and limb mobility. Although the actuator being described and developed in this paper is specifically designed for knee straightening assistance during squatting positions, much can be learned from the material selection and fabrication methods used for the fabrication of these bending actuators.

When considering the design for lower limb mobility assisting wearable devices, there are a number of constraints and limiting factors that need to be considered. For most average

healthy adults, roughly 80-85% of body weight is located above the knee joint. Because of the considerable weight above the knee, wearable devices designed for lower limb mobility need to have significant output force while retaining a lightweight and comfortable form factor. The straightening force output requirements of an actuator also depend on the specific application and purpose of the device, much of which are outlined by Thalman et al. [5]. For example, wearable devices that are designed to move the leg with limited assistance or even resistance from the user, require much higher force outputs from the actuator when compared to devices that are meant for supplemental mobility assistance during common knee bending and straightening sequences like walking or squatting. However, because of the high baseline force output requirements for lower limb mobility assistance, many wearable soft actuator devices designed for knee joint mobility use multi bladder wrinkle actuator designs. Some examples of these actuators can be seen in the works done by Park et al. and Kabir et al. [6] [7]. The actuators described in these two works use similar actuator design structures with an array of identically stacked air pocket cavities that expand vertically in one direction when pressurized. Although this design has high force and torque output, they suffer from two major issues. First, the large cross sectional area of each actuator bladder unit makes the device protrude a considerable amount from the leg which adds significant excess bulk. This large form factor isn't an issue for basic medical rehabilitation purposes that don't require dynamic motion or outdoor use. However, for individuals who require day to day mobility assistance while participating in athletic activities, the large form factor of these actuators becomes a major problem. The second issue with the multi bladder wrinkle actuator design is the inflation/deflation time and consistency. Because the actuator needs to act as a singular unit, it is really important for each bladder to pressurize and depressurize as expected. Also, since each bladder is stacked vertically

along the leg, the circulation of air between cavities needs to be consistent and reliable. There is a lot of research currently being done to mitigate some of these issues, but there is still work to be done. The straightening actuator presented in this report, will consist of individual air pocket units that can be stacked horizontally across the circumference of the leg to increase the maximum force output. This actuator design not only creates a slimmer profile against the leg, but also allows for easier fabrication and testing, which is ideal for prototyping.

The basic actuator being developed in this research will use a similar singular cavity, two layered textile sleeve design as seen in the works of Capello et al., in combination with the material selection and pressurization straightening behavior of the actuators made by O'Neill et al. [3] [2]. The fabrication process for each actuator will also be similar to the bending actuator from Capello et al. but with stiffening woven fabric on both sides of the sleeve, as well as the addition of removable 3d printed flat end caps at the sleeve opening.

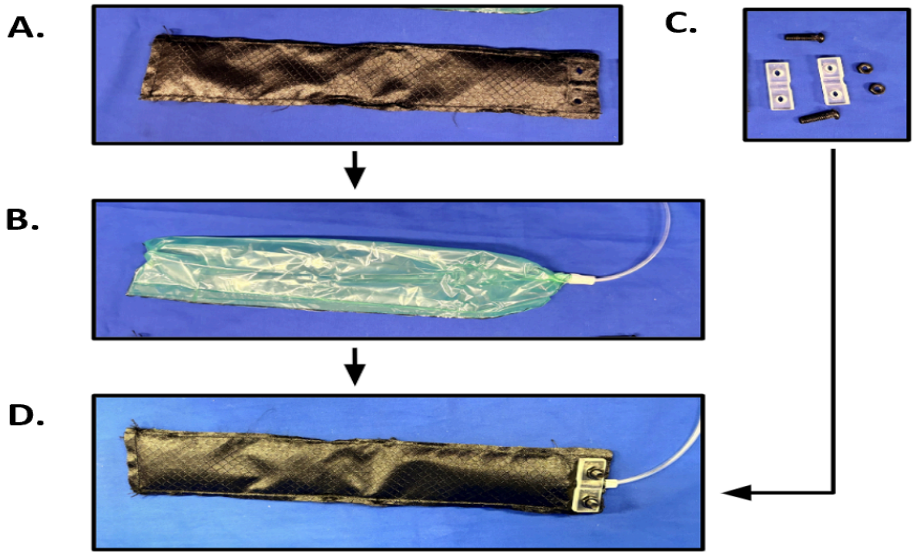


Figure 1: Graphic showing fabrication process of the basic actuator design. **(A)** Rigid woven textile sleeve. **(B)** Airtight TPU bladder with 4mm silicone tube. **(C)** Resin 3d printed end caps with 2 M6 screws. **(D)** Fully assembled actuator.

The basic construction of the soft pneumatic actuator consists of an inflatable TPU tube, rigid outer fabric sleeve, 3d printed resin end caps, and two M6 screws. The fabric sleeve is a simple two layered sheet design with two identical 30 cm by 5 cm strips of rigid woven textile, cut to size with fabric scissors. These textiles were chosen because they have an interlocking fabric structure that makes them bidirectionally stiff. This bidirectional stiffness results in a straightening behavior when the sleeve is stretched. After cutting, both sheets are sewn together along the edge with an opening at the top for an inflatable thermal plastic tube to be inserted later. Two holes were punched equidistant from each other at the opening to provide a secure attachment point for the 3d printed end caps. This fabric sleeve will provide rigidity to the elastic bladder when inflated and prevent it from tearing from pressurization.

Next, as seen in part (B) of figure 1, an inflatable bladder will be made with a thin TPU thermal plastic material and a 4 mm silicone air feed tube. The construction of the TPU bladder is similar to the fabric sleeve in that a sheet of the material will be cut and connected along the edges to create an air cavity. Because the TPU sheet being used is thin, it has limited stretchability and needs to be cut larger than the fabric sleeve it will be inserted into. After observing the expanding properties of the TPU sheets through repeat pressurization testing, a size of 31 cm by 6 cm, which is 1 cm greater on both length and width dimensions of the fabric sleeve, was determined to be ideal. Each sheet was cut to size using a x-acto knife. To create the air tight seal, the TPU sheet was first thermal welded on all edges with a small opening left at one end. A 4 mm silicone air feed tube was then inserted into the opening and the remaining gaps between the tube and sheet were sealed using a combination of CA glue and electrical tape. The purpose of this bladder is to prevent any air leakage during inflation which is vital for

accurately controlling pressurization and maintaining constant pressure conditions inside the actuator.

After the TPU bladder is inserted into the fabric sleeve, 2 identical flat endcaps were made to seal the opening and prevent the bladder from expanding out the top during inflation. The basic design for the 3d printed end caps is a two part, flat rectangular shape with two holes large enough to fit two M6 screws. The final end cap design was modeled on Solidworks (CAD software) and printed using a SLA 3d printer (Saturn 2). SLA 3d printing was used to make the end caps because of the material properties of resin that make it strong against flexing. To finish the assembly of the actuator, the fabric sleeve opening is sandwiched closed by the two flat end caps which are clamped shut using the M6 screws inserted through both the sleeve and the end caps. Although this seal can be created with additional sewn stitching along the opening after inserting the TPU bladder, the end caps allow for easy removal of the bladder in case there are any air leaks or potential design iterations that need to be done. Possible future market production model designs for these actuators would likely not include this endcap as it was only meant for prototyping purposes. The purpose of the 3d printed end caps are to create a tight seal between the TPU tube, fabric sleeve, and air feed tube.

As mentioned previously, multiple of these actuators can be integrated into the leg to act as a singular unit in order to increase maximum resistive force from pressurization. Combination of multiple actuators can be done through stitching two separate actuators together along one side, or cutting a larger sheet of fabric and sewing multiple cavities between the two edges. An important consideration when sewing multiple lines within close proximity to each other is the possibility of weakening the fabric. The process of sewing creates small tears in the fabric that can accumulate if sew lines are close or overlapping. Therefore, when adding multiple air

pockets for the actuator design, a distance of 0.5 centimeters between stitch lines was determined to be an acceptable gap.

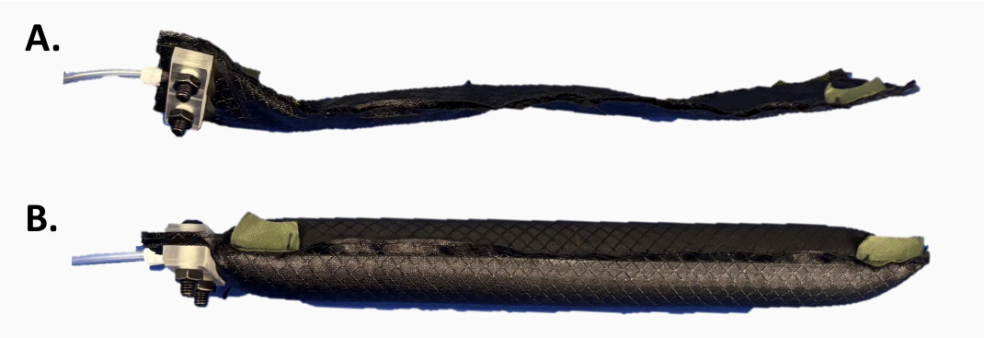


Figure 2: Visual representation of actuator straightening behavior.

With the main actuator design realized, simple characterization of the actuators inflation and deflation behavior can be observed. Figure 2 shows an image representative of the expected straightening behavior described previously. At the deflated state of the actuator in (A), there is little to no resistance against bending. The actuator essentially acts as a piece of fabric at this stage. However, when the actuator is inflated and pressurized in (B), the actuator straightens and there is high resistance against bending. This resistance is expected to increase linearly with a similar increase in pressure. Enlarging the cross sectional area of the actuator at the location of folding/bending is also expected to increase the resistive force output of the actuator.

Since this actuator is meant to be used in a wearable device, an integration method to attach it to the lower limb of a user had to be designed as well. The two main expected issues with integration stemmed from non uniform surfaces of contact and actuator bend location/direction. First, the non uniform contact surfaces between the actuator and the leg posed an interesting problem of slipping. Since the geometry of the human leg is naturally tapered towards the ground, most wearable devices struggle to maintain their original location after repeated movements due to gravitational effects causing slip. A simple strapping design can be

used, but without proper placement, the straps will have to be tightened too much for comfortable use. The second issue related to the bending location/direction of the actuator arises from the inconsistent bending angles and speeds of the knee. After testing multiple design iterations, a simple solution to these problems was a triple strap approach.

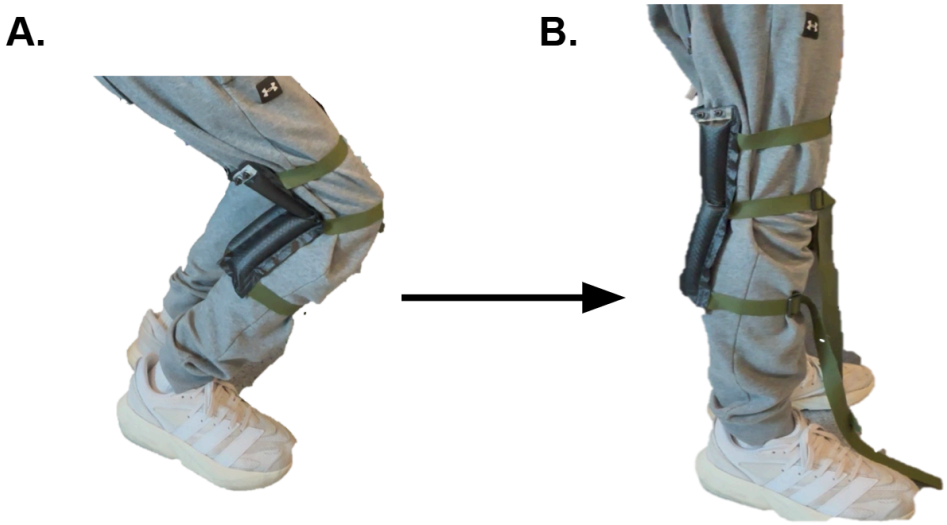


Figure 3: Image of actuator integrated onto the leg behind the knee joint. (A) to (B) shows knee bending to straightening sequence.

In order to integrate the actuator behind the knee joint of a user, the strapping system shown in figure 3 was developed. First, three 25 mm wide straps were cut using a woven fabric that is thicker than the ones used for the actuator. Two straps were sewn at each end of the actuator and were placed at the top and bottom. The third strap was similarly sewn at each end of the actuator and placed at the middle. One end of each strap was sewn onto a plastic length adjustable triple loop backpack buckle. This backpack buckle was used to secure each strap loop around the leg and prevent the actuator from moving. This design solved the previously mentioned slipping and bending issues with the actuator because of the middle strap. The strap located in the middle of the actuator solved the issue with slipping by being attached right above

the knee cap. This location of the strap was ideal because it not only allowed the actuator to hook onto the knee cap, which prevented slipping, but it also kept the actuator at a constant height relative to the length of the leg. The middle strap also solved the inconsistent bending problem with the actuator by creating a slight crimp in the middle of the actuator. This crimp is expected to weaken the straightening force of the actuator, but considering the inability of the device to function properly without a consistent bending location, this was a sacrifice worth making. This crimp design was also implemented during testing of the force output vs input pressure using a rubber band around the midpoint of the actuator.

CHARACTERIZATION AND RESULTS

After the design and fabrication of the actuator is complete, and the subsequent leg integration method is determined, the force output of the actuator will need to be characterized to verify the viability of the actuator in meaningfully assisting lower limb mobility. An evaluation of the resistive force against bending at all angles and pressure levels relevant to the expected knee bend and actuator strength conditions was chosen as a good test to begin understanding the true performance of the actuator. In order to begin testing, a custom test rig must be made. This rig will need to mimic lower limb bent knee postures between angles of 90 degrees to 165 degrees in order to get a range of input pressure vs force output values at all expected real world use case parameters. Although the actual force output of the actuator during normal bending and straightening cycles when mounted on the leg is expected to be different, using this test rig can help gauge approximate force values. This will help determine ideal actuator size, material, or geometry for potential future design iterations.

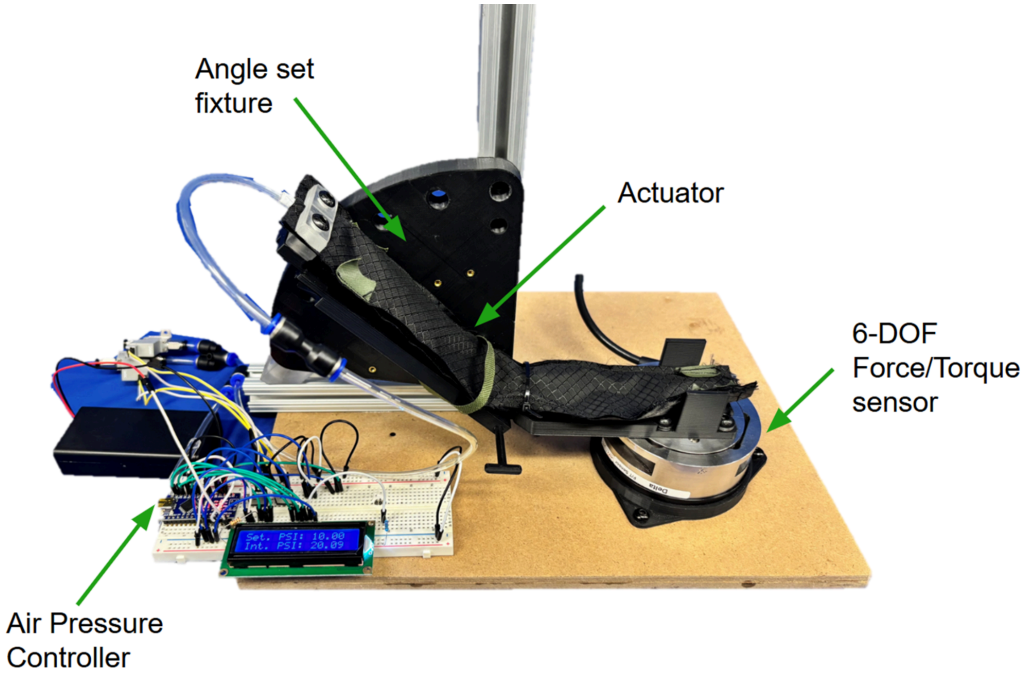


Figure 4: Experimental setup test rig. The actuator is secured onto a rotatable arm that has adjustable angle set points between 90 to 165 degrees in increments of 15 degrees. A force/torque sensor and custom built pressure controller will be used to characterize actuator performance.

To begin characterization of the actuators force output, a fluid pressure controller that can maintain constant pressure conditions within the actuator will need to be made for testing actuator performance. This pressure controller consists of a Arduino nano micro controller, 100 psi rated Honeywell pressure sensor, 7.4 volt battery, rotary encoder, 1602 lcd screen module, 2 SMC pneumatics Series SY100 solenoid valves, and an external air compressor which will be used as the main pressure source. The 2 solenoid valves used for the pressure controller are 3 way valves solenoids with the release valve exit blocked. This makes it possible to have total control over the inflow, outflow, and blockage of air flow into the actuator. The arduino microcontroller will read the pressure values given by the Honeywell pressure sensor and initiate the opening and closing of the two solenoid valves to match the desired pressure value. A simple user interface to set the actuator to a specific pressure level was made with the rotary encoder

and lcd screen module. Using the rotary encoder, the pressure inside the actuator can be set from a range of 0 to 20 psi. Both set pressure and the constantly updating current pressure will be visible on the lcd screen. The arduino microcontroller will be powered using the 7.4 volt battery and all other components controlled by the arduino are 5 volt components that can be powered through the arduino power pins.

The main construction of the experimental setup testing rig was made using a laser cut mdf board, 8020 aluminum extrusion, screw clamps, a 6-dof force/torque sensor, and custom 3d printed parts for the angle set fixture and actuator attachment points. The angle set fixture was designed as a quarter circle with peg holes located at angles between 90 to 165 degrees in increments of 15 degrees. A rotatable, 2 part jointed arm was fdm 3d printed with peg attachment points for the angle set fixture on one end, and screw holes for force/torque sensor attachment points on the other end. A slim fabric strap located at the joint, in combination with zip ties and velcro was used to keep the actuator from lifting off the arm and bending at a consistent location. A single M3 screw heat set insert will be placed between the two peg holes at each angle to secure the jointed arm flush against the surface of the angle set fixture. The 6-dof force/torque sensor will be fastened to the mdf board using 3 M6 screws. The 3d printed angle fixture will be mounted onto the L shaped 8020 aluminum extrusion with the 90 degree set point perpendicular to the surface of the mdf board. Before building, the entire test setup will first be 3d modeled in Solidworks CAD software to ensure its viability and for better visualization of components.

With the test rig complete, the actuator can be attached to the jointed arm and experimentation can begin. The test procedure begins by adjusting the angle set to 90 degrees. At 90 degrees, the actuator is first pressurized to 5 psi for 10 seconds and depressurized until the force sensor returns to a stable 0 N reading. This pressurization and depressurization sequence is

repeated 3 times to get a more accurate result. A plot with three peaks and three valleys representative of the 3 pressurization and depressurization stages of the test procedure is made using the data collected from the force sensor. The flat plateaued section of each peak is used to get three separate average force output readings. Using these three data points, an average force output can be calculated along with an estimation of the max and min error. The same process used to collect the force output data at an actuator angle set at 90 degrees and pressure of 5 psi is repeated for every angle at pressure levels between 0 to 20 psi in increments of 5 psi. A 2d and 3d mesh plot comparing the force, pressure, and angle of the actuator was made using matlab to help visualize the results of the test and behavior of the actuator at each stage.

After the initial testing, a few issues arose that needed to be remedied. First, the force output of the actuators near the middle angle sets between 120 degrees and 150 degrees began merging together at all pressure points. This made it so that the force output at higher angles were measured as greater than the force output at lower angles. The second issue observed after initial testing was inconsistent force increasing patterns across all angle sets. Some angle sets resulted in a concave up, exponentially increasing relationship between force output and pressure input, particularly at the smaller angles near 90 degrees. At the lower and middle angle sets, the exact opposite concave down relationship or the expected linear relationship between force and pressure was observed. The first issue related to the merging force output near the middle angle sets was determined to be caused by a combination of inconsistent constant surface area between the actuator and test jig arm as well as inconsistent bend location of the actuator. Additional straps were added around the jointed hinge section of the arm to help maintain constant contact between the actuator and arm surface. Also, two zip ties were wrapped around the actuator at the midpoint, tight enough to create a slight crimp when pressurized, so that the actuator will

naturally bend at the middle. The second issue related to the inconsistent increasing force behavior of the actuator was determined to be caused by imprecise solenoid valve open and close sequences from the pressure controller. Because the solenoid valves used to control the input and output flow of the air into the actuator act as a two way valve, the solenoid valves have the potential to block air flow at an incorrect pressure if the acceptable max and min pressures are set too far apart. This means that the actual pressure inside the actuator can be as much as 2 to 3 psi difference from the actual desired pressure. The large range of acceptable max and min differences from the desired pressure was initially programmed into the microcontroller to help conserve air in the pressure source, as it helps limit the exhaust of air from the actuator. This choice was made with the eventual goal of making a portable pressure controller device in mind. However, since the actuator does not need to be portable while testing, it can be hooked up to a larger air compressor pressure source. This allowed the max and min pressure variance range to be significantly decreased to get as close to the desired pressure inside the actuator as possible. After adjustment of the max and min parameters, the actual pressure level measured from the pressure sensor was much closer than before with a variance of only around .1 to .3 psi. For the portable pressure device currently in development, the pressure source will likely have to hold a larger volume of air to account for the increased exhaust sequences, or the precision of the pressure control will have to be sacrificed in favor of conserving air to a certain degree. After fixing these two issues, the results of the actuator evaluation came out a lot better. The data collected from these revised tests is represented in the two figures shown below.

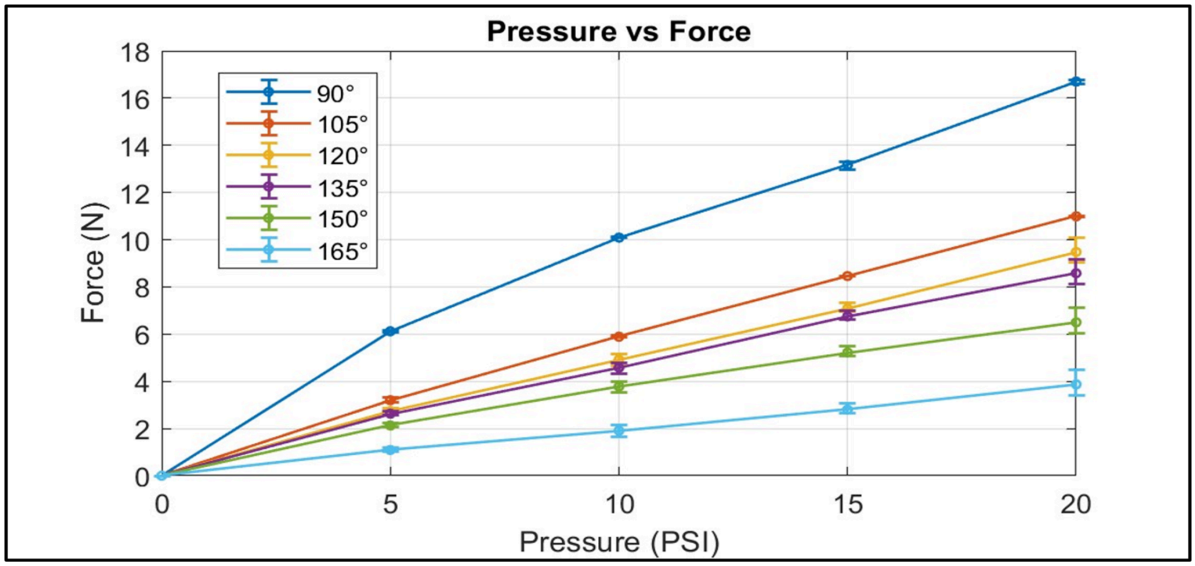


Figure 5: 2 axis plot of force characterization of actuator at pressures (0-20 PSI) with angles represented by different colors.

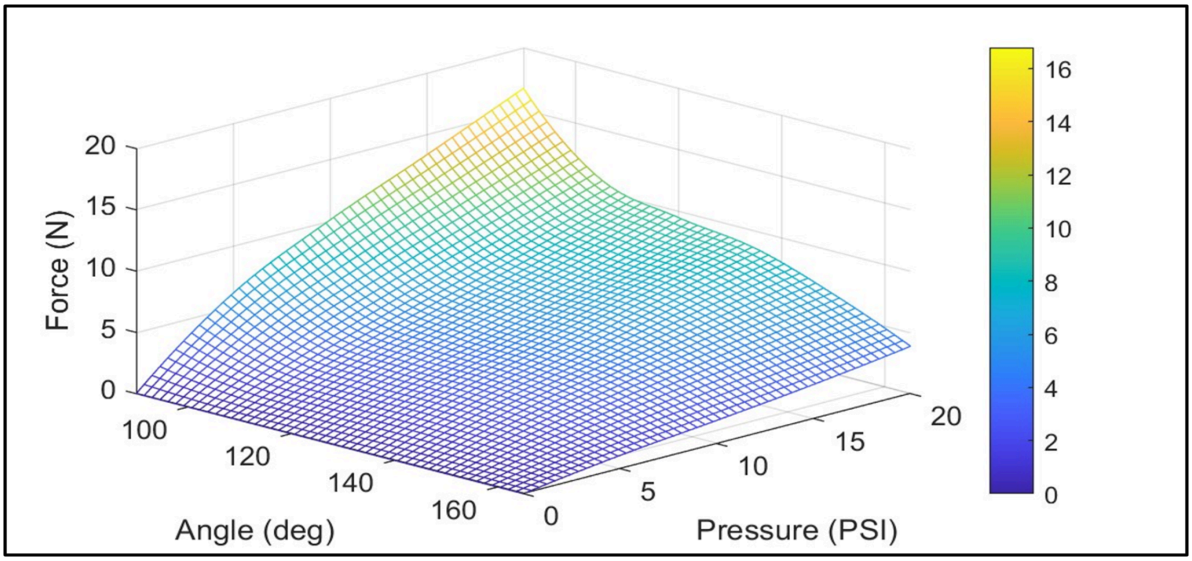


Figure 6: 3 axis plot mesh of force characterization of actuator at angles (90-165 degrees) and pressures (0-20 PSI).

When examining the actuator response of other straightening actuators tested with similar angle varying testing rigs, such as the testing done by O'Niell et al, there are certain characteristics that are expected to show up [2]. The first expected observation is different actuator creasing and shearing properties at different angles. The second expected result from bend resistance force testing for straight actuators is a positive linear relationship between force

output and pressure input. The overall results of the actuator force output characterization in this work resulted in the expected outcomes for the most part. As expected, the highest force output of the actuator was observed at around 16.7 N at the max pressure of 20 psi and at the smallest angle set of 90 degrees. On the other side, the lowest force output of the actuator was observed at around 1.09 N at the lowest pressure of 5 psi and at the largest angle set of 165 degrees. When looking at the plot shown in figure 5, the positive linear relationship between the pressure and force output of the actuator at all angle sets is easy to see. The relative smooth transition between the high and low points of the 3d mesh comparing angle, pressure, and force shown in figure 6 is representative of this expected behavior as well. Also, the plots show a smaller gap in force output particularly in the middle range of angle sets. This small difference in force between all pressure levels for these middle angles is expected as the actuator is at an angle large enough to show some shearing effects. Although the force output of the single actuator is relatively low for lower limb wearable device applications, as mentioned previously in this report, the design of the device is meant to be a modular system. Additional actuator units can be added in sequence with another to effectively double or triple the total force output of the entire device. The only limiting factor for the addition of actuator units is the circumference size of the user's leg. Overall, the force characterization of the basic actuator design produced results that were within the expected range of outcomes.

CONCLUSION

In conclusion, this report details the development of a soft pneumatic actuator based wearable device that is specifically designed for lower limb mobility assistance during athletic activities. The basic actuator consists of a fabric sleeved, single cavity design that straightens when pressurized. A triple strap system was implemented to integrate the actuator behind the

knee joint of a user to make it wearable. After fabrication, a custom test rig was used to characterize the actuator force output performance across a range of input pressures and bend angles. Currently, a portable pressure controller is being developed to complete the final form factor of the wearable device. For future work, additional experimentation with body tracking using positional data from IMU sensors and camera tracking, or electromyography (EMG) may be used to study energy consumption and muscle fatigue while wearing the device. Additionally, IMU sensors may be integrated into the actuator and controller for more adaptive pressure control based on the body position changes of the user.

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