Soft robotic glove for assistive and at-home rehabilitative use

Abstract

Soft robotics offers many advantages over traditional robotics as an assistive and rehabilitative technology due to its cost-effectiveness and ability to mimic physiological movements. Because of the importance of grip on a patient’s quality of life, we leveraged soft robotic technology to create a flexible, exoskeleton glove for both therapy and assisting daily living for patients with hand impairments. Our device targets muscular dystrophy patients because these patients have a limited selection of devices for their needs. Our design consists of soft robotic actuators attached to a glove that mimic finger movement by bending upon inflation and surface electromyography (SEMG) sensors on the arm to predict the user’s intention to grip or release. Our device effectively recapitulates the forces generated during the grip of a healthy individual using three soft robotic actuators, with each actuator generating 2.83 ± 0.44 N (n = 6). The actuation frequency is 0.2 Hz (12 grips/minute) due to delays in deflation. The EMG sensors, MyoWare and OYMotion Analog EMG Sensor, were not sensitive enough to pick up graded changes in muscle activity, meaning we could only replicate an on-off response through a thresholded spike detection of processed EMG signals in two muscle groups that corresponded to grip and release. Going forward, we will use more sensitive EMG sensors, such as the Biosignalsplux Electromyography Muscle Sensor, increase actuation frequency by actively deflating actuators and include a pressure - solenoid feedback system to have controllable grips.
Section 1. Introduction

Section 1.1 Motivation

Hand functionality is a vital factor in living an independent life. A partial or complete loss of hand motor ability results in patients being unable to perform activities of daily living (ADL) and considerably reduces quality of life and independence [5]. In the United States alone, 6.7 million people report difficulty performing a grasping motion [6]. Of those individuals, there are around 206,700 stroke survivors every year who require assistance with ADL due to upper extremity disability [6]. An estimated 714 males are born with Duchenne/Becker muscular dystrophy (DBMD) each year and struggle with hand mobility throughout their lifetime [6]. Based on an overall incidence rate of 1 in 3500, this translates to over 90,000 DBMD patients in the United States alone [6]. Over 200,000 spinal cord injury (SCI) survivors live with sustained neurological damage and require assistance with grip [6]. Given the aging population and increased risk of diseases such as stroke, there is an increased need for devices that can rehabilitate hand function to meet this growing market.

Though there are numerous diseases that lead to hand impairments, we chose to focus on the rare genetic disease, Duchenne/Becker muscular dystrophy. These patients lack normal dystrophin protein, which links muscle fibers to the extracellular matrix [10]. For most of these cases, patients then experience early onset muscle weakness and partial or total absence of hand motor ability [6], [10]. The effects of this disease are lifelong, beginning in adolescence with progressive degeneration over time. A better assistive and therapeutic device will have a great impact on the patient’s quality of life, since therapy can reduce the rate of progression of muscular degeneration [4],[5]. Though our device will focus on DBMD patients with partial hand motor ability, it can be easily adapted to treat other conditions such as cerebral palsy, stroke, and incomplete spinal cord injury and benefit those patients.

Section 1.2. Current DBMD Treatment Options

For muscular dystrophy patients with hand impairment, there exist several potential options to help increase hand mobility and grip ability. These options can be broken down into three groups: physical therapy, low tech devices, and robotics. These three options can either be used in a clinical setting or at home [4],[5],[6]. Typically, most of these devices are created for the clinic, with the low-tech devices being used at home.

Physical therapy has been used to improve hand function through the involvement of repetitive task practice. By breaking down a task into individual movements and practicing those individual movements, physical therapy improves the hand strength, accuracy, and range of motion of the patient [4],[5]. Task specific training with the affected hands has been shown to reorganize the cortex faster, producing better functional improvements [4],[5]. Current rehabilitative methods, however, are costly, slow, and labor-intensive, placing a high demand on the training and availability of the physical therapist [4],[5]. This option typically requires patients to travel to a clinic, making it a challenge for patients to take the time to travel between physical therapy and home.

Low-tech device options for DBMD patients include braces and tensile-driven assistive devices. Such examples include hand orthoses and the SaeboGlove. Braces are made to support
weakened muscles, helping to keep the muscles flexible [7]. Though a recent study found that overnight hand braces can help improve DBMD patients’ wrist extension mobility, grip strength and fine motor function, these options require further study before reaching market viability [7]. The SaeboGlove, on the other hand, is the market leader for tensile-based rehabilitation systems with gloves that allow an individual to incorporate hand therapy at home [8]. The SaeboGlove’s proprietary tension system extends the clients’ fingers and thumb following grasping, allowing the device to be used as both an assistive and rehabilitative device [8]. Priced at $399, the device is designed for patients with arm and hand weakness caused by neurological or orthopedic injuries. Individuals prescribed this by a licensed therapist must be able to close or squeeze their hand minimally. This restricts the patient population for the device to those with higher functionality and thus limits the indication for use [8].

Robotic devices can range from clinical-use, advanced robotics such as the Hand of Hope and home-use devices such as the Hand Tutor. Methods for implementing robotic devices to assist with rehabilitation of the upper extremity have not shown significant advantages over traditional approaches in a review of over 120 robotic devices [1]. This lack of evidence regarding efficacy of robotics over conventional therapies is thought to be attributed to rigid materials such as steel, which can serve to restrict motion in non-actuated directions and lead to joint alignment issues [1]. The Hand of Hope is a powered glove exoskeleton for rehabilitation in clinics and is used as a teaching device for relearning hand movements [3]. Unfortunately, the high cost of $25,500 for this device makes its implementation impractical for the clinical setting, and the device’s lack of mobility prevents it from being used as an assistive device [3]. Meanwhile, the Hand Tutor is an at-home exercise device that incorporates a computer and robotics to involve the patient in the rehabilitation [2]. The robotic aspect of the device sends feedback to the telecommuting therapist and allows the therapist to give objective feedback about the motion [2]. While the cost is a moderate $2,000 per device, this device is not reimbursed by insurance, thus limiting its availability to patients [2]. In addition, the Hand Tutor does not incorporate any assistive technology and the patient occasionally requires outside help to do certain movements.

Section 1.3. Impact of the Grip Glove

Our solution, with its use of soft robotic actuators, can help change the way we approach rehabilitation and therapy, all at the low target price of $600. Our solution can improve the benefits of rehabilitative therapy by providing greater affordability, portability, lower weight, easier customization, increased range of motion, safer human-robot interactions, and the ability to conduct task specific training or exercises that simulate ADL [4],[5]. By expanding patient access to therapy, we can increase the quality of life for patients through our personalized treatment plans programmed into the device that are consistent with contemporary rehabilitation exercises. Thus, our device would allow patients to take their therapy home, which improves patient compliance and outcomes. Moreover, the ability to use this device as both a therapeutic and assistive device will allow our prototype to be adaptable for patients with different indications in a universal, low-cost manner. This not only helps familiarize the patient with the device and lower the probability of patient rejection, but also provides high-quality care for all patients, regardless of their financial situation [4],[5]. As such, our prototype’s improved patient outcomes can reduce the societal costs for individuals with limited hand mobility.
Section 2. Objectives and Overview

Soft robotics has emerged in recent years as an alternative to traditional robotic rehabilitative therapies. Due to their compliance, ease of deformation, ability to conform to the contours of the human body and tunable mechanical properties, they possess better biomimetic capabilities than conventional materials [4]. Grip Glove aims to leverage the versatility of soft robotic technologies to assist with the gripping of non-deformable objects.

The main objective of our project is to utilize fiber-reinforced soft robotic actuators composed of a silicone main body, a Kevlar fiber radial strain limiting layer, and a fabric linear strain limiting layer, which are materials that have been used successfully in the literature for this purpose [5]. These soft robotic actuators are used to generate and sustain flexion of the fingers to assist with gripping everyday objects. The actuators contain a hollow semi-circular bladder that inflates when pressurized to create extending, bending, and twisting motions, or any combination thereof. By weaving the Kevlar fibers around the length of the actuator, radial expansion is restricted upon pressurization, forcing expansion in the axial direction. By applying a fiberglass strain limiting layer along the bottom of the actuator, linear expansion is restricted along one side of the actuator which serves to generate the bending motion trying to be replicated. The basic procedure for the fabrication of the soft-robotic actuators is adapted from the Soft Robotics Toolkit and is optimized for the materials and equipment presently available to us [5].

In order to detect an intended motion of the user, an electromyography (EMG) cuff will be placed on the forearm with sensors placed along both flexor and extensor muscles. This signal will be read into and processed by a microcontroller. This will regulate an air pump to apply a pressure to the soft-robotic actuators located on the top of the glove. Because of our device’s air pump use, we will have the potential to help patients with both extension and flexion, which creates better assistive qualities than that of the Saebo glove.

Although proof-of-concept devices for grip have been developed by teams at Harvard’s Wyss Institute for Biologically Inspired Materials, none have yet reached a marketable phase of production [4],[5]. However, these current devices have some limitations. The first is that their actuation system relies on pumping liquid through the device instead of air. Since the pump is already an external component to the glove itself, the liquid reservoir would only serve to add to the total weight of the device. The use of air pumps in our design can cut down on the complexity and weight of the external mechanisms of the device. Other gloves currently on the market utilizing soft robotics such as the Exo-Glove Poly only provide mechanical support for the thumb, middle, and index fingers, whereas our device can be easily expanded to facilitate motion in all 5 fingers.

Our prototype is intended for use as an assistive and rehabilitative device in both a clinical and at-home setting. Rehabilitative therapy in the home environment has been shown to increase psychological performance, which leads to shorter, more functional therapy sessions [1]. For our design, we integrated pre-programmed therapy regimens intended to target patient populations that can benefit from at-home therapy. These automated sessions can reduce the total time spent transitioning between a patient’s home and a rehabilitation center, ultimately leading to fewer hours of rehabilitation for which the patient is billed. The pre-programmed routines can also be adapted to suit each individual patient’s needs by accounting for the degree of motor impairment. Since the degree of hand impairment will vary for DBMD patients based on age and rate of
progression, the personalized regimens, decreased weight, and increased degrees of freedom of motion offered by a soft-robotic glove will allow for better patient outcomes.

**Section 3. Project Illustration**

An illustration of the concept of Grip Glove’s design is displayed *Figure 1*, which clarifies the key concept and solution.

![Figure 1. A Brief Pictorial Overview of Prototype’s Design:](image)

*Figure 1. A Brief Pictorial Overview of Prototype’s Design:* (Left to right) [1] The surface EMG (SEMG) signal from the user is read from sensors on the wrist and upper arm. [2] The SEMG signal integrates with the microcontroller to control the activation of the air pump. When the air pump is turned on, air flows through the tubing to provide air pressure to the soft robotic actuators mounted on the glove, causing the soft robotic actuator and subsequently the glove to bend. [3] This bending motion helps the user grip objects. This entire process can be used to aid user rehabilitation by automating the soft robotic actuators to make pre-programmed motions.

**Section 4. Specifications and Standards**

**Section 4.1. Specifications**

In order to meet our goal of creating a device that can be used as both an assistive and therapeutic device, there were three main categories of specifications we needed to evaluate: the fabrication and testing of the soft robotic actuators, the EMG signal acquisition and processing, and the overall glove design.

In order for our device to target patients with hand impairments, we needed to be able to effectively compensate for loss of muscle strength. The average gripping force encountered when performing everyday tasks is in the range of 10-15 N. Thus, our actuators were required to apply and sustain forces within this range over many cycles of actuation.

Though force is an important metric for our device design, the actuators also needed to be able to bend such that the physiologic range of motion is recapitulated. Thus, the actuators corresponding to the pointer finger, middle finger, and thumb had to be able to bend within 10% of the physiological range of motion of 180° (162°) for the thumb and 10% of the physiologic range of motion of 270° (243°) for the non-thumb fingers [5].
In order to use EMG signals to turn the pump on and off, we first needed to select muscle groups for both flexion and extension such that the characteristic signals were differentiable. We ultimately chose the brachioradialis muscle of the wrist and the flexor carpi ulnaris muscle of the upper forearm. This selection allowed for a grip-based trigger for the initiation of the actuators and an extension-based mechanism of release. Since our device is intended for both assistive and rehabilitative use, the actuators needed to be able to go through cycles of repeated flexion and extension that represent the iterative tasks often applied in physical therapy. Thus, an ideal actuation frequency of 30 cycles per minute (0.5 Hz) was the target for our device based on previous designs support such therapy routines [5].

On the processing side, our device should have a short latency between sensing the SEMG and the actuation of the soft robotic actuators. Because myoelectric arms and equivalent devices have a very small lag time, our device must have the time between processing and a completed motion to be less than 2.2 seconds [5]. Since our device is an assistive device, the device must allow the user to pick up everyday objects including, but not limited to, cups, food, and writing utensils. We tested this aspect of the device by attempting to use its functionality to write messages, make a cup of coffee, and pick up a bottle.

In addition, our complete prototype’s weight should be less than 0.5 kg for all components of the hand, which is based on the literature and previous devices [5]. To account for this, the only materials that were on the hand include the soft robotic actuators, the glove, the tubing and the sensors; all other materials (pump, battery, and electronics) were mounted in a small mobile device that can be carried or placed on a table. The glove design should be easily modifiable to account for different hand sizes and potential future scaleup.

Furthermore, to make this device accessible to patients, we sought to keep costs of manufacturing low, specifically below $350 in order to create typical profit margins of 40%. Thus, the device can be priced at $600, which can be reasonably covered by medical insurance. This will allow patients to have greater access to the device once it reaches the market in the future.

Our overall specifications for the device have not changed much from the previous semester, although we did not use the Box and Blocks test to evaluate our device since we did not have an IRB-approved pre-clinical study that would evaluate this metric. The Box and Blocks test is a functional test that access improvement of a patient’s grip by measuring dexterity. In addition, we made a more specific bending angle specification between actuators for thumb and non-thumb fingers due to the difference in physiology; the new specifications now take this difference into account.

Section 5. Design and Testing

Section 5.1 Design Process - Design Goals

Our design process was categorized based on specific design goals for each of the main device component. These components include: fabricating soft robotic actuators, controlling the movement of the soft robotic actuators using SEMG and the air pump, and creating the glove design.
For the soft robotic actuators, steps to achieve the goal of creating a durable soft robotic actuator included:

- Developing a process of soft robotic actuator fabrication to consistently meet the specific designations of a 20 mm radius and 10 mm height (from the cross-section of the soft robotic actuator) and can be made with variable lengths to accommodate different finger sizes

- Creating a method of attaching a radial-strain limiting layer to prevent the soft robotic actuator from expanding radially without interfering with the soft robotic actuator’s overall ability to bend

- Generating a method of adding the linear-strain limiting layer that is durable, prevents air leaks, and maximizes bending

To control the movement of these soft robotic actuators, we broke these steps down into two main components—reading SEMG and controlling the air pump. The steps included:

- Reading SEMG:
  - Targeting key areas on the arm that will reliably generate SEMG signals for patients with hand impairments
  - Interfacing the SEMG sensors with an Arduino for signal processing
  - Developing filters that will both eliminate noise for SEMG signals and distinguish between signals that correspond to flexion and extension
  - Creating a comfortable, durable cuff to incorporate SEMG sensors

- Air Pump Control:
  - Using digital switches and Arduino to control individual solenoid valves independently
  - Developing a reliable method using solenoid valves to control the air flow to the soft robotic actuators
  - Creating a process to imitate tasks commonly used in physical therapy sessions through controlling air flow to create pre-programmed routines
  - Fabricating a container to hold the air pump and other components to prevent the items from being damaged during transportation or device use

To create a glove to mount the actuators, the steps included:

- Developing an attachment system where the soft robotic actuators can be mounted
- Producing a glove design where the components are easily adjustable and replaceable

Section 5.2 Design Solution

Section 5.2.a Soft Robotic Actuators Production and Selection

The soft robotic actuator manufacturing process we used was derived from Galloway et al. [5] that would allow the actuators to better model the finger’s configuration and allow for us to customize the actuators’ size and length. The overall process is displayed in Figure 2.

Figure 2. Soft Robotic Actuator Manufacturing Process [A] Silicone rubber body with hollow core is created with steel half-round rod being placed inside of 3D printed mold filled with silicone. [B] Linear strain limiting layer adhered to the bottom of the rubber body via Silpoxy (Smooth-on, Macungie, PA). [C] Radial strain limiting layer is wrapped in a single-helical wrapping around the rubber body for a thumb soft robotic actuator or a double-helical wrapping around the rubber body for all other fingers. [D] Rubber body is placed in 3D printed mold with silicone to create the rubber layer encapsulating reinforcements. [E] Ends of rubber body are then capped with silicone and a vented screw is installed to allow air into the soft robotic actuator’s inner chamber. [5, p. 138].

To test the different properties of varying silicone components, we created actuators utilizing different silicones for the rubber body and rubber layer encapsulating reinforcements and tested different strain limiting layers. We utilized DragonSkin 30 (Smooth-On, Macungie, PA) and Ecoflex 30 (Smooth-On, Macungie, PA) for the silicone components; 4oz S Glass fiberglass (U.S. Composites, Inc., West Palm Beach, FL) and Sew Classics Spandex Knit Fabric (Jo-Ann Stores, Inc, Hudson, OH) for the linear strain limiting layer; and Kevlar string (McMaster Carr, Robbinsville, NJ) for the radial strain limiting layer due to their use in the literature. The silicone selected followed ASTM F2038 standards regarding silicone use in medical applications, ensuring that the silicone selected was biocompatible and safe for our intended use and users.

Safety factors such as durability and lack of over inflation of soft robotic actuators were observed and noted as the soft robotic actuators were inflated and deflated. We also tested for tensile strength and elongation at break according to ASTM standards to finalize material selection.
for fabrication. Our testing procedures adhered to ASTM standard D5053 for spandex fabric, ASTM D412 for elastomers, and ASTM D2105 for fiberglass. All of these standards outline tensile testing procedures for these different materials.

We used a set of five specimens for each tested fabric and fiberglass material (25 mm wide, 150 mm long) on an Instron® (Illinois Tool Works, Inc, Norwood, MA) test machine and averaged the values among the five specimens [17]. We fabricated the hybrid silicone/fabric composites with 3D printed molds to create silicone with a uniform thickness of 2.0 ± 0.2 mm, adhering the fabric to the silicone specimen through Sil-Poxy (Smooth-On, Macungie, PA), and then cut the silicone/fabric composite into dumbbell-shaped specimens. The specimens were tested 2 months after initial manufacturing.

After testing the material properties of the raw and hybrid silicone/fabric materials, the manufactured soft robotic actuators were created and evaluated by their force generation, bending degree angle, and safety factors (i.e. no over inflation and durability). Force generation was measured via a force sensor that was placed in front of a fixed soft robotic actuator and measured the force generated at the tip of the actuator.

To measure bending degree angle, we fixed an actuator at its inlet end and induced pressurization of the actuator. The motion capture software Kinovea was then used to quantify the bending angle of the actuator.

| Table 2: Tensile Strength and Elongation at Break for Different Testing Materials |
|-------------------------------|-------------------------------|------------------|
| Testing Material              | Tensile Strength (MPa)       | Elongation at Break (%) |
| Fiberglass                    | 25.4237                       | 51%               |
| Spandex Fabric                | 3.5714                        | 235%              |
| Ecoflex 30                    | 1.2805                        | 246%              |
| Dragon Skin 30                | 3.0861                        | 127%              |
| Dragon Skin + fiberglass      | 7.3582                        | 46%               |
| Ecoflex + fiberglass          | 6.3992                        | 59%               |
| Ecoflex + fiberglass + DragonSkin 30 | 6.5637       | 158%              |
| DragonSkin 30 + spandex fabric | 3.5789                       | 160%              |
| Ecoflex + spandex fabric      | 2.7493                        | 175%              |

Our testing showed lower than expected values for elongation at break compared to other observations in the literature and data sheet [17]. This may have been due to the long period of time between manufacturing and testing. However, our tests had comparable tensile strengths for the silicone materials, which is relevant to overall durability with regards to permanent plastic deformation. The tensile strength observed for the samples showed that these materials have potential to cyclically expand and contract upon pressurization from an air pump.
Overall, the purpose of examining tensile strength and elongation at break shows that adding high modulus fabric increases load capacity in silicone such as Ecoflex 30 and DragonSkin 30 that has a relatively low elastic modulus, meaning higher elongation but lower tensile strength for increased strength and durability of the silicone. The combination of the Ecoflex 30, Fiberglass, and DragonSkin 30 showed a noticeable increase in tensile strength without sacrificing elongation, unlike the Ecoflex 30 type and fiberglass combination alone. We believe this is due to the second layer of silicone preventing the less elastic fiberglass from breaking. Because of this, we decided to focus our attention on using fiberglass as the linear strain limiting layer moving forward in the testing phases. The combinations tested are shown in Table 3.

After conducting trials, with results recorded in Table 3, we noted that the DragonSkin 30 was more resistant to expansion, so serving as the rubber body of the soft robotic actuator was not ideal. However, this property made the DragonSkin 30 ideal for the rubber layer encapsulating reinforcements. Ecoflex 30 rubber bodied actuators without DragonSkin 30 reinforcements over inflated.

Table 3: Different Soft Robotic Actuator Fabrication Methods

<table>
<thead>
<tr>
<th>Silicone Type</th>
<th>Linear Strain Limiting Layer</th>
<th>Radial Strain Limiting Layer</th>
<th>Rubber Layer Encapsulating Reinforcements</th>
<th>Bending Angle for Non-Thumb Actuator (°)</th>
<th>Force Generation for Non-Thumb Actuator (N)</th>
<th>Bending Angle for Thumb Actuator (°)</th>
<th>Force Generation for Thumb Actuator (N)</th>
<th>Safety Factor Observations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ecoflex 30</td>
<td>Fiberglass</td>
<td>Kevlar String</td>
<td>Ecoflex 30</td>
<td>260.8 ± 5.1</td>
<td>1.75 ± 0.15</td>
<td>170.8 ± 3.5</td>
<td>1.45 ± 0.39</td>
<td>Force of the air pressure caused over inflation that broke the rubber layer</td>
</tr>
<tr>
<td>Ecoflex 30</td>
<td>Fiberglass</td>
<td>Kevlar String</td>
<td>DragonSkin 30</td>
<td>255.8 ± 3.7</td>
<td>2.12 ± 0.26</td>
<td>165.8 ± 4.2</td>
<td>2.01 ± 0.56</td>
<td>Force of the air pressure caused over inflation that broke the rubber layer after repeated trials</td>
</tr>
<tr>
<td>Ecoflex 30</td>
<td>Fiberglass</td>
<td>Kevlar String</td>
<td>DragonSkin 30 (2 Layers)</td>
<td>253.6 ± 3.1</td>
<td>2.95 ± 0.32</td>
<td>163.4 ± 2.9</td>
<td>2.79 ± 0.34</td>
<td>Secure under air pressure</td>
</tr>
<tr>
<td>Ecoflex 30</td>
<td>Spandex Fabric</td>
<td>Kevlar String</td>
<td>EcoFlex 30</td>
<td>259.2 ± 3.8</td>
<td>1.20 ± .33</td>
<td>172.2 ± 4.2</td>
<td>1.07 ± 21</td>
<td>Force of the air pressure caused over inflation that broke the rubber layer</td>
</tr>
<tr>
<td>DragonSkin 30</td>
<td>Fiberglass</td>
<td>Kevlar String</td>
<td>None</td>
<td>30.4 ± 1.5</td>
<td>1.92 ± .25</td>
<td>29.3 ± 2.7</td>
<td>1.76 ± .11</td>
<td>Secure under air pressure</td>
</tr>
</tbody>
</table>

Based on the force generated and bending angle, the fabrication that had the highest force generation and met the degree angle specification was the combination of an Ecoflex 30 Rubber Body, Fiberglass linear strain limiting layer, and Dragonskin 30 rubber layer encapsulating reinforcements. Thus, the finalized design methods included the aforementioned combination of materials in addition of the Kevlar string radial strain limiting layer. Actuators fabricated using
this method possessed the ability to maintain shape without over inflating and accurately modeled finger flexion. A paired t-test between thumb and non-thumb actuators yielded a p value > 0.05, thus allowing us to not reject the null hypothesis and show that there is no significant difference in force generation between the thumb and non-thumb actuators. The force generation at the tip for this fabrication method was $2.83 \pm 0.44$ N ($n = 6$) and bending angles of $253.6 \pm 3.1^\circ$ ($n = 3$) for non-thumb actuators and $163.4 \pm 2.9^\circ$ ($n = 3$) for thumb actuators.

After finalizing the materials, we evaluated the reproducibility and actuation frequency of the final design through the following methods.

The reproducibility of the soft robotic actuator fabrication process was also assessed by measuring actuator dimensions with a caliper. The manufactured actuators had a $21.3 \pm 0.1$ mm radius and $10.3 \pm 0.1$ mm height ($n = 30$). This showed that our actuators were produced consistently with dimensions that are similar in length to a finger and have a low profile.

To quantify the actuation frequency, we fixed the actuators at their inlet end and turned on the air pump. Once the actuators were fully extended, the pump was shut off and the actuators were allowed to return to their equilibrium position, before the pump was turned on again and the process repeated and timed. This yielded an actuation frequency of 12 cycles per minute (0.2 Hz) which was below our intended specification. This was primarily due to the fact that we were not actively deflating the actuators with a second pump, which led to a longer actuator relaxation time.

Section 5.2.b EMG Signaling

To actively process the EMG signals, we initially used two MyoWare Muscle sensors placed on the wrist and the upper arm, which mapped to muscle groups involved in flexion and extension, respectively. The MyoWare sensor utilized gel-based electrodes that produced a better-quality signal compared to metal strip-based sensors. The MyoWare sensor outputs a rectified and integrated EMG signal and contains an onboard adjustable-gain potentiometer. For each signal, we used a digital high pass filter with a cutoff frequency of 2 Hz to remove baseline drift. EMG signals are intrinsically susceptible to noise.

One of the largest sources of noise we encountered at this stage of the design process was motion artifacts. Motion of the cable attaching the electrodes to the amplifier, and variable contact at the skin-electrode interface are two ways in which motion artifacts can be introduced into the system, as muscle contractions will cause a concomitant shift in the skin and electrodes. This typically creates noise from motion in the frequency band between 1 and 10 Hz [11]. We mediated the issue of variation in the contact interface through the addition of elastic cuffs that served to cover the sensors and secure the cable. We also eliminated false positive readings through the addition of an accelerometer (Adafruit, New York, NY) that would disregard a grip/extension detection based on an acceleration threshold.

After filtering of the signal, we calculated the line length across a moving 10-sample window using Equation 1. Line length is defined as the sum of the absolute values of the differences between neighboring data points over a specified time interval [16]. Line length was chosen as a classifier for muscle activity due to its simplicity and prior use in the detection of high activity electrical events that occur during seizures [15]. The line length for each signal was then
smoothed using a moving average filter with a 5-sample window length to reduce the chance of obtaining secondary peaks.

\[ \text{Equation 1:} \sum_{i=2}^{N} |x_i - x_{i-1}| \]

We first calibrated each sensor for ~10s and took a 0.75x scaled version of the maximum peak value during that calibration to use as a threshold. Instructions were displayed on an LCD display. Once these values were finalized, any peak that occurred above the grip or extension thresholds would trigger the actuators to bend or relax by means of controlling power to the pump, respectively as displayed in Figure 3. However, we noticed that it was difficult to define locations such that grip and extension peaks did not occur nearly simultaneously. In the case that both readings had peaks above their respective thresholds within a certain window of time, we needed to prioritize extension as the final interpretation to the system when making the decision to keep the pump on or shut it off when signals for extension was followed quickly by a random spike in the grip signal.

![Figure 3. Representative graph illustrating the basics of the detection algorithm. Once the grip signal (black) exceeds its threshold (blue line) and peaks (red dot), the pump is turned on and the actuators will bend accordingly. The system will wait for the extension signal from the OY Motion sensor (magenta) to pass its threshold (green) and spike (cyan dot) before the pump is shut off, allowing the actuators to relax.](image)

In order to mitigate this issue, we replaced the MyoWare sensor for extension with an OYMotion Gravity EMG sensor for both its ease of use and because it was not as susceptible to motion in the upper arm, thus yielding a more distinguishable result. This sensor functioned by rectifying the signal and then applying an envelope function that set all values below a
predetermined threshold to zero. Thus, we selected one MyoWare muscle sensor to monitor for grip and one OYMotion Gravity EMG sensor for extension.

In order to test the accuracy of the EMG sensors we conducted 4 sets of 20 trials with three users, with each set being taken on a different day to account for day-to-day variation in electrode placement. An individual trial was considered successful if, after calibration, the device was able to successfully detect a grip followed by an extension. Any response that was not consistent with an intended motion was classified as a failure. Failures and successes for trials were classified as 0 and 1 respectively. One-way ANOVA tests between different users on the same day and paired t-tests between the same user on different days generated p values > 0.05, thus showing that there is not significant difference in accuracy for individual users on different days or different users on the same day. Through this method, we thus calculated that the algorithm was 75 ± 5% accurate. Since we were not able to test the device on DBMD patients, test subjects for the device were in otherwise healthy condition and only attempted to initiate the actuators through a minimal degree of exertion and muscle use to better represent the target population.

Section 5.2.c Air Pump Selection

<table>
<thead>
<tr>
<th>Pump</th>
<th>Air Flow (Liters per minute)</th>
<th>Cost ($USD)</th>
<th>FDA Approved?</th>
<th>Number of Actuators Pump is capable of simultaneously fully inflating</th>
<th>Time to Actuators Inflating (seconds)</th>
</tr>
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<tbody>
<tr>
<td>D2028</td>
<td>13</td>
<td>$14.95</td>
<td>No</td>
<td>3</td>
<td>1.1 ± 0.1</td>
</tr>
<tr>
<td>TTC Miniature Diaphragm Pump</td>
<td>8.8</td>
<td>$321.00</td>
<td>Yes</td>
<td>2</td>
<td>1.4 ± 0.1</td>
</tr>
<tr>
<td>BTC-IIS Series</td>
<td>11</td>
<td>$321.00</td>
<td>Yes</td>
<td>3</td>
<td>1.2 ± 0.1</td>
</tr>
</tbody>
</table>

We evaluated different air pumps using the above specifications in Table 4 to find the most efficient and cost-effective air pump. Based on the results in Table 4, we decided to use the D2028 air pump. Even though it was limited to 3 actuators, the D2028 air pump was able to outperform the FDA approved pumps at a much more economical price. Because the non-FDA pump had vibrations that may disturb a user, we secured the pump and reduced the vibrations through the use of the dampening material Sorbothane® (Sorbothane, Incorporated, Kent, OH) to meet cost and comfort specifications.

Due to the limitations of the pump, we ultimately decided to utilize 3 actuators in our final design. With the 12 V air pump only being rated with a flow rate of 12-15 liters per minute, actuator performance declined significantly when the air flow was split amongst 4 actuators. This would have required the addition of a second pump and 12 V battery, thereby increasing the total weight of the device.

To control the flow of air, we used 12 V solenoid valves (U.S. Solid, Cleveland, OH) to stop the flow of air for the rehabilitative aspect of the glove. We used 3 solenoid valves to control each individual finger by using a transistor to act as a digital switch and allow the microcontroller to turn the solenoid valves on and off. We used this same transistor setup to control power to the air pump in order to regulate the glove’s assistive capabilities, since the EMG sensors were not
sensitive enough to allow us to control each individual finger. Using an Arduino, a push button, and LCD screen, we allowed users to set the number of grips or finger curls that they could use for rehabilitation. After the user set the number of exercises, the Arduino then controlled the solenoid valves and the air pump to allow an individual to perform those exercises.

Section 5.2.d Glove and Cuff Design

![Glove Design](image)

To achieve a more versatile design that could allow for easy replacement of soft robotic actuators and adjustments to different users, the design displayed in Figure 4 was created. The double-sided velcro sewn into the main body of the glove and the soft robotic actuator and finger attachments allows the user or an assistant to modify the glove easily, making the design more universal. Based on the results of the air pump selection, we sewed on three pieces of double-sided velcro to accommodate the three soft robotic actuators. The Sew Classics Spandex Knit Fabric (Jo-Ann Stores, Inc, Hudson, OH) was chosen due to its lightweight properties, flexibility, and comfort. The accompanying adjustable cuff was used to secure the EMG sensor to the user and contain the wires and air pump tubing.

Section 5.3.a Final Product and Process Flow

Based on Section 5.2, the final design (pictured in Figure 5) was fabricated. The process flow of the final design is pictured in Figure 6, Figure 7, and Figure 8, which explain the SEMG assistive device processing, the pre-programmed grip control rehabilitative program, and the pre-programmed finger curl rehabilitative program respectively.
Figure 5: Final Design Prototype [A] Glove with mounted soft robotic actuators and SEMG sensor components on the upper arm and wrist [B] Mobile device containing the air pump, microcontroller, and other electrical components. Note: Wires and tubing (not pictured) connect our prototype and the mobile device.

Figure 6. Process Flow for SEMG Controlled System
Figure 7. Process Flow for Pre-programmed Grip Control Rehabilitative Program

Figure 8. Process Flow for Pre-programmed Finger Curl Rehabilitative Program
Section 5.3.b. Testing and Results for Current Prototype

The current prototype meets most of our specifications. In total, the cost of the current prototype is $279.26, which is lower than our target cost of $350. In the current iteration of the design, the total weight of all hand components amounted to 151.3 g, which was well below our target specification. This ensures that the device is portable and reduces the amount of physical exertion the patient has to output when using the device.

We used a hand dynamometer to quantify the total force generated by the final prototype. Since grip strength can vary with the angle of the arm, we followed recommendations set forth by the American Society of Hand Therapists (ASHT) and have the dynamometer measurements taken with the elbow at a 90-degree angle [9]. Measurements were carried out by inflating the glove with 1, 2, and 3 actuators attached to the glove while gripping a Camry Digital Hand Dynamometer (Camry Scale, South El Monte, CA). This was performed without a user wearing the glove and any additional human-aided grip. When tested, this experimental design generated forces of 3.2 ± 0.2 N, 6.6 ± 0.5 N, and 10.3 ± 0.3 N, for the 1, 2, and 3 actuator configurations, respectively, showing that we are able to meet force generation specifications. The actuators utilized in the final prototype exhibited bending angles of 253.6 ± 3.1° (n = 6) for non-thumb actuators and 163.4 ± 2.9° for the thumb actuators, thus continuing to meet the bending angle specifications.

In order to ensure that latency between signal detection and a fully-completed motion was below the specification of 2.2 s, we video-recorded the device and EMG readout (using Arduino serial plotter) simultaneously. When the grip EMG signal passed above its threshold value, we timed how long it took to reach a fully-actuated state; this was found to take 1.1 ± 0.1 s showing that our device’s response is sufficiently quick.

Unfortunately, due to time constraints, we were unable to launch an IRB approved, pre-clinical study with a muscular dystrophy patient. As a result, we tested this device on able-bodied users to show the proof of concept.

Using the final prototype, we tested the accuracy of the prototype using the ability of a user to grip and then perform tasks of writing messages with a pen, picking a cup of coffee, and picking up a water bottle. For each of the above actions, we conducted 4 sets of 20 trials across three users. Success of the action was measured by a user’s ability to grip and release objects in these tasks within a time period of 5 seconds to match our prototype’s actuation cycle. The user conducted these trials with minimal human activity. We controlled for the experiment by conducting a control trial in which the user was not wearing the glove but still controlling the glove’s movements via SEMG. The user would cause the glove to grip around an object, another person would pick up the glove to simulate the user manipulating object before setting the glove back down, and the user would then release the grip. The control trials were as successful as the non-control trials, with the inaccuracy being due to the EMG sensors. A single-trial ANOVA test for different users on different days performing different tasks and paired t-tests for the same user performing a specific task on the same day also yielded no differences across users in terms of accuracy and across actions, with p values > 0.05. Our accuracy tests were similar to the SEMG sensors’ accuracy, with an accuracy of performing tasks within the time frame 71 ± 7% of the time. Ultimately, this showed that any potential activity of the user was not contributing to the device generated grip and
our device is able to aid patients grip and the accuracy of the final prototype did not vary across users.

We also tested the rehabilitative aspects, the pre-programmed gripping motions and the SEMG grip control. The pre-programmed gripping motions and grip control ran successfully without any errors; however, without clinical data from patients, we were unable to quantify the pre-programmed action’s effectiveness.

Section 5.2.c. Discussion

Though we did not have time to launch an IRB approved pre-clinical study with a muscular dystrophy patient, we were encouraged by the design’s preliminary performance. The final device was able to aid grip and potentially help with ADLs. However, given that our test subject was in an otherwise healthy condition, only qualitative user feedback of improvement in the grasping and manipulation of these objects was possible. To employ a more quantitative approach to detect improvement over the patient population’s baseline dexterity, we would like our future users to improve their ability to grip by demonstrating a 40% increase in picking blocks on a Box and Blocks test, a standard test to measure gross manual dexterity of a patient, based on the pre-clinical results of similar devices [5].

The pre-programmed microcontroller responses that simulate grip and extension routines could impact the key stakeholders by making rehabilitation more convenient and accessible, which can, in turn, improve outcomes [11],[12].

Though the SEMG control is not yet accurate enough, the current accuracy, with enough training, could potentially allow hand impaired users to be able to use this as an assistive device by moving the SEMG sensors to another location, such as the shoulder, which would generate higher SEMG signals and be gesture based release as opposed to extensor based release. Because the design is adaptable, this implementation is possible in the future. Ultimately, this device will allow users to improve their grip ability, thus, providing users with a higher quality of life and better patient outcomes. The next steps would be to test this device with a patient using the Box and Blocks test with the SEMG sensors and pre-programmed microcontroller responses to examine improvement for both aspects respectively.

Section 7. Conclusions and Recommendations

We were successfully able to create a soft robotic exoskeleton glove that can mimic finger movement to help those with impaired grip. We were able to meet key specifications, such as force generation, movement, weight, and cost. Our device effectively recapitulates the forces generated during the grip of a healthy individual using three soft robotic actuators, with each actuator generating 2.83 ± 0.44 N (n=6). Though the SEMG sensors on the arm were not as accurate as we hoped, only identifying a grip-extension cycle 75 ± 5 % of the time, and the soft robotic actuators had lower than expected actuation frequencies of 0.2 Hz, the device still has the potential to be used as an assistive device due to the mechanical fit and as a rehabilitative device through pre-programmed actions. This device has the potential to deliver better patient outcomes for those with hand impairments, such as muscular dystrophy or stroke patients, by assisting their grip and providing potential for at-home rehabilitation. Future directions for this design may also require
replacement of the SEMG sensors in lieu of more sensitive, but more expensive SEMG sensors such as the Biosignalsplux Electromyography Muscle Sensor. In doing so, we will be able to use more sensitive SEMG signals to allow a user to modify grip strength accordingly. Additional features would be to actively pump out air as the soft robotic actuators deflate in order to increase actuation cycle frequency and to include air pressure sensors to create more nuanced pre-programmed responses for the device’s rehabilitative properties. In the future, we would ideally also include a pressure - solenoid feedback system to have controllable grips and varying grip strength.
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References:


