Soft Robotic Exoskeleton for Elbow Assistance

A Technical Report submitted to the Department of Mechanical and Aerospace Engineering

Presented to the Faculty of the School of Engineering and Applied Science

University of Virginia • Charlottesville, Virginia

In Partial Fulfillment of the Requirements for the Degree

Bachelor of Science, School of Engineering

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Spring, 2022

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On my honor as a University Student, I have neither given nor received unauthorized aid on this assignment as defined by the Honor Guidelines for Thesis-Related Assignments

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Technical Report

Soft Robotic Exoskeleton for Elbow Assistance

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***Abstract*— Wearable robotic upper-limb exoskeletons can improve the mobility of individuals with neuromuscular disorders and allow them to complete daily tasks with greater independence. We are working with soft robotic technology, a lightweight and flexible alternative to bulky motorized exoskeletons, in designing and fabricating an exoskeleton with six degrees of freedom to be used by patients at the University of Virginia (UVA) hospital. The robot will sense electromyographic signals from the muscles in the wearer's arm and respond accordingly with assistive motion to help the wearer achieve their intended motion. The design of our exoskeleton is textile-based and actuated by pneumatic artificial muscles, specifically McKibben actuators. This technology is simple to manufacture and can be used to produce more organic motion than motors as it more closely imitates real muscle fibers. A major focus of our work is soft actuation in the rotational degree of freedom of the forearm, which has limited precedent.**

Keywords*— wearable technology; exoskeleton; rehabilitation; Thin McKibben actuator; Inverse Pneumatic Artificial Muscle

I. INTRODUCTION

The topic of this paper is robotic upper-limb exoskeleton technology with a focus on soft textile designs. The overall goal of the project is to help patients with neuromuscular disorders gain autonomy and achieve daily tasks. Prior literature informs the basis of this topic and the research question.

Prior research for an exoskeleton is primarily based around the McKibben muscle. The McKibben muscle is a type of artificial muscle that has two major parts. The first part is the outer wire mesh, and the second part is the internal bladder. When the bladder is expanded with air, the outer wire mesh contracts. The combination of these parts creates an artificial contracting muscle. For our design we went through three different muscles until we decided on our final design. We tested a multifilament muscle created using multiple small McKibben muscles, a larger singular McKibben muscle, and an inverse pneumatic artificial muscle. With the multifilament muscle we intended for the design to be less bulky than prior designs and to imitate natural muscle fibers better. The other two designs we tested also are less bulky than previous designs we have researched. Additionally, prior research tends to lack a rotational component to the upper limb exoskeleton. Our exoskeleton will aim to remedy this problem by adding two additional muscles for pronation and supination.

When conducting research on the proposal topic, there were multiple case studies that provided useful information and designs that could be incorporated into our soft robotic exoskeleton arm.

A. Carry

Carry is a soft, lightweight, upper-limb exoskeleton that is used in the assistance of elbow flexion when holding and carrying loads. The exoskeleton combines a soft human-machine interface and soft pneumatic actuation, making it a desirable design to model after. *Carry* was created with the purpose of assisting those in occupational fields that require a great amount of load handling, such as nursing aides, orderlies, and construction workers. The total weight of the exoskeleton is 1.85kg and includes one tethered soft pneumatic actuator for each elbow, as can be seen in Fig. 1. The actuators are made using a Thermoplastic polyurethane bladder that is encased by a textile tube. The textile's specifications are: Twill-Polyester, 90/180 den, impregnated, 180 g/sqm [1].

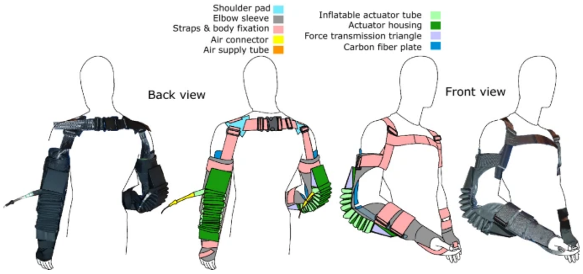

Fig. 1. *Carry* exoskeleton diagram [1].

To make *Carry* autonomous, there are three possible setups, as shown in Fig. 2. In Set Up A the exoskeleton is powered by a compressor, in Set Up B it is powered by a pressure tank, and in Set Up C it is powered by both a compressor and a pressure tank [1]. Similar to the Carry design, we plan to have a harness that spans across the chest to support the actuators.

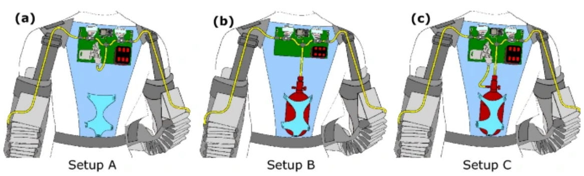

Fig. 2. Power source for *Carry* exoskeleton [1].

B. Forearm Rehabilitation Exoskeleton

The Forearm Rehabilitation Exoskeleton is used to assist in the rehabilitation of the rotational motion of the forearm. This exoskeleton uses an application of the novel Extensor-Contractor Pneumatic Artificial Muscle (ECPAM) [2]. Previously there was not a pneumatic actuator that could generate both contraction and extension force in response to applied air pressure; a pneumatic actuator could only be designed to do one of the two. In order for the system to do both, a contraction actuator and an extension actuator were put parallel to one another. In the case of the ECPAM, the two actuators are combined into one design, where the contractor muscle is placed within the extension muscle, taking up dead space, meaning that the actuator will use slightly less air than if they were parallel to each other. The ends of the actuators are formed by two end-caps. The contraction muscle attaches to the thin central section of the end-caps, and the extension muscle connects to the larger diameter section of the end-cap [2]. One of the end-caps contains two holes for air supply; the one for contraction is in the center of the thin section, and the one for extension is in the larger diameter section, as can be see in Fig. 3 [2]. The actuator is placed diagonally across the forearm and is secured at the terminal and central sections of it using adjustable elastic straps as seen in Fig. 4 [2].

Fig. 3. Parts of novel Extensor-Contractor Pneumatic Artificial Muscle (ECPAM) [2].

Fig.4. Rotational actuator for forearm [2].

During our research, this design was the only one found that focused on rotation, rather than flexion/extension. The design of the exoskeleton looks quite bulky, but is actually lightweight, only having a total weight of 1.8 kg [2]. Despite its unappealing aesthetic, the methods used to create the rotation in the forearm will be useful in designing our robotic exoskeleton, however, we will separate the extensor actuator and contractor actuator to make our design less bulky.

C. Elbow Rehabilitation Exoskeleton

This Elbow Rehabilitation Exoskeleton is used in the recovery of flexion and extension movement in the elbow joint and was developed based on a Extendor-Bending Pneumatic Artificial Muscle (EBPAM) [2]. The EBPAM is based on linearly extending McKibben artificial muscles. Along one side of the muscle it is reinforced, keeping it at a fixed length regardless of pressure, while the other side is left free. When it is pressurized, the reinforced side does not change, but the free side grows longer, so rather than extending linearly, the muscle bends, as can be seen in Fig. 5 [2]. In this design, two bending actuators are used [2]. To reinforce the exoskeleton to the user's hand and each bending muscle, plastic handles with Velcro were used (See Fig. 6). To produce the desired bending angle, the actuators are pressurized simultaneously with the same amount of pressure.

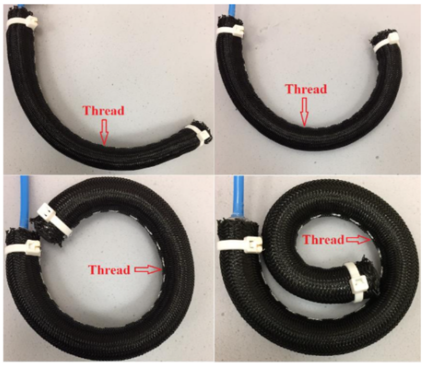

Fig. 5. Image of Extendor-Bending Pneumatic Artificial Muscle (EBPAM) [2].

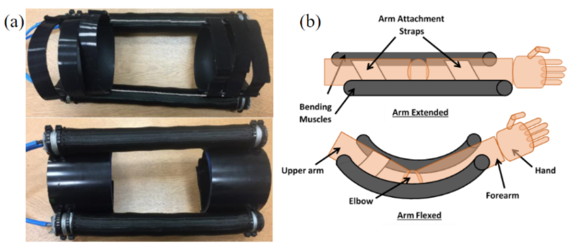

Fig. 6. Image and diagram of Elbow Rehabilitation Exoskeleton [2].

D. Exosuit Using Butyl Rubber Tubes and Textiles

This Soft Exosuit for Elbow Assistance was developed for flexion and extension actuations. For the pneumatic actuator, butyl rubber tubes that are commercially available were used. A lightweight polyester fabric envelops the tubes in order to eliminate nonhomogeneous expansion. The fabric of the actuator is then mounted on the clothing fabric of the exoskeleton (González-Vargas et al., 2016, pp. 420-421). The full design looks similar to an elbow pad used for safety. The actuators organized in a vertical pattern are used for extension, and the actuators organized in a horizontal zigzag pattern in the center are used for flexion (González-Vargas et al., 2016, pp. 420-421). The total weight of the skeleton is 0.5 kg, and it has dimensions of 0.4 by 0.18 meters. It provides a range of motion of 95 degrees, and showed a 45% EMG signal reduction during testing (González-Vargas et al., 2016, pp. 423).

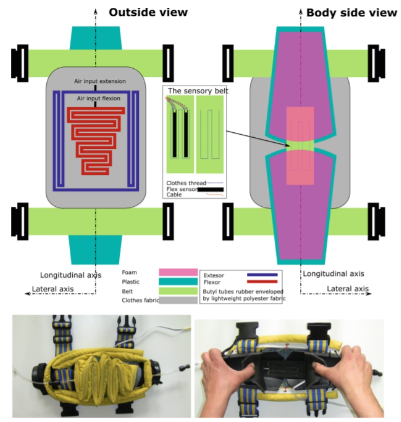

Fig. 7. Image and diagram of Soft Exosuit for Elbow Assistance [3].

The largest gap in literature is that there is a lack of successful exoskeleton that includes the rotational degree of freedom. Therefore, this challenge will most likely take the most time and focus of the team.

The goal of this project is to develop wearable technology that can assist upper limb muscular movements for patients with neuromuscular disorders. Noticing the gap in research on textile exosuits modeling both of the elbows degrees of freedom, the research question we plan to answer is: *How can a soft textile robotic arm be designed to actively assist patients with neuromuscular disorders to achieve daily tasks?* More specifically: *How can we use pneumatic actuators on a flexible, wearable frame to accurately model and replicate the two degrees of freedom observed in the elbow joint?* The plan is to gain a comprehensive understanding of the current technology in the wearable exoskeleton field, and use this collection of resources in conjunction with technical knowledge gained from our engineering courses to sufficiently answer these research questions. A comprehensive answer to this research question will include theoretical knowledge and design, physical prototypes, and prototype testing in order to verify if the design does actually satisfy the research question at hand.

As previously stated, the goal of this project is to create an upper limb exoskeleton in order to assist patients with neuromuscular disorders during daily tasks. We hope to accomplish this by contributing a lightweight and sleek upper limb exoskeleton that provides flexion and extension for the elbow along with pronation and supination for the wrist without being overbearing in size or design.

II. DESIGN PROCESS

A. Methods

After adequate research on past exoskeletons and applicable technical engineering details, the first task is to develop a design plan for the initial prototype. In order to sufficiently answer the research question, the design can be broken down into smaller tasks that allow us to achieve the end goal of a functioning soft upper-limb exoskeleton. The first task is to design an actuator for the first and main degree of freedom in the elbow, flexion and extension. This actuator design needs to utilize pneumatic artificial muscles to accurately replicate a human's range of motion while remaining lightweight and comfortable for the user. The second task is to design an actuator for the second degree of freedom of the elbow, torsion of the forearm. Similar to the first task, this actuator design needs to utilize artificial muscles and be lightweight and comfortable for long term use. The third task is to combine these two actuator designs onto one wearable sleeve or framework, and cooperate with the other groups to integrate the elbow design into the shoulder and sensor designs.

In designing an upper-limb exoskeleton, the first and main consideration is deciding which style of actuator to use to provide motion assistance. Pneumatic actuators, also known as artificial muscles, are ideal for this application because they do not need a rigid framework, they imitate natural muscle contraction accurately, they are lightweight, and they are easy and cheap to manufacture. More specifically, a style of pneumatic actuators called "thin McKibben muscles". Thin McKibben muscles are comprised of a thin silicone tube nested inside of a braided sheath, both of which have a very small diameter, only a millimeter or so. The thin McKibben muscle functions in the same manner as a normal McKibben muscle; air pressure expands the inner bladder radially, and due to the braided sheath, radial expansion results in lengthwise contraction. Due to the small diameter of a thin McKibben the force exerted is less than that of a thicker McKibben muscle, so a series of many thin McKibben muscles are used together in parallel to achieve high forces. There are multiple benefits to this style of actuation over using a single large McKibben muscle. For one, thin McKibbens can be much less bulky than the traditional style. Instead of one large cylinder on an arm, having a series of smaller tubes allows flexibility in the arrangement, such as fanning each individual fiber out along a surface to make the overall muscle shape thinner and wider as opposed to just cylindrical. This decrease in bulk is important for maximizing comfort for long term use of the exoskeleton. In addition, thin McKibbens can more accurately imitate natural muscles, because they both function based on a series of smaller fibers. While a single large McKibben muscle can only actuate linearly, a muscle composed of many smaller fibers can have fibers in multiple directions, allowing for more complex and more efficient motion. These benefits are described well in Sunichi Kurumaya's research article titled "Musculoskeletal lower-limb robot driven by multifilament muscles" [4]. In this article, examples of thin McKibben muscles were produced in order to attempt to imitate lower-limb muscle movements. Pictures of their thin McKibben design are shown below.

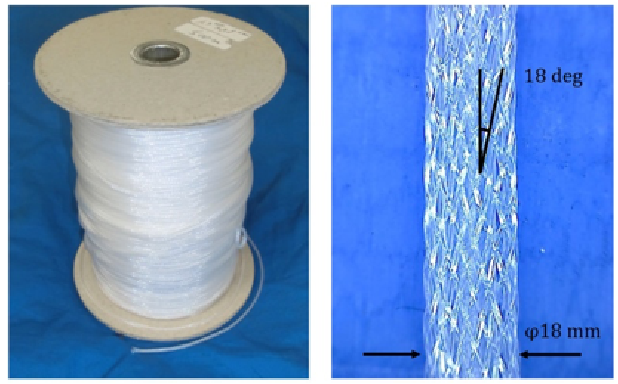

Fig. 8. Mass production of Thin McKibben muscles [4].

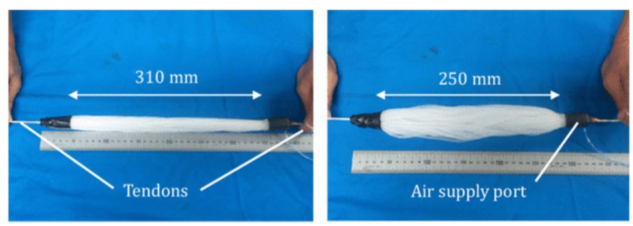

Fig. 9. Contraction of multifilament muscle [4].

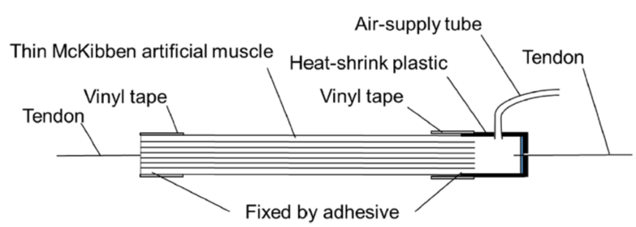

Fig. 10. Multifilament connection diagram [4].

Fig. 8 shows the simple mass manufacturing of the thin McKibben muscles, where a spool of braids can be combined with a spool of silicone tubing to quickly create a long line of thin McKibben muscles. Fig. 9 depicts the actuation of a muscle made up of many thin McKibben fibers, and Fig. 10 depicts the way that each multifilament muscle is connected. This research article proved the functionality and benefits of thin McKibben muscles, and for these reasons thin McKibbens will be used in this project for an upper-limb exoskeleton. They are more efficient spatially, can be arranged for more complex actuation, and are easily mass produced.

In order to accurately design these artificial muscles to achieve specific forces and ranges of motion, the force per thin McKibben fiber and the contraction ratio must be known. The theoretical equation modeling the force of a McKibben muscle is shown below.

$$F = \left(\frac{\pi D_0^2 P}{4} sin^2\theta_0\right)\left[3(1 - \frac{\Delta L}{L_0})^2 cos^2\theta_0 - 1\right] \qquad (1)$$

In equation (1), F is the force exerted by the muscle, D_0 is the relaxed outer diameter, P is the air pressure, θ_0 is the relaxed braid angle, and $\frac{\Delta L}{L_0}$ is the contraction ratio. While this is helpful for understanding how certain parameters and variables affect the performance of McKibben muscles, experimental data may vary from this theoretical equation due to variables such as inner bladder material, complex braid patterns, and the fact that not all fibers in each muscle will be perfectly linear; some will inherently be curved or turned slightly. For these reasons, designing for specific forces and ranges of motion was done using experimental data as opposed to this equation.

B. Materials

With a style of actuator decided on and a plan made for how to achieve specific forces and ranges with said actuators, a more comprehensive design plan was created for the first design iteration. The goal of this design was to achieve the desired motions for the elbow joint while being lightweight and flexible, and for this reason there will be no solid framework. Instead, the framework will be made up of flexible materials such as nylon straps. A 3D visualization of the framework can be seen below. Note that the full design is only one shown for one arm, for simplicity of visualization, but the final product will be on both arms.

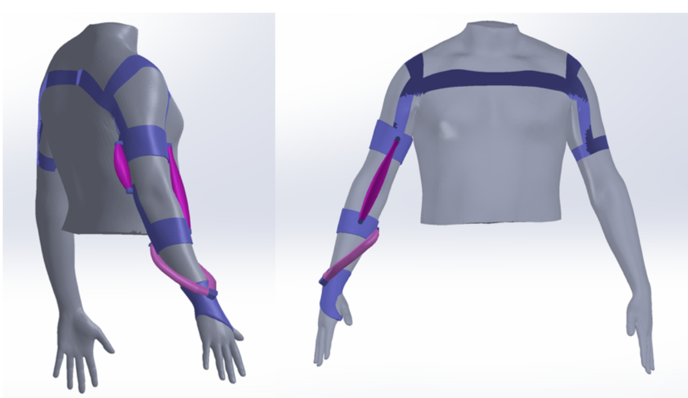

Fig. 11. 3D Visualization of elbow joint actuators and framework.

Each muscle shown represents where a series of thin McKibben muscles would be placed to operate as one muscle. There is one actuator over the bicep area, connecting an anchored point on the shoulder to the upper forearm, providing flexion of the elbow. Another muscle is placed over the triceps area, which connects from an anchored point behind the shoulder to a piece of strap that wraps around the elbow. By pulling on this strap, it pulls on the backside of the forearm, causing an extension of the elbow. Two muscles are also placed wrapping around the forearm, each one causing torsion in opposite directions when each one is actuated. This provides the pronation and supination of the forearm. The actual dimensioned drawings of each piece of this framework are shown below in Fig. 12.

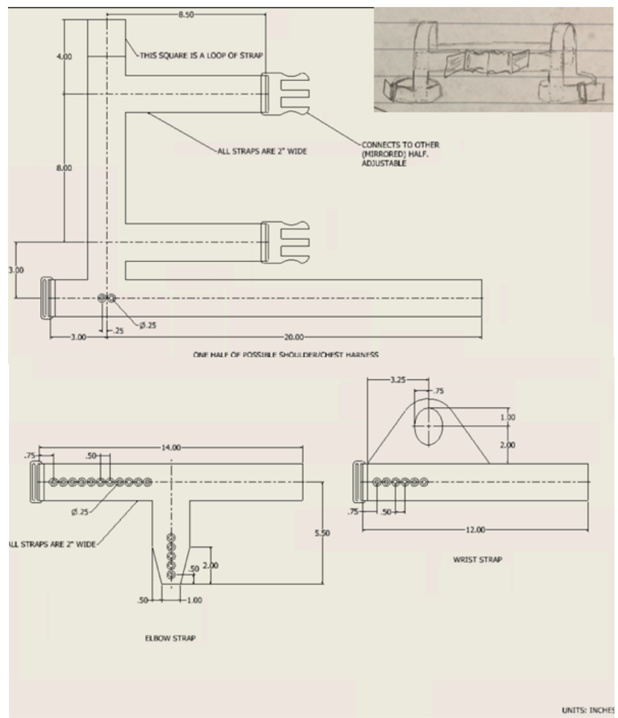

Fig. 12. Dimensioned drawing of framework.

The picture on the left depicts one half of the chest/shoulder harness. The buckles connect over the chest and one over the back. The long strap at the bottom of the drawing is the strap that wraps around the upper arm, going through the loop on the other side, and connects back around. This is the location of the connection point for the flexion and extension muscles. The “T” shaped piece on the right is the elbow strap. The top of the “T” is the long strap that wraps around the upper forearm, while the protruding strap wraps around the elbow, connecting to the triceps muscles. The rightmost drawing is the wrist piece, which wraps around the wrist and has a thumb hole for support.

There are multiple things to notice on these drawings. For one, each piece is comprised completely of 2 in thick nylon straps, stitched together in certain ways. This makes the production of the prototype simple and cheap. The straps use a simple adjustable synching mechanism in order to achieve variation for different sizes of patient arms. The connection points for the muscles are small metal eyelets through the straps. The end of the muscle tendon will have a small metal clip which clips into these eyelets, keeping the connections simple and adjustable. Each connection point has a line of multiple eyelets to allow the user to connect the muscle to slightly different points, based on their specific arm dimensions. In addition, it is important to note that although the chest/shoulder harness was designed to provide an anchored upper connection point, this design can easily be adapted to utilize other chest/shoulder harnesses, such as one designed by the shoulder team. As long as there is a location to connect a muscle on the front and back, the design can function correctly.

With a dimensioned design, it was now possible to design the force ranges of each muscle in order to achieve certain performance goals. For each muscle, a certain load is given. Using a moment analysis, the necessary torque is found to achieve that load. Using the dimensions of the prototype, the radius of the line of action to the axis of rotation can be estimated, which is then used to determine how much force the muscle itself will need to exert in order to achieve this desired torque. Table 1 shown below shows these estimations.

Table 1. Estimated Design Range and Load

Motion	Range	Contraction Distance Needed	Max Payload	Required Actuator Force
Flexion	-10° to +145°	~3 +/- 1”	15 lbf	~70 lbf
Extension		~3 +/- 1”	15 lbf	~70 lbf
Pronation/Supination	-90° to 90°	~1.5 +/- .5”	7 in*lbf	14 lbf

The “Max Payload” column lists how much force is desired to be lifted not including the weight of the arm itself, for example the goal for the max payload for flexion of the elbow would be for the user to be able to lift a 15lb weight in their arm. This weight has a radius equal to the length of the arm, so using this along with the weight of the arm and an estimation of its center of gravity provide a max torque for the design. This max torque can be used along with the dimensions of the prototype to determine that the force that the bicep artificial muscle would need to exert is about 70 lbf. Based on results from Sunichi Kurumaya’s research article on thin McKibben muscles, this is a realistic force to achieve. Note that the desired maximum payloads are not extremely high. This is because the goal of this exoskeleton is not to give the user superhuman strength, but instead is just to help the user accomplish daily motions, such as lifting a cup of tea or a laptop.

With the desired force ranges for the muscles decided, it would be easy to use the measured force per thin McKibben to decide how many fibers will make up each muscle. However, the full force of the maximum torque is not needed for every motion, so the Arduino code will be used to control how much voltage is given to each air pump to achieve the desired output force.

```
void setup() {
  // put your setup code here, to run once:
  pinMode(3,OUTPUT); //sets pin D3 to output
  Serial.begin(9600); //opens serial port at 9600 bps
}

void loop() {
  // put your main code here, to run repeatedly:
  if (Serial.available() > 0){ //if there is any available info to be read from the serial port:
    int Force = Serial.parseInt(); //reads incoming Force desired from serial port to find integer
    // Insert equation relating force to the psi necessary to achieve that force. This depends on the measured
    // values of forces vs psi
    int mapped = map(PSI,0,MAXPSI,0,255); //mapping the range 0-MAXPSI to 0-255 to match the voltage
    // scale to our psi scale
    analogWrite(3,mapped); //gives pin 3 the output of a PNW with the value we just mapped
    Serial.println(Force); //prints the input Force to the serial port
    //delay(500); //delays 500 ms
    }
}
```

Fig. 13. Example Arduino code.

This code takes the desired muscle force as an input. Once sample thin McKibbens are measured experimentally, an equation can be deduced that relates the force of the artificial muscle with the psi given to the muscle. Note that this is not included in the code yet, because it depends on the experimental data that will be collected in the future. For now, there is just a placeholder comment for where this equation will go. The maximum psi possible by the air pump is also shown as an undefined variable because it depends on which air pump that ends up being used. Using the ratio between the desired psi and the maximum psi, this psi output can be mapped to a voltage that is the same fraction of the max voltage. This voltage is then given as a PWM to the air pump to allow it to give the desired air pressure to the muscles.

C. Prototype Construction and Iterations

To construct the first prototype, the sample materials ordered for the McKibbens were used to test different thin McKibben options. The inner inflatable tubes used were 1/32” inner diameter and 1/16” thickness, with hardnesses of 30A and 50A. The braided sleeves tested were the “FLEXO Thin” 1/15” diameter with 300% expansion, and the FLEXO PET ⅛” with 150% expansion. Coupled with either of the inner tubes, the FLEXO PET proved to have much too large of a diameter to be effective. When air pressure was applied to either inner tube, there was no expansion, so more samples were ordered of a thinner tube (.058” ID, .077” OD, .009” thickness). This thinner tube did achieve radial expansion, and when coupled with the FLEXO Thin, axial contraction was achieved. With these materials chosen, a method of sealing one end and supplying air pressure to the other end was needed. In an effort to utilize materials on hand and not buy more, the sealed end was tied off with itself, and the open end was stretched over a bike pump needle then zip tied on. This makeshift setup proved to be very finicky, as the sealed end would leak air, and the air supply end would either fall off the needle or leak as well. Another problem arose with the braiding; the fibers are so thin that the ends quickly fray when working with them, and once the weave pattern loses its integrity, it becomes a weak point in the McKibben. When air pressure is applied, the inner bladder would be able to expand more where the braid was compromised, leading to bubbling and popping in these areas.

Fig. 14. Singular Thin McKibben muscle attached to a force tester.

Despite said issues, force testing and contraction testing was still achieved with a single thin McKibben fiber. The next step was putting multiple together to create a multifilament muscle. The first effort at creating a multifilament muscle utilized heat shrink end caps to seal the ends, with a bike pump needle attached to the supply end by shrink wrap as well. The heat shrink proved difficult to implement because as heat was applied to the heat shrink wrappers, the heat would burn and melt the braided sheath as well, compromising the braided sheaths’ structural integrity. In addition, air was able to escape through the small holes between each fiber in the bundle. In effort to fix this second issue, hot glue was applied in the crevasses of each end in an attempt to seal all air holes. This fix did not prove effective, as air still leaked at all times. This led to the creation of a new multifilament design, where custom parts were 3D printed to provide airtight attachments for each individual fiber. An SLA printer was used for this to achieve the necessary level of detail in such small prints. The design includes attachment tubes for each fiber that all connect to form an airtight connection between the air supply and each fiber. The inner tubing of the muscles are stretched over the connecting tubes using fine-tipped tweezers and secured with a second printed piece that fits firmly over the ends. Several iterations of this design are shown in Figure 15. The final design is the prototype farthest to the right in the figure.

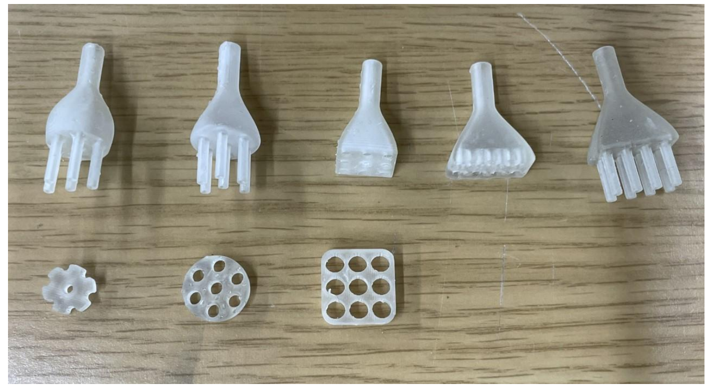

Fig. 15. Several iterations of the multifilament Thin McKibben connector.

During the testing process, the shape of the connector was changed from a circle to a rectangle to more easily accommodate changes to the number of muscle fibers in each overall muscle. One issue that arose with the first prototypes (farthest left) was the pegs breaking off. This was in part due to the difficulty of placing each thin McKibben individually on the pegs in such a confined space. A proposed solution was to print the pegs separately from the main connector, attach the muscles to the pegs, then screw the pegs into the connector. This proved infeasible with such a small design. A second solution was to use thin aluminum tubing as pegs, attach the muscles, and use a press fit to attach them to the connector. This again proved infeasible due to air leaking out at the location of the press fit. To reduce the number of non-airtight regions, the final design uses the original 3D printed peg design, with some slight adjustments to fillet radius and length to decrease the chance of breaking. The circular design had one peg in the center that was completely surrounded, making it difficult to attach the muscle. Switching to the rectangular design solved this problem. Additionally, a major problem with the concept of securing the muscle fibers by stretching them over a peg was that the tweezers used would often poke small, often unnoticed holes in the inner tubing of the muscles. Any hole in any of the tubes would greatly affect the ability of the multifilament muscle to contract. The solutions used were dulling the tips of the tweezers, taking great care to not stretch the muscles past the necessary point, and coating the ends of the muscles in resin to seal any unnoticed holes once they had been placed. The resin also helped prevent the braided sheath from fraying, and was used to seal the free ends of the multifilament muscle. One solution that would theoretically solve the problem of holes in the muscles would be a specialized tool that would work similarly to an aperture, fitting inside and stretching the tubing uniformly. This tool was not available to this group, due to the small size of the tubing, but would likely solve this problem effectively. The final design of the multifilament muscle is shown below. This prototype was able to successfully and stably contract with no air leakage.

Fig. 16. Final design of the multifilament Thin McKibben muscle.

Concurrently with this design being developed, the harness system was also being created and tested. In order to construct the harness system before the thin McKibbens were ready, single larger McKibbens were used in place of the multifilament muscles for testing the mechanical system. This is because these larger McKibbens were easier to produce and had already proven to provide strong contraction forces. These were intended to be placeholders for the thin McKibben multifilament muscles while the design was being developed. The large McKibbens used a ⅜" mesh braided sheath and a ¼" latex inner tube. The sealed ends are capped with shrink wrap caps, and the air supply end is shrink wrapped with a bicycle pump needle.

The first iteration of the harness system was based on the CAD drawing shown earlier. It was made up of 2" thick nylon straps. One piece wrapped the chest and upper arm, one piece was on the upper forearm, and the last piece went around the wrist connecting to a glove. The large McKibben actuators were attached around the forearm for pronation and supination, and over the biceps and triceps for flexion and extension. A picture of this setup can be seen in Figure 17.

Fig. 17. First harness system design.

The first issue with this design arose when testing the pronation and supination. The actuating design of having the muscles pull diagonally around the forearm worked well, but the upper connection at the top of the forearm tended to slide as it was too loose. In order to fix this, new buckles were ordered which better secured the strap. Next, there was an issue with the flexion of the elbow joint (bicep curl motion). Since the straps were fabric and not solid, the first portion of the muscle contraction just pulled the edges of the fabric up without moving the arm, so only the last bit of contraction actually moved the arm. For this reason, there was only a small angle of motion from this setup. The solution to this would either be a muscle with a larger contraction ratio, or a more rigid harness.

This issue led to the next iteration of the harness design. While discussing the optimal method of integrating the shoulder DOF design and the elbow DOF design, it was decided that a solid arm brace would be used as the backbone of the mechanism as opposed to soft straps. This is the same mechanism that was being used in the shoulder DOF design, but it also proved beneficial for the elbow DOF because this solved the problem of needing a more rigid harness for flexion. A picture of the actuators with this harness system is shown in Figure 18.

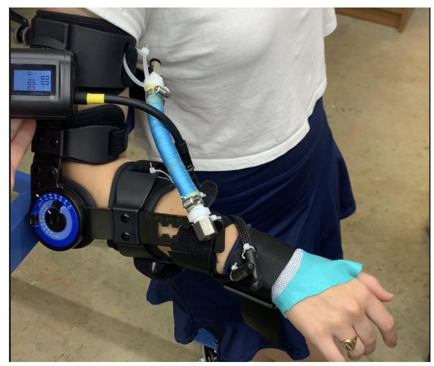

Fig. 18. Final harness system design.

Using this new rigid brace, the actuators were placed in the same layout. The supination and pronation still worked as they did in the original harness using diagonal muscles around the forearm. The flexion motion worked better than the last harness, however the contraction ratio was still too small to achieve a full range of motion. To achieve a larger range of motion, the McKibben design was abandoned for a different style of pneumatic actuator. This new design features a latex inflatable bladder with a nylon string wound around it, based on the design developed by Hawkes, Christensen, and Okamura [5]. This is depicted in Figure 19.

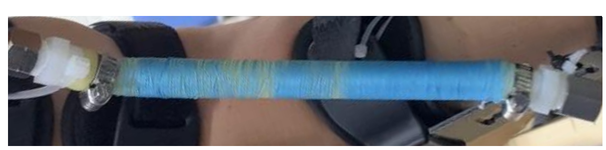

Fig. 19. Inverse Pneumatic Artificial Muscle.

When air pressure is applied the inner bladder inflates. The string prevents the latex bladder from expanding radially, so instead it expands axially, lengthening. Once air pressure is removed, the elastic property of the latex takes over and shrinks the muscle back to its original length. In the context of elbow flexion and extension, air pressure would need to be applied and held while the elbow is straight, then releasing the pressure pulls the muscle and pulls the arm up. The benefit to this style of muscle is that it can achieve 200% extension, meaning when air is released it achieves 67% contraction from the lengthened state. This solves the issue of the McKibben's small contraction ratio, and when applied to the solid brace, a fuller range of motion was achieved. The extension motion was still achieved using a normal McKibben muscle, as were the forearm rotations.

While this new style of actuator is very effective when working properly, it is very inefficient in production and causes many issues. It is much more difficult to make, as the winding is very tedious and has to be perfect to avoid bubbling and popping. Even when a muscle is wrapped correctly initially, after actuating a few times the wrappings tend to move slightly allowing for vacant spaces which lead to bubbling and pops. In an attempt to lessen this issue, a line of super glue was applied to one side of the muscle to prevent the windings from sliding around as much. While this technique does help the muscles last longer, they still only last a few rounds until the windings start to fray apart. It is important to be very careful handling these muscles in order to not ruin the windings. It is possible in the future to come up with varying designs to the same style of actuator to achieve the same contraction ratio with more durability.

Despite the shortcomings with this muscle style, it still remained in our final prototype due to its effectiveness in elbow flexion. This hard arm brace mixed with the aforementioned muscles became the final prototype for the elbow degrees of freedom. After this was created, this mechanism was integrated with the design for the shoulder degrees of freedom. The arm brace and muscles remained the same, and it was added to the shoulder team's shoulder harness/mount. This included one large McKibben which attached to a point on the upper arm, but did not interfere with the design of the elbow mechanism. It is also important to note that although the goal was to replace the larger McKibben muscles with the thin McKibben multifilament muscles eventually, the large McKibbens remained in the final prototype due to their ease of construction and higher strength comparatively. A picture of the final joined design is depicted in Figure 19.

Fig. 20. Combined elbow exoskeleton and shoulder exoskeleton setup.

Currently, this mechanism functions on a binary input, either on or off. This makes the motions binary, either all the way contracted or all the way released. This is due to time constraints of the project, but in the future a programmable air supply will be applied in order to achieve motions at variable speeds and variable positions.

III. TESTING AND RESULTS

A. McKibben

The McKibben actuator was force tested using the setup seen below:

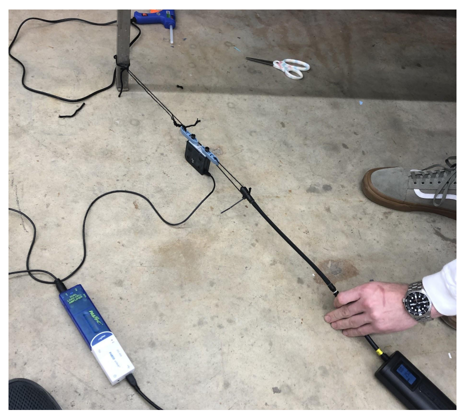

Fig. 21. Force testing for McKibben muscles.

Even with low pressures, the tube yielded high forces, as seen in Table 2, and although not recorded quantitatively, the contraction ratio was observed to be very high.

Table 2. Pressure vs. Force results for McKibben Muscle

Run	Air Pressure (kPa)	Force (N)
1	100	31 (incomplete)
2	50	70.4
3	25	79.7
5	10	yarn broke

As a result of this testing, it was determined that a single McKibben tube produced the necessary force for the pronation/supination degree of freedom.

B. Thin McKibben Single Fiber

The force tests on the single thin McKibben fiber were done in a similar manner to the larger McKibben. Two ends were secured, at a fixed length, air pressure was applied, and the resulting force was recorded. As shown in Table 3, the fiber was tested at many pressures for fewer trials in order to get a range of values before the muscle failed.

Table 3. Pressure vs Force results for Single Fiber Thin McKibben

Pressure (kPa)	Force (N)			
	Run 1	Run 2	Run 3	Avg.
200	4.2	7.2	4.2	5.2
250	5.5	-	-	5.5
300	5.5	-	-	5.5
350	5.2	-	-	5.2
400	4.4	-	-	4.4
450	5.3	-	-	5.3

The average force output did not seem to increase linearly with increased pressure. Despite this, these results are promising because a multifilament design would combine the force output of each fiber, so manually changing the number of fibers in a muscle can achieve the desired amount of force. There is not sufficient data regarding the contraction ratio, but the ratio seemed to increase with increased pressure, somewhere around 20%.

C. Thin McKibben Multifilament

The thin McKibben multifilament muscle was force-tested by securing both ends of the muscle and pressurizing the tubes to set values and measuring the corresponding force, as seen in Figure 22. The results of this testing are shown in Table 4.

Fig. 22. Force testing setup for multifilament muscles.

Table 4. Pressure vs. Force results for Multifilament Muscle

Pressure (kPa)	Force (N)			
	Run 1	Run 2	Run 3	Avg.
30	19.4	15.6	16.8	17.27
40	18.2	18.2	21.3	19.23

These results show that pressure has a direct relationship with actuation force. However, at safe pressure values, the muscle would not produce a sufficient force for pronation and supination according to the predicted values.

Contraction testing yielded a contraction of 1.5 inches, or 17.6%, from 8.5 inches to 7 inches. This is sufficient for pronation and supination, however durability testing while applied to the actual arm is yet to be determined.

D. Inverse Pneumatic Artificial Muscle

The inverse pneumatic artificial muscle was ideal for elbow flexion because of its high contraction ratio, which was approximately 67%. The muscle was effective in this sense, but not durable. The windings are easily moved, which allows for bubbles and popping of the tube. The windings also had to be wrapped around the tube with enough pressure to keep the tube from expanding radially, but not so tight as to obstruct air flow. This makes the creation of one tube quite difficult. Overall, the muscle provided an ideal amount of flexion by bending the elbow approximately 30 degrees.

E. Degrees of Freedom

The exoskeleton achieves two degrees of freedom: pronation and supination, and flexion and extension. The forearm can be rotated up to 80 degrees for pronation, and up to 80 degrees for supination using the McKibben muscles. This encompasses the full rotation of the forearm, making this degree of freedom successful. The elbow can be extended 15 degrees with the McKibben muscles. The Inverse Pneumatic Artificial Muscle allowed for 30 degrees of flexion of the elbow. This degree of freedom is overall considered successful. Specifically, we consider the flexion to be a success. Extension is semi-successful and will require more work to be deemed a complete success.

F. Air Supply Performance

An electric air pump was used to supply air to the artificial muscles. The air pressure was manually adjusted and applied to the muscles by one of use. Once the device deemed that the muscle had reached the maximum pressure it shut off, leaving the muscle fully inflated. To release the air from the muscle, it had to be removed from its attachment to the air pump. The air pump was able to supply the necessary amount of air for the muscles to reach maximum contraction, but an ideal air supply would be able to inflate and deflate the muscles automatically according to sensing data.

IV. CONCLUSION

A. Discussion

While thin McKibbens were originally chosen to be the muscle, difficulty in implementation led regular McKibben muscles to stand in for the thin McKibbens until they could be created more consistently. These problems included braid fraying, air leakage, popping, and difficulty combining many McKibbens. Despite these issues, a consistently working thin McKibben setup was eventually created.

The results were mostly successful, particularly for pronation and supination. Flexion also works relatively well, though not as well as our group had hoped at the beginning of the project. Extension is the motion that needs the most work. The natural motion can be performed, but with a very limited range of motion.

The numerical results shown in tables two and three suggest that the single McKibben muscle is preferable to the multifilament thin McKibbens. Despite this difference in force output, it is possible that the thin McKibbens may have a better overall performance if comfort and a versatile shape are necessary for a future muscle. In many cases, the inverse pneumatic artificial muscle can also be considered as an option, as a well constructed version can hit a contraction ratio that our current McKibben muscles simply cannot achieve.

B. Conclusion

Our elbow design has achieved the original goal of two degrees of freedom. We have gotten pronation, supination, flexion, and extension to work through the use of the McKibben muscle. From these degrees of freedom, pronation and supination work extremely well. Flexion and extension also work, but the range of motion is not currently as high as our team had originally hoped it would be. Additionally, we did not have enough time to move the design into the hands of patients. Despite these hiccups, our team has created an upper-limb exoskeleton that successfully filled the gap in previous research by creating functional pronation and supination alongside working flexion and extension.

C. Future Work

Future work on this project will focus on further testing and perfection of the design. In order to analyze the product's effectiveness, a human subject study will be completed, involving allowing UVA Hospital patients with neuromuscular disorders to test the arm's design. A survey questionnaire will then be conducted to compile feedback information, through which subsequent redesigns can be made. In order to begin human subject studies, an application will be submitted to the Institutional Review Board, and CITI training must be completed by all group members. The survey will ask a comprehensive set of questions with the goal of gaining feedback on both the ease of muscular use and the comfort of wearing the device for long periods of time. With this information, we will be able to review our designs and make adjustments accordingly, resulting in a redesigned product with increased utility. This testing and redesign process will be repeated as many times as time permits, or until the product design seems to be optimized.

To evaluate the effectiveness of the actuator in the future, the output force should be measured throughout the process, attempting to find the maximum amount. Initially, the force of the elbow actuator individually will be assessed, and then the force of the exoskeleton as a whole. Sensors should be put on the user to see how much effort and force they have to exert while wearing the device in comparison to the force exerted without the exoskeleton in use. Other evaluations include the range of motion and the air intake of the actuators, which will be measured in CFM using an air mass flow meter. The power usage of the actuator and exoskeleton as a whole should also be considered to determine the most efficient design. Once it is proven that the actuator and exoskeleton are successful in these parameters, other metrics may be considered, such as comfort, weight, and adjustability. These final metrics will be evaluated through working with doctors and patients at the UVA hospital to iterate and refine the design through patient-cooperative feedback.

Other future work includes the creation of a more versatile air pump. Our current system is controlled by a simple on or off air pump. In the future, a system that is more than binary on or off would give the actuators many more options regarding range of motion and strength. Additionally, we would like to make more efficient muscles in the future. A more effective tricep muscle would give the arm more range and strength in that degree of freedom. This could be assisted by an inverse pneumatic artificial muscle. A more soundly constructed inverse pneumatic artificial muscle should be worked on in the future to better assist both the bicep and tricep motions.

ACKNOWLEDGMENT

The authors acknowledge Sarah Sun for assisting us in the development of the wearable exoskeleton design.

REFERENCES

[1] Nassour, J., Zhao, G., & Grimmer, M. (2021). Soft pneumatic elbow exoskeleton reduces the muscle activity, metabolic cost and fatigue during holding and carrying of loads. Scientific Reports, 11(1), 12556. https://doi.org/10.1038/s41598-021-91702-5

[2] Al-Fahaam, H. S. H. (2019). Wearable exoskeleton systems based-on pneumatic soft actuators and controlled by parallel processing. Retrieved November 11, 2021, from https://core.ac.uk/reader/228133021

[3] González-Vargas, J., Ibáñez, J., Contreras-Vidal, J. L., Kooij, H. van der, & Pons, J. L. (2016). Wearable Robotics: Challenges and Trends: Proceedings of the 2nd International Symposium on Wearable Robotics, WeRob2016, October 18-21, 2016, Segovia, Spain. Springer.

[4] S. Kurumaya, K. Suzumori, H. Nabae, and S. Wakimoto, “Musculoskeletal lower-limb robot driven by multifilament muscles,” *ROBOMECH Journal*, vol. 3, no. 1, p. 18, Sep. 2016, doi: 10.1186/s40648-016-0061-3.

[5] E. W. Hawkes, D. L. Christensen and A. M. Okamura, "Design and implementation of a 300% strain soft artificial muscle," 2016 IEEE International Conference on Robotics and Automation (ICRA), 2016, pp. 4022-4029, doi: 10.1109/ICRA.2016.7487592.