Classification of Fabric Based Soft Actuators and Feedback Controller for At-home Hand Rehabilitation

by

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ABSTRACT

With an aging population, the number of later in life health related incidents like stroke stand to become more prevalent. Unfortunately, the majority those who are most at risk for debilitating heath episodes are either uninsured or under insured when it comes to long term physical/occupational therapy. As insurance companies lower coverage and/or raise prices of plans with sufficient coverage, it can be expected that the proportion of uninsured/under insured to fully insured people will rise. To address this, lower cost alternative methods of treatment must be developed so people can obtain the treated required for a sufficient recovery.

The presented robotic glove employs low cost fabric soft pneumatic actuators which use a closed loop feedback controller based on readings from embedded soft sensors. This provides the device with proprioceptive abilities for the dynamic control of each independent actuator. Force and fatigue tests were performed to determine the viability of the actuator design. A Box and Block test along with a motion capture study was completed to study the performance of the device. This paper presents the design and classification of a soft robotic glove with a feedback controller as a at-home stroke rehabilitation device.

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INTRODUCTION

1.1 Background

In a world where people are living longer, it follows that there would be coupled increase in later life ailments (Virani *et al.* (2021)). One such ailment is a stroke. Virani *et al.* (2021) found that in the US alone there are 800,000 new stroke patients a year and this value will increase as the number of people living to over 85 years increases. A stroke is an event where there is a focal and/or global interruption of cerebral function, generally having a vascular cause, lasting over 24 hours or until death. When this occurs, blood flow to the brain is interrupted/reduced preventing the circulation of oxygen and nutrients to the brain/part of the brain for a period of time. There are two common types of stroke: hemorrhagic and ischemic which are respectively caused by a bleed in the brain or a blocking of blood vessel in the brain. There are multiple causes of stroke with the long term affects being primarily based on the location and extensiveness of the affected area. One of the most common disabilities caused by stroke is paralysis or hemiplegia. This can cause other long term issues like the development of clots which can again cause a stroke. Given that 16% to 18% of stroke survivors will have another stroke within 5 years, those who suffered from one stroke are at a higher risk of having another causing further disability or even death (Feng *et al.* (2010);Virani *et al.* (2021).

A patient recovering from stroke can be expected to require extensive physical therapy and/or occupational therapy to regain mobility in their upper and/or lower extremities. This is a lengthy process where the amount of received care directly affects the speed of recovery (Zhang *et al.* (2017); Sasaki *et al.* (2013)). Kwakkel *et al.* (2004) found that a period of 6 months was the optimal general length of treatment for stroke. The issue with the current model of care is that patients rely on their insurance to cover the costs of therapy, but policies and coverage are constantly changing. Asaithambi *et al.* (2021) found that of the 800,000 people who may suffer from stroke in a given year, 55% of them may be under-insured or have no insurance. This requires them to pay for healthcare expenses out of pocket or receive treatment with strict restrictions, but those with sufficient insurance can still face these restrictions. There are two ways insurance covers therapy: allotted number of visits a year and length of reimbursement period. The issue being that both of these methods face decreasing coverage year to year (Medford-Davis *et al.* (2016)). Insurance generally covers 7-15 therapy sessions a year which causes the therapist to ration sessions, potentially impacting the recovery of the patient. From discussions with therapists at Barrow Neurological Institute, therapists will front load therapy sessions since a long time between sessions initially can negatively impact the tone (muscle tension) of the patient. If the there is a set window of time a patient can see a physician, this may be limited to three months or even less. Since not every patient is the same, not everyone can undergo the same amount of therapy within this coverage period. While a therapist can request for more time, insurance rarely approves extensions without sufficient progress being made with the patient due to concerns pertaining to insurance fraud.

To improve patient outcomes, therapists have begun adopting robotic devices to assist with their regular exercises. While still new, the most commonly seen robotics in therapy are lower limb robotic exoskeletons to aid in relearning how to walk, and robotic hand coverings to aid in regaining hand ability. The hand robotic devices employed now are confined to medical offices due to their complexity of use and cost. Many of the current devices have mechanical drawbacks as well, usually lacking in one or more desired qualities. The main robotic systems used are either classically rigid mechanic or soft actuator driven. Both machine types can be around $20k and have individual problems as well. Rigid devices are large and do not conform well to a user's. The weight and supporting machinery limit the uses of the devices as their operational space is restricted to large medical facilities. Most soft robotic devices avoid this problem by using smaller support systems and simpler articulation methods. However, most soft devices cannot be used in a practical setting since they many do not track the movement of the individuals hand or have dynamic controls; limiting the devices to a few movements such as open, close, pinch, and full grip.

Recently, a shift has been occurring in healthcare delivery methods driven by covid impacts in in-person activities and people seeking convenient methods to receive medical care. This has driven the telehealth market and has permanently shifted how healthcare can be delivered to patients. While adoption was accelerated by covid, McFarland *et al.* (2021) found that telehealth has been helpful for overcoming access to care issues such as lack of specialists in an area and transportation to and from therapy. By allowing specialists to see patients online, more people can receive the care they need.

1.2 NSF I-Corps Program

This project was fortunate to be chosen by the NSF as a participant in their national NSF I-Corps program. The I-Corps program is a seven week course where project teams work with stake holders and mentors to develop a business model canvas. Primarily, this consisted of interviewing at least 100 stake holders, which translated to therapists, therapy center leadership, and medical billing specialists. After completing these interviews, it became apparent that there are ubiquitous problems with respect to receiving care within the United States. The main problems found were a lack of reliable transportation for patients to and from facilities, lack of uniform availability of specialists and facilities, an at-home device needs to be able to track patient progress. Also, insurance is increasingly difficult to deal with as it rarely approves extension of care requests and while also lowering allotted sessions per year or decreasing the window of reimbursable care. As therapists did not see insurance changing its current behavior, they desire devices that may help augment or accelerate the treatment they can give patients to fit within their insurance plan.

1.3 Problem Statement

There is a growing need for convenient healthcare as the population ages and the healthcare system becomes overwhelmed. Currently, a majority of people who may have a stroke event from will not be able to receive the care they need due to the lack of insurance coverage. Therefore, there is a demand for at home treatment tools that can be used to augment in person healthcare visits.

However, current devices do not meet this need. This is primarily attributed to the high cost of devices making them a liability for hospitals to loan out to patients and the devices requiring physician oversight due to lack of patient tracking. Several soft robotic gloves have been developed by research groups, but these systems have limited controllability and actuator performance. These hurdles may be able to be overcome by employed fabric based soft actuators with a feedback control scheme. By taking steps to produce a low-cost device that allows for remote care, it can further the current state of at-home focused medical treatment devices.

1.4 Research Questions

This section contains the research questions that will guide the development of a soft robotic glove for at-home hand rehabilitation described in this thesis.

- What are the functional requirements for these actuators to assist hand function?
- What information should be used for feedback control of the actuators?
- Can these actuators be reliably controlled with a dynamic controller rather than with pre-defined states?
- Can these actuators be effectively used in rehabilitation?

1.5 Objectives

The primary objectives are to design and construct a glove that employs fabric based soft actuators and soft sensors that can perform hand therapy exercises. This will consist of the design of fabric based soft actuators and their integration into a glove with embedded soft sensors. After the creation of such actuators, they can be used in force and fatigue tests to determine their viability in a practical setting. With the inclusion of the glove, the device can be used in a physical therapy exercise and motion capture study. This data will help determine if this device can he used to perform rehabilitation exercises.

1.6 Scope

One soft robotic glove was developed to be used in this research. This thesis covers the development of a soft robotic glove to aid hand rehabilitation for an at-home application through the analysis of design and controls testing. Encompassed within is the design and classification of soft pneumatic actuators and soft capacitive bend sensors which drive the functionality of the device. The Hardware and Design section covers the development of the physical systems and the Software and Control section explains the controllers used. The results of performance tests are further described and explained with their relation towards the desired characteristics of an at-home rehabilitation device.

1.7 Limitations

A limitation of this study is the availability of commercial soft sensors that can gauge angle. Since the scope of this project does not encompass soft sensor creation, it was decided to use easily available products on the market. Current products fall within two categories: resistive and capacitive. Resistive sensors were not chosen due to the undesirable characteristics of their drift, low resolution, and lack of robustness to external forces. Capacitive sensors were chosen due to their accuracy and path independent readings. Since there are very few commercially available capacitive flex sensors, the flex sensor from Bend Labs was used in this study.

Given the scope and timeline of the project, a three finger glove was designed to provide a two or three finger pinch test. While this does not provide the user with the ability to perform gross grasp exercises however, the current design can easily be extended to five fingers. This is due to the modular design of the controller allowing a user to simply add another regulator to the electronics and copy previous blocks of code with slight changes per additional actuator.

1.8 Outline

Chapter two presents a review of previous work in the area of soft robotic actuators. This is followed by a discussion on the bio-mechanics of a human hand and theory behind rehabilitation practices. The section concludes with a discussion of the current state of soft robotics gloves. The body of knowledge from these sources laid the foundation for the developments of this thesis.

Chapter three covers the development of the soft robotic hardware. This covers the design of the glove and actuators as well as describing the electronics used in the device. The reasoning for design and manufacturing methods is also explained.

Chapter four discusses the creation of the sensor data collection software as well the controllers used in this study. This includes why certain methods were used and how problems were solved during development.

Chapter five present the tests and results done on the device and its components. Force delivery of the actuators and their performance in a rehabilitation exercise is verified along with the performance of the glove with and without a human hand. Further, the reason for the each test is explained as well as their setup.

Chapter six completes this thesis with the discussion of the results and conclusion.

Challenges found during testing are explained as are the potential impact of these results. Based on the results of the tests, future development and tests are also described.

Chapter 2

LITERATURE REVIEW

2.1 Soft Actuators

In the field of soft robotics there are many differing modes to achieve actuation. Common methods include dielectric electrically responsive, magnetically responsive, thermally responsive, and pressure driven. Electrically driven actuators encompass dielectric elastomers, piezoelectric, and motor driven cable bowden systems (El-Atab *et al.* (2020)).

Dielectric actuators, HASEL actuators, operate based on Coulombic attraction between two flexible electrodes which contract the actuator when an electric signal passes through the dielectric liquid. While these offer fast actuation speeds and have compounding strength when stacked together, they require large voltages to articulate (Shintake *et al.* (2016)). Piezoelectric actuators, when an electric field is applied, produce a mechanical deformation however these also require large voltage to articulate (Sohn and Choi (2017)). Cable Bowden systems are a common method of articulation where motors control the spooling of cables to mimic the behavior of tendons (Yurkewich *et al.* (2019)). These offer precise control of actuators; however the large support system hinders the adoption of these systems outside of stationary devices.

Magnetically responsive actuators deform when a magnetic field is applied, causing a deformation of a material or fluid such as ferrofluid. Jeon *et al.* (2019) demonstrated a ferromagnetic material based soft actuator on a sub-millimeter scale which can steer and navigate in any direction using magnetic actuation. A problem with these actuators is patterning the magnetic materials to achieve desired behaviors but, Wang *et al.* (2020) developed a technique for patterning magnetic microparticles within an elastomer. This has led to the development of planar magnetic actuators that are much easier and cheaper to produce. The largest drawback of these actuators is the need of significantly sized magnets to produce strong enough magnetic fields to achieve actuation ie. 800μm material thickness requiring a 20mT magnetic field (Wang *et al.* (2020)).

The most common thermally responsive actuators are shape memory alloy (SMU) and shape memory polymers (SMP). In both of these implementations, deformations occur when heat is applied to the system. This can be done by running a current through the materials, causing heat to be generated which in turns causes the material to revert to its original shape. While these actuators display precision motion, their slow articulation speed and weak generated force prevent their use in most settings (Rodrigue *et al.* (2017)).

Pressure driven actuators exploit radial expansion in order to generate desired movements (Connolly *et al.* (2015)). These actuators use hydraulics or pneumatics to increase the pressure within their bladders. To create directed motion, actuators employ a restrictive material to limit expansion in chosen direction. The actuators are usually created with compliant fabrics or silicone with embedded restrictive materials such as non-extensible threads, fabrics, or thicker layers of silicone. By changing the patterning of these non-extensible layers within the actuator, one can create different behaviors such as bending, twisting, extension, and contraction. These patterns can also be combined to have non continuous bending or compound movements such as twisting and bending. This offers an attractive level of customization while retaining an ease of manufacturability and and low cost.

2.2 Biomechanics of the Hand

Before designing the soft robotic device, product requirements needed to be determined from prior biomechanical studies on a fully functioning human hands. Since this device means to improve the grip and mobility of a patients hand, the main variables needed to be followed were grip strength and the joint space of each finger. There are two difference values of interest with regard to grip strength: the overall average force required to hold everyday objects and force delivery at the fingertips. While grip strength is necessary to know for the holding of everyday objects, this study primarily is focused on the restoration of pinch force in patients.

Blennerhassett *et al.* (2006) studied the strength of pinch grip post stroke recovery and found that a range of 4-6 N of force delivery at the fingertip was necessary for the successful gripping of objects in a two finger pinch test. Below 4N slippage can occur causing the grip to fail. In gross grasp of sub-maximal static grasping, for a 1kg object, the force delivery at each finger tip can be 5.7N, 3.8N, 2.9N, and 2.6N for the Index, Middle, Ring, and Small finger respectively (Kim *et al.* (2014)). Matheus and Dollar (2010) found that a grip strength of 9-15 N is sufficient to hold everyday objects without slippage. For a healthy adult hand, the functional range of motion can be defined as the range required to perform 90% of activities. Bain *et al.* (2015) found the correlated functional range of motion was 19° - 71°, 23° - 87°, and 10° - 64° at the metacarpophalangeal (MCP), proximal interphalangeal (PIP), and distal interphalangeal (DIP) joints respectively. Therefore, a device should be to apply a force of 9N-15N and articulate in the minimum range of 0° to 181°

2.3 Rehabilitation

From Wade and Hewer (1987), one of the leading causes for paralysis is stroke; 33.7 percent of people who suffer from some form of paralysis. Of those that suffered from a stroke, 90% of people will be affected by paralysis to some degree (van Kuijk *et al.* (2009)). Those with impacted hands, can be filtered into those with spastic or non spastic conditions (Warlow (1998)). The increase in spasticity is due to abnormal increase in the natural tension in a muscle that resists stretching, also referred to as tone. With the increase in tone, muscles become stiff and tighten up which makes their use impossible or painful i.e. spastic. The opposite of this is someone who has complete or partial paralysis of the hand and wrist where they cannot make a strong enough grip or articulate their fingers to make full use of their hand(s). The way tone is assessed is with the Modified Ashworth Scale which assesses the resistance during passive range of motion (Pandyan *et al.* (1999)). This is a six point scale where lower scores reflect normal tone while higher represent greater spasticity.

To help affected people, rehabilitation is used to relearn dexterous motions and strengthen ones grip. This can be done with a variety of exercises with a focus on different grip patterns with different objects and speed/thoroughness of task completion. A main influence factor on a patients recovery is repetition (Zhang *et al.* (2017); Kwakkel *et al.* (2004)). As with any skilled activity, the more one practices the more fluid/natural one becomes. Without repetition, there can be little to no increase in synaptic strength when learning a new activity (Kwakkel *et al.* (2004)). Since motor skills are usually affected by stroke, teaching the body how to do certain tasks is necessary for recovery. According to de Sousa *et al.* (2018), data shows that repetition as an intervention was able to influence motor recovery. Thus, relearning through repetition can be a key element in strengthening synaptic connections in stroke patient recovery (Chen *et al.* (2017)). From this, it becomes apparent that a device that runs highly repetitive exercises can be beneficial to a patient.

2.4 Soft Robotic Gloves

Current soft robotic gloves primarily use either pressure driven or motor driven actuation. Pressure driven gloves either use hydraulic or pneumatic systems to drive radial expansion in their actuators. Motor driven soft gloves will use either a cable bowden system or linear actuators that manipulate artificial tendons (Yurkewich *et al.* (2019)). While mechanically driven systems offer precise control as well as flexion and extension articulation, the larger support systems required for their operation make them undesirable. This makes the devices not suitable for an at home device for independent use. Also, these actuators exhibit constant curvature which is not a desirable motion since this is not how the human finger bends.

Pressure driven gloves on the market such as the ESO Glove Pro by Rosco Technologies also employ constant curvature bending actuators. Besides this, current commercial products are also limited in their controls. Since these devices do not have sensors to provide them with proprioceptive abilities, the control method is limited to open loop controls(Zhou *et al.* (2019); Polygerinos *et al.* (2013); Heung *et al.* (2019) ;Cappello *et al.* (2018)). Proprioception, also known as kinesthesia, is the internal ability to sense body movement and position. This means that the devices must directly write a pressure value to a achieve a desired configuration; which was determined with external measurement techniques such as motion capture(Heung *et al.* (2019); Polygerinos *et al.* (2015c)). Since these rely on external measurement methods, they can not act as stand alone devices for mobile use. As these devices cannot use internal sensor feedback to control their position, there is a lack dynamic control meaning they cannot be programmed with custom exercises easily. Also, by lacking the ability to track real time position current devices not track patient performance. So, if these devices were used at home there is no way to verify that a patient is completing their exercises nor if they are progressing over time.

Recently, there has been further progress with soft robotic gloves however they have yet to be commercialized. These devices have included resistive soft sensors to read strain. By reading strain they can relate these values to angles, however motion capture is still used to verify the bending angles and the strain sensor output is not used in the controller (Polygerinos *et al.* (2015b); Correia *et al.* (2020)). More devices have also incorporated segmented bending into their actuator design to provide more natural bending. To overcome the need of an operator, Zhou *et al.* (2019) employed electromyography (EMG) sensors to gauge user intent specifically for tests such as the Box and Blocks test for the opening and closing of the glove. However, results have been poor due to the sensor reading threshold not being met due to the weakened electrical currents generated in the muscle of post stroke patients (Delph *et al.* (2013); Thielbar *et al.* (2016); Zhou *et al.* (2019); Polygerinos *et al.* (2015a)).

Chapter 3

HARDWARE AND DESIGN

3.1 Glove design

Given the patient population intended for this device, there were unique requirements for the design of the glove. Primarily, the ability to easily don and doff the glove for the user. Since the patient has lost the dexterity of their hand, simple tasks like putting a glove on can be very difficult due to the shape of gloves. To overcome this, finger gloves were used which only require the tip of the finger to be inserted into the end of the glove, leaving the rest of the finger and palm exposed. This is secured with a Velcro strap around the wrist, allowing the user to easily don/doff the glove. Since the users strength will also be affected, the glove needs to be light to limit muscle fatigue. Therefore, no electronics are mounted on the glove besides the soft sensors via sown on sheaths. To attach the actuators, compliant straps are used to mount each actuator to the glove. This allows the propagation of actuator motion to the hand while not limiting the work space of the actuator. The bend sensors are attached to the glove via flexible sheaths sown onto the glove. The straps are made with compliant material so bending strain is transferred to the sheath rather than sensors. This is done due to the possibility of the sensor providing false readings under strain. In total the components mounted onto the glove are three soft sensors and three actuators with a total weight of 64g (fig.3.1).

Figure 3.1: Donned Soft Robotic Glove

3.2 Electronics

This device used an existing platform designed for the control of soft actuators. To run the control software, two raspberry pi's are used to run a low level controller and a high level controller. Both raspberry pi's can communicate via ethernet cables connected to a network switch. The raspberry pi running the low level controller is connected to an I2C multiplexer (Adafruit TCA9548A), where each signal is sent to three digital to analog converters (DAC) (Adafruit MCP4725) to convert the digital signal before being sent to the three pressure regulators (SMC ITV1000) (fig. 3.2). The pressure regulators are powered with a 24V power supply which connects to the platform by an adjustable voltage regulator. It is of note that SMC ITV1000 regulators has a built in low level controller which maintains a constant pressure depending on the input. The second raspberry pi is connected to an Arduino nano slave via USB serial communication which is connected to the three soft sensors. The air supplied in the current platform originates from an air line in the testing lab, not a separate pump.

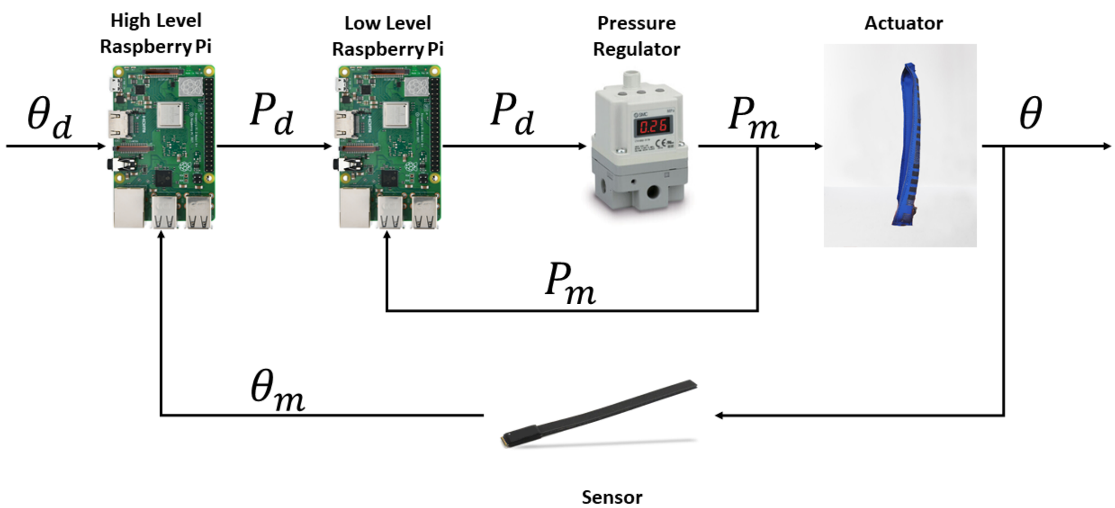


Figure 3.2: Signal Generation with Physical Controller Components

In future work, this platform will become mobile by condensing the electronics and including an air pump. The condensing of the platform includes using a single raspberry pi to initialize the low level controller while also running the high level controller, and mounting the electronics onto two 15cm x 30cm Lexan sheets. The top level will be the mounting plate for the regulators and pneumatic tubing, while the lower level has the electronics mounted to the bottom. The air pump (California Air Tools (1P) 90945 Quiet, Oil-Free .6 hp Motor/Pump) provides flow rate of 40

L/min at 40 psi or 34 L/min at 90 psi. This pump was chosen primarily due to the audible output of 56 decibels which is about the volume of a normal conversation (60 decibels) (Daniel (2007)). Tests will need to be done to determine if the flow rate is sufficient for 3 – 5 actuators.

3.3 Sensors

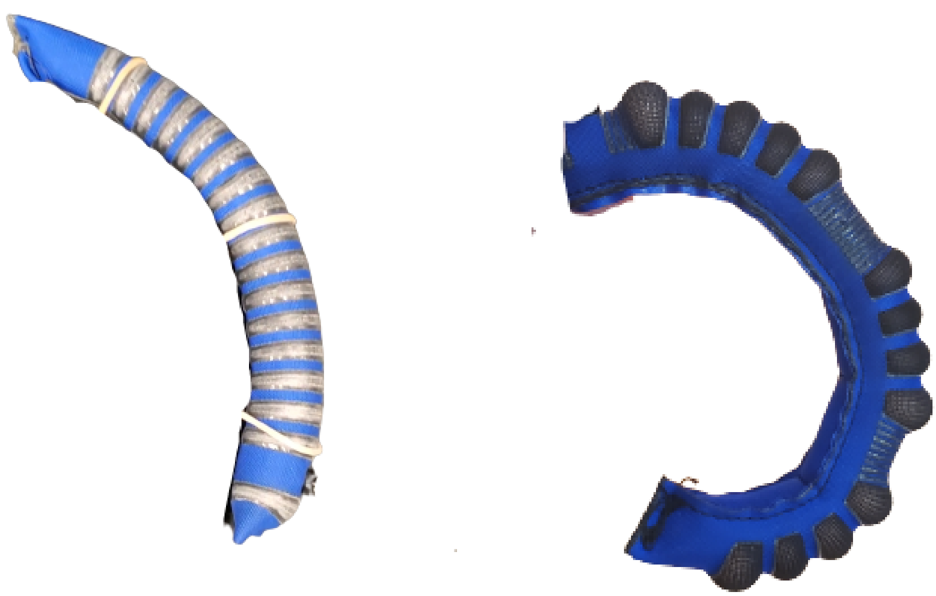
The flex sensors used are one-axis capacitive bend sensors from Bend Labs (fig. 3.3). These sensors measure angular displacement via a differential capacitance measurement, meaning common mode signals such as temperature fluctuations, strain, and noise are rejected. This provides a high fidelity measurement of angular displacement at very low voltage, which lends itself to a highly stable signal which does not drift over time. These sensors are also advertised as being path independent, meaning that the angle reading is between the two terminal ends only with the shape of the sensor between these points not affecting the reading. Each sensor is connected to an Arduino nano slave over I2C with a 3.3v input from a raspberry pi. A proprietary software library published by Bend Labs is used to read the angle values of the sensor.

3.4 Actuators

The actuators used in this device are fabric based pneumatically driven soft actuators. They are based on the design put forth in “Design and Computational Modeling of Fabric Soft Pneumatic Actuators for Wearable Assistive Devices” by Nguyen *et al.* (2019). For initial prototypes this design was used, however new actuators were developed for the final device for reasons. The design of the actuators by Nguyen *et al.*

Figure 3.3: Bend Labs 1-Axis Bend Labs Flex Sensor

(2019) starts with the layering of fabrics which will act as the actuator body. The body was comprised of 3 compliant layers: an inextensible layer (200 denier nylon pack cloth with one side thermoplastic polyurethane coating, Seattle Fabrics, Seattle, WA) cut into a specific pattern to provide the desired movement, an extensible layer of bi-directional high-stretch knitted fabric (24350, Darlington Fabrics, Westerly, RI), and a layer of thermoplastic polyurethane (TPU) to join the extensible and inextensible fabrics. Inside the fabric shell is a TPU bladder which applies a force onto the shell when filled with air. This bladder starts as a TPU sheet joined to the laminated fabrics on the extensible fabric side via an air intake port secured with a sealing washer and nut. Once secured the TPU sheet is folded into the desired configuration and each edge is heat sealed to form a bladder. The theory behind these actuators follows the exploitation of the restricted radial expansion of the bladder by restricting expansion to only occur past a certain point in the desired cut away areas of the inextensible layer. While these actuators offer multiple different types of movement depending on the patterned cuts of the inextensible layer, bending was the primary movement used in the device. From Nguyen and Zhang (2020), there can be two different types of bending: constant curvature and segmented. Constant curvature bending occurs when the pattern of the inextensible layer is cut with a single repeated shape to create a single bending curve. Segmented bending takes advantage of the same design, but instead of having a single repeated pattern there are areas of restriction creating a zone of straight extension between bending curves (fig. 3.4).



(a) Previous Design **(b)** Current Design

Figure 3.4: Constant Curvature vs Segmented Bending

Both bending types require an inextensible spine to restrict the direction of extension into a curve (fig. 3.5), as without this there would only be extension rather than bending. To create this spine, the laminated fabrics are folded in half with the extensible material on the outside surface and sown (40wt 100% polyester thread) together along the edge opposite of the fold. The shell is then inverted so the extensible material is now on the inside of the actuator, creating a seam. After the ends are sown shut, the actuator is usable for bending. The manufacturing process from Nguyen and Zhang (2020) can be see in Appendix A.

Originally these actuators were used for a gripper in the form of a human hand. Nguyen *et al.* (2019) found that these actuators could provide a max force delivery at the distal end of 7.75N at 30psi (Nguyen and Zhang (2020)). While this may work for a pure soft actuator gripper, when this style of actuator was attached to a hand problems arose. Since a human hand has natural internal resistance, these actuators could only slightly articulate the MCP when attached to a hand. It was initially observed that the holistic bending angle decreased from 180° to 45° meaning there was a significant decrease in force delivery at the distal tip of the actuator when applied to a human finger (Nguyen and Zhang (2020)). Another drawback of this design is the manufacturing method. By relying on a single seam to control the direction of bending there was a high variation of bending capability determined by the straightness of the sown seam and the distance of the seam to the parallel edge folded actuator shell. Finally, while these actuators can pressurize up to 30psi, expansion was great to be safely used when mounted to a glove due to a risk of rupturing as well as the expanded size being too bulky to comfortably use. These actuators primarily ruptured due to the inextensible seam/spine being ripped apart by the force acting along the shell, however the inextensible layer would also tear where the restrictive bands over the extension area would meet the seam causing a blowout.

The final actuators used in the device followed the same principles and manufacturing method as the design put forth by Nguyen *et al.* (2019) up to the assembly of the inextensible spine. Due to the drawbacks of the sow and invert method, I decided to use an inextensible fabric spine being of the same material as the inextensible fabric layer (fig. 3.5). Rather than folding, sowing, and inverting the actuator shell, two parallel edges of the laminated shell (inextensible layer facing outward) are sown along their complementary edges of the spine. This makes the manufacturing process quicker as it takes 9.5 minutes to assemble verses 20 minutes for the precious design. A 10 minutes difference is due to the inversion step of the previous actuators. The resulting ends are then sown together creating an actuator of a smaller width (16.5mm compared to 25mm) which does does not expand to such a degree as previous designs, and provides a flatter surface along the finger when inflated. The inextensible material was also changed to 400 denier nylon pack cloth with one side TPU coating from Seattle Fabrics to prevent tearing of the restrictive layer during actuation. The weight of these actuators were found to be 6.4g given the size of 17mm x 16.5mm x 150mm.

Figure 3.5: Current Actuator Design

Chapter 4

SOFTWARE AND CONTROL

4.1 Software

4.1.1 Data Collection

Two scripts are used to facilitate the sending of sensor readings to the high level controller:an Arduino script reads the sensor output while a python script polls the Arduino to receive the sensor readings. Since the bend sensors communicate via I2C, multiple sensors can be connected to the SDA/SCL pins of the Arduino without occupying multiple input/output pins as long as their addresses are unique. Therefore, I had to change the I2C addresses of each sensor. This was completed by using the proprietary library from bend labs which changes the sensor I2C address of a sensor to a desired open address. The Arduino script starts with initializing each sensor before the main loop waits for a message from the high level controller to send the sensor readings over serial communication. Initially I2C communication from the raspberry pi directly to the sensor was attempted however, due to the drivers created for activation and reading of the sensors, it was not possible to obtain sensor readings using this method. This is due to the sensor drivers requiring an Arduino master with the sensor as a slave, rather than a raspberry pi master and the slave sensor. After numerous attempts it was not possible to adapt the drivers to work in a pythonic environment. To send a reading, each sensor is initialized and sends the output as a grouping of all three sensors three times. This is done because it was found that these sensors only converge to a reliable sensor reading after being initialized and pinged for a reading after three sensor activations. If the controller only used the initial reading, then the reading would be far below the actual angle measurement.

In the python script, to obtain sensor readings, a message is passed to the Arduino which activates the sending of data. In development, it became apparent that on the first run of the system after powering on that when a message was sent to the Arduino no data would be sent as a response. After confirming that the Arduino was receiving a message and the sensors were working, it was assumed that this was due to a timing error between the raspberry pi and Arduino even if the baud rate of 115200 was used across all platforms. Since this would break the system, a check was implemented to make sure a proper message was being sent by the Arduino before initializing processing of the readings. As serial communication passes the readings as a string, this message has to be turned into a list of signed floats, with only the last three of nine values being saved for use in the high level controller.

While serial communication works with the current set up, as more sensors are added it will be necessary to integrate multithreading into the data collection. This would decrease response time while potentially eliminating the need to send data in chunks and check to make sure a usable message is being sent from the Arduino Cheng *et al.* (2015).

4.2 Control

4.2.1 Low Level Controller

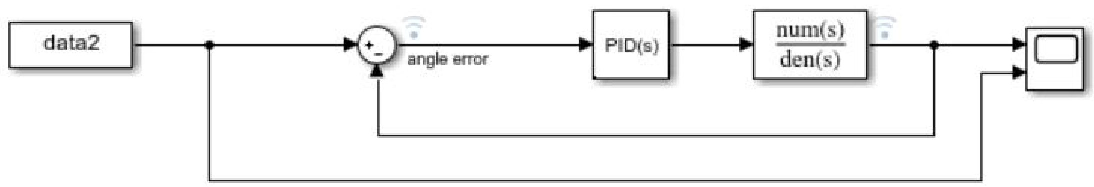
A low level controller primarily exists to initiate the ability to write pressure values to the regulators and then calibrate the regulators before use. Pressure values can be written and read by using the pickle, zlib, and zmq libraries. Pickle was used to serialize/deserialize i.e. converting a pythonic object such as pressure into a byte stream for use/storage before being compressed by zlib. Using zmq, this byte stream can then pass to the regulators easily as zmq allowed for the creation network sockets for low-level message passing. This works for reading and writing pressure values, however when a value is received, the message is uncompressed then unpickled. The calibration function takes a voltage a voltage reading from the ADCs and calculates an offset by multiplying the readings by 31.13 then subtracting 29.33. This makes sure that there is no bias in the signal being sent, preventing a disparity between the desired and actual regulator pressure.

4.2.2 High Level Controller

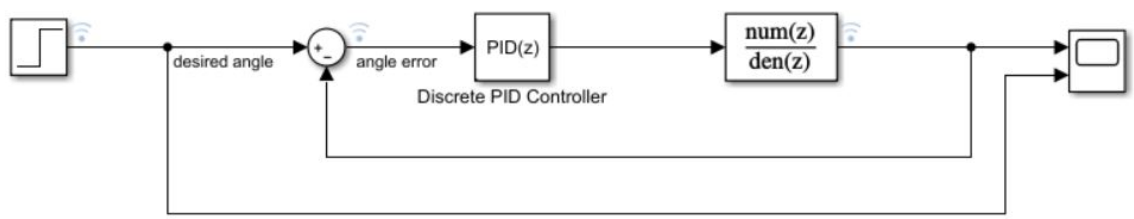
A high level controller was created primarily to facilitate the tests for the device. This controller includes two different control methods: open loop and closed loop. The open loop controller was designed for the reason of testing the actuators physical abilities, ensuring that the actuators can safely be used in a common rehabilitation exercise. A closed loop controller was designed to test if the addition of proprioceptive abilities could lend itself to a robust dynamic control scheme. This would also allow a physician to dynamically control a patient's figures, or create custom exercises if desired. When the controller is initialized, the user is prompted to decide if they want to use the open or closed loop method. If the user decides to use open loop, then the option to close or open the hand is presented. If closed loop is chosen then the user is asked to input the desired bending angle values of each finger for the exercise. The method to activate closing and opening of the hand is the same as the open loop method.

The open loop controller follows Polygerinos *et al.* (2015b) where a pressure is directly written to the actuators to either open or close the hand to a desired grip pattern. From the force test, a pinch force of 14N was used to provide an adequate grip on the blocks. This translated to a pressure of 15psi, 25psi, and 25psi for the thumb, pointer finger, and middle finger respectively. When the controller received the command to close the hand, these pressure values would be written to the regulators. When the command to open the hand is received, a zero array is sent to the regulator.

The process model/plant was first modeled in MATLAB/Simulink (fig. 4.1). Taking recorded pressure data from the regulators and related angle reading from the bend sensors, a continuous time transfer function was created relating an input pressure to an angle output. To obtain bending data an actuator was inflated to 33psi with the sensor attached to the spine of actuator. The angle reading was read from the Arduino serial data into an excel sheet while the pressure regulator data was recorded in a separate csv file. This was done three times to provide multiple datasets to model this relationship. Since the sample time of the angle data was greater than the regulator, the angle data was down sampled and refit to match the number of pressure data datapoints. As the angle data was offset, a value of 36.55 was subtracted from



(a) Continuous Time Block Diagram



(b) Discrete Time Block Diagram

Figure 4.1: Simulink Block Diagrams

each angle value. This was done so a pressure value of 0 did not correspond to an angle reading of 36.55, skewing the controller results. This offset came from the initial position of the actuator generated by the difference in angle between the distal and proximal ends of the actuator. Once the data was processed, the MATLAB sysid toolbox was used to generate three transfer functions. The transfer function generated from the first trial was used as at provided the best fit to the estimation data. Since there was a slight oscillation of at steady state pressure (fig. 4.2), a 2nd order system was used to avoid overfitting the estimation to the data:

$$\frac{1.257s + 0.08562}{s^2 + 0.3509s + 0.0328} \tag{4.1}$$

.

While this led to a lower fit of 79% and higher MSE of 52.66 when compared to higher order systems, it would create lower chance of error when at the min/max of the system. Also, it would be simpler to tune a lower order system by hand than a higher ordered system.

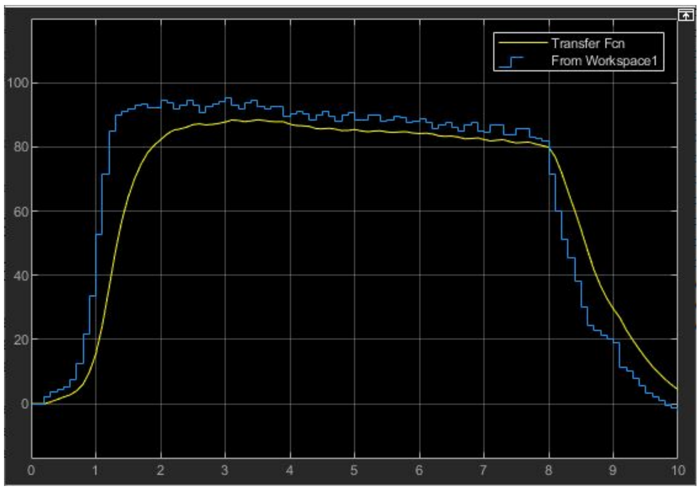


Figure 4.2: Continuous Time System Plant Output

A PI controller was chosen to shape the signal closer to the provided data add as it is intuitive to tune for a preliminary control method. A proportional gain acted to increase the output signal from controller and an integral gain was used to correct for steady state error. A derivative term was not used in this controller as the signal has no oscillation. After tuning the signal, proportional gain of 2.25 was used with an integral gain of 0.25. This created a system closely mirroring the system while avoiding over fitting (fig. 4.2). The poles of this system are:

$$-0.1754 + 0.0449i$$

$$-0.1754 - 0.0449i$$

meaning that the system is stable as they both reside on the left hand of the complex plane.

This continuous system was then turned into a discrete system using the MATLAB c2d command:

$$\frac{0.1239z - 0.1231}{z^2 - 1.965z + 0.9655} \tag{4.2}$$

. After tuning the PI controller, a proportional gain of 5.2 and integral gain of 8 was used to provide the response (fig. 4.3). The poles of this system are:

$$0.9826 + 0.0044i$$

$$0.9826 - 0.0044i$$

which means the system is stable as it is within the unit circle.

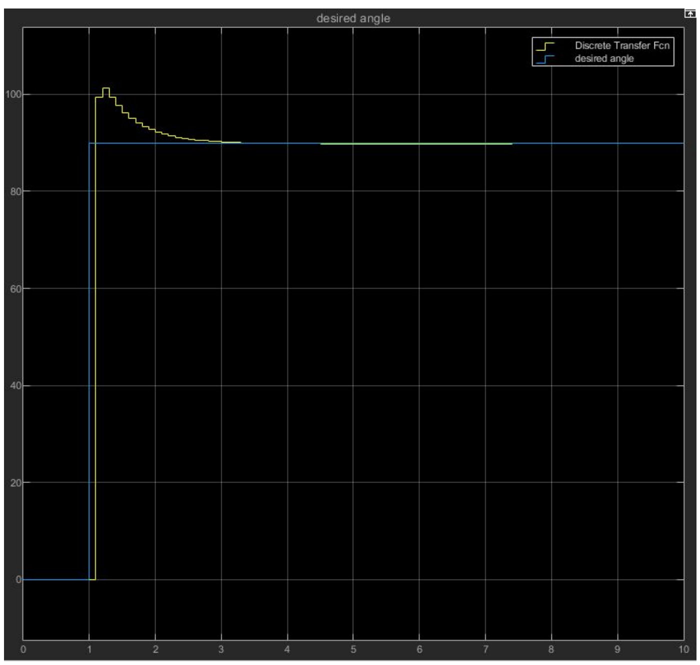


Figure 4.3: Discrete Time System Plant Output

After creating the discrete time model in Simulink, a controller can be developed in python. Two classes were created to control the behavior of the actuators: PI and Actuate. The class PI describes the plant created in MATLAB with the main function being control. With an angle reading and the desired bending angle as input an error signal is generated. This error signal is then passed through the PI controller, returning a generated signal. Actuate controls the open and closed loop bending with the functions "open_loop_bending" and "closed_loop_bending" respectively. It also initializes the zmq socket for publishing pressure data to the regulators.

The open loop controller function writes the pressures as mentioned previously. The closed loop bending collects the sensor readings to send to the control function, summing the output values over each cycle. This was done to account for inflation and deflation signals generated by the control function. This signal is then sent to a clamp to limit the minimum and maximum pressure. The lower bound of 0 psi and upper bound of 30 psi were used, which is explained in the Discussion section. The resulting angle is compared to the desired angle with an tolerance of +/- 1 degree giving a range of 3 degrees, which is explained in Sensor Classification. Since exact angle bending is not necessarily as important achieving a sufficient pinch, a tolerance makes the system more robust while not decreasing functionality. If the angle condition is met, then the pressure value is frozen and a flag is raised signaling that bending is complete. This is done independently for each actuator, only allowing for the release of the grip once each has met its angle condition. After tuning the system, a proportional gain of 0.2 and an integral gain of 1.2 were used.

Chapter 5

TESTING AND RESULTS

5.1 Sensor Classification

5.1.1 Background

To ensure the bend sensors were accurate in their readings, a motion capture study was done to validate the angle reading of the sensor. Since this device as the goal of not being used in the presence of a medical professional, it is imperative that the device is precise. This is due to the physician needing to be able to trust that the device is articulating the patients hand into the desired shape for each exercise so no unintended actions occur. While a window of 3 to 5 degrees does not change the overall behavior of the device, vastly inaccurate readings would cause the controller to not function properly by skewing the generated error.

5.1.2 Method

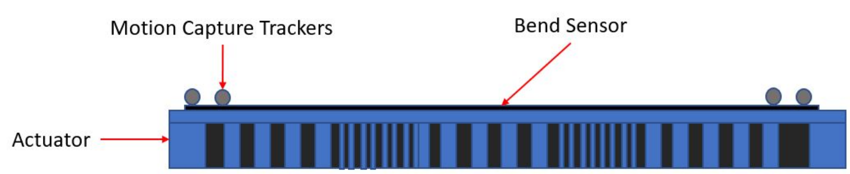


Figure 5.1: Sensor Classification Test Setup

In this study, a bend sensor was attached to the bottom of an actuator via double sided tape. Tape was used as this would not transfer any force due to radial expansion onto the sensor. Four motion capture trackers were attached in pairs of two to each terminal end of the actuator. Each pair was attached to a 3d printed square rod so the individual markers would not shift their position. These rods were then affixed with double sided tape to the spine of the actuator with the markets being directed to one side of the actuator (fig. 5.1). The actuator was placed on the motion capture table with the actuator affixed to the table from the base of the actuator. The actuator was inflated to a sensor reading of 80 degrees and held there at that pressure value for 4 seconds over three trials. To calculate the angle of bending from motion capture, the data from each pair of trackers were used to create vectors whose dot product was used to calculate the angle between them:

$$\frac{u \cdot v}{||u||||v||} \tag{5.1}$$

. Where u and v are the generated vectors.

5.1.3 Results

After plotting the sensor and motion capture results it became evident that there is a slight offset at steady state readings between the sensor and motion capture results (fig. 5.2). Since this offset was no more than 5 degrees it was determined to be within an acceptable range which was be accounted for with a tolerance in the controller. Also, as the bending angle changed over the 4 seconds of constant pressure it is possible to see some hysteresis behavior as the actuator settles into a final position.

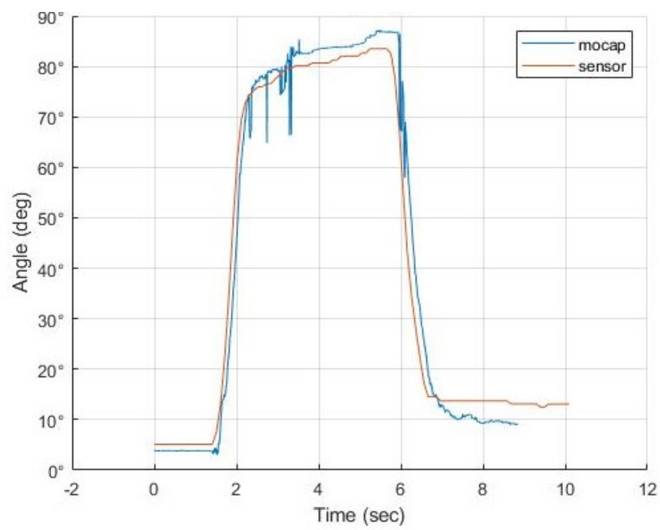


Figure 5.2: Motion Capture and Sensor Reading Classification Comparison

5.2 Force Testing

5.2.1 Background

As mentioned prior, the grip force needed to pick up most every day objects is 9N-15N (Matheus and Dollar (2010)). Therefore, it was important to find what the force output of these actuators when attached and not attached to a finger. This distinction due to a drop off in force translation seen in other soft actuators when attached to a user (Polygerinos *et al.* (2015c); Brokaw *et al.* (2011); Polygerinos *et al.* (2013)). By learning the force applied by the finger, it can be determined if the force delivery of these actuators is sufficient to provide the 9N-14N force.

5.2.2 Method

The design for this experiment focused on reading the force applied by the tip of the actuator. Using a set up similar to (Polygerinos *et al.* (2015c)), the actuator

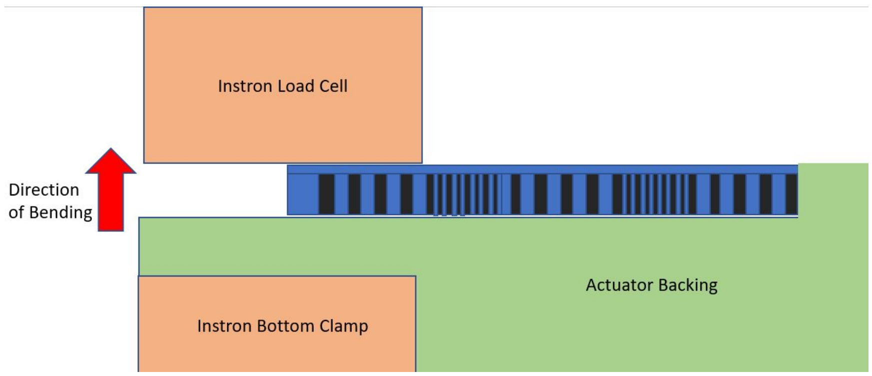


Figure 5.3: Force Test Setup

was mounted to an Instron 5944 so the inside tip of the actuator rested at the load cell (fig). This was done by securing the actuator, from the base, upside down on supporting material attached to the lower mount of the Instron machine (fig. 5.3). This supporting material followed the length of the actuator to drive the force transfer to the load cell. The applied force was recorded starting from 0psi up to 35psi, every 5psi, besides a recording at 33psi due to this being the expected max operating pressure observed during development. This setup and the steps were then repeated for when the actuator is attached to the hand via the designed glove.

5.2.3 Results

The force delivery of the actuator alone were 0N, 0.68N, 1.6N, 3N, 5.44N, 8.6N, 12.52N, and 16N for 0psi, 5psi, 10psi, 15psi, 20psi, 25psi, 30psi, 33psi, and 35psi respectively. When attached to a hand the forces were 0N, 0.8N, 1.8N, 2.8N 4.11N 5.52N, 7.5N, 8.68N, and 9.2N over the same order of pressure values previously. The pressure values are similar in both applications from 0psi to 15psi due to a pressure range of stiffening before bending occurs (fig. 5.4).

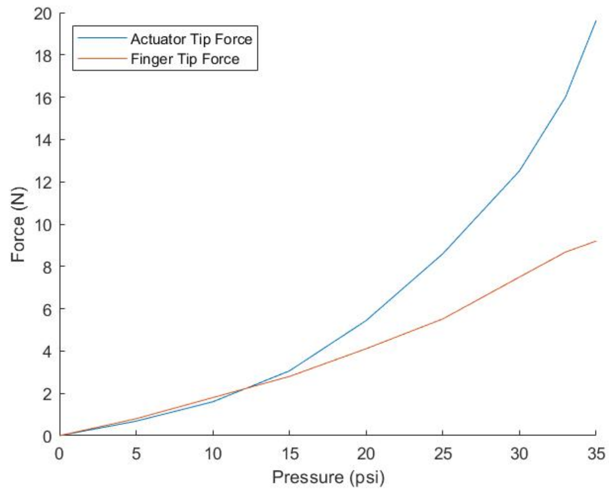


Figure 5.4: Actuator Tip Force Generation

5.3 Fatigue Testing

5.3.1 Background

The aim of this device is to act as a rehabilitation device when outside a medical setting. This means that the actuators need to be safe and reliable at operating pressure. Therefore, it was necessary to perform a fatigue test on these actuators to determine their max safe operating pressure and how long they can reliably be used. Since these actuators are fabric based, they are cheap to produce and therefore sell. So, their lifespan is not a primary concern. Ideally, these actuators would be able to safely function over five, one hour therapy sessions before being replaced. The term bubbling is used to describe the degree of expansion where high bubbling is undesirable.

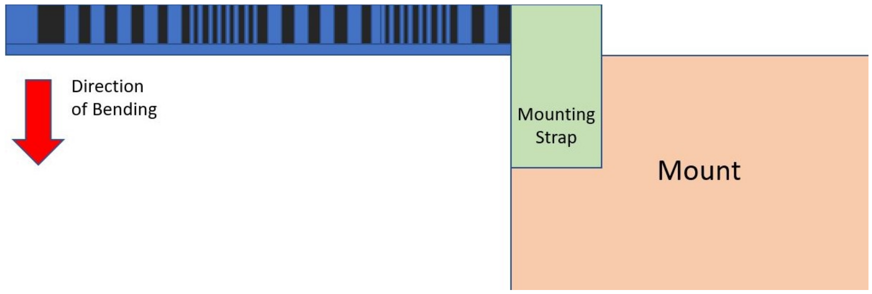


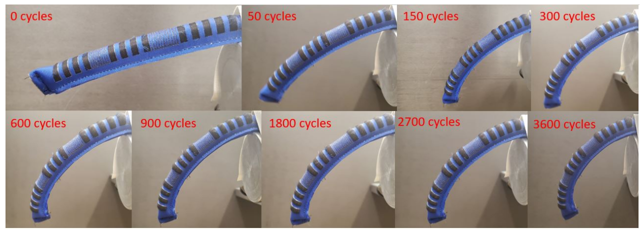
Figure 5.5: Fatigue Test Setup

5.3.2 Method

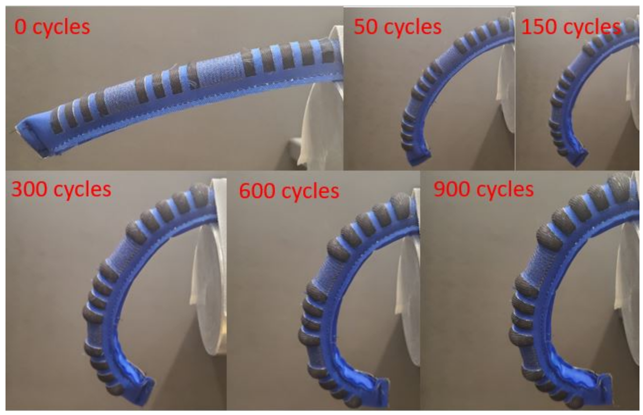
Three new actuators were used in this study for testing 25psi, 30psi, and 33psi. These values were chosen based on the results of the force testing previously. 25 and 30 psi were chosen based on their force output while 33 psi acted as a max operating pressure. Each test was set up to run for 1 hour or until failure, comprising of a 4 second cycle alternating every two seconds between a high pressure and 0 pressure. The actuator was affixed to a mount, allowing for the free bending of the actuator every cycle (fig. 5.5). Photos were taken at regular points every 0, 50, 150, 300, 600, and 900 cycle except in the case of failure or if the test went past one hour (900 cycles). Failure was determined by the visible separation of the extensible layer and inextensible layer from one another, by the rupture of a bladder, or the tearing of seams/shell material.

5.3.3 Results

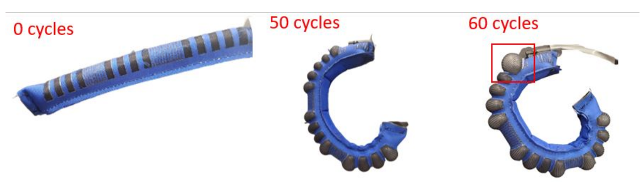
When testing at 25psi, the actuator continued use with only slight of degradation of the strength of the elastic woven material layer after 4 hours of use. There was slight bubbling which increased very slightly until 200 cycles, where this stopped increasing.



(a) 25psi



(b) 30psi



(c) 33psi

Figure 5.6: Actuator State During Fatigue Tests at Various Pressures with Highlighted Failure Point

This slight increase in bubble led to a slow progression of the bending angle while it occurred (fig. 5.6). After four hours there was no separation of the material in the areas of expansion. Insert image For 30psi, the bubbling behavior was the same as 25 psi. however, the expansion stopped increasing at 300 cycles. Expansion was seen to be uniform along the actuator, except at the region of expansion at the base closest to the air input/MCP joint. After the full hour there was only the degradation of the elastic response of the extensible layer. At 33psi, failure was seen after 60 cycles. This failure was located at the MCP bending region directly in front of the air intake port. At 33psi there was not an increase in bending angle as seen at 25psi and 33psi until the there was a failure due to the separation of the extensible and inextensible layers. This caused the restrictive layer to fold back on itself and greatly expand bubbling, creating a risk of the bladder rupturing (fig. 5.6c).

5.4 Box and Blocks

5.4.1 Background

The Box and Blocks test is a standard test which measures unilateral gross manual dexterity of a test subjects hand (Kontson *et al.* (2017)). Due to the simplicity of the test and its reliability, it has become a widespread method to obtain these objective measurements from a large patient population including stroke. Kontson *et al.* (2017) found that a healthy adult male aged 20 to 80, the average score for the right hand is 77 blocks (SD ±11.6) and 75 blocks (SD ±11.4) for the left cite. It was also found that a healthy adult female aged 20 to 80 has an average score of 78 blocks (SD ±10.4) with the right hand and 76 blocks (SD ±9.5) with the left hand. When looking at our target age group, 60+ years old, these values for healthy men and women decrease respectively to 61 to 70 blocks and 63 to 76 blocks cite. Since the affect of stroke can vary wildly, there is no averages for those suffering from its after-effects. This is a test to gauge dexterity and patient improvement over time. By choosing this test, it was possible to see how these actuators may function in a real world setting.

5.4.2 Method

This test uses the following standardized equipment: one 53.7cm x 25.4cm x 8.5cm 5 sided box, a partition with height 15.2cm, and 150 2.5cm3 wooden cubes. The partition is used to separate the box into two 25.4cm sections to house the cubes. All of the cubes should originate in one of the box sides, to be placed into the other side. Once the test is assembled the user should sit in front of the box with the blocks on the side of the hand being tested. Starting with their hands next to the sides of the box perpendicular to themselves, the user has 60 seconds to move as many blocks as possible from one box to the other (fig. 5.7). A block is counted as long as it is moved over the partition and is dropped. The block does not need to be placed in the opposing box to count nor does it need to land in the box, just cross over the partition. If multiple blocks are picked up and dropped then they will collectively count as a single block. A patient may have a 15 second warm up before initiating the test if desired.

The box and block test was run with two healthy adult males between 25 and 30 years old (fig. 5.7). To simulate a stroke patient with value of 0 on the Modified Ashworth Scale, each subject was instructed to keep their hand and wrist completely relaxed throughout the duration of the experiment. Each subject underwent 5 trials using their right hand with the number of blocks being recorded afterwards. Since this test used the open control method, an external operator opened and closed the hand as the subject reached for block and was in a state for a successful release pic maybe here of setup, or somewhere in this section.

(a) Box and Block Test Setup

(b) Box and Block Test in Progress

Figure 5.7: Box and Block Test

5.4.3 Results

Subject 1 scored 10, 16, 17, 19, and 20 blocks over the five trials and subject 2 scored 9, 15, 16, 18, 18 (table 5.1). The initial increase in performance may be attributed to the user and operator both becoming more familiar with the device; beginning to plateau after the 3rd trial. I close of of the pinch generated during this test can be seen in Appendix B.

5.5 Human Inclusion Study

5.5.1 Background

A motion capture test of the controller was done to determine the accuracy of bending with the glove not attached and attached to a hand. The goal of these tests

Table 5.1: Box and Block Test Results

	Subject 1	Subject 2
Trial 1	10	9
Trial 2	16	15
Trial 3	17	16
Trial 4	19	18
Trial 5	20	18

were to see if there would be a significant difference in performance with the addition of a human hand to the system. As human disturbance can greatly affect a system, it was important to identify if the addition of a hand may affect the behavior of the controller.

5.5.2 Method

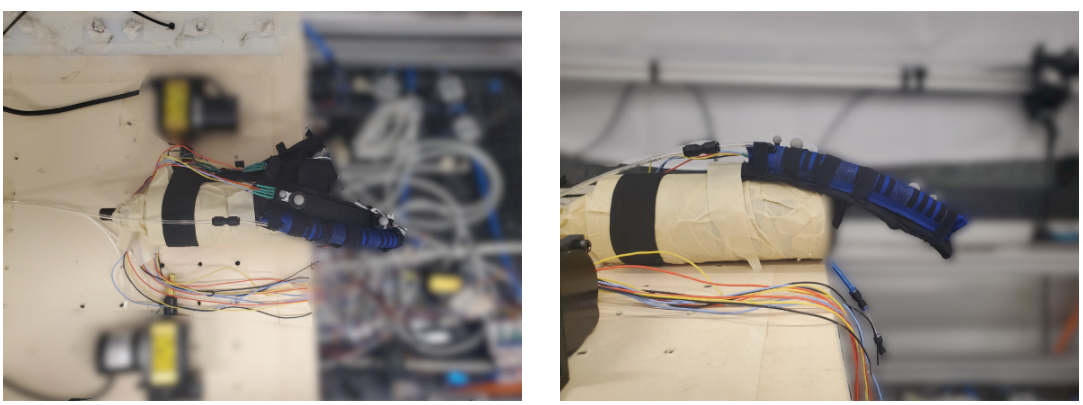
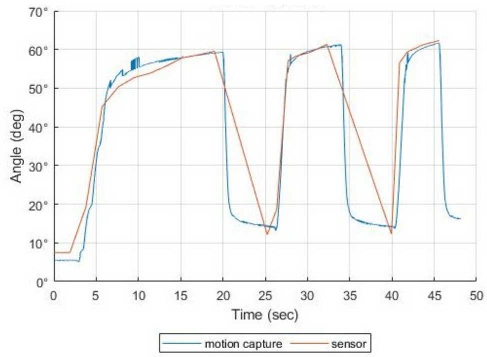


Figure 5.8: Motion Capture Test Setup Without Human Hand (Middle Finger)

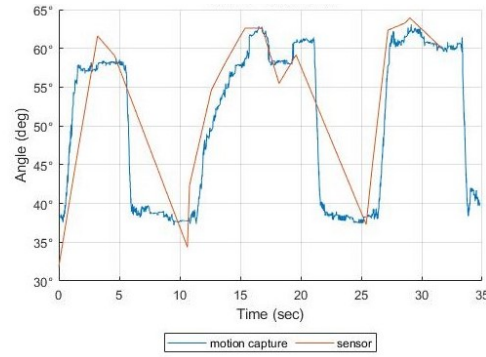
For this test the thumb, point finger, and middle finger were inflated three times to a goal of 60° +/- 1°, 75° +/- 1°, and 90° +/- 1° respectively. These values were chosen at random, but within the actuators operable range of 0° - 200° at 30psi. Each finger was inflated individually, with and without the glove being attached to a hand. Following the same method as the sensor classification test to obtain and calculate bending angle, motion capture markers were placed each terminal end of the bend sensor. This was done to ensure the motion capture results are as reflective of the sensor value as possible. For trials without a hand, the glove was attached to a cylindrical mount via the elastic strap (fig. 5.8). After the bending angles were calculated for each case, the values were plotted against the sensor readings. Each sensor reading was recorded at each loop of the controller, with the time of each reading being recorded as well.

5.5.3 Results

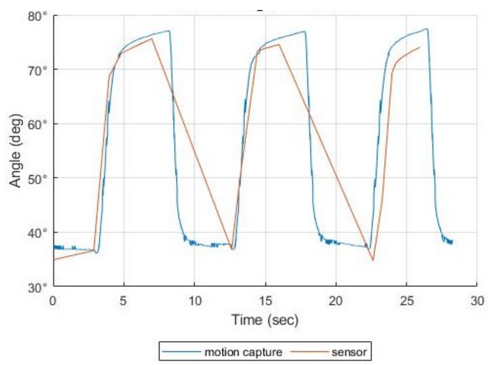
After plotting the results it became apparent that there is a significant difference in performance given the presence of a hand when compared to no hand present figures (fig. 5.9). When a hand is not present, the sensor readings are similar to the corresponding calculated motion capture angles figures. The plots show a smooth inflation to the desired angle without overshooting or significantly inaccurate sensor readings (fig. 5.9a,c). With the addition of a hand, the thumb continued to show accurate bending results figure while the middle and pointer finger differed from the previous tests. When the hand was incorporated into the pointer finger test, there was significant overshoot of 40° generated from initial pressurization figure (fig. 5.9d). A spike in sensor reading led an increase in the motion capture data meaning that the actuator is bending to this degree after the sensor signal is read. While this initial spike in sensor reading skews the error generation, it can be seen that the controller is functioning properly as this increase is corrected and the device converges to the desired angle. A similar behavior was seen when a hand was incorporated into the middle finger tests. During the initial inflation of each bending task, there is a spike in sensor readings not reflected in the motion capture data figure. Since the angle value was continuously increasing, these trials were stopped when the actuator was visually 90° with a keyboard interrupt writing the pressure to 0psi (fig. 5.9f). If the pressure was not released, the increasing angle value would keep the actuator inflated at max pressure rather than correct for overshoot.



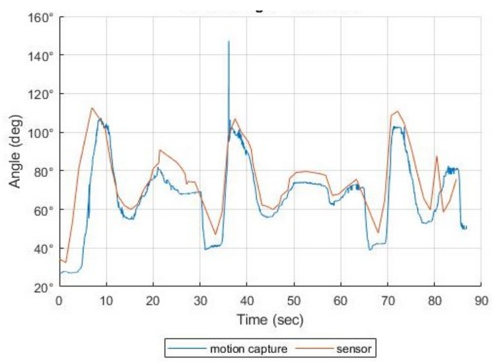
(a) Thumb - No Hand



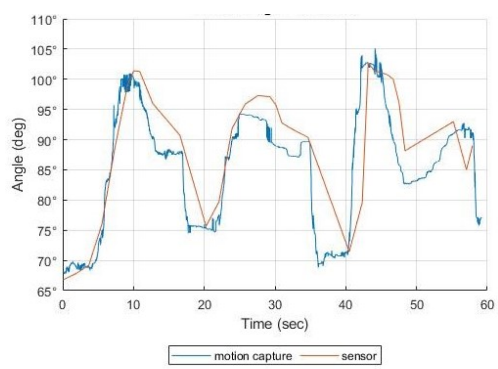
(b) Thumb - With Hand



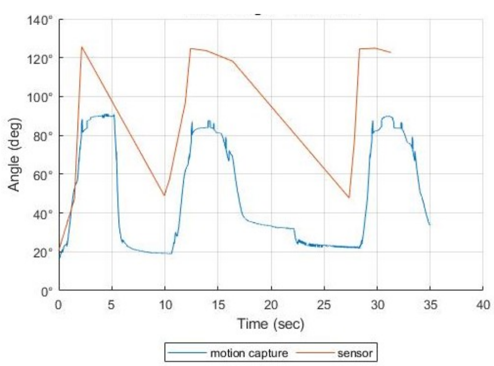
(c) Pointer Finger - No Hand



(d) Pointer Finger - With Hand



(e) Middle Finger - No Hand



(f) Middle Finger - With Hand

Figure 5.9: Comparison of Glove Performance With and Without Human Presence

Chapter 6

DISCUSSION AND CONCLUSION

6.1 Discussion

While the sensor classifications shows that this sensor does provide accurate angle readings, this study started to bring to problems with the sensors that became more prevalent as development continued. The first problem observed with these sensors is their lack of robustness to added strain. Originally these sensors were attached to the top of the actuators, however it was found that the added force from the bands attaching the sensors to the actuator would invalidate the sensor readings by seemingly returning random values. From Bend Labs, this should not occur as the sensors are advertised to not be resistant short bursts of 75% strain and extended use at 30% strain. Therefore the sensors were attached to the spine of the actuator with double sided tape so there would be no force transference to the sensor. Once this problem was overcome it was possible to see an continuous, accurate, drift free angle reading. With this said, the total precision of bending is not necessarily the most important as long as a desired grip pattern is created. While the device needs to be accurate in its readings, a tolerance in readings in allowable as long as it is within an acceptable range of the target angle.

As aforementioned, a grip force of 9N-14N is required to pick up most everyday objects. When using the initial actuator design by Nguyen and Zhang (2020), it quickly became apparent that while they may work as an end-effector they do not transfer enough force to a hand to provide the requisite grip strength. Other implementations of soft actuators similar to this also see a low force delivery of 3N when applied to a user's hand which is a 50% decrease from when not attached to a hand (Polygerinos *et al.* (2015c); Yurkewich *et al.* (2019); Zhou *et al.* (2019)). This is viewed as acceptable since when this force is multiplied by each finger, a total grasp force of 15N is generated and a 3 finger pinch force can be expected to be 9N. This decrease in force delivery can be due to the internal resistance of the hand which works against the motion of actuation as well as the deforming of the soft tissue on the palm of the finger which absorbs some of the force being transferred. After testing, it can be seen that the force generation of these actuators is much greater than previous designs by providing a max grip force of 40N at 30psi and a 2 finger pinch force of 15N at 30psi. Similar to other actuators, there was also a 50% loss in force when transferred to the hand however this is not a problem as the tip force output is high enough for performance to not be impeded. This means that a user of this device should be able to pick up most everyday objects with just a 2 finger pinch.

From the fatigue test, a maximum regular operating pressure was found to be 30psi rather than the expected 33psi. This was decided because the main driver of fatigue seemed be the supplied psi rather than the number of cycles. When the pressure of 33psi was used, the inextensible bands would be pushed back onto themselves peeling away from the extensible material. While the bladder did not rupture, it is no longer constrained by the shell of the actuator. Since this was not seen at 30psi, this became the maximum operating pressure implemented in the previously mentioned clamp function. Another behavior exhibited by these actuators is the steady increase in expansion/bending angle over 200-300 cycles. Initially I believed this to be due to degradation of the layers, however this behavior stopped after the aforementioned cycles. Therefore, since these were each fresh actuators I believe this to be a breaking in period where an actuator gradually stretches to its maximum bending angle over time for a given pressure. Since the designed controller does not map a single pressure to a bending angle, this does not pose a problem. This gradual change was not seen as extensively at 25psi, meaning that there is little effect on the materials when operating at this pressure and it may be able to operate at this level more much longer than four hours. The failure case of these actuators may lie in the lack of TPU in the areas of extension for a segmented bending actuator. However, when there was a continuous sheet of TPU in the laminated shell, the bending was highly restricted making the actuators perform more similarly to constant curvature. While this may shorten the life span of the actuators, it lends itself to better bending behavior. In summary, this actuator design has an operating lifetime of at least 3600 cycles at 25psi or at least 900 cycles at 30psi; lasting for their desired lifespan.

The Box and Blocks test has shown that these actuators can provide sufficient force to articulate a human hand into a 3 finger pinch and reliably hold the blocks. This shows that at a base level with a similar control scheme to commercial products, this device can be used for simple rehabilitation exercises. There was no failure of the actuators however, it can be difficult to align the hand into a proper pinch without wrist support. No blocks slipped out of the pinch but it was difficult to manipulate the blocks if there was not sufficient space for the device to fit around the block. Both subjects scores increased over the duration of testing, however this could be due to operator timing variance in opening and closing the hand. In comparison to Polygerinos *et al.* (2015b), where their glove was able to achieve 10 blocks, the glove presented offers better given similar control methods. However, the actuators in this test were silicone based hydraulic actuators. To remove the variable of the operators response time, user intent estimation will need to be incorporated so the device can open and close automatically. While there may be improvements to make, this test has shown that these actuators can be used in a physical/occupational therapy exercise.

By performing a motion capture study on the effect of human inclusion on the controller, it was possible to see that that controller is working as designed. While some angle readings may be inaccurate, the actuator did move to what it perceived to be the intended point multiple times fig. In addition, when the actuator overshot its intended angle, the controller corrected for this motion and was able to converge to the desired angle fig. It is also apparent that overshooting and false angle readings primarily occur with the presence of a hand. While this does mean that there is a significant difference in performance with the glove attached to hand, I believe this to be a sensor based problem. These sensors claim to be path independent as long as the angle between the sensing terminal and middle section is above 2x the width of the sensor. However, when a hand is inserted into the glove the sensors shift slightly causing there to be a slight bend to the pointer and middle finger sensor. This shift occurs due to the top layer of the glove slightly contouring to the curve of the finger, generating a small contortion of the sensor for the pointer and middle fingers but not the thumb. When the pointer and/or middle finger actuator inflates and the glove bends, the sensor occasionally sends an inaccurate value which skews the error signal causing the actuator to overshoot. It was surprised to see this behavior given this mounting scheme was determined after being consulted by Bend Labs with regard to slight strain affecting sensor reading. The current placement of the sensors was determined after being consulted by Bend Labs after slight strain was found to negatively affect sensor readings Therefore, it can be concluded that while the controller functions properly and the addition of proprioception is viable, there is still work to be done in the development of robust soft sensors for the application to soft actuator driven devices.

6.2 Conclusion

With the goal of creating a device using fabric based soft actuators that can allow an individual undergo rehab exercises, this has been largely successful. These actuators provide sufficient force and range of motion to manipulate a hand into various configurations. Due to the magnitude of force generation, a user has the ability to pick up most everyday objects with the force generated by a 2 finger pinch (14N). The device can also be continuously running for at least four hours at 25psi, the pressure used for the Box and Block test, showing that the actuators can function reliably over extended use. Therefore, the research question pertaining to if this device can be used effectively in rehabilitation can be confirmed given the results of these two tests and the performance of the device in the Box and Blocks exercise. However, intent estimation must be added so a user does not need an external operator such as an assistant or therapist to control the opening and closing of the glove.

Rather than relying on an external measuring system like motion capture: Zhou *et al.* (2019), Polygerinos *et al.* (2015c), Heung *et al.* (2019), and Cappello *et al.* (2018) this controller directly reads the angle difference between the distal end of the DIP and the proximal end of the MCP. This adds proprioceptive capabilities to the device by giving each actuator a sense of its state rather than relying on estimations, lending itself to a system easily programmable for custom therapy exercises. Answering the research question with concern towards the performance of a dynamic controller, this device has shown to be able to reliably articulate an actuator to a desired bending bending however it is dependent on the quality of the input signal. Unfortunately, the Bend Labs sensors used are not robust enough to reliably operate a device unless under ideal conditions. This may be able to be fixed by replacing these sensors with small inertial measuring units (IMUs) into the tip of the fingers and top of the palm to record movement over time which can be turned into angle measurements.

In its current state, this glove is focused on rehabilitation rather than being an assistive device. However, this device can easily be used to table top assistive application such as writing or eating. What limits this device to rehabilitation is the supporting systems. Unless a patient has a wheelchair that the controlling electronics can be mounted to with a sufficient power supply, this device is not mobile in the sense of a patient walking around their house with it. Over time this device could be adapted into a mobile assistive device, however further work must be done to minimize the electronics and power requirements.

6.2.1 Future Work

To further the functionality of the device, future work may attempt to add intent estimation with an accelerometer or IMU. An EMG could be considered however, as previously mentioned their use has been shown to not produce desirable results on patients with muscle impairments (Delph *et al.* (2013); Thielbar *et al.* (2016); Zhou *et al.* (2019); Polygerinos *et al.* (2015a)). If IMUs are used, then it may be possible to use their readings to track and control the state of the actuators as well. This would require the addition of a sensor fusion algorithm to calculate bending angle.

For future testing, it would be beneficial to use impaired subjects for the Box and Blocks test to provide a more accurate understanding of the performance of the device with its intended patient population. These patients would first be asked to perform the test with no assistance at least three times so an average can be found. The patients would then repeat the tests with the addition of the glove. For both tests the patient will be allowed the standard warm up time to familiarize themselves the test/device. It may be beneficial to also hold a trail with the glove attached but not powered on, to see if there is any passive effect of the glove. The number of blocks moved with and without the glove would them be recorded and compared. Each patient would then be asked to fill out a qualitative survey about their experience with the device.

The device developed in this paper offers a promising solution to the lack of at-home rehabilitation devices. With further development into soft sensors a more reliable system can be made to bridge the gap of care between the fully insured and under/non insured patient populations.

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APPENDIX A

PREVIOUS ACTUATOR MANUFACTURING METHOD

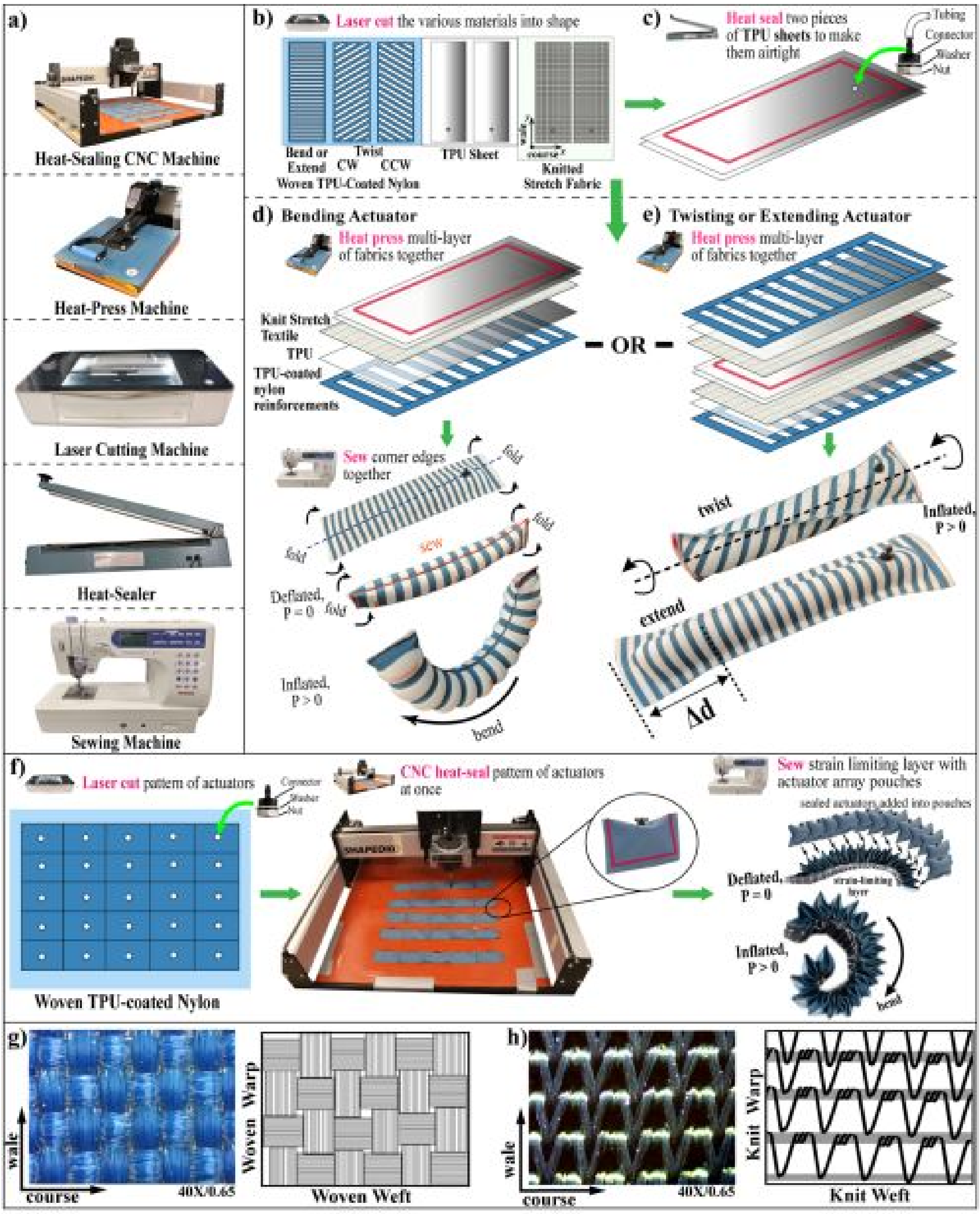

Figure A.1: Manufacturing Method of Previous Actuator Design (Nguyen *et al.* (2019); Nguyen and Zhang (2020))

APPENDIX B

CLOSE UP OF THREE FINGER PINCH

Figure B.1: Close Up of Three Finger Pinch From Box and Blocks Test