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ON THE DESIGN, MODELING, FABRICATION, AND APPLICATION
OF FIBER-REINFORCED SOFT PNEUMATIC ACTUATORS

BY

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DISSERTATION

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Abstract

Fiber reinforced elastomeric enclosures (FREEs) are soft pneumatic actuators with performance characteristics that can be customized during fabrication. In the decades since their initial conception, FREEs and their predecessors have been applied in a multitude of linear force generation applications. A variety of theoretical models have been developed to predict their force and displacement output. However, existing research has predominantly focused on FREEs as singular actuators in near-linear configurations. There is a vast and relatively unexplored design space accessible when one deviates from this paradigm.

The work in this dissertation expands the FREE design space by using them as building blocks in unique architectures and applying them in wrapped and curved configurations. Two exoskeleton applications served as platforms for exploration of new FREE architectures designed to actuate revolute joints. A nested linear architecture and a helical wrapped architecture were combined for effective joint actuation and stiffening in a soft elbow sleeve exoskeleton. A cable-driven variation of the nested linear architecture was compared to a bioinspired pennate architecture for a shoulder exoskeleton application. These two investigations served to demonstrate the ability of FREE architectures to deliver tailored solutions to unique mechanical challenges.

The centerpiece of this research is a medical training simulator uniquely enabled by novel curved FREEs. Transseptal puncture (TP) is the technique used to access the left atrium of the heart from the right atrium via the interatrial septum in increasingly common cardiac catheterization procedures. There is a need for a realistic TP task trainer to help cardiology

fellows learn to recognize key haptic cues felt through the catheter as they navigate to the correct puncture site on the fossa ovalis. To create low-risk training opportunities for new TP operators, we developed a Soft Active Transseptal Puncture Simulator (SATPS), designed to match the dynamics, static response, and visualization of the heart during TP. The SATPS includes three main subsystems: (i) An anatomically accurate soft right atrium uses FREEs to mimic atrial contraction and provide realistic dynamic force feedback to the operator. (ii) A replaceable, puncturable fossa ovalis simulates the tissue properties of the real fossa to provide accurate force feedback during tenting and puncture. (iii) A simulated intracardiac echocardiography environment gives the user live visual feedback representative of an ultrasound monitor during an actual TP procedure. Subsystems of the SATPS were validated with benchtop tests. In a system validation study, experienced clinicians rated the SATPS as realistic and useful for training fellows in TP.

Novel FREE designs will be enabled by new tools in the form of a parametrized finite-element model and low-cost additive manufacturing. The curved FREEs introduced in the SATPS atrium were investigated using finite-element analysis (FEA). A custom software was developed to procedurally generate FREEs with user-defined parameters and three-dimensional centerline paths. The program interfaces with commercial FEA software for analysis and extracts FEA output for post-processing, allowing for exploration of kinematic trends. Prototyping curved FREEs or other geometrically-complex soft actuators will be enabled by a low-cost soft materials 3D printer designed to print elastomers of multiple stiffnesses with a water-soluble support material.

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Chapter 1

Introduction

1.1 Background

1.1.1 Fiber-Reinforced Elastomeric Enclosures (FREEs)

Soft robotic actuators offer unique advantages over their traditional rigid counterparts, including lower weight, fewer components, versatility in movement, and increased safety for human interaction [1], [2]. The studies presented in this dissertation focus on a class of soft actuators called fiber-reinforced elastomeric enclosures, or FREEs [3]. The working principle of FREEs is inspired by hydrostatic skeletons, which are biological features that produce movement and force through the interplay of fluid pressure and connective fibers embedded in flexible chambers. FREEs consist of an elastomer bladder surrounded by helical networks of inextensible fibers, with air supplied through a pneumatic fitting attached to one end. The angle of the fibers relative to the longitudinal axis of a FREE control the FREE's deformation when pressurized (Fig. 1.1). Typically FREEs include two networks of fibers, defined by fiber angles α and β . During fabrication, FREE fiber angles can be set to produce contraction, extension, axial rotation, rigidity, or bending [3]. The versatility of FREEs make them adaptable to a variety of soft robotic applications.

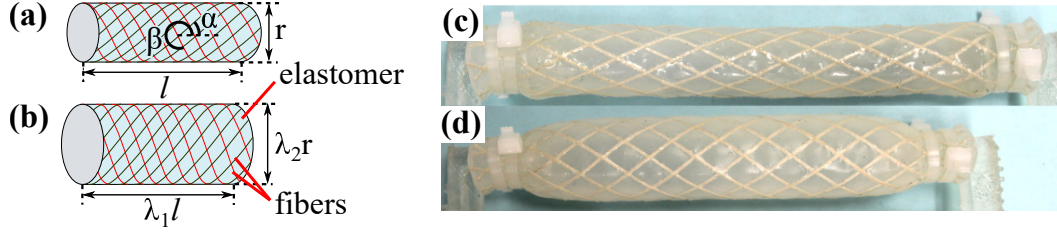


Figure 1.1: (a) A FREE is defined by its fiber angles, α and β . (b) When $\alpha = -\beta < 54^\circ$, the FREE contracts axially and expands radially under pressure. (c) Unpressurized and (d) pressurized contracting FREE.

1.1.2 FREE Design

Fiber-reinforced soft actuators were first patented in the 1950s and their design was virtually static for decades to follow, but in recent years there has been accelerated evolution of FREE designs fueled by increased interest in soft robotics [4], [5]. The predecessor and a subset of FREEs, the pneumatic artificial muscle (PAM), also known as the McKibben muscle, was a contracting actuator consisting of a rubber tube covered by a braided sheath with symmetric fiber angles. Commercial PAMs were eventually developed based on the same principle [6]. FREEs with symmetric fiber angles were the only form of FREE until the 2010s, when Bishop-Moser, Krishnan, Kim, and Kota coined the term FREE and expanded the FREE design space with the introduction of parallel architectures and asymmetric fiber angles [3], [7], [8]. Shortly after, Connolly et al. designed segmented FREEs with regions of fiber angles that were programmed to perform task-specific motions, and developed an optimization-based design strategy for trajectory matching with multi-segmented FREEs [9], [10]. Multi-FREE architectures have been developed to further tailor actuation performance, including wrapped and nested architectures, explored in Chapter 2, and pennate architectures inspired by muscle fibers, explored in Chapter 3 of this dissertation [11]–[16]. Until recently, FREEs were exclusively linear actuators, largely due to manufacturing constraints. Hu et al. recently developed a pre-curved extending FREE that produces bending using a strain-limiting material [17]. Chapter 4 of this dissertation introduces novel single-material pre-curved FREEs, which are developed further in Chapter 5. This dissertation focuses

predominantly on FREEs with symmetric fiber angles ($\alpha = -\beta$) that are less than 54° , which produce axial contraction and radial expansion with no axial rotation.

1.1.3 FREE Modeling

FREE kinematics have been modeled extensively since the actuator’s conception, with most of the research focus placed on contracting and extending FREEs. Several notable contributions are listed here. Kothera et al. conducted experimental characterization of three leading models: a simple kinematic derivation from Chou et al. that incorporates the virtual work principle, a model from Klute et al. that treats the bladder as a Mooney-Rivlin solid to account for elastic energy storage, and a model from Ferraresi et al. that derives actuator force using Newton’s second law [18]–[21]. Trivedi et al. used Cosserat rod theory to apply transverse bending loads to extending fiber-based actuators in parallel bundles [22]. Goriely and Tabor modeled the deformation of elastic cylinders with asymmetric fiber angles [23]. Krishnan et al. were the first to publish an experimentally validated model for the kinematics of generalized FREEs with any fiber angle configuration [3]. Singh and Krishnan derived a constrained maximization of the bladder volume that, while computationally expensive, accurately predicts the deformed shape and force output of FREEs with any general fiber angle for a given pressure [24]. The FREE application described in Chapter 3 of this dissertation used Singh and Krishnan’s constrained maximization model for FREE force and displacement predictions. The model was used again in the early design stages of the soft robotic heart simulator presented in Chapter 4.

Finite-element analysis (FEA) of FREEs is a common and effective method for validating analytical models and FREE-based designs [25]. FEA can model nonlinearities that are difficult to capture with analytical models. Existing FREE fabrication methods are time-consuming, and FEA offers the ability to verify actuation performance before producing physical prototypes. It can also act as a substitute for experiments requiring many iterations of FREEs. To predict the pressurization response of FREEs embedded in a soft elastomer,

Roche et al. used a simplified FEA model that used temperature and orthotropic coefficients of thermal expansion to model kinematics of FREEs during pressurization [26]. When Krishnan et al. expanded the FREE design space to include all possible fiber angles, the authors used FEA to verify their kinematic model and to capture the limitations imposed by FREE end cap constraints that could not be included in the kinematic model [3]. Connolly et al. similarly used FEA to investigate new kinematic possibilities enabled by asymmetric fiber angles [9]. Satheeshbabu and Krishnan used FEA to estimate material parameters for a pseudo-rigid-body that captures axial and bending stiffness of pressurized FREEs [27]. Chapter 5 of this dissertation demonstrates the use of FEA as a design tool for pre-curved FREEs, an application for FEA that is increasingly viable as computing technology advances.

1.1.4 FREE Fabrication

There are several known methods for fabricating FREEs. The contracting subset of FREEs called PAMs are fabricated from a rubber bladder surrounded by a braided sleeve [4], [19], [28], [29]. Bishop-Moser and Kota produced FREEs with custom fiber angles by manually wrapping individual fibers around a latex rubber bladder, and coating the assembly with latex rubber [30]. The FREE actuators used in the exoskeletons in Chapters 2 and 3 were manufactured with this manual wrapping method. Manually-wrapped FREEs can be made with more consistency and better fiber angle accuracy from silicone rubber cast into the desired bladder size with grooves for each fiber [27] (Fig. 1.1c-d). The fibers are then sealed to the bladder with additional silicone. The curved FREEs used in Chapter 4 and further developed in Chapter 5 are made from cast silicone. Sangian et al. 3D-printed fibers on a rotating mandrel for linear contracting FREEs, taking into consideration the need for fiber separation at fiber intersections [31]. Recently, Schaffner et al. developed UV-curable silicone inks with a range of stiffnesses that they used with a direct-ink-writing (DIW) 3D printer to produce soft actuators resembling FREEs [32]. Multi-material elastomer 3D printing, explored further in Chapter 6, could unlock vast possibilities for networks of FREE actuators

that are impossible or highly impractical to fabricate with established methods.

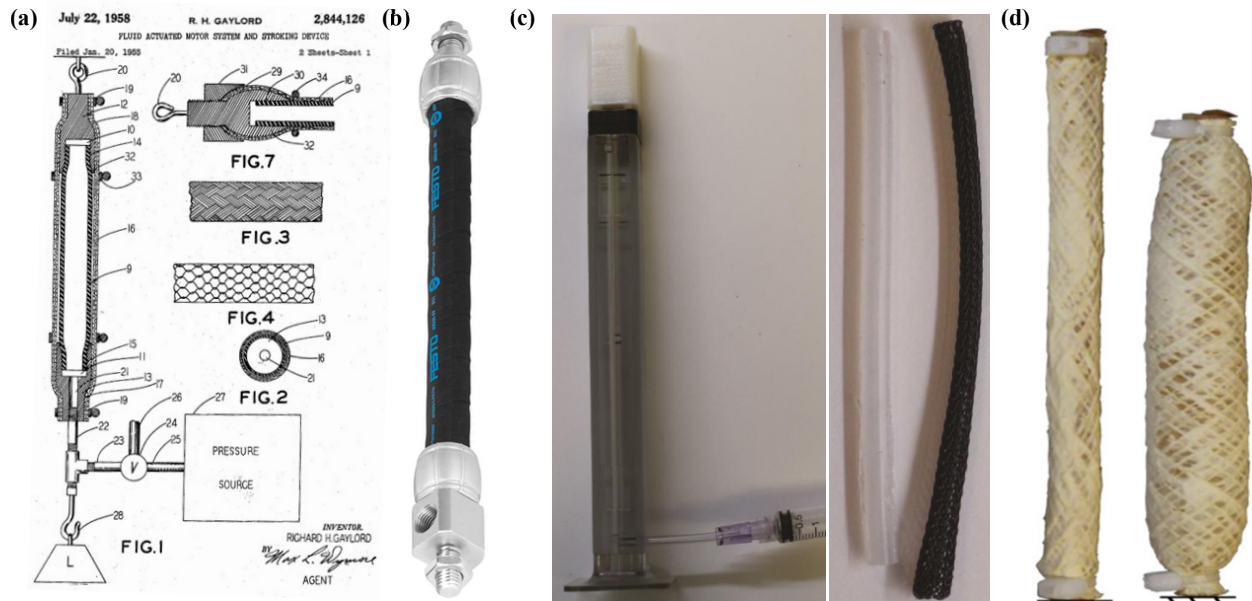


Figure 1.2: (a) The original PAMs [4], (b) modern commercial PAMs [29] are constructed from a braided mesh tube encased in rubber. (c) Cast silicone rubber can also be used for PAMs with braided fiber networks [26]. (d) Manually-wrapped latex FREE [33]. A cast silicone FREE made with preset fiber grooves can be seen in Fig. 1.1c–d.

1.1.5 FREE Applications

FREEs, most commonly in the form of PAMs, have been used in a variety of applications that take advantage of FREEs' compliance and high power-to-weight ratio. PAMs are a popular choice for wearable applications since their low weight and compliance make them portable and safe for human interaction [34], [35]. Tentacle-like soft manipulators make use of parallel bundles of PAMs to reach a large range of motion and wrap themselves around objects with a variety of form factors [22], [33], [36]. Past works have applied PAMs as bioinspired analogues for human muscle in applications such as a ventricular assist device [26]. Chapters 2 and 3 use architectures of FREEs in exoskeleton applications. Chapter 4 presents curved FREEs applied in a bioinspired medical training simulator.

1.2 Objectives and Significance

The overarching goal of this research is to advance the field of soft robotics through novel deployment of FREE actuators. Wide-scale adoption of FREEs has been limited by their achievable force and stroke characteristics when used alone in purely linear configurations. For example, exoskeletons for upper extremities require high forces over large ranges of motion. Designing a single FREE to meet these requirements may result in unreasonably large and bulky geometries. The work in this dissertation seeks to expand the FREE design space by introducing additional design variables in the form of topological architectures and curvatures. Below is a brief summary of the objectives and significance of each chapter of this dissertation:

Chapters 2 and 3 detail two completed studies in which bioinspired architectures of linear contracting FREEs for upper limb exoskeleton applications were established to tailor the achievable force, stroke and stiffness. The studies provided useful insight into these architectures and provided a benchmark for their use for revolute joint actuation.

Chapter 4 includes the design and validation of a FREE-based Soft Active Transseptal Puncture Simulator (SATPS) to provide low-risk training opportunities to novice cardiologists in a technique called transseptal puncture (TP). The device is intended to reduce the need for training on actual patients and thereby lower risk of dangerous complications by teaching new TP operators the haptic skills necessary for accurate and swift completion of the procedure. The FREEs in the SATPS follow the contours of the heart while exerting dynamically equivalent forces on the heart walls. This new type of FREE is explored further in Chapter 5.

Chapter 5 presents a novel type of FREE that is cast in a predetermined curved shape to produce axial length change along a curved path, as well as bending. Previously, FREEs have been linear in shape due to manufacturing limitations. This new type of FREE can be used in designs demanding contraction along curved paths without buckling. This work includes a finite-element model that is used to explore the fundamental kinematics of curved FREEs.

Chapter 6 describes the development of a 3D printer capable multi-material printing of

silicone-based inks. This work seeks to further enable the inclusion of the new FREE design variables introduced in the chapters above. 3D printing with soft materials and water-soluble support structures will enable soft robot designs unattainable with traditional manufacturing methods, such as complex networks of pneumatic chambers, and active materials with embedded actuators.

Conclusions and potential future work are presented in Chapter [7](#).

Chapter 2

Augmented Joint Stiffness and Actuation using Architectures of Soft Pneumatic Actuators ¹

Abstract

Soft robotic actuators are well suited for use in exoskeleton applications due to their innate compliance and low weight. We have developed a wearable soft robotic sleeve that uses fiber reinforced elastomeric enclosures (FREEs) to provide actuation and stiffness at the elbow for augmented lifting and carrying ability. The sleeve includes novel linear and helical actuator architectures to induce and resist joint movement respectively, and is intended to be comfortable, lightweight, and low profile. We developed test protocols to measure actuation and stiffness performance of different helical and linear architectures, and to compare helical and linear actuator groups when used individually and together. Our findings indicate that nested linear actuators have superior contraction ratios compared to parallel linear actuators, resulting in greater angular displacement. Stiffness from helical actuators increased with

¹This work has been published in [14].

pressure and number of parallel actuators. A combined linear-helical actuator configuration considerably outperformed helical and linear actuator groups when used on their own.

2.1 Introduction

Soft robotic actuators offer unique advantages over their traditional rigid counterparts, including lower weight, fewer components, versatility in movement, and increased safety for human interaction [1], [2]. Pneumatic soft actuators have been used previously to control revolute joints in various robotic applications for their simplicity and high strength-to-weight ratio [37]–[40]. Their innate compliance and low weight make pneumatic soft actuators particularly well suited for use in wearable devices. Previous designs have primarily focused on contracting or expanding linear actuators, without much consideration given to stiffness modulation [35], [41], [42]. Wearable devices such as these can be used to restore or augment lost mobility in a medical application or improve lifting and carrying capacity and reduce risk of injury in an industrial setting. Soft pneumatic wearable devices require an air source, which can be in the form of a portable compressed air tank or a tethered air supply.

This paper documents our most recent progress in developing a wearable soft robotic sleeve that is actuated using a class of soft pneumatic actuator known as fiber reinforced elastomeric enclosures (FREEs) [3]. FREEs utilize networks of inextensible fibers wrapped around elastomer tubing to control deformation when the tubing is pressurized, producing movement that is determined by the fibers’ angle with respect to the longitudinal axis of the actuator (Fig. 2.1a). Fiber angles can be set to produce contraction, extension, axial rotation, rigidity, and bending. FREEs can be manufactured with small diameters, allowing them to be used in discreet and comfortable wearable applications.

The performance of FREEs (contraction, and force) are known to drastically reduce with miniaturization [12]. The goal of this work was to develop architectures of FREEs to improve the user’s load bearing ability and assist in joint actuation, while maximizing

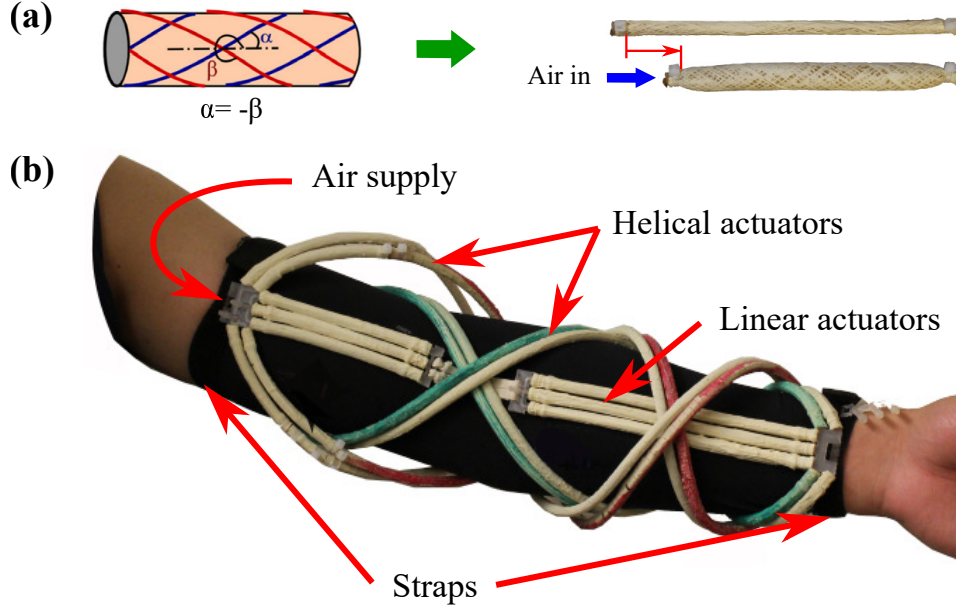


Figure 2.1: (a) A FREE with equal fiber lengths less than 54° contracts when pressurized. Reproduced with permission [3]. (b) Layout of one embodiment of the FREE-actuated sleeve.

compactness, discreetness and wearability. Our sleeve prototype included helical and linear actuator architectures that provided torque to the elbow joint to reduce loading on the elbow during lifting and carrying (Fig. 2.1b). Specifically, we will present our experimental evaluation of the actuation and stiffness performance of helical and linear actuators with different architectures, when used separately and together. Our testing was based on a human elbow joint, modeled by a 3D printed arm with one degree of freedom, but the behavior of the FREE actuators can be extrapolated to any rotational joint, human or robotic. Stiffness was determined by loading the joint and measuring its resistance to angular displacement. Actuation performance was measured as the achieved joint angle at a given actuator pressure. This paper will provide an overview of the sleeve design, specific design parameters based on FREE actuator characterization, experimental protocols, and finally our results and their implications.

2.2 Sleeve Design Overview

2.2.1 Sleeve Structure

The base layer of our device was a black fabric sleeve, secured to the arm with straps at the wrist and biceps (Fig. 2.1b). FREE actuators were installed with 3D printed fittings that were then stitched onto the two straps. The actuators, which were all contracting types, fall into two categories in this design: helical actuators and linear actuators.

Helical FREE actuators have been shown previously to provide stiffness to a simplified elbow joint [11]. In our sleeve design, there were two groups of helical actuators, designated as Group 1 (red in Fig. 2.2) and Group 2 (green in Fig. 2.2). Each group contained two subgroups of long contracting FREEs were wrapped around the upper arm and forearm forming a symmetric double-helical structure, such that the intersection of a double-helix crossed the back (Group 1) or front (Group 2) of the elbow. Each helical actuator subgroup formed 1.5 turns around the arm. The lengths of the two actuator groups were designed independently to ensure they could conform to the arm when pressurized at any joint angle from 0–90°, but would not impede movement of the joint when not in use. The length of the actuators within each group were all the same. The stiffness performance of each subgroup can be improved with increased number of parallel actuators, as demonstrated in Section 2.6.1.

Linear actuators provide contraction along a single axis. This design used two linear actuators: one located along the bicep and a longer one located along the forearm (Fig. 2.1b). A flexible string connector with negligible extensibility connected the two actuators at the elbow. Using two separate actuators allowed for unimpeded movement when the actuators were not pressurized. The lengths of the actuators and the connector were determined by the arm geometry. The proximal end of the upper arm actuator and the distal end of the forearm actuator were attached to the sleeve straps. The distal end of the upper arm actuator and the proximal end of the forearm actuator should nearly touch when the arm has a joint angle of

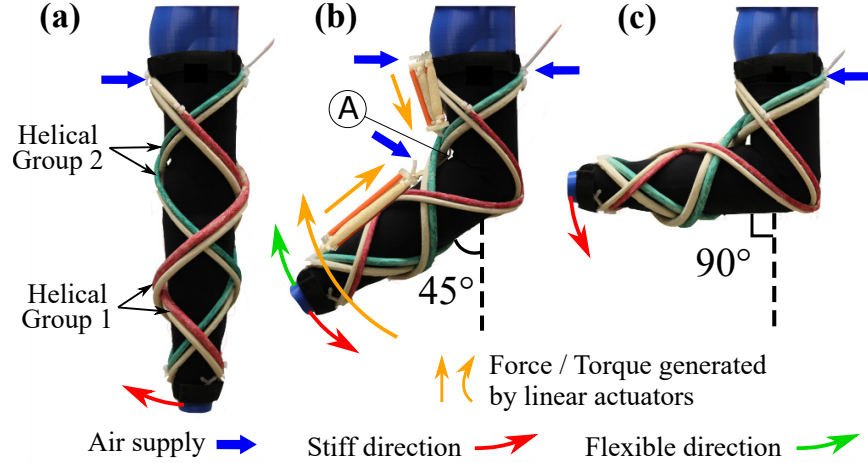


Figure 2.2: Lateral view of sleeve prototype, installed on a 3D printed arm. (a) Helical Group 1 provides stiffness in flexion. (b) Helical Group 2 provides stiffness in extension while linear actuators provide displacement in flexion. (c) Helical Group 2 provides stiffness in extension.

90°. The entire linear actuator assembly should have no slack at a joint angle of 0°. As with the helical actuator groups, multiple FREE actuators can be used in each linear actuator assembly to improve performance.

2.2.2 Working Principle

In the current sleeve design, there was a single air inlet for each actuator group, allowing each of the actuator groups to be pressurized independently. A qualitative demonstration of the sleeve function is shown in the supplemental video.

During sleeve operation, Group 1 helical actuators provided stiffness in the direction of flexion, i.e. resist loading in flexion, while Group 2 provided stiffness in the direction of extension. When pressurized, the helical actuators contract to conform to the shape of the arm within the sleeve at any joint angle from 0–90°, creating joint stiffness in one or both directions for any arm position. If only Group 1 actuators are pressurized, the wearer is free to move the arm in extension, and movement in flexion will be resisted (Fig. 2.2a). Similarly, if only Group 2 actuators are pressurized, the wearer will have free movement in flexion, but will have increased stiffness in extension. If both helical groups are pressurized, the joint is

effectively “locked”, facing increased stiffness in both directions.

The linear actuators’ primary function is to actuate the wearer’s arm at the elbow. When the actuators are pressurized, the design mimics the working principle of the human bicep, producing angular displacement in flexion by generating a contracting force (Fig. 2.2b). Improved actuation performance, i.e. increased angular displacement at maximum pressure, can be achieved by maximizing the contraction ratio and the overall length of the linear actuator group. By using two linked actuators on the forearm and upper arm instead of a single actuator on the upper arm, we more than double the stroke of the actuator group. When the linear actuators and the Group 2 helical actuators were used together, the Group 2 actuators provided an anchor point for the linear connector at the elbow, increasing the joint angle at a given linear actuator pressure (point A in Fig. 2.2b). Given the constraint imposed by the physical dimension of the arm, we were also required to introduce new nested architectures to our linear actuators to improve linear stroke.

2.3 FREE Actuator Characterization

2.3.1 Nested Multi-Stage Linear Actuators

Nested linear actuators use FREEs in combination with 3D printed assemblies to improve the stroke compared to equal-length parallel linear actuators [12]. Nested and parallel designs have the same actuator length for each section, denoted as l_1 for the upper arm actuators and l_2 for the forearm actuators, and the same connector length l_c (Fig. 2.3).

A two-level nested actuator contains two connecting stages, which couple two parallel sections of contracting actuators (Fig. 2.3b). Two 3D printed posts are fixed between the two stages. When the FREEs contract, the posts act in compression to transmit displacement from one stage to the other, allowing the two sections of contracting actuators to work in a serial fashion. The connecting stages and the posts have internal air channels to allow all four actuators to receive air pressure from a single air input channel. Upon pressurizing a

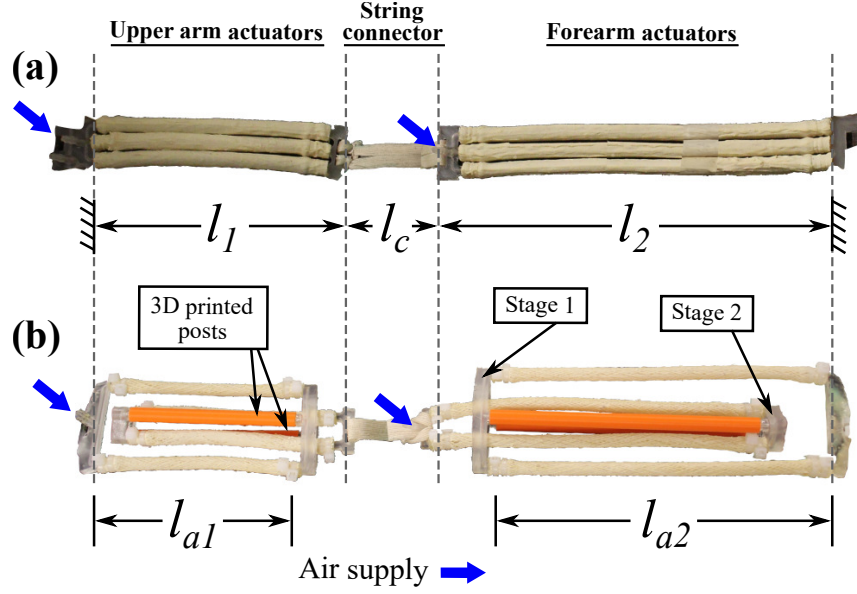


Figure 2.3: Structure of equal length linear actuators using (a) single actuators in parallel and (b) two-level nested actuators.

nested actuator, the overall system deformation is the sum of the displacements generated by each section of contracting FREEs. Consider the following equation:

$$N(l_a C_r a) > l_s C_s \quad (2.1)$$

where l_a and $C_r a$ indicate the length and contraction ratio of each contracting actuator within the nested structure, while $C_r s$ denotes the contraction ratio of a single actuator with length l_s . Here, contraction ratio is defined as $\Delta l / l_0$, where Δl is the axial length reduction and l_0 is the original length. N indicates the number of nested levels, which is two in this case.

Nested structures were designed in a centerline symmetric fashion so that torque generated by the transmitting load is balanced (Fig. 2.3b). As a result, each nested level inherently has two actuators arranged in parallel. Thus, the actuation force generated by a nested actuator is theoretically equivalent to two single actuators working in parallel.

In our sleeve application, all actuators used both in the single case and in the nested case had the same fiber angles and tubing dimensions. From our analysis of the nested structure, to maximize the overall system contraction ratio, l_a was chosen to be around $0.8l_s$.

Under such circumstance, there was virtually no deviation between $C_{r,a}$ and $C_{r,s}$ since the contraction ratio of actuators with same specifications decreases dramatically only with large differences in length. Thus, Eq. (2.1) held true, indicating that nested actuators would provide an advantage in terms of stroke.

2.3.2 FREE Contraction Ratios

In this subsection, we characterize our contracting FREE actuators using experimental data. Prior research has shown that a FREE actuator’s contraction ratio and force production have an upper limit as length increases [12]. The FREEs used in the sleeve prototype were all longer than this threshold length, allowing us to present contraction ratios as percentages of initial length.

We characterized individual FREE actuators by measuring their contraction ratio using a custom test apparatus (Fig. 2.4a). Air was controlled with a manual regulator, flowing through a digital pressure gauge (Parker, model K4517N14160) and into the actuator, which was fixed at the inlet end. For contraction tests, the other end of the actuator was unconstrained, allowing for measurement of the FREE at different pressures. The setup could also be equipped with a load cell (FUTEK, model LRF400) to measure contraction force of the FREEs. We measured the contraction ratios of FREEs used in helical actuators as well as two types of linear actuators (Fig. 2.4b). The helical actuator was smaller in diameter than the linear actuators, and had a thicker outer layer of latex (Table 2.1). For this reason, the helical actuators had a smaller contraction ratio than the linear actuators. The two-level nested actuators had 80% greater contraction ratio than single actuators with the same FREE parameters, verifying that nested linear actuators provide greater stroke than parallel actuators of equal length.

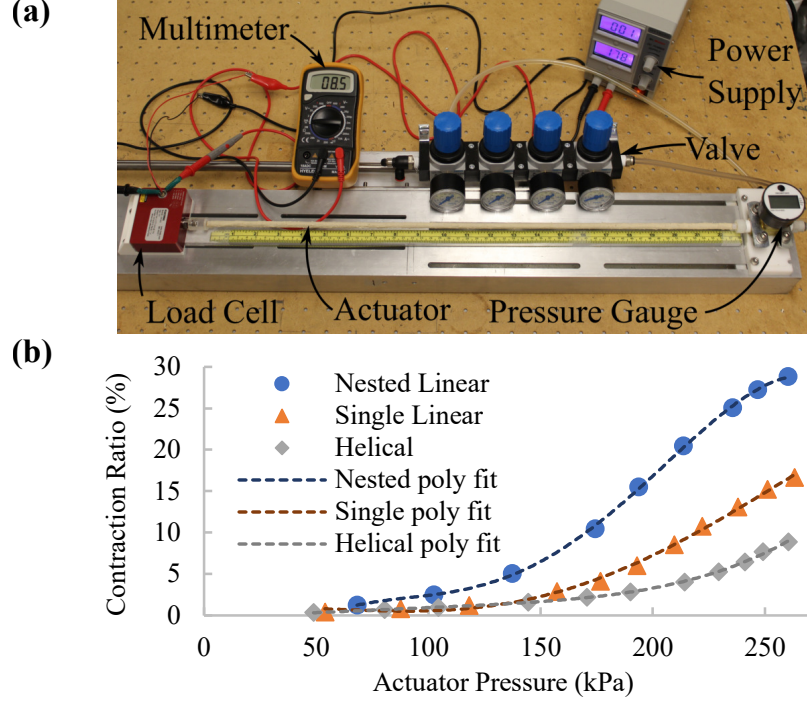


Figure 2.4: (a) Experimental setup for FREE contraction measurement. Reproduced with permission [11]. (b) Mean contraction ratio at different pressures for a two-level nested linear actuator, a parallel FREE linear actuator, and a helical actuator. Fourth-order polynomial fit lines are included for extracting contraction ratios for any actuator pressure. $n = 3$ for all tests.

2.4 Sleeve Design Parameters

In this section, we present the specific design considerations for the sleeve and illustrate the underlying principle for parameter selection. We restate and emphasize the functionality of the helical and linear actuator groups as our design constraints:

1. The undeformed length of the two helical actuator groups must be long enough to impart negligible resistance to movement for joint angles from 0° – 90° .
2. The absolute value of contractions (contraction ratio multiplied by initial length) generated by the helical actuators should allow for the actuators to contract to the contours of the arm at any joint angle from 0° – 90° under a certain pressure below 310 kPa.

3. The length of the two linear actuators must allow for full extension (0°) without stretching, as well as full flexion (90°) without coming into contact with each other. The string connector may buckle during flexion.

The path length of each actuator group changes with joint angle (Fig. 2.5). As the elbow flexes, increasing the joint angle, the helical Group 1 actuators must increase in length to continue conforming to the arm. If these actuators are pressurized, they will resist increases in length, resulting in increased joint stiffness in that direction. This principle applies to all three actuator groups. Similarly, if the actuators are subjected to a pressure corresponding to an unblocked actuator length smaller than the path length for the current joint angle, then that actuator group will produce a moment at the elbow in the direction of decreasing length in Fig. 2.5b.

We chose the length of each actuator using Fig. 2.5b. The undeformed length of each actuator must equal the maximum value of the corresponding curve in this plot to prevent undesired restriction of movement throughout the operating range. The length of the linear actuator connector was approximately equal to the overall change in path length for the linear actuator group. This guideline maximizes the length of the linear actuators within the confines of design constraint 3. FREE fiber angles must be less than 54° to produce contraction [3], [19], and the specific angle determines the actuator's contraction ratio. Small fiber angles produce large contraction ratios, but also result in larger radial expansion under pressure. Since the intent was to create a wearable device, we wanted to limit radial expansion to minimize

Table 2.1: FREE Actuator Parameters for Sleeve Device

Helical Actuators			
Fiber Angles	$\pm 15^\circ$	Outer Diameter	4.8 mm
Latex Thickness	~ 0.8 mm	Group 1 Length	534 mm
Wall Thickness	0.8 mm	Group 2 Length	499 mm
Linear Actuators			
Fiber Angles	$\pm 15^\circ$	Outer Diameter	6.4 mm
Latex Thickness	~ 0.8 mm	Upper Length	89 mm
Wall Thickness	0.8 mm	Lower Length	148 mm
Stage Thickness	6.0 mm	Connector Length	89 mm

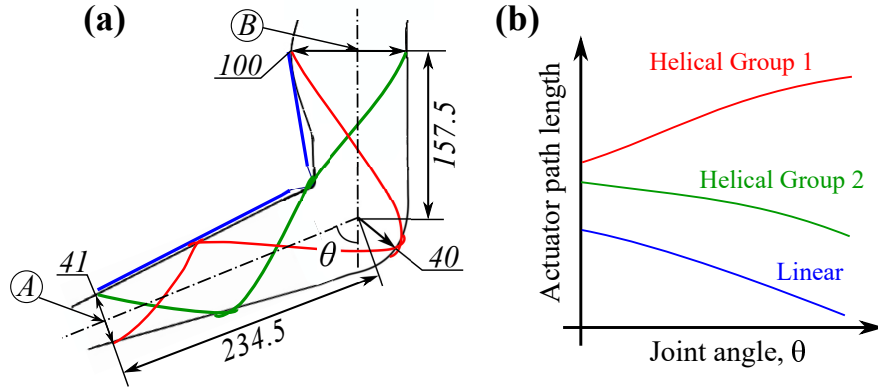


Figure 2.5: (a) Arm model and sleeve dimensions, in mm. Group 1 helical actuators are shown in red, Group 2 helical actuators are green, and linear actuators are blue. Points A and B represent the ends of the sleeve at the wrist and bicep, respectively. (b) Generalized behavior of each actuator group’s path length with respect to joint angle.

bulkiness and discomfort from excessive constriction. Keeping all design considerations in mind, and through some trial and error, we arrived at the parameters in Table 2.1.

2.5 Experimental Methods

The goal of our testing was to measure the ability of helical and linear actuator groups to actuate and provide stiffness to a rotating joint for joint angles from 0° — 90° . Our focus was on applications involving lifting and holding, so actuation tests measured displacement in flexion, while stiffness tests resisted loading in extension. For this reason, Group 1 helical actuators were excluded from our setup. The two helical actuator groups have the same working principle and are expected to work symmetrically.

To measure the sleeve’s performance under different air pressures and loading conditions, we attached the sleeve to a 3D printed arm. We conducted all testing using a 3D printed model of a human arm with one degree of freedom at the elbow joint, modelled after a male subject (height 175 cm, mass 70 kg). We fixed the upper arm to a rigid aluminum frame (Fig. 2.6a). The frame suspended the arm to allow free movement of the forearm. Weights placed in a container fixed to the wrist of the arm provided loading in the extension direction.

Manual regulators controlled airflow into the linear and helical actuators independently. We used an electronic goniometer (Baseline, model 12-1027) to measure joint angle.

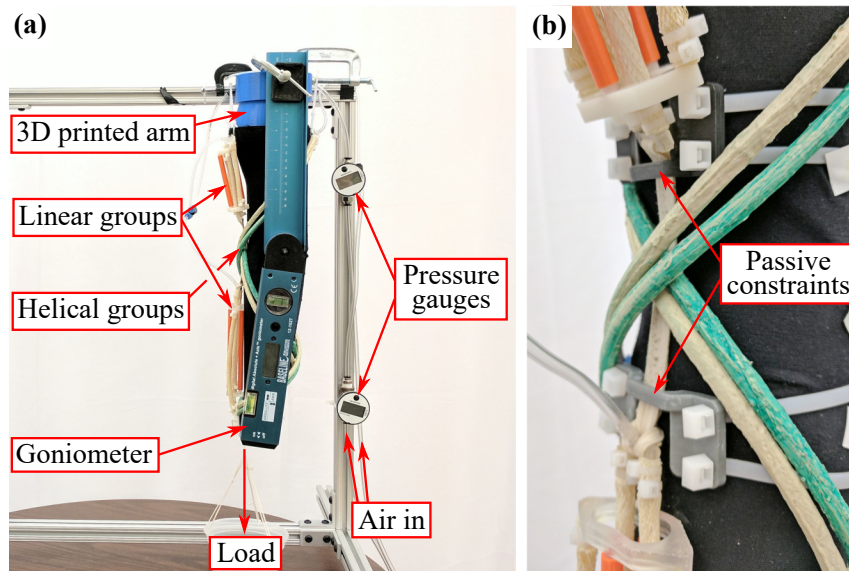


Figure 2.6: (a) Experimental setup for stiffness and actuation testing. An electronic goniometer was fixed to the aluminum frame in line with the joint pivot point for measuring joint angle. (b) Passive constraints used in L configuration, attached at the proximal end of the forearm and the distal end of the upper arm. These constraints were removed for H and LH configurations.

2.5.1 Actuation Testing

We evaluated the joint stiffness of each actuator configuration by loading the arm model incrementally from a bent position with the actuators filled to their maximum pressure. First, we held the arm at a joint angle of 90° while pressurizing the actuators to 262 kPa (linear) or 276 kPa (helical), and recorded the starting joint angle. We added weight at the wrist in 200 g increments up to 2600 g, recording the pressure and joint angle after each addition. All actuators were depressurized between each trial. Stiffness was defined as the slope of the resulting torque vs. joint angle curve. Although the slopes of these curves are negative due to the direction of loading, stiffness values are presented here as positive scalar values for clarity.

2.5.2 Actuation Testing

To measure the ability of each actuator configuration to provide flexion about the elbow, we started with depressurized actuators and the arm at a nominal 0° joint angle. We then incrementally increased pressure to the actuators, recording the pressure and resultant joint angle at each step. We repeated the tests for 0 g, 500 g, and 1000 g loads, depressurizing between each test.

2.5.3 Test Configurations

Multiple assessments of different actuator configurations were examined. To identify the contribution of each FREE actuator to the sleeve’s stiffness and actuation performance, we tested three different actuator configurations: linear actuators (L), helical actuators (H), and a combination of both (LH). The L configuration used two passive constraints to maintain correct actuator positioning (Fig. 2.6b). In the LH configuration, this position was maintained by helical actuators pressurized just enough to assure contact between the surface of the arm and the connector between the two linear actuators. The L and LH configuration used two-level nested linear actuators, and the H and LH configurations used double parallel helical actuators. To examine the effect on joint stiffness of number of actuators used in a helical configuration, we used a modified protocol of the stiffness test described above. We compared multiple helical actuator types using one, two, and three actuators in parallel at 276 kPa and 310 kPa and with loads up to 1500 g. We used the actuation test protocol to compare two-level nested linear actuators with single, double, and triple parallel linear actuators in the LH configuration.

2.6 Results and Discussion

2.6.1 Stiffness Testing

Comparison of mean stiffness values found that stiffness from helical actuators increased with increasing number of actuators and actuation pressure (Fig. 2.7). The increase in stiffness between 276 kPa and 310 kPa actuation pressure was amplified as the number of parallel actuators increased. For both tested pressures, we found a greater increase in stiffness between two and three parallel actuators than we saw between one and two. While it appears that the trend of increasing stiffness would continue with additional parallel helical actuators, there was an upper limit imposed by space constraints. For this prototype sleeve, using more than three actuators in each helical group would create too much interference between actuator groups, inhibiting motion when the actuators are not active.

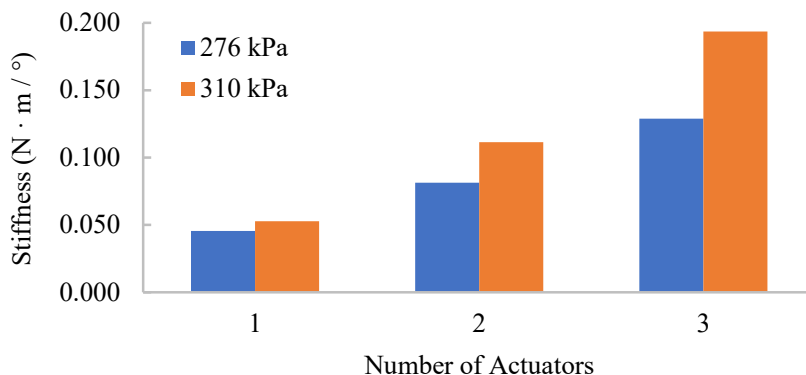


Figure 2.7: Calculated mean stiffness for helical configurations with different numbers of parallel actuators at two pressures. $n = 3$ for all tests.

Results from stiffness testing of L, H, and LH actuator configurations showed that the L configuration provided less stiffness than the H and LH configurations, as indicated by the shallower slope of the L curve when compared to the H and LH curves (Fig. 8a). When the arm was released from a 90° joint angle, the L configuration was only able to provide stiffness for joint angles less than approximately 70° , due to limited contraction length of the selected linear actuators. This threshold marked the upper limit for the contribution of the linear

actuators to the LH configuration. The LH configuration's stiffness was determined by the stiffness of the helical actuators, as indicated by the overlapping H and LH curves for joint angles greater than 70° . For joint angles less than 70° , there was a separation of the H and LH curves due to the added stiffness from the linear actuators in the LH configuration. This trend became more apparent when we plotted the instantaneous stiffness of each configuration at all tested joint angles (Fig. 2.8b). We observed decaying stiffness with decreasing joint angle for the LH and H configurations at similar rates until their joint angle reached approximately 70° . Below 70° , the stiffness of the LH configuration decayed at a slower rate due to the input from the linear actuators, as evidenced by the linear fit lines in Fig. 2.8b. Additionally, while the linear actuators offered less stiffness to the joint than the helical actuators, the stiffness they provided was nearly independent of joint angle, within their working range.

The stiffness results indicate that helical actuators were better suited for providing stiffness to a joint than linear actuators, and were the primary contributor to stiffness in a combined linear-helical actuator configuration. However, stiffness from helical actuators decreased as joint angle decreased. Linear actuators provided less stiffness to the joint, but their stiffness was not reduced as loading increased. The decreasing stiffness effect from the helical actuators can be reduced or eliminated using linear actuators in a combined configuration. Sizing of the linear actuators determined the joint angle at which the linear actuators began to contribute to joint stiffness. For a wearable exoskeleton application, we would likely desire the highest possible stiffness to assist in load carrying. In a more general robotic application, there may be a need for modulated stiffness values. While varying the pressure to the actuators can change the stiffness of the joint, it also reduced the maximum achievable joint angle. Combining actuator types can allow for more customization of stiffness across a wide range of joint angles.

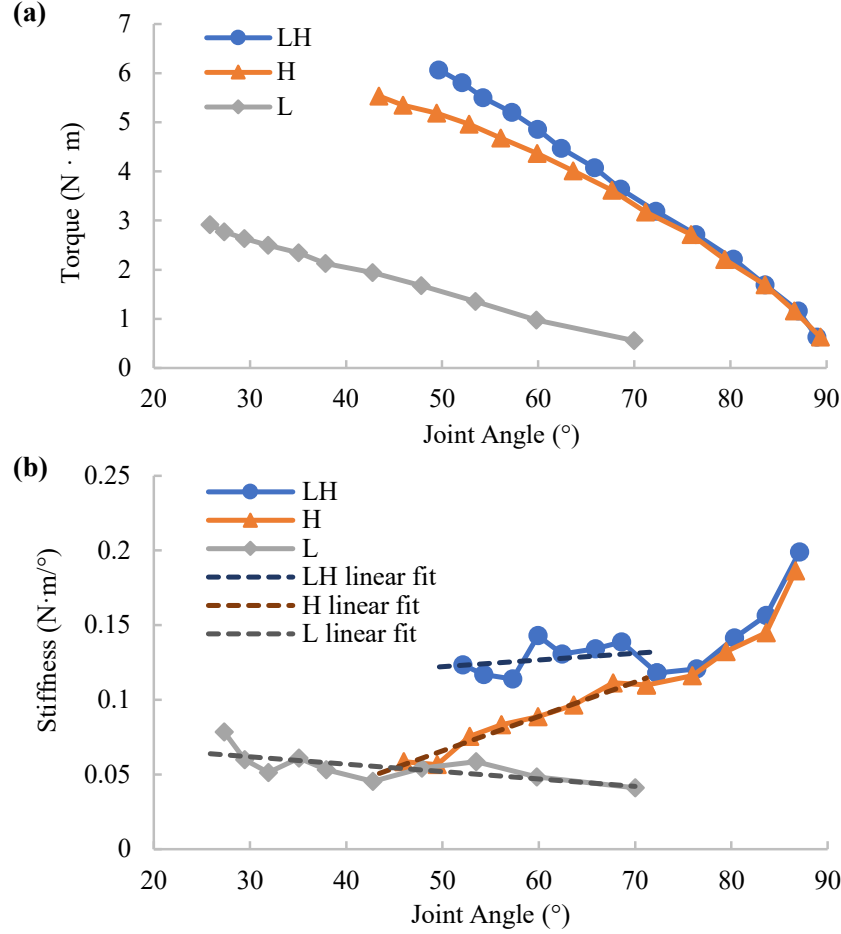


Figure 2.8: (a) Mean applied torque and resultant joint angles for three actuator configurations at 276 kPa. $n = 3$ for all tests. (b) Instantaneous stiffness at all tested joint angles, found using the centered difference method and the data from (a). Fit lines apply to joint angles less than 70°

2.6.2 Actuation Testing

Results from actuation testing of four types of linear actuators demonstrated increased angular displacement with greater numbers of parallel actuators (Fig. 2.9). Nested linear actuators achieved greater maximum joint angles than parallel linear actuators under all loading conditions. As expected, maximum joint angle decreased as external loading increased.

While increasing the number of parallel actuators did not change the contraction ratio of the actuators themselves, it did improve the force generation of the actuator group, allowing for small improvements to displacement. The nested actuators provided a greater increase to

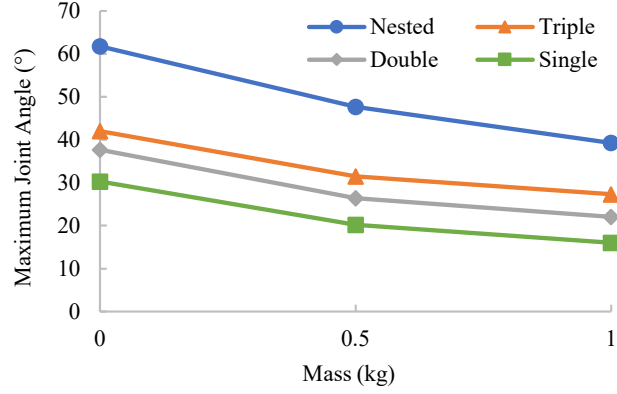


Figure 2.9: Mean maximum joint angle under three loads for LH configurations with different types of linear actuators. $n = 3$ for all tests.

maximum joint angle due to their larger contraction ratio compared to single actuators. As with the helical actuators, there was an upper limit on the number of parallel linear actuators that can be used in this type of application due to space constraints.

Comparison of the L, H, and LH actuator configurations revealed similar angular displacement for the linear and helical actuators when used alone under all loading conditions (Fig. 2.10). When linear and helical actuators were combined in the LH configuration, the sleeve provided greater angular displacement across all operating pressures, excluding those that produced negligible displacement for any configuration. The LH configuration also had the smallest decrease to maximum joint angle when subjected to increasing loads.

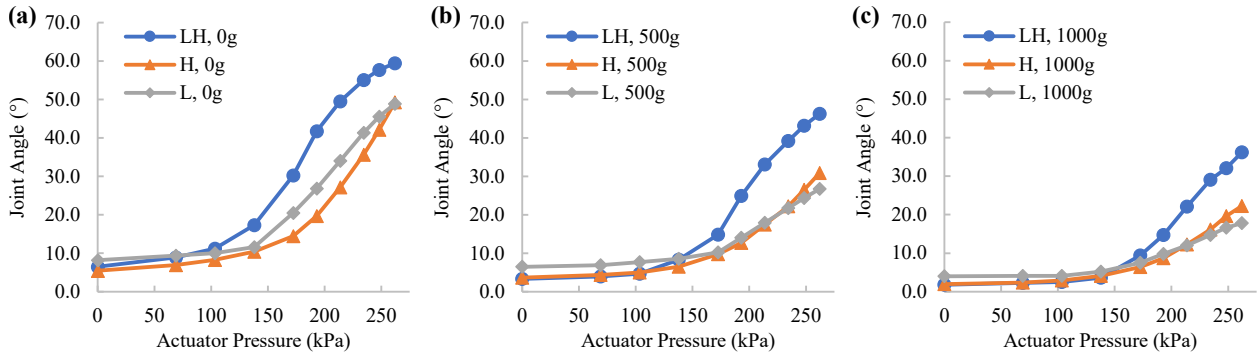


Figure 2.10: Mean joint angle under increasing pressure for three actuator configurations under three different loads. $n = 3$ for all tests.

From these results, we see that, as with stiffness modulation, a combined linear-helical

actuator configuration outperformed linear and helical actuators when used individually. This result was likely due in part to reduced loading on both sets of actuators as a result of the combined activation. Reduced tension on each individual actuator allowed for greater contraction ratio at a given pressure, which in turn produced a greater joint angle.

2.7 Conclusions

The purpose of this study was to characterize the stiffness and actuation provided by helical and linear FREE actuators when used on a single joint. Our results demonstrated that nested linear actuators have higher contraction ratios and therefore produce greater angular displacement than equal-length actuators connected in parallel, and that stiffness from helical actuators increases with pressure and number of parallel actuators. We found that for both stiffness and actuation, a combined linear-helical actuator configuration considerably outperformed either actuation group when used alone. Future work will include revising the linear nested actuator design to produce greater angular displacement, up to joint angles of 90° or more. We are also developing an analytical model for predicting stiffness from helical and linear actuators, which will allow for optimization of actuator configurations as well as provide a step toward electronic control of a FREE-actuated joint.

Chapter 3

Characterizing Architectures of Soft Pneumatic Actuators for a Cable-Driven Shoulder Exoskeleton ¹

Abstract

Low weight and innate compliance make soft pneumatic actuators an attractive method for actuating wearable robots. Performance of soft pneumatic actuators can be tailored to an application by combining them in novel architectures. We modeled and constructed nested linear and pennate architectures using fiber-reinforced elastomeric enclosures (FREEs) with identical manufacturing parameters and total effective lengths to compare their suitability for a cable-driven exoskeleton for augmenting shoulder flexion. We determined actuator performance requirements using a static model for the transmission of actuator forces to the upper arm via Bowden cables. We experimentally characterized the architectures by measuring their force-displacement curves at a range of pressures, yielding greater force and displacement from the nested architecture in the domain required by our exoskeleton. Results

¹This work has been published in [16].

also indicated a force threshold above which the pennate structure produced greater force at any given displacement. We validated the nested linear architecture using a prototype exoskeleton installed on a passive mannequin. Measured joint angles at varying pressures were close to predicted values, adjusted for measured losses due to cable anchor movement.

3.1 Introduction

Soft robotic actuators offer advantages over their traditional rigid counterparts, including lower weight, fewer components, versatility in movement, and increased safety for human interaction [1], [2]. Pneumatic soft actuators including bag-type actuators and pneumatic artificial muscles (PAMs) have been used to drive revolute joints in various robotic applications for their simplicity and high strength-to-weight ratio [14], [37], [40], [43]. Soft pneumatic devices require an air source in the form of a portable compressed air tank or a tethered air supply. Their innate compliance and low weight make pneumatic soft actuators well-suited for use in exoskeletons. Bowden cables are a popular means of force transmission in soft exoskeletons as they allow actuators to be located away from the joint they are actuating [44]–[47].

Our work focuses on a class of soft pneumatic actuator called fiber-reinforced elastomeric enclosures (FREEs) [3]. FREEs utilize networks of inextensible fibers wrapped around elastomer tubing to control deformation when the tubing is pressurized, producing movement that is determined by the fiber angles with respect to the FREE’s longitudinal axis. The FREEs used here have symmetric fiber angles, causing them to contract under pressure like a traditional PAM. FREEs can be arranged in novel architectures to further customize force and stroke output [12], [13].

In this study, we evaluate the feasibility of two different FREE architectures (Fig. 3.1) for actuation of a shoulder exoskeleton device (Fig. 3.2). The exoskeleton is intended to reduce the risk of work-related musculoskeletal disorders (WMSDs) of the shoulder by reducing joint

and muscle loads [48]. Tasks requiring prolonged elevation of the upper arms increase the risk for WMSDs [49]. Shoulder and neck WMSDs are common among working populations, especially those undertaking repetitive tasks, and have a lasting impact on affected individuals [49]–[52].

We will combine the strength and compliance of FREEs with the convenience of Bowden cables to reduce the net torque on the shoulder during tasks requiring elevated upper arms. The exoskeleton prototype will provide a supplemental flexion torque about the shoulder to offset the extension torque from the weight of arm. Our primary goal is to characterize two different architectures of FREEs and assess their performance in the context of our exoskeleton application.

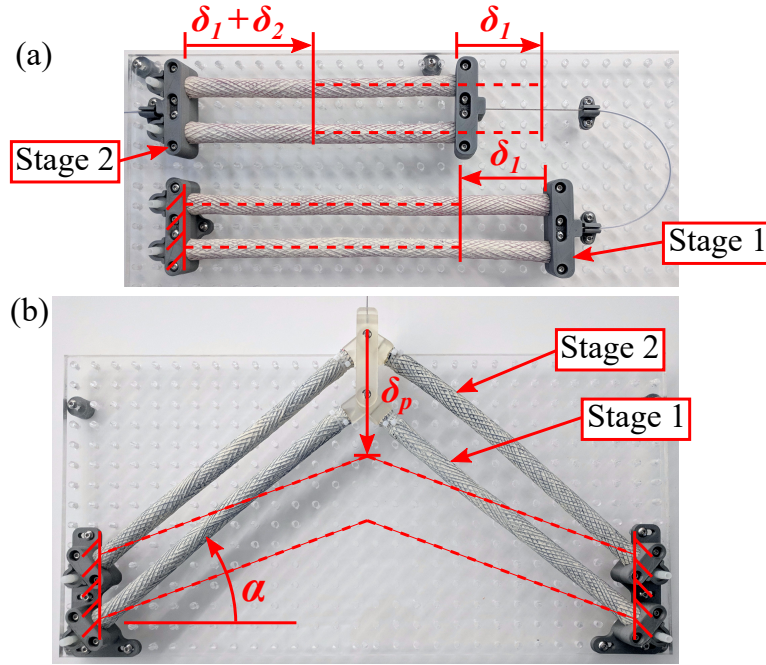


Figure 3.1: (a) Nested linear and (b) pennate FREE architectures. δ_1 , δ_2 , and δ_p represent the contraction displacements of the first nested stage, second nested stage, and pennate architecture, respectively.

3.2 FREE Architectures

3.2.1 Nested Linear Architecture

Nested linear FREE actuators use multiple contracting FREEs connected in series and in parallel to improve the stroke and force output compared to equal-length parallel actuators. Previously, nested linear actuators have been constructed using stiffening FREEs or rigid posts in compression to transfer displacement and force between the serial stages [12]. The nested actuator presented here instead uses a Bowden cable to transfer the load in tension, maintaining compliance between the two ends of the actuator assembly without the possibility of buckling (Fig. 3.1a). One end of the first stage is fixed, while the other end is free to displace axially. The force and displacement from the first stage is transferred to the second stage via a Bowden cable with fixed anchors axially aligned with each nested stage, allowing the second stage to be positioned independently of the first stage. For this study, the second stage is positioned parallel to the first stage to minimize the footprint of the actuator assembly.

We used the energy-based model presented in [24] to predict the force and contraction displacement of the nested actuators. This numerical model involves a constrained maximization of the FREEs' volume using calculus of variations. We used the model to construct an estimation of the force-displacement curve for individual FREEs from the initial length to the fully contracted position for various pressures. We modelled the nested actuator as a single FREE with length equal to the total effective length of the nested architecture, multiplying the force by the number of parallel FREEs in each stage. This approximation ignores any effects from friction in the Bowden cable connecting the two stages.

3.2.2 Pennate Architecture

Pennate FREE actuators, first proposed in [13], are inspired by the bipennate fiber arrangement in some biological muscles. The pennate architecture consists of symmetrical

arrangements of FREEs at some angle of inclination α from the horizontal, joined at a vertical centerline and fixed at their lateral ends (Fig. 3.1b). Multiple parallel FREEs can be stacked to increase the force output of the actuator. A pennate architecture has the maximum possible stroke when its FREEs have $\alpha = 0^\circ$ at maximum pressure. The critical unpressurized α_c corresponding to maximum stroke is dependent on the contraction ratio, and therefore the fiber angle of the individual FREEs. As α increases, the actuator tends to output more force and less displacement.

Rather than use the pseudo-rigid body model presented in [13], we constructed a simplified model of the pennate architecture with pin joints at both ends of each FREE. In doing so, we ignored effects from bending forces that are present in FREEs with fixed ends. We treated the architecture as a linkage with identical contracting links and a single degree of freedom. Static analysis of this simplified model allowed us to use only the force and displacement from the individual FREE model presented in [24] to estimate the output from the whole architecture.

3.2.3 Design Considerations

We characterized each FREE architecture by measuring its force-displacement curve over the full range of possible operating pressures, up to 350 kPa for the latex FREEs used here. In addition to the maximum stroke and force output, an important performance metric of the FREE architectures is the stiffness of the actuator. This paper defines positive displacement in the direction of contraction, so stiffness here is equal to the negative slope of the force-displacement curve.

Since the chosen application is a wearable device, there are also geometric and ergonomic considerations when comparing these two architectures. The overall footprint of the actuator assembly must be kept to a minimum due to the limited amount of space available on the exoskeleton. As tested, the nested actuator was the more compact of the two architectures. Reaction forces at fixed ends of the actuator can affect user comfort, particularly in the case

of a soft exoskeleton without a rigid frame to absorb the loads.

The direction of the output cable impacts force transmission. While the cable can be redirected using cable sheaths, minimizing the overall bend angle of the cable reduces friction inside the sheath, improving force transmission efficiency [47]. With the architectures as shown in Fig. 3.1 affixed to the user’s back, the nested actuator can be positioned with the output cable aligned with the longitudinal axis, pointing up over the user’s shoulder. The pennate architecture would have an output cable aligned with the mediolateral axis, pointing laterally near the user’s elbow.

3.2.4 FREE Parameters

The FREEs used in both architectures for this study are identical in diameter, wall thickness, and fiber angle (Table 3.1). To isolate the effects of architecture on performance, we held the total effective FREE length L_e for each architecture constant. Effective length represents the length of FREE that inflates during pressurization, and does not include any material enclosed by pneumatic fittings. The chosen FREE diameter and lengths maximize force and stroke within the architecture footprint allowed by the exoskeleton. For the pennate architecture, we selected α corresponding to the maximum stroke as described in Section 3.2.2.

Table 3.1: FREE Actuator Parameters

	Nested Linear	Pennate
Fiber Angles	$\pm 20^\circ$	$\pm 20^\circ$
Inner Diameter	12.7 mm	12.7 mm
Wall Thickness	1.6 mm	1.6 mm
Stage 1 L_e	334 mm	290 mm
Stage 2 L_e	246 mm	290 mm
Angle of Inclination (α)	-	40°

3.3 Shoulder Exoskeleton

3.3.1 Description of Exoskeleton Concept

The prototype shoulder exoskeleton device is constructed from a modified lumbar support brace (Miami Lumbar Proteo, Össur), consisting of a stiff aluminum strut and adjustable waist and shoulder straps (Fig. 3.2). The brace provides a comfortable and secure base for the actuators and Bowden cable anchors. The aluminum strut runs along the center of the back brace, protecting the spine from reaction forces induced by the actuators when they are pressurized.

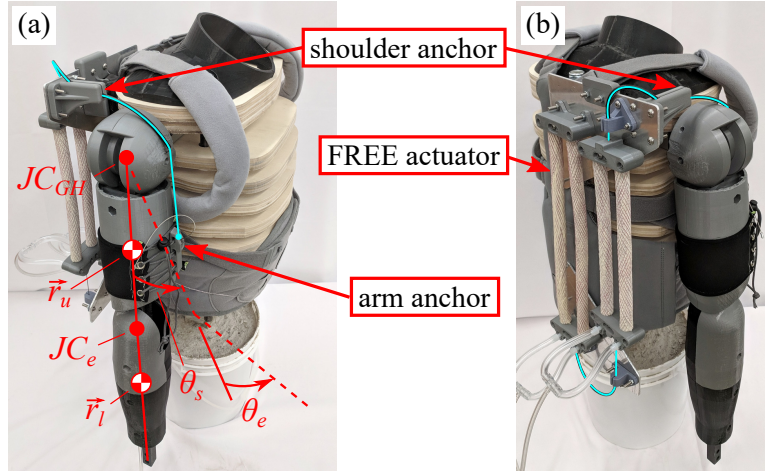


Figure 3.2: (a) Front and (b) rear view of the shoulder exoskeleton prototype with nested linear actuators, installed on a passive mannequin. JC_{GH} and JC_e are the GH and elbow joint centers. Cable path is shown in cyan.

Two aluminum plates attached to the strut serve as mounting panels for the FREE actuators and cable anchors. The shoulder anchor is cantilevered from the upper plate over the user's shoulder. The arm anchor is attached to a neoprene arm strap, worn on the upper arm near the elbow and secured with a nylon cord and a spring-loaded cord lock. The neoprene strap is lined with a soft foam bandage fabric (SuperWrap, FabriFoam), providing a comfortable but high-friction interface between the arm anchor and the user's arm.

The output cable from the FREE actuator enters a cable sheath, secured to the upper aluminum plate with a cable anchor. The other end of the cable sheath is fixed to the

shoulder anchor. The cable exits the shoulder anchor and terminates at the arm anchor. Since the cable is in contact with the shoulder at small joint angles, an open-ended cable sheath surrounds the cable for part of the span between the shoulder anchor and the arm anchor to protect the shoulder from chafing as the cable displaces. Additional padding under the sheath would prevent discomfort from cable pressure during human testing or implementation.

3.3.2 Actuator Performance Requirements

The exoskeleton offsets the extension torque about the shoulder produced by the mass of the arm with a flexion torque from a FREE actuator acting on the arm in tension via a Bowden cable. To estimate the torque from the arm, we used a static model of the arm and shoulder of a 50th percentile American male. The shoulder is approximated as a ball joint located at the joint center of the glenohumeral (GH) joint. Arm segments are considered point masses located at the centers of mass for each segment. Joint center locations and segment masses are as described in [53]. All segment centers of mass and joint centers are assumed to be in the same sagittal plane, and the wrist is assumed to be in a neutral position. We can define a coordinate system with the origin at the GH joint center, positive x in the anterior direction, positive y in the upward longitudinal direction, and positive z in the lateral direction. Extension torque $\vec{\tau}_a$ about the shoulder from the arm's weight is then

$$\vec{\tau}_a(\theta_s, \theta_e) = \vec{r}_u(\theta_s) \times \vec{w}_u + \vec{r}_l(\theta_s, \theta_e) \times \vec{w}_l$$

where θ_s and θ_e are the shoulder and elbow flexion angles. $\vec{r}_u(\theta_s)$ and $\vec{r}_l(\theta_s, \theta_e)$ are the positions of the centers of mass of the upper arm and the lower arm, including hand (Fig. 3.2a). $\vec{w}_u$ and $\vec{w}_l$ are the force vectors from the weight of the upper arm and the lower arm.

If we take $\theta_e = 0^\circ$, we can express τ_a as a scalar:

$$\tau_a(\theta_s) = m_a g r_a \sin(\theta_s) \tag{3.1}$$

where m_a is the combined mass of the upper arm, forearm, and hand, g is the acceleration due to gravity, and r_a is the position along the arm of the combined center of mass.

The model for cable force and displacement is based on the model derived in [44]. The force required to produce a flexion torque equal in magnitude to τ_a depends on the locations of the shoulder and arm anchors relative to the GH joint center.

Positions of the shoulder and arm anchors are represented by $\vec{a}_s$ and $\vec{a}_a(\theta_s)$, respectively. For large θ_s , the cable extends directly from the arm anchor to the shoulder anchor. Below some joint angle θ_c , dependent on $\vec{a}_s$ and $\vec{a}_a(0)$, the cable may be in contact with the shoulder. If both anchors are in a shared sagittal plane with the GH joint center, i.e. if $\vec{a}_s \cdot \hat{k} = \vec{a}_a \cdot \hat{k} = 0$ with $\hat{k}$ as a unit vector in the z -direction, we can model the cable path by approximating the surface of the shoulder as a circle, centered on the origin with radius r_s . The length of the cable between the shoulder anchor and the arm anchor is then

$$L(\theta_s) = \begin{cases} \|\vec{a}_s - \vec{s}_s\| + \|\vec{s}_s - \vec{a}_a\| + r_s\phi, & \theta_s < \theta_c \\ \|\vec{a}_s - \vec{a}_a\|, & \theta_s \geq \theta_c \end{cases} \quad (3.2)$$

where $\vec{s}_s$ and $\vec{s}_a(\theta_s)$ are the contact points from the shoulder and arm anchors, respectively, and ϕ is the contact angle around the shoulder (Fig. 3.3a). Cable displacement is the change in cable length between these two anchors:

$$\delta(\theta_s) = L(0) - L(\theta_s) \quad (3.3)$$

Torque about the GH joint generated by the cable acting on the arm is

$$\vec{\tau}(\theta_s) = \vec{a}_a \times f\hat{c}(\theta_s) \quad (3.4)$$

where f is the magnitude of the cable force and $\hat{c}(\theta_s)$ is the normalized cable direction as it

leaves the arm anchor:

$$\hat{c}(\theta_s) = \frac{\vec{c}(\theta_s)}{\|\vec{c}(\theta_s)\|}$$

$$\text{with } \vec{c}(\theta_s) = \begin{cases} \vec{s}_a - \vec{a}_a, & \theta_s < \theta_c \\ \vec{a}_s - \vec{a}_a, & \theta_s \geq \theta_c \end{cases}$$

Since both anchors share a sagittal plane with the GH joint center, and because we are interested in producing a flexion torque, we can isolate the flexion component of the torque vector in (3.4) to determine the cable force f_{req} required to counteract the extension torque from the weight of the arm:

$$f_{req}(\theta_s) = \frac{\tau_a}{(\vec{a}_a \times \hat{c}) \cdot \hat{k}} \quad (3.5)$$

From (3.3) and (3.5), we obtain force-displacement curves for supporting a 50th percentile male's arm for arbitrary shoulder anchor positions in the x-y plane (Fig. 3.3b). The simulated angles $0^\circ \leq \theta_s \leq 120^\circ$ represent the functional range of motion for shoulder flexion [54]. Based on this model, arm anchor position has little effect on requirement curves. After simulating shoulder anchor positions with $-60 \text{ mm} \leq x \leq 60 \text{ mm}$ and $20 \text{ mm} \leq y \leq 130 \text{ mm}$, we selected $(-30 \text{ mm}, 70 \text{ mm})$ for its relatively small peak force and displacement requirements: 168 N and 130 mm, respectively.

3.3.3 Actuator Architecture Selection

Plotting the force requirement curve with output curves from the FREE architectures at multiple pressures lets us predict whether an architecture will generate enough force and stroke to support the arm of the exoskeleton's user (Fig. 3.4). The output curves for FREE actuators are dependent on pressure. Intersections of the output curves and the force requirement curve represent equilibrium points between the weight of the arm and the force generated by the FREEs.

The actuator stiffness, defined by the slope of the force-displacement curve, determines the resistance felt by the user when they move their arm while the pressure is held constant.

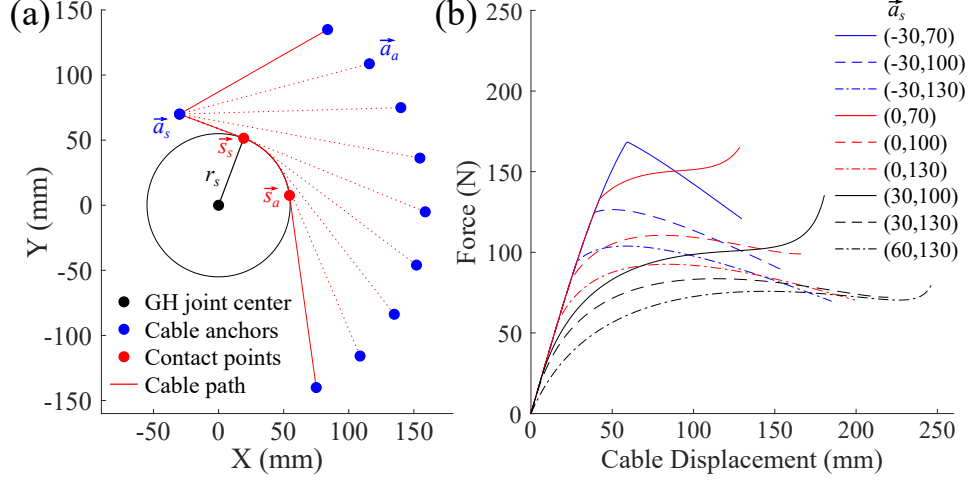


Figure 3.3: (a) Cable path for $0^\circ \leq \theta_s \leq 120^\circ$ with shoulder anchor located at $(-30 \text{ mm}, 70 \text{ mm})$ and arm anchor at $(75 \text{ mm}, -140 \text{ mm})$ when $\theta_s = 0^\circ$. (b) Theoretical force-displacement curves for various shoulder anchor positions in a shared sagittal plane with the GH joint.

High stiffness translates to a small range of supported motion at a constant pressure, since the force will increase sharply as the user extends their arm from the equilibrium position, and decrease sharply if the user moves their arm in flexion. By modulating the pressure to the FREEs as the user flexes or extends their upper arm, we can support the arm in any position on the force requirement curve.

Based on our theoretical models, the nested linear architecture is better able to traverse requirement curves with moderate force and high displacement, such as for supporting the weight of a 50th percentile male's arm through a large range of motion (Fig. 3.4). However, for each pressure there is an intersection between the curves of the two architectures, above which the pennate architecture produces higher forces than the nested architecture at a given displacement. The pennate architecture is therefore better suited for applications requiring high force and small displacement, such as providing a flexion torque at the shoulder for a lifting task with a small range of motion and flexed elbows. The pennate architecture produces also greater predicted maximum stroke, usable only if forces are kept to a minimum (Fig. 3.4).

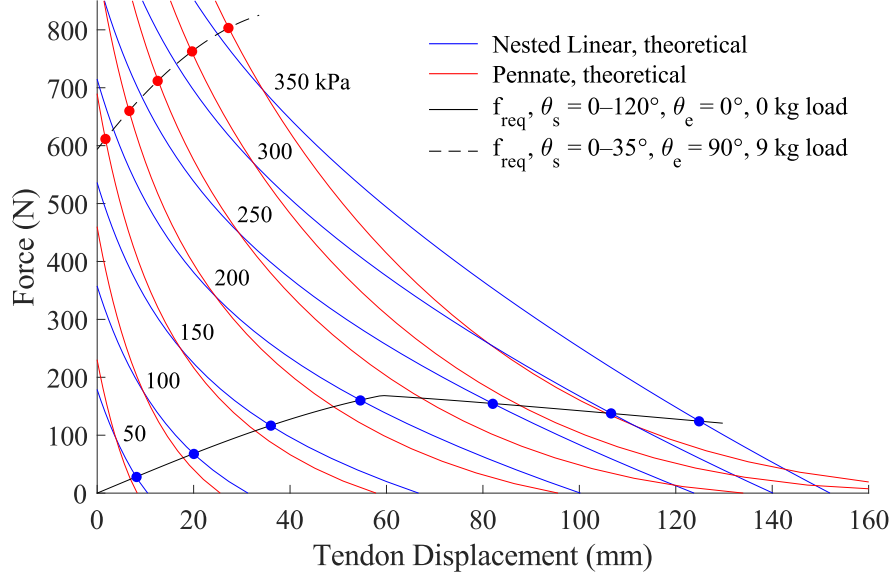


Figure 3.4: Theoretical force-displacement curves for nested linear and pennate FREE architectures overlaid with two arm requirement curves. Red intersection points represent equilibrium between the force from the pennate actuator and the weight of the arm when the elbow is flexed and there is a 9 kg load on the hand. Blue intersection points represent equilibrium between the nested linear actuator and the unloaded arm when the elbow is extended.

3.4 Experimental Methods

3.4.1 Architecture Force-Displacement Test Methods

We built a custom apparatus to measure force-displacement curves for each architecture. The test setup consists of two assemblies: the actuator plate (Fig. 3.1) and the measurement unit (Fig. 3.5). The measurement unit includes a 3D printed linear stage, positioned with a NEMA 17 stepper motor (SY42STH38-1684A, SOYO) and an 8 mm lead screw. An axial load cell (LCM300, FUTEK) attached to the stage is connected to an adjustable Bowden cable that runs parallel to the lead screw. At the end of the lead screw distal to the stepper motor, the cable enters a PTFE-lined, steel reinforced cable sheath (M-System, Shimano), through which the cable is attached to the actuator system. The Bowden cable transmits tensile force from the FREE actuator on the actuator plate to the load cell on the measurement unit.

An Arduino Uno microcontroller board controls the measurement unit and executes the

test protocol. The controller operates the motor via a stepper motor driver (ROB-09402, SparkFun). Linear stage displacement is calculated from the number of steps the motor has taken during testing. The load cell signal is amplified by a handheld sensor display (IHH500, FUTEK) before being sent to the Arduino. Air pressure is set using a manual regulator and a digital pressure gauge (K4517N14160, Parker), and toggled with a solenoid valve (VUVG-L10, Festo) controlled by the Arduino.

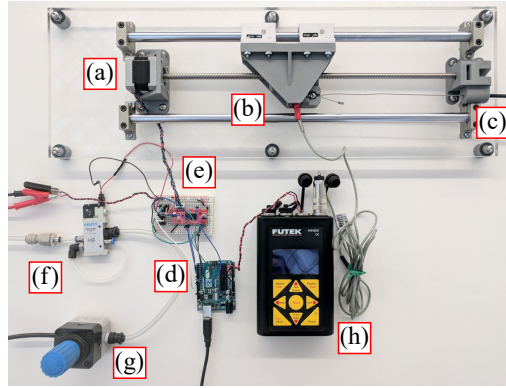


Figure 3.5: Measurement unit for force-displacement testing. A stepper motor with lead screw (a) displaces the linear stage with an axial load cell (b) attached to the input cable (c) from the FREEs. A microcontroller (d) drives the motor via a motor driver board (e) and toggles air pressure via a solenoid valve (f). Pressure is controlled with a manual pressure regulator (g), and load cell readings are captured with a handheld sensor display (h).

We measured force-displacement curves for each FREE architecture at 50 kPa increments up to 350 kPa. For each set of trials, we began by zeroing the linear stage and finding the zero-force positions for the pressurized and unpressurized states by displacing the stage until the load cell registered a nominal 10 N pretension. Starting from the pressurized zero-force position, we pressurized the FREEs, recorded the tensile load, depressurized, and displaced the stage toward the unpressurized zero-force position by a small increment. Each trial, this cycle repeated until the stage reached the unpressurized zero-force position or the load reached 350 N.

3.4.2 Exoskeleton Validation Methods

We validated the exoskeleton with a passive articulated upper-body mannequin and a custom pressure control board. We built the mannequin using relevant anthropometric data for a 50th percentile American Caucasian male [53]. The mannequin simulates the shape and size of a human torso, upper arm, and lower arm. Joint center locations, arm segment masses, and arm segment center of mass locations are consistent with the values presented in [53] (Table 3.2). The shoulder is approximated as a universal joint, and the elbow as a single-axis revolute joint. The torso is constructed from layered cross sections of plywood on an aluminum frame. The arm consists of a steel frame with a 3D printed shell.

The pressure control board consists of a pneumatic proportional valve (LS-V05S, Enfield Technologies), a pressure sensor (SP5, Enfield Technologies) and a data acquisition device (Q8-USB, Quanser). A LabVIEW-based PID controller regulates a pressurized air supply to actuate the FREEs.

After outfitting the mannequin with our exoskeleton prototype, we placed markers on the GH joint center and the centerline of the upper arm for measuring the shoulder flexion angle. We also placed a marker on the output stage of the actuator to measure cable displacement. A camera aligned with the mediolateral axis of the mannequin and supported by a tripod captured video of each test, which we processed with video tracking software. An LED allowed us to synchronize the angular displacement with the pressure readings from the control board's pressure sensor.

We validated our cable displacement model by manually displacing the cable near the actuator output and recording the resultant flexion angle. This test allowed us to isolate losses in stroke due to shoulder and arm anchor movement.

To validate the equilibrium points between the actuator force and the arm weight, we used two protocols measuring flexion angle of the mannequin with the FREE actuators pressurized in 50 kPa increments, and a maximum pressure of 300 kPa. For one test, we increased the pressure from zero starting with the arm in a neutral position. For the other,

Table 3.2: Mannequin parameters, 50th percentile male [53]

Upper Arm Mass	m_u	1.98 kg
Lower Arm Mass	m_l	1.63 kg
Upper Arm Center of Mass	$ \vec{r}_u $	163 mm
Lower Arm Center of Mass	$ \vec{r}_l $	182 mm
Upper Arm Segment Length	l_u	269 mm

we decreased the pressure from 300 kPa with the arm starting in a flexed position. Separating these protocols let us identify directional effects from friction in the Bowden cables.

3.5 Results and Discussion

3.5.1 Architecture Force-Displacement Results

Experimental force-displacement curves for the nested linear architecture closely matched those predicted by the model in Section 3.2 for pressures above 150 kPa (Fig. 3.6a). The theoretical model overestimated forces for low displacements for pressures up to 200 kPa, resembling discrepancies for individual contracting FREEs measured in [24], where the error is attributed to fiber stiffness not included in the model. However, there was a larger error at small displacements between the experimental and theoretical curves for the nested architecture than for the individual actuators in [24]. Static friction in the Bowden cable connecting the two linear stages likely reduces the transmitted output force from the fixed stage. Using rigid compression stages as in [12] would eliminate frictional losses at the cost of compliance.

Force-displacement results for the pennate architecture also resembled predicted curves from Section 3.2 for pressures above 150 kPa (Fig. 3.6b). Experimental results again showed lower output force than the model at low pressures. Additionally, the pennate results had reduced maximum stroke for tests above 200 kPa compared to theoretical values. This error is due to the pin joint assumption in the simplified pennate model, which does not account for bending forces in the FREEs when α approaches 0° at large actuator displacements. A

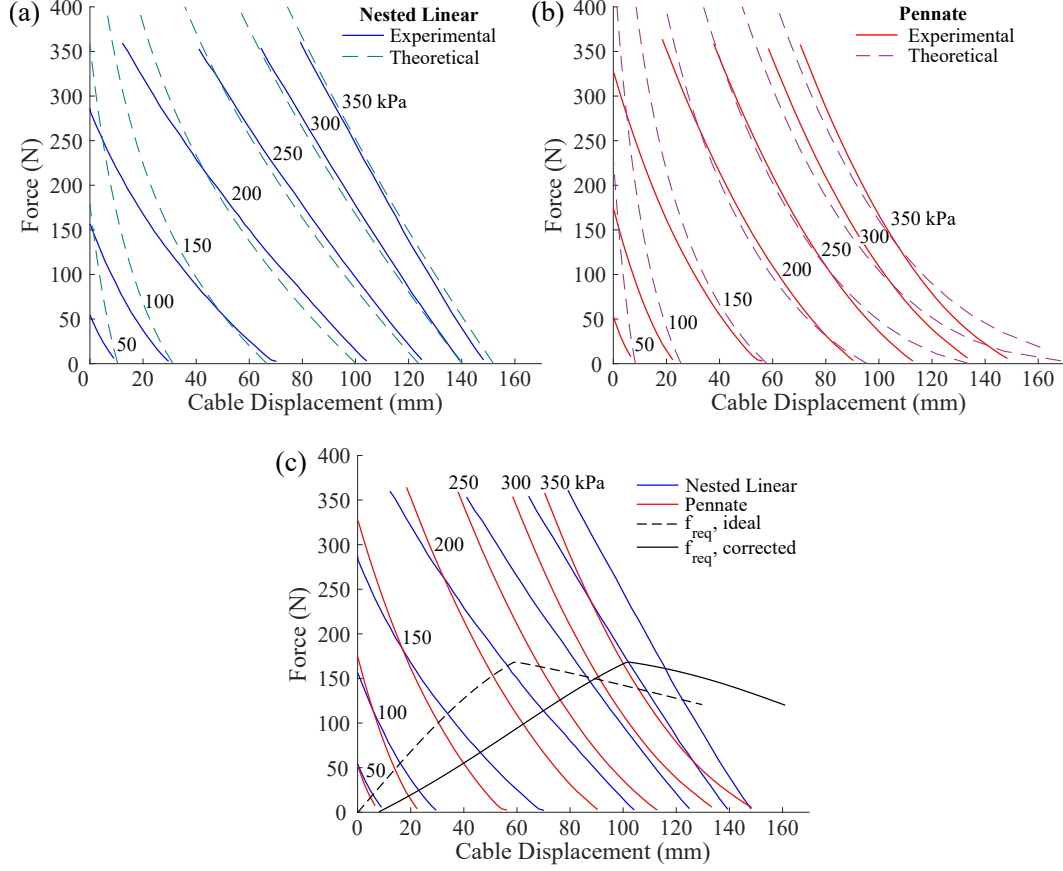


Figure 3.6: (a) Nested linear and (b) pennate architecture theoretical and mean experimental force-displacement curves at 50 kPa increments. (c) Experimental force-displacement curves for both architectures with ideal and corrected arm requirement curves. $n = 5$ for all tests.

pennate architecture with pin joints at both ends of each FREE would more closely match the force-displacement curve predicted by the simplified model.

As predicted in Fig. 3.4, the nested architecture achieved greater force at any given displacement compared to the pennate architecture in the region specified by our requirement curve from Section 3.3 (Fig. 3.6c). Force-displacement results indicated that in an ideal case, the nested architecture can provide sufficient force and stroke for nearly the entire 120° range of motion when pressure is modulated between 0 and 350 kPa. The pennate architecture can only offset the weight of the arm up to a cable displacement of 110 mm, corresponding to a flexion angle of 105° . These results led us to choose the nested actuator for our exoskeleton prototype. It is important to note that the attainable range of motion for either architecture

is reduced once we correct for losses due to cable anchor movement.

The nested architecture produced more linear force-displacement curves, indicating less variation in stiffness than the pennate architecture. The pennate architecture had high stiffness at small displacements that decreased as the FREEs approached full contraction.

3.5.2 Exoskeleton Validation Results

The manual cable displacement test revealed losses in stroke due to shoulder and arm anchor movement that resulted in a discrepancy between theoretical and experimental flexion angles. Using polynomial fits of the manual cable displacement results and the theoretical cable displacement described in Section 3.3.2 (Fig. 3.7a), we mapped the theoretical cable displacement to the experimental displacement and produced a corrected force requirement curve to include losses due to anchor movement (Fig. 3.6c).

Results from the exoskeleton validation testing highlight losses that are not accounted for in the theoretical model (Fig. 3.7b). At 300 kPa, the increasing pressure trials produced joint angles 32° less than those produced by the decreasing pressure trials. This discrepancy is likely caused by friction in the Bowden cables between the two linear stages and extending from the actuator output to the shoulder anchor. As the pressure increases from 0 kPa and the arm moves in flexion, friction in each cable sheath opposes movement of the cable in the positive displacement direction, toward the actuator. Conversely, as the pressure decreases from 300 kPa with the arm moving in extension, friction opposes the cable as it displaces negatively to allow the arm to extend.

The effects from friction described above suggest that the predicted equilibrium joint angle at a given pressure should fall between the joint angles found during the increasing and decreasing pressure trials. While this proved to be the case for pressures up to 200 kPa, and marginally for 250 kPa, the theoretical curve overpredicted joint angle for the highest test pressure, 300 kPa. This error was likely caused by deformation of the back plates that the actuators are attached to. Any reduction in distance between the actuator end points results

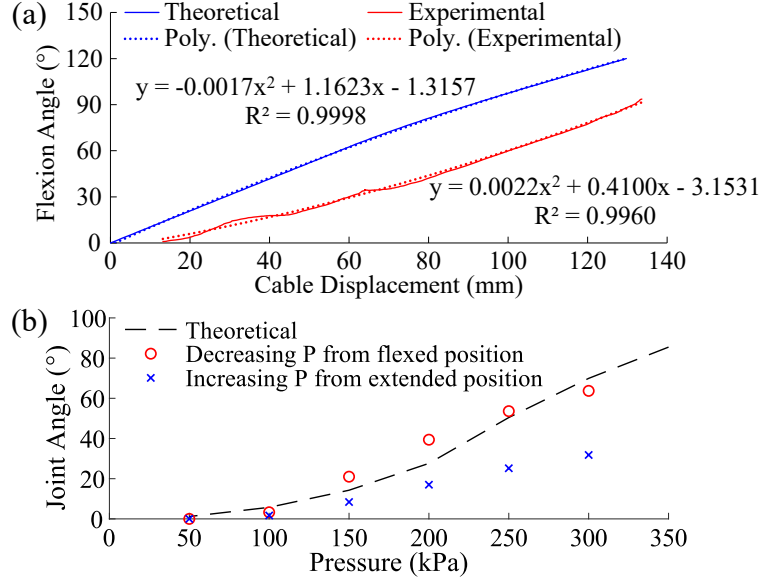


Figure 3.7: Exoskeleton validation results. (a) Theoretical and experimental joint angle vs. cable displacement, with respective polynomial fit lines. (b) Theoretical and mean experimental joint angle for the nested linear architecture installed on the shoulder exoskeleton prototype at 50 kPa increments. $n = 3$ for both tests.

in a reduction in stroke and therefore a decrease in joint angle. This deformation is more prevalent at high pressures due to higher actuator forces.

A more rigid exoskeleton frame would reduce unwanted deformation, but would also add weight to the device. An actuator with a larger stroke could compensate for deformation losses from the back plate and the anchors as well as frictional losses from the Bowden cable without sacrificing compliance. Functionally, the decreasing pressure trials are more relevant to the intended use of the exoskeleton, since the device is meant to augment healthy users rather than replace lost muscle function as in a rehabilitation device.

3.6 Conclusions

Our results demonstrate that novel architectures of FREEs are a means to alter the “gear ratio” of an actuator assembly without changing the structure and properties of the FREEs themselves. This is a valuable property of FREE actuators to designers seeking to tailor soft actuators to a specific application because it provides an avenue for actuator customization

beyond fundamental manufacturing parameters. Nested linear actuators were better suited for moderate stroke and moderate force applications such as our shoulder exoskeleton concept. Pennate actuators proved to be better suited for applications requiring large forces with small displacements, though stroke performance could be improved for dual pin joint pennate structures.

To further develop the shoulder exoskeleton design, we would need to make additional considerations for user comfort, such as weight, contact pressures, and shear forces at the body interfaces. Human subject testing would be needed to validate the device as a means for reducing risk of WMSDs. Common methods include comparison of muscle input via surface electromyography and estimation of joint loads via motion capture and inverse dynamics.

Future work will include further development of actuator architectures, including generalized actuator topologies for broader performance customization.

Chapter 4

A Soft Robotic Simulator for Transseptal Puncture Training

Abstract

Transseptal puncture (TP) is the technique used to access the left atrium of the heart from the right atrium via the interatrial septum in increasingly common catheter-based procedures such as atrial fibrillation ablation. Through repetition, electrophysiologists and interventional cardiologists experienced in TP develop manual skills to manipulate the transseptal catheter assembly inside the right atrium to their target on the fossa ovalis. Cardiology fellows and cardiologists that are new to TP currently train on actual patients to develop this skill, resulting in increased risk of dangerous complications. To create low-risk training opportunities for new TP operators, we have developed a Soft Active Transseptal Puncture Simulator (SATPS), designed to match the dynamics, static response, and visualization of the heart during TP. The SATPS includes three main subsystems: (i) An anatomically accurate soft right atrium, fitted with pneumatic artificial muscles, mimicks the dynamics of the heart felt by the operator through the catheter assembly. (ii) A replaceable, puncturable fossa ovalis simulates the tissue properties of the real fossa to provide accurate force feedback during

tenting and puncture. (iii) A simulated intracardiac echocardiography environment gives the user live visual feedback representative of an ultrasound monitor during an actual TP procedure. Subsystems of the SATPS were validated with benchtop tests. Face and content validity were established through evaluation by experienced clinicians. Passive and active actuation states were deemed suitable for simulating different patient cardiac conditions. Participants rated the SATPS as realistic and useful for training fellows in TP.

4.1 Introduction

Transseptal puncture (TP) is a common procedure used by cardiac electrophysiologists and interventional cardiologists, wherein the left atrium of the heart is accessed by puncturing the interatrial septum from the right atrium. The procedure is conducted using a catheter assembly consisting of a sheath, dilator, and Brockenbrough needle inserted into the femoral vein and passed through the inferior vena cava into the right atrium. The most prevalent treatment that uses TP is atrial fibrillation (AF) ablation. AF cases are rising, expected to affect 6–12 million people in the US by 2050 [55]. Thus, the use of AF ablation is expected to also increase and more TP operators will need to be trained. TP is also used in an increasing number of other cardiac procedures, such as left atrial appendage occlusion and mitral valve repair [56].

New operators, i.e., electrophysiology fellows, interventional cardiology fellows, and cardiologists without TP experience, train in TP on live patients under the careful observation of experienced mentors. Until they can perform with the efficiency and accuracy of veteran operators, trainees conduct TP with higher failure rates and increased procedural time [57], [58]. Skilled operators, i.e. electrophysiologists and interventional cardiologists experienced in TP, develop the ability to navigate the atrium by judging the force felt through their catheters when in contact with the atrial wall and interatrial septum [59]. Prior to live patient training, new operators can review relevant fluoroscopy or ultrasound images, but this preparation

provides no haptic feedback, leaving them to rely on mentors and live-patient training for crucial haptic skill development. The cardiovascular workforce is aging and retiring, leaving a shortage of experienced guiding hands to pass this skill to the next generation [60]. Thus, there is a clear need for the development of a training simulator to revolutionize the way TP training is conducted to reduce the need for actual patient training and to standardize the training process.

Realistic simulators for training and surgical planning must be equivalent to their biological counterparts in terms of (i) dynamics, (ii) static response, and (iii) visualization. Dynamic equivalence requires matching the time-varying muscle forces and their effect on the walls of the organ. Static response equivalence requires accurate surface-level force responses that are functions of geometry and material properties. Finally, visualization entails near-equivalent imaging such as fluoroscopy or ultrasound that help guide the operator during a procedure. Existing TP simulators from literature range from a virtual reality training environment to static anatomically correct designs with simulated transesophageal echocardiography and fluoroscopy images [61]–[63]. Commercial endovascular simulators such as the ANGIO Mentor™ and the Mentice VIST® offer simulated haptic feedback and visualization for a variety of catheter-based procedures, including transseptal puncture [64], [65]. These devices have reported significant levels of realism, but none incorporate actuation and tailored material properties in a single device.

The overarching goal of this work was to provide low-risk training opportunities to new TP operators, allowing them to develop the haptic awareness and skill of more experienced operators before performing TP on a live patient. To that end, we developed a Soft Active Transseptal Puncture Simulator (SATPS), designed to match the dynamics, static response, and visualization of the heart during TP (Fig. 4.1). Subsystems were validated with benchtop tests. System validation of the SATPS by clinicians experienced in TP addressed three key hypotheses:

- Hypothesis 1: An active TP simulator is more realistic than a passive simulator.

- Hypothesis 2: The SATPS realistically simulates the tactile and visual feedback experienced during TP.
- Hypothesis 3: The SATPS is a useful training device for transseptal puncture.

Medical simulators are commonly evaluated in terms of face validity and content validity [66]–[70]. According to McDougall, face validity establishes the realism of the simulator, and content validity is a judgement of the appropriateness of the simulator as a teaching modality [68]. Face and content validity are subjective, and usually established through input from experienced clinicians [70]. In this study, face validity was represented by Hypothesis 2 and content validity by Hypothesis 3.

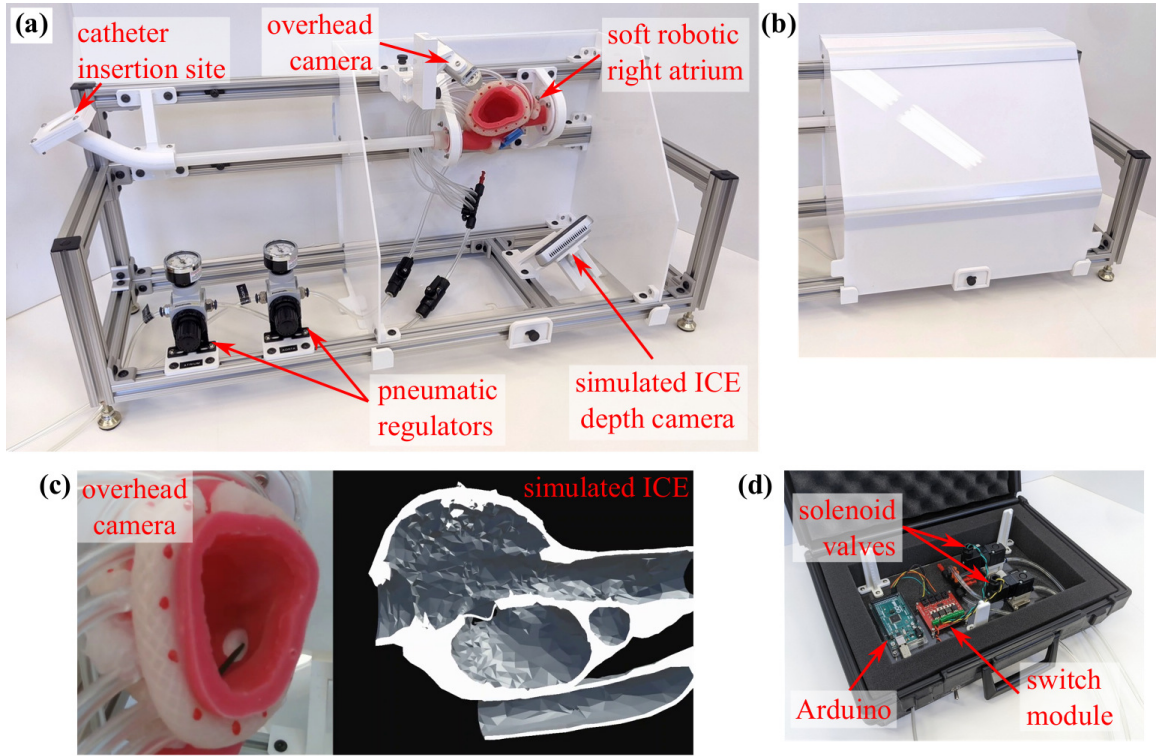


Figure 4.1: (a) Soft Active Transseptal Puncture Simulator (SATPS). (b) SATPS with its shroud in the closed position. (c) Overhead camera view and real-time simulated ICE. The laptop is not shown. (d) Pneumatic control board in insulated case. The pneumatic compressor is not shown.

4.2 System Design

4.2.1 Design Considerations

The basic sequence of events during transseptal puncture is as follows: (i) A guidewire is inserted into the femoral vein, through the inferior vena cava (IVC), past the right atrium, and into the superior vena cava (SVC). (ii) A sheath and dilator are advanced over the guidewire into the SVC, the guidewire is removed, and a Brockenbrough needle is inserted into the dilator, stopping short of the dilator tip. (iii) The catheter assembly (sheath, dilator and Brockenbrough needle) is retracted from the SVC, over the aortic mound, and onto the fossa ovalis, an elliptical membrane on the interatrial septum [71] (Fig. 4.2). (iv) Once tissue deformation from the applied force (i.e. “tenting”) is verified on an ultrasound monitor, the needle is advanced from the dilator, puncturing through the fossa and into the left atrium. Then the dilator and catheter are advanced over the needle into the left atrium, completing the TP.

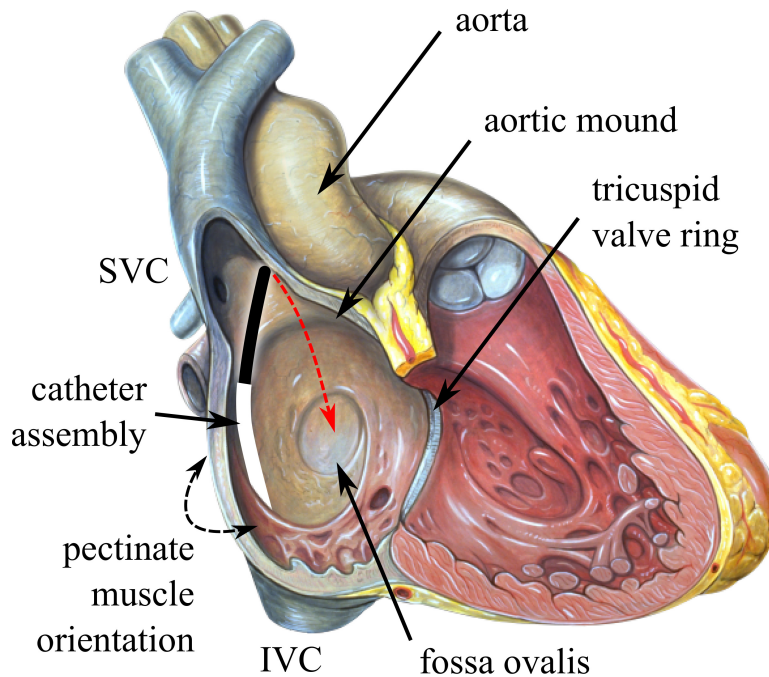


Figure 4.2: Cutaway image of the right atrium. A transseptal puncture catheter assembly is positioned in the SVC, ready for navigation to the fossa ovalis. Labels and catheter were added to the heart illustration drawn by Patrick J. Lynch, used under CC BY 2.5.

Training operators to navigate a patient’s heart with a TP catheter assembly requires consideration of three critical steps during the procedure: (i) As the catheter assembly is retracted back from the SVC into the right atrium, it encounters the pulsating aortic mound [72], [73]. The operator must avoid puncturing the aortic mound, which would lead to excessive blood loss. (ii) The operator uses a combination of visual cues and dynamic force feedback to locate the fossa ovalis. (iii) The operator applies the correct force on the catheter to produce tenting before advancing the needle to puncture the fossa ovalis. This last step usually requires force feedback and visual observation of the fossa tenting using ultrasound via intracardiac echocardiography (ICE). Historically, fluoroscopy has also been used to visualize catheter position before puncture, but reduced radiation exposure and the ability to locate specific puncture sites on the septum make ICE the preferred visualization tool for modern TP procedures. To capture these critical steps, the SATPS required systematic integration of bioinspired actuators on an anatomically accurate atrium, stretchable materials with the same properties as the fossa, and visual simulation tools.

4.2.2 System Description

The SATPS includes three main subsystems. An anatomically accurate soft right atrium incorporates compliant pneumatic actuators to mimic the dynamics of the heart felt by the operator through the catheter assembly (Fig. 4.3). A replaceable, puncturable fossa ovalis simulates the tissue properties of the real fossa to provide accurate force feedback during tenting and puncture (Fig. 4.3d). Lastly, a simulated intracardiac echocardiography environment gives the user live visual feedback of the fossa ovalis that is representative of an ultrasound monitor during the real procedure (Fig. 4.4).

Soft Robotic Right Atrium for Dynamic Equivalence

The soft right atrium was modeled from a CT heart scan (The Lynn & Arnold Irwin Advanced Perioperative Imaging Lab) and cast from silicone rubber (EcoFlex 00-50, Smooth-On Inc.)

in a 3D printed mold with a water-soluble polyvinyl alcohol (PVA) mold core (Fig. 4.3. We modified the digital heart model using 3D modeling software (Blender 2.8, Blender Foundation, Amsterdam, Netherlands) to incorporate functional elements such as a pneumatic chamber for an actuated aortic mound, a pocket for inserting a replaceable fossa ovalis, and grooves for the pneumatic actuators. After casting, a low friction silicone coating (MED-6670, NuSil) was applied to the interior of the atrium to simulate the lubricity of blood between the catheter assembly and cardiac tissue.

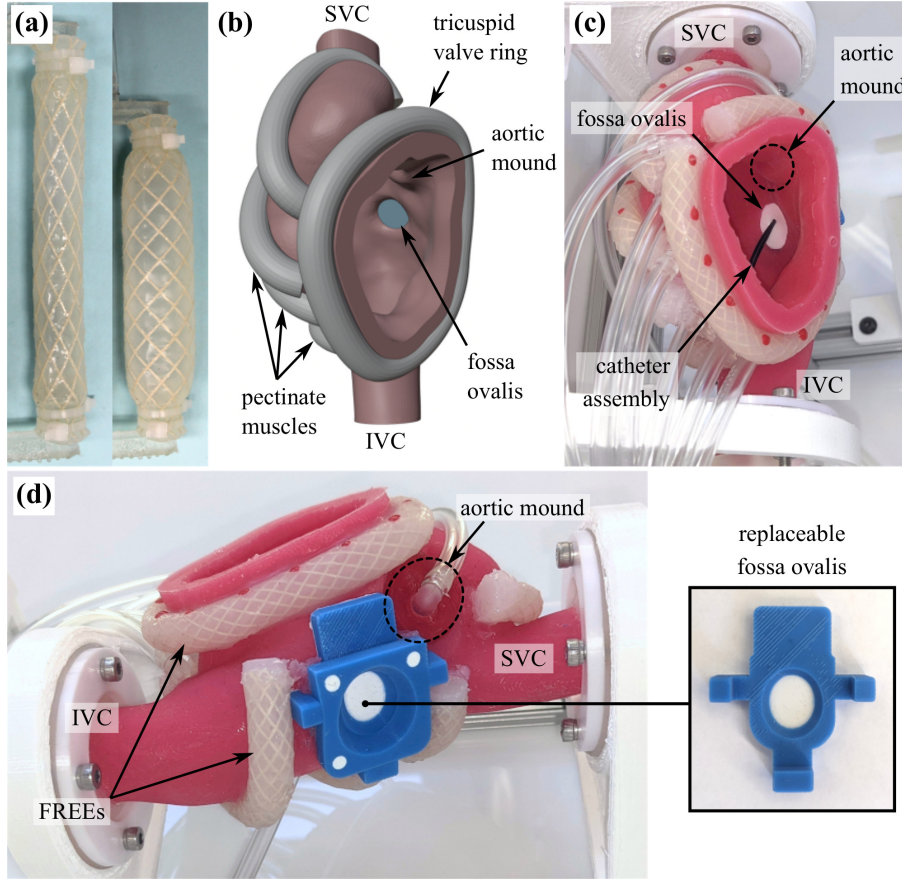


Figure 4.3: (a) Unpressurized (*left*) and pressurized (*right*) contracting FREE actuator. (b) CAD rendering of soft right atrium with FREEs representing pectinate muscles and tricuspid valve ring. (c) Cast silicone right atrium with catheter assembly in tenting position. (d) Cast silicone right atrium with replaceable fossa ovalis.

The SATPS simulates atrial contraction using soft pneumatic actuators called fiber-reinforced elastomeric enclosures (FREEs) [3]. FREEs consist of an elastomer bladder surrounded by helical networks of inextensible fibers. The angle of these fibers relative to the

axis of the FREE determines the FREE’s behavior when pressurized. The FREEs used in the SATPS have fiber angles of $\pm 30^\circ$, causing them to contract axially and expand radially when pressurized, similar to biological muscle (Fig. 4.3a). Contracting FREEs have been used previously to simulate contraction from ventricular muscles in a soft robotic device for ventricular compression [26].

Achieving realistic atrial compression in the SATPS demanded FREEs that could be wrapped around the tight radius of the atrium without buckling, necessitating the development of custom pre-curved FREEs. The actuator bladders were cast in 3D printed molds from silicone rubber (EcoFlex 00-50, Smooth-On Inc.). This silicone rubber was also used to bond the FREEs to the cast atrium. The axis of each FREE follows a three-dimensional cubic Bézier curve approximating a muscle path along the surface of the right atrium to produce realistic atrial dynamics. Specifically, our FREEs simulate the pectinate muscles and the tricuspid valve ring (Fig. 4.2 and 4.3b). The FREE bladder castings also include grooves for fibers, which are arranged in helical paths around the curved FREE axis. The pitch of the helical fiber path is equal to the pitch of a fiber in a straight FREE with the desired contraction ratio. The fibers used in the SATPS FREEs are cotton thread. We verified the FREEs’ ability to produce realistic atrial contraction by measuring the change in volume during actuation, and comparing the results to physiological values (see Section 4.3.1).

Air is supplied to the actuators by a pneumatic control system consisting of a small compressor (TC-326T, Master Airbrush), two pneumatic regulators (PPR2-N02BG-2, PneumaticPlus), two solenoid valves (VT307W-5DZ1-02T-F, SMC Corporation), a MOSFET switch module (IRF540, NOYITO), and a microcontroller board (Mega 2560, Arduino). The regulators are mounted on the frame of the SATPS to allow for pressure adjustment. All other components aside from the compressor are stored in an insulated case to reduce noise from the solenoid valves (Fig. 4.1c). The control system toggles air supply to the FREEs and the aortic mound chamber, both of which are actuated at a rate of 75 BPM.

Replaceable Fossa Ovalis for Static Response Equivalence

The SATPS atrium is fitted with a housing that holds a puncturable material that simulates the fossa ovalis (Fig. 4.3d). This component allows for continued use of the SATPS without the need to replace the entire right atrium. The fossa insert is cast from silicone rubber (Dragon Skin 10 Medium, Smooth-On Inc.), chosen to mimic the tenting stiffness and the puncture force of real fossa tissue during TP (see Section 4.3.2). Matching these properties will help new operators learn the correct force to apply during the final steps of the procedure. Application of excessive force could result in accidental puncture of the left atrial wall due to overextension of the needle into the left atrium.

Simulated Intracardiac Echocardiography for Visualization Equivalence

Transseptal puncture operators rely on intracardiac echocardiography to verify dilator position on the fossa ovalis during tenting before puncture. ICE uses a probe inserted into the right atrium along the same vascular pathway as the transseptal catheter to create a cross-sectional image of the fossa ovalis (FO) and atrial walls.

To simulate ICE imaging, the SATPS uses a depth camera (RealSense D415, Intel Corporation) to detect the deformation of the simulator’s fossa ovalis as it moves with the beating right atrium. The depth camera data are used to produce real-time simulated ultrasound images in a 3D content development software (Unity 2019, Unity Technologies) displayed on a laptop that is facing the user. The animated simulation shows a deforming virtual fossa ovalis that is embedded in a CT heart scan model [74]. The digital heart is animated to simulate beating. The simulated heart motions are timed with the signals opening and closing the pneumatic solenoid valves for the soft robotic right atrium. A cross-section through the simulated interatrial septum of the 3D heart is displayed to the user in a configuration representing a typical ICE view during TP (Fig. 4.4a).

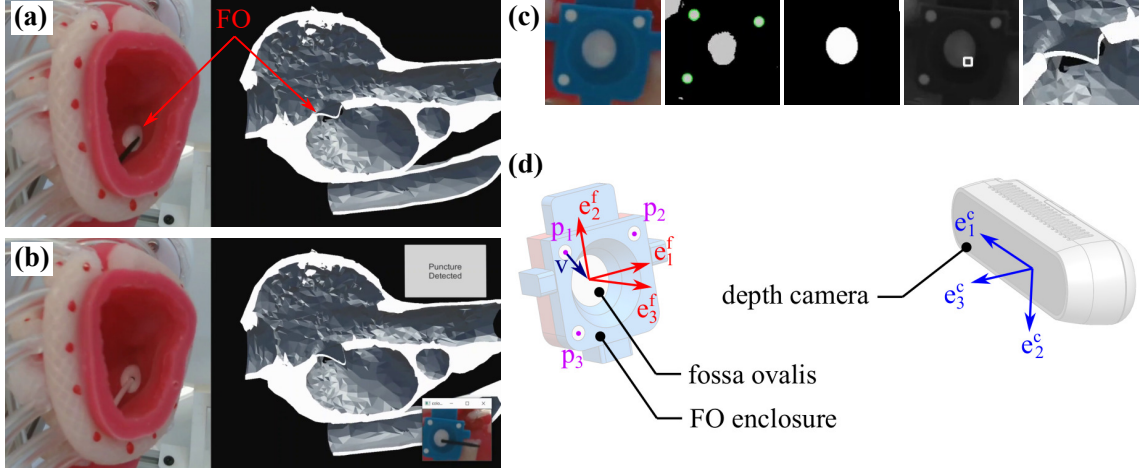


Figure 4.4: (a) Simulated ICE environment displays deformation of the fossa ovalis (FO) during tenting and (b) notifies the user after puncture has been detected. Color images on the left of (a) and (b) are from an overhead camera that provides the user with a real-time interior view of the atrium. (c) Computer vision sequence for the simulated ICE subsystem. From left to right: RGB view, location of reference points, isolation of FO pixels, identification of tenting peak, and animation of tenting on the digital FO. (d) Definition of camera and FO orthonormal bases.

The 3D depth camera data of the physical FO are used to create the 2D animation of the virtual FO in real-time. Each frame of the animation begins by establishing an instantaneous local coordinate system for the physical FO, which is constantly moving due to the actuation of the right atrium. Positions of all pixels detected by the depth camera are represented in the orthonormal camera basis $\{e^c\}$ positioned on the front face of the camera, and must be converted to an orthonormal basis $\{e^f\}$ aligned with and centered on the undeformed FO. Colored markers on the exterior of the FO enclosure located at points p_1 , p_2 , and p_3 are identified using an open source computer vision library (OpenCV) and used to define 3×1 reference vectors $\vec{p}_1$, $\vec{p}_2$, and $\vec{p}_3$ in the camera basis (Fig. 4.4c–d). The 3×1 unit vectors making up $\{e^f\}$ are then defined as follows:

$$\hat{e}_1^f = \frac{\vec{p}_2 - \vec{p}_1}{\|\vec{p}_2 - \vec{p}_1\|}, \quad \hat{e}_2^f = \frac{\vec{p}_1 - \vec{p}_3}{\|\vec{p}_1 - \vec{p}_3\|}, \quad \hat{e}_3^f = \hat{e}_1^f \times \hat{e}_2^f \quad (4.1)$$

The change of basis for any 3×1 position vector $\vec{r}$ from $\{\mathbf{e}^c\}$ to $\{\mathbf{e}^f\}$ is then:

$$[\vec{r}]_{\mathbf{e}^f} = x\hat{e}_1^f + y\hat{e}_2^f + z\hat{e}_3^f = \mathbf{A} ([\vec{r}]_{\mathbf{e}^c} - \hat{p}_1 - \vec{v}), \quad (4.2)$$

$$\text{where } \mathbf{A} = \begin{bmatrix} \hat{e}_1^f & \hat{e}_2^f & \hat{e}_3^f \end{bmatrix}^T \quad (4.3)$$

and $\vec{v}$ is the known vector pointing from p_1 to the center of the undeformed FO, which is constant and determined by the geometry of the FO enclosure.

The last steps in each frame are to identify and transmit the tenting peak location, and to recognize when puncture has been completed. The color boundary between the FO and the edge of the FO insert is identified using OpenCV (fig. 4.4c). The coordinates with the maximum z value within the FO boundary according to (Eq. 4.2) represent the tenting peak. This peak location is sent to Unity, where it is used to position the peak of the virtual FO mesh. The nodes radiating out from the virtual peak are positioned with a second-order polynomial approximation, terminating at the edge of the FO mesh. SATPS users receive an on-screen notification that puncture has occurred if the color of the TP dilator is detected by the depth camera (Fig. 4.4b). The above process repeats for every frame captured by the depth camera, resulting in a real-time ICE-style representation of tenting of the simulator's FO.

4.3 Subsystem Validation

The aim of the design validation was to use benchtop tests to verify the accurate simulation of the right atrium during TP by the SATPS. Atrial dynamics of the SATPS were evaluated by measuring the volume displacement of the soft robotic right atrium during actuation. Static response of the atrial septum was validated with static force testing of several possible materials for the replaceable fossa ovalis subsystem. Accuracy of the simulated ICE visualization was verified by comparing measured tenting peak locations to known constants in a dynamic test.

4.3.1 Atrial Volume Displacement

The aim of this test was to verify the soft actuators' ability to simulate atrial muscle contraction. A soft robotic right atrium was sealed at the entrance to the SVC and the tricuspid valve opening with 3D printed plugs attached with silicone adhesive (Sil-Poxy, Smooth-On, Inc.). A graduated cylinder was attached to the IVC using a 3D printed adapter. The atrium assembly was mounted on an aluminum frame with the graduated cylinder oriented vertically. The atrium was filled with water and the baseline volume was recorded from the graduated cylinder. Actuator pressure was increased in 10 kPa increments up to 100 kPa, and the change in atrial volume was recorded while maintaining constant pressure at each increment (Fig. 4.5a). The right atrial (RA) ejection or emptying fraction (EF) was calculated as:

$$EF = 100\% \times \frac{\Delta RAV_P}{RAV_0} \quad (4.4)$$

where ΔRAV_P is the change in RA volume at pressure P and RAV_0 is the volume of the unactuated atrium. RAV_0 was estimated as 90.6 mL using the CAD model for the inner core of the atrium mold. Calculated EF values were compared to physiological RA stroke EF values from literature [75].

The right atrium was displaced an anatomically realistic volume of water when actuated. Atrial ejection fraction reached a maximum of 23% at a pressure of 100 kPa. This result is nearly within one standard deviation of mean healthy RA stroke ejection fractions measured by Li et al. ($35 \pm 11\%$, $n = 135$) (Fig. 4.5b) [75]. Gorter et al. [76] determined that RA EF is reduced in patients with atrial fibrillation, indicating the SATPS could be adjusted to simulate atrial dynamics of AF patients by decreasing pressure to the actuators.

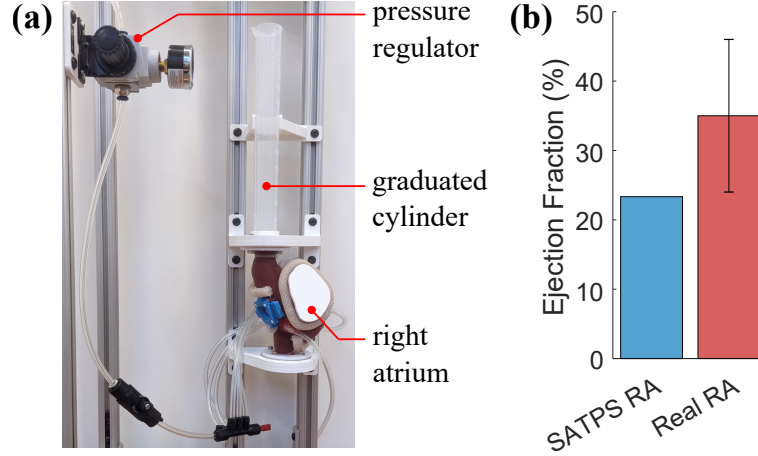


Figure 4.5: (A) Atrial volume displacement testing setup. (B) SATPS right atrial ejection fraction at 100 kPa compared to right atrial ejection fraction measurements from healthy adults [75] (mean $\pm$ SD).

4.3.2 Fossa Ovalis Tenting and Puncture Force

To provide realistic haptic feedback during tenting and puncture, the material used for the replaceable fossa ovalis subsystem needed to have the same tenting and puncture behavior as real atrial septum tissue. Testing protocol and physiological values for comparison were based on the work of Howard et al. [77], who measured tenting and puncture response of *ex vivo* human atrial septum tissue. We evaluated the potential fossa ovalis materials based on three characteristics. Tenting force, which here is the force needed to tent the fossa ovalis by a distance of 8 mm, provides a measure of the force feedback felt during tenting. Puncture force, or the force required to pass the needle tip, dilator tip, and sheath through the fossa ovalis, represents the force feedback felt during puncture. Needle tenting distance, which is the depth of tenting by the needle before it punctures the fossa ovalis, provides a measure of the required tenting depth before puncture is feasible during TP.

Three types of silicone rubber were tested: Ecoflex 00-30, Dragon Skin 10 Medium, and Dragon Skin 30 (Smooth-On Inc.). Cast material samples (2 mm $\times$ 25 mm $\times$ 25 mm) were secured with a 3D printed fixture for testing (Fig. 2). The fixture compressed a sample between two flat plates secured with a screw and nut. Openings in the top and bottom plates

of the fixture represent the mean dimensions of an adult fossa ovalis, $14.1 \text{ mm} \times 12.1 \text{ mm}$ [78]. Three samples of each material were tested. Each sample was subjected to one tenting force test followed by one puncture force test.

Tenting and puncture force testing were conducted using a modified Brockenbrough needle (EP003994S, Medtronic, Inc.) and Mullins sheath and dilator assembly (EP008591, Medtronic, Inc.). The outside diameters of the needle tip, dilator, and sheath were 0.8 mm, 2.6 mm, and 3.0 mm, respectively. The needle was straightened and trimmed to a length of 130 mm, measured from the needle tip. The sheath was been trimmed to 95 mm, and the dilator to 110 mm. The sheath and dilator were bonded together at the trimmed base to prevent relative displacement during testing, leaving 15 mm of dilator extended past the tip of the sheath. The needle was allowed to slide inside the dilator so it could be retracted for tenting tests or extended and fixed in place for puncture tests. Material testing was conducted on an electromechanical force testing machine (Insight 1, MTS) fitted with clamp-style grips (Advantage Screw Action Grips, MTS) to secure the sample fixture and modified catheter assembly (Fig. 4.6a).

Tenting tests measured the force response as the dilator was advanced into the sample without puncture. The sample fixture was held in the lower grip of the testing machine and the catheter assembly was held in the upper grip, centered on and perpendicular to the material sample with the needle tip retracted into the dilator. The tool head was positioned such that the dilator tip was just above the top surface of the material sample. Force and displacement data were then collected while advancing the dilator tip into the sample at a rate of $254 \text{ mm} / \text{min}$ to a depth of 8 mm. Tenting force was recorded as the force measurement at the final 8 mm depth.

Puncture tests measured the force response as the needle, dilator, and sheath were advanced through the sample. The sample fixture and catheter assembly were positioned in the testing machine grips in the same manner as for tenting force testing. However, for puncture force testing, the needle was extended and constrained to prevent retraction into

the dilator during testing. The tool head was positioned such that the needle tip was just above the top surface of the material sample. Force and displacement data were collected while advancing the needle through the sample at a rate of 254 mm / min to a depth of 40 mm. Local maxima in the force-displacement plots indicated the puncture force for each of the TP catheter assembly components. The displacement value at the peak representing needle puncture force was recorded as the needle tenting distance.

One-way ANOVA and Tukey-Kramer tests were used to compare mean values for tenting force, all puncture forces, and needle tenting distance. Only comparisons between silicone samples and human tissue are reported.

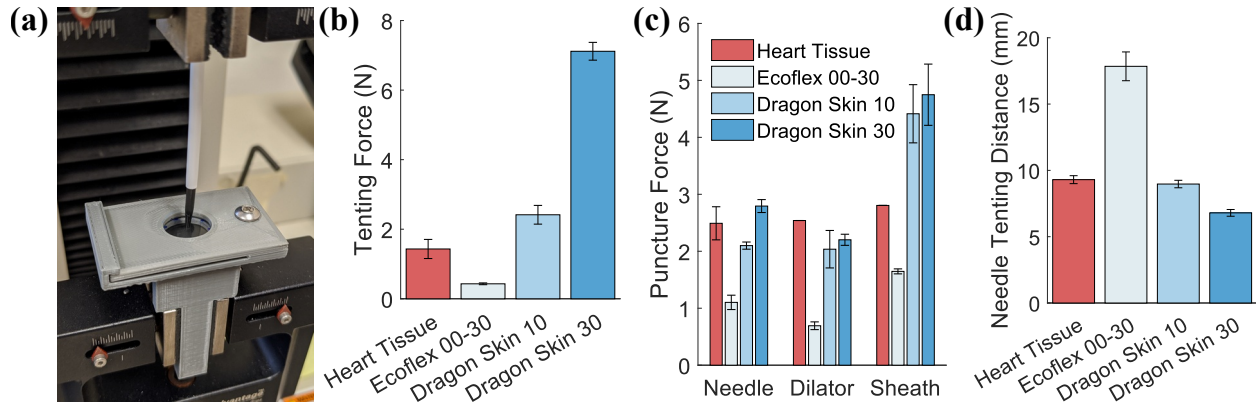


Figure 4.6: (a) Fossa ovalis material testing setup. (b) Force required to tent the fossa ovalis a distance of 8 mm. (c) Force required to pass each component of a TP catheter assembly through the fossa ovalis. (d) Tenting distance by needle before puncture. Human tissue data is from [77]. Bar plots represent mean $\pm$ SE with $n = 8$ for human tissue and $n = 3$ for each type of silicone.

Fossa ovalis material evaluation indicated Dragon Skin 10 Medium was the closest simulation to real human fossa ovalis tissue. Although human tissue samples in [77] were found to have mean thickness 0.68 ± 0.27 mm (mean $\pm$ SD, $n = 8$), due to fabrication limitations of ± 0.3 mm the simulated tissue samples were thicker. Silicone samples had mean thickness 1.87 ± 0.18 mm (mean $\pm$ SD, $n = 9$). Mean tenting force for Dragon Skin 30 was significantly higher than that of human tissue ($p < 0.0001$) (Fig. 4.6b). Mean tenting forces for Ecoflex 00-30 and Dragon Skin 10 were not significantly different than that of human tissue despite large differences between mean values due to large variance in the human tissue

data. Needle puncture force for Ecoflex 00-30 was significantly lower than that of human tissue ($p < 0.05$). Standard deviations were not published for human tissue puncture force for the dilator tip and sheath insertion, but mean values suggest a closer match to Dragon Skin 10 and Dragon Skin 30 compared to Ecoflex 00-30 (Fig. 4.6c). Mean needle tenting distances for Ecoflex 00-30 and Dragon Skin 30 were both significantly different than that of human tissue ($p < 0.0001$ and $p < 0.05$, respectively)(Fig. 4.6d). Dragon Skin 10 was the only tested material that did not show significant difference with human tissue in any of the measured properties, and thus it was chosen as the material for the replaceable fossa ovalis insert.

4.3.3 Simulated ICE Measurements

For the simulated ICE subsystem to provide accurate visual feedback, the location of the tenting peak displayed on-screen must match the location of the physical tenting peak relative to the dynamic right atrium. The simulated ICE algorithm locates and measures the position of the tenting peak on the moving fossa ovalis, and converts the position to a local basis centered on the FO (Section 4.2.2). To evaluate the accuracy of the algorithm’s output, a validation study was performed using 3D-printed test cones attached to the exterior surface of fossa ovalis inserts. The cones were designed to simulate the shape of the tented FO with peaks in 2 mm increments and centered at $x = y = 0$ (Fig. 4.7a). Each assembled test insert was installed in the soft robotic atrium and the calculated x , y , and z coordinates of the tenting peak, as defined in (Eq. 4.2), were recorded. The right atrium was actuated by the FREEs for the duration of each test to verify accurate motion tracking of the FO. Each test cone was measured for 3000 frames, or approximately 30 seconds. The calculated coordinates were then compared to the known actual coordinates, which were determined by the test cone geometry.

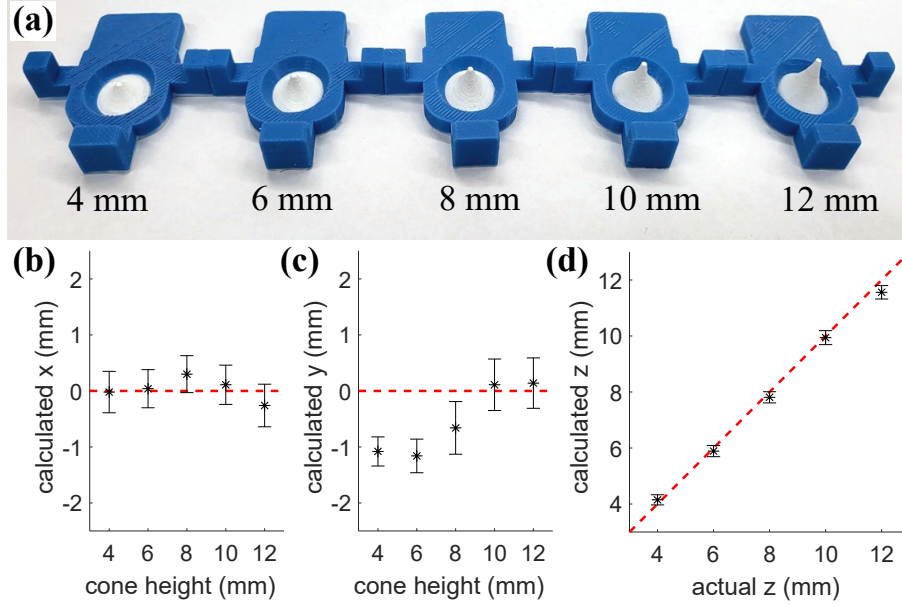


Figure 4.7: (a) Test cones with varying heights attached to FO inserts. Calculated (b) x , (c) y , and (d) z peak coordinates (mean $\pm$ SD, $n = 3000$ for each cone). Red dashed lines represent perfect accuracy.).

Results of the simulated ICE validation test showed that the depth camera algorithm accurately measures the three-dimensional position of the physical FO tenting peak relative to the local basis on the moving atrium. Mean calculated x coordinates were within 0.5 mm of the true value for all test cones (Fig. 4.7b). The least accurate coordinates were in the y direction, with a maximum mean discrepancy of 1.16 mm (Fig. 4.7c). Mean tenting distance, represented by the z coordinate was within 0.5 mm of the true value for all test cones (Fig. 4.7d). A possible source of error in mean x and y coordinates is misalignment during attachment of the cones to the FO inserts. Variance in the data is mostly due to noise in the raw camera measurements. The demonstrated accuracy from the simulated ICE algorithm is sufficient to allow users to position the needle in the desired region of the FO and unambiguously confirm tenting before puncturing the FO.

4.4 System Validation

The purpose of system validation was to test the three hypotheses presented in Section 4.1 through hands-on evaluation by experienced clinicians.

- Hypothesis 1: An active TP simulator is more realistic than a passive simulator.
- Hypothesis 2: The SATPS realistically simulates the tactile and visual feedback experienced during TP.
- Hypothesis 3: The SATPS is a useful training device for transseptal puncture.

Hypotheses 1 was tested through blind and disclosed comparison of the simulator in “active” and “passive” states. Hypothesis 2 was tested through face validity assessments of the SATPS that asked participants to rate the simulator’s realism. Hypothesis 3 was tested through a content validity assessment that asked participants a series of questions about the training utility of the simulator.

4.4.1 Methods

This study was conducted at the Jump Trading Simulation and Education Center in Peoria, IL and OSF Saint Anthony Medical Center in Rockford, IL, and was approved by the University of Illinois at Urbana-Champaign Institutional Review Board (Appendix A). A convenience sample of seven clinicians with a minimum of 100 career TPs were recruited to perform the system validation assessments (Table 4.1). All participants gave informed consent before beginning their session.

Validation sessions were conducted individually. Each trial in this study consisted of the participant performing a transseptal puncture on the SATPS in one of two states: passive (actuators off) and active (actuators on). Pneumatic control valves were cycled at 75 BPM in both states, but airflow to the actuators could be toggled on or off using ball valves hidden under an opaque shroud. This configuration allowed the pneumatic control valves to cycle

and create pneumatic actuation sounds during either actuator state, thus preventing the user from guessing the difference between the two states using auditory cues. Before each trial, research personnel positioned the catheter assembly with the tip of the dilator in the superior vena cava (SVC) with the needle retracted. Participants then navigated the catheter assembly to the fossa ovalis, induced tenting, and punctured through the fossa, completing the trial. Participants were allowed to reset trials at any time.

Participants conducted a practice trial with the SATPS in the passive state before proceeding to a blind assessment, followed by a disclosed assessment, and finishing with a content validity questionnaire. For the disclosed assessment, a real-time overhead view from a second depth camera (RealSense D415) was displayed on a laptop next to the simulated ICE view to assist the user during right atrial navigation before tenting (Figs. 4.1a and 4.4a–b).

Blind Assessment

The first stage of system validation was a blind assessment, in which the actuator state was unknown to the participant. The purpose of the blind assessment was to obtain an unbiased comparison of the system in the passive and active states by blocking the participant’s view of the atrium under an opaque shroud (Fig. 4.1b). The overhead camera view was omitted from the blind assessment trials to prevent participants from visually identifying the difference between the two states. Presentation of the passive or active test condition was randomized between participants, and a new fossa ovalis insert was installed before the start of each test condition. A blind assessment questionnaire included questions evaluating

Table 4.1: System validation participant background information

Participant	Primary Field	Years of Practice	TPs per Month	Career TPs
1	Electrophysiology	6	16	> 1000
2	Electrophysiology	2	6	250
3	Electrophysiology	10	12	2000
4	Electrophysiology	18	10	> 700
5	Electrophysiology	> 20	9	> 1000
6	Electrophysiology	> 20	12	> 200
7	Interventional Cardiology	7	4	> 150

each step of conducting TP on the SATPS for realism on a five-point Likert scale (Table 4.2). Participants completed the questionnaire after each test condition. After completing both test conditions, participants were given the option to retry their first test condition and modify their questionnaire answers for the first test condition based on the experience of both conditions. After finishing the blind trials, participants answered one additional question asking which test condition felt more realistic in terms of tactile feedback (Table 4.2).

Disclosed Assessment

The second stage of system validation was a disclosed assessment, in which the participant knew the state of the actuators. The disclosed assessment provided a direct comparison of the two actuator states without the hindrance of being unable to see the catheter assembly position prior to tenting. First, research personnel explained and demonstrated the difference between the active and passive actuator states. A new fossa ovalis insert was installed, and participants conducted at least one trial each for the passive and active states. The overhead camera provided participants with a view of the interior of the atrium for the entire duration of each trial. After all disclosed trials, participants completed a series of questions directly comparing the passive and active states in terms of realism and user preference on a five-point Likert scale (Table 4.2).

Content Validation

After completing all system validation trials, participants completed a questionnaire designed to evaluate the content validity of the SATPS. The questionnaire included a task training assessment to evaluate the utility of the SATPS for training users in each critical step of TP on a five-point Likert scale (Table 4.2), and a system assessment with yes/no questions regarding various use cases for the SATPS (Table 4.6).

Table 4.2: Questions and score key for system validation

Questions	Score Key				
	1	2	3	4	5
Blind Assessment					
How realistic was...					
...the feel of navigating the right atrium with the catheter?					
...the feel of the catheter on the aortic mound?					
...the feel of tenting the fossa ovalis?					
...the feel of puncturing the fossa ovalis?	very unrealistic		neutral		very realistic
...the representation of tenting in the simulated ICE imaging?					
...the representation of the rest of the heart in the simulated ICE imaging?					
...performing transseptal puncture on the SATPS, overall?					
Which felt more realistic in terms of tactile feedback?	passive much more realistic		no difference		active much more realistic
Disclosed Assessment					
Which state was more realistic in terms of...					
...the feel of navigating the right atrium with the catheter?					
...the feel of the catheter on the aortic mound?					
...the feel of tenting the fossa ovalis?	passive much more realistic		no difference		active much more realistic
...the feel of puncturing the fossa ovalis?					
...performing transseptal puncture on the SATPS, overall?					
Which felt more realistic in terms of tactile feedback?	strongly prefer passive		no preference		strongly prefer active
Content Validation: Task Training Assessment					
How useful is the SATPS for training how to...					
...correctly locate the fossa ovalis with the catheter?					
...verify catheter position on the fossa ovalis before puncture using ICE?	very useless		neutral		very useful
...puncture the fossa ovalis with appropriate force?					

Statistical Analysis

Statistical analyses of compiled validation questionnaire scores were used to evaluate each hypothesis. To test Hypothesis 1, paired t-tests were used to evaluate whether active realism scores were greater than passive realism scores for each participant during the blind assessment. Further, one-sample t-tests were used to evaluate whether the mean scores from the blind and disclosed assessment direct comparison questions were greater than the neutral score of 3. Hypothesis 2 was tested using averaged passive and active blind assessment realism scores for each participant. One-sample t-tests were used to determine if the mean of all averaged participant scores for each question was greater than the neutral score of 3. Hypothesis 3 was evaluated by using one-sample t-tests to compare mean content validation scores to the neutral score of 3 (Table 4.2). All t-tests used $p < 0.05$ as the threshold for significance. Statistical analysis was performed using data analysis software (OriginPro 2020, OriginLab Corporation). Results are presented as mean $\pm$ standard error (SE) unless otherwise noted.

4.4.2 Results and Discussion

Results of system validation generally confirmed the overall realism and training utility of the SATPS. However, results also suggest that Hypothesis 1 did not account for the high variability in patient cardiac conditions. Participant feedback yielded valuable insights for improving the SATPS.

Passive and Active State Comparison

Results from the blind assessment found no statistically significant difference in realism scores for the features of the simulator between passive and active states ($p > 0.05$, Table 4.3). Mean scores comparing the realism of the two states after the blind and disclosed assessments were not significantly greater than the neutral score of 3 (3.2 ± 0.6 and 3.3 ± 0.4 , respectively). None of the disclosed assessment direct comparison scores were significantly greater than 3 (Table 4.4). Thus, these results led us to reject Hypothesis 1, which suggested that an active

Table 4.3: Blind Assessment: Realism scores for passive, active, and average of passive and active (n = 7)

How realistic was...	Score (mean $\pm$ SE)		
	Passive	Active	Average
...the feel of navigating the right atrium with the catheter?	3.7 ± 0.5	3.6 ± 0.2	$3.6 \pm 0.3^*$
...the feel of the catheter on the aortic mound?	3.7 ± 0.3	3.1 ± 0.3	3.4 ± 0.3
...the feel of tenting the fossa ovalis?	4.3 ± 0.3	3.7 ± 0.4	$4.0 \pm 0.3^{**}$
...the feel of puncturing the fossa ovalis?	4.4 ± 0.2	3.7 ± 0.4	$4.1 \pm 0.3^{**}$
...the representation of tenting in the simulated ICE imaging?	4.1 ± 0.5	3.6 ± 0.4	$3.9 \pm 0.4^*$
...the representation of the rest of the heart in the simulated ICE imaging?	3.1 ± 0.5	3.3 ± 0.5	3.2 ± 0.5
...performing transseptal puncture on the SATPS, overall?	3.6 ± 0.6	3.4 ± 0.3	3.5 ± 0.4

* $p < 0.05$, ** $p < 0.01$

TP simulator is more realistic than a passive simulator.

Comments from participants indicated that the reason for the rejected hypothesis was not due to a lack of perceptible difference in the tactile feedback of the two actuator states. Six of the seven participants were electrophysiologists (EPs), meaning they most commonly performed transseptal puncture during atrial fibrillation ablation procedures. Multiple participants stated that the active heart felt more like a healthy young heart than that of an AF patient. Patients with atrial fibrillation have reduced atrial ejection fractions [76], [79], implying decreased atrial wall displacement during atrial systole. Therefore, the beating of the right atrium in a patient with atrial fibrillation is less likely to be felt by the operator through the TP catheter assembly compared to patients without AF. For clinicians who primarily operate on AF patients, a passive simulator could then be considered more realistic than an active one.

Table 4.4: Disclosed Assessment: Mean direct comparison scores (n = 7)

Which state was more realistic in terms of...	Score (mean $\pm$ SE)
...the feel of navigating the right atrium with the catheter?	3.7 ± 0.4
...the feel of the catheter on the aortic mound? ¹	2.8 ± 0.4
...the feel of tenting the fossa ovalis?	3.1 ± 0.1
...the feel of puncturing the fossa ovalis?	3.0 ± 0.0
...performing transseptal puncture on the SATPS, overall?	3.1 ± 0.3

¹n = 5 for this question, since two subjects do not use the aortic mound as a tactile cue during TP.

Results and feedback from the passive and active state comparison suggest that the active state might be considered more realistic by pediatric EPs or interventional cardiologists than the primarily adult EPs included in this study. An expanded validation study with a more diverse participant group would help distinguish the simulation preferences of different specialists. Even within our participant group there was variability in actuator state preference, suggesting that a TP simulator should offer versatile actuation settings to train new users for different TP techniques and applications. The ability of the SATPS actuators to be adjusted via pressure regulation make it possible for the SATPS to simulate variety of cardiac conditions.

Face Validity

Face validity of the SATPS, represented by Hypothesis 2, was largely supported by the results of this study (Table 4.3). Participants rated the realism of the tactile feedback from the soft robotic atrium and the replaceable fossa ovalis favorably. The feel of navigating the right atrium was significantly more realistic than neutral (3.6 ± 0.3 , $p = 0.046$). The realism of the feel of the catheter on the aortic mound was borderline significant (3.4 ± 0.3 , $p = 0.086$). The feel of tenting and puncturing the FO received the highest mean realism scores among all subsystems, and both were significantly higher than neutral score of 3 (4.0 ± 0.3 , $p = 0.005$ and 4.1 ± 0.3 , $p = 0.003$, respectively).

The most common detractors for tactile realism scores were FO position and friction. The FO in the soft robotic atrium was more posterior than expected for most participants, resulting in increased difficulty locating the fossa during the blind assessment. While the position of the fossa was anatomically correct in the atrium model according to mean FO position from literature [78], it is likely that the compliance of the soft robotic atrium caused it to sag, rotating the FO posteriorly. This rotation error was increased during actuation. Adding an additional rotational constraint to the assembly could help maintain correct FO positioning in both actuation states. Although friction between the TP catheter assembly

and the walls of the atrium was reduced by the low-friction coating, there was still noticeably more friction between the catheter and the silicone than there would be between the catheter and heart tissue lubricated by blood. This friction caused excessive resistance during axial translation of the catheter and during “clocking”, or axial rotation of the catheter. This friction exacerbated the perceived FO position error, since participants had to apply extra rotation of the catheter to overcome friction during clocking.

Representation of tenting in the simulated ICE subsystem had a mean realism score that was significantly greater than the neutral score, but representation of the rest of the heart did not (3.9 ± 0.4 , $p = 0.031$ and 3.2 ± 0.5 , $p = 0.34$, respectively, Table 4.3). Participants felt the on-screen FO deformation during tenting was accurate and useful for confirming the puncture location, but the lack of pre-tenting needle tracking and the static view were regarded as major limitations of the current system. Without the ability to locate the needle inside the atrium with simulated ICE or fluoroscopy, some participants became disoriented during the blind assessment trials. Participants had varying preferences for the orientation of the heart in the simulated ICE. The view of the FO and the surrounding anatomy during real TP is dependent on the ICE technology used (radial or phased-array) and the specific procedure for which the TP is being performed [80]. These results highlight a need for expanded simulated ICE capabilities in the SATPS to improve realism, including a controllable ICE view and pre-tenting needle imaging.

The mean realism score for performing TP on the SATPS overall was not significantly higher than the neutral score (3.5 ± 0.4 , $p = 0.14$, Table 4.3)). When discussing their responses, participants who gave low scores for overall realism primarily focused on the absence of sophisticated imaging they typically use during TP. These participants emphasized that they considered tactile feedback secondary to visual feedback, giving their perception of the simulated ICE realism more weight than the other two subsystems when assigning the overall score.

Content Validity

Content validity of the SATPS, represented by Hypothesis 3, was supported by the results of this study. In the task training assessment, the utility of the SATPS scored significantly higher than the neutral score of 3 for training users how to locate the FO and puncture it with appropriate force (4.0 ± 0.4 , $p = 0.019$ and 4.1 ± 0.3 , $p = 0.008$, respectively) (Table 4.5). The utility of the simulator for training how to verify catheter position on the FO using ICE scored greater than neutral with borderline significance (3.9 ± 0.5 , $p = 0.056$). The system assessment portion of the validation study also supported Hypothesis 3, with six out of seven participants stating that the SATPS could be useful for training fellows in TP, and five out of seven stating they would recommend the SATPS for training in TP (Table 4.6). Participant responses suggest that in its current form, the SATPS is primarily useful as an introduction to TP for fellows before operating on a real patient, and less suitable for use in testing or certification processes.

Suggestions from participants to improve content validity focused on more advanced visual feedback capabilities and representation of a larger variety of patient anatomies and pathologies. While many experienced operators are able to navigate the atrium relying primarily on tactile feedback, most still rely on fluoroscopy or ICE to assess the situation if they encounter unexpected difficulties. Expanding the function of the simulated ICE environment and adding a simulated fluoroscopy environment would help trainees to learn how to troubleshoot with both imaging technologies when the procedure does not go as planned. Participants also pointed out that the SATPS is a good representation of a normal, healthy heart, but that the SATPS would better prepare trainees for difficult TP scenarios

Table 4.5: Content Validation: Mean task training assessment scores (n = 7)

How useful is the SATPS for training how to...	Score (mean $\pm$ SE)
...correctly locate the fossa ovalis with the catheter?	$4.0 \pm 0.4^*$
...verify catheter position on the fossa ovalis before puncture using ICE?	3.9 ± 0.5
...puncture the fossa ovalis with appropriate force?	$4.1 \pm 0.3^{**}$

* $p < 0.05$, ** $p < 0.01$

Table 4.6: Content Validation: System assessment. Number of responses are shown in each column.

	YES	NO
Could the SATPS be useful for training residents? ¹	3	3
Could the SATPS be useful for training fellows?	6	1
Could the SATPS be an appropriate training and testing format before catheterization lab experience for residents / fellows?	4	3
Could the SATPS be used as part of the privileging or certifying process for transseptal puncture?	1	6
Would you recommend using the SATPS for training in transseptal puncture?	5	2
¹ n = 6 for this question because one participant felt it was not applicable for residents to train in TP.		

if it represented atypical, but not uncommon abnormalities, such as redundant atrial septa and thick or calcified septa. These two example scenarios require a modified technique for puncturing through the septum, and would be safer for trainees to encounter for the first time in a simulator instead of a real patient.

4.5 Study Limitations and Future Work

4.5.1 Study Limitations

System validation highlighted several design limitations of the SATPS (Section 4.4.2). The current soft robotic right atrium and the replaceable fossa ovalis seem to accurately simulate a typical healthy heart, but the SATPS would be more realistic and a more effective training tool if there were additional atria and FO inserts to represent the high variability in patient heart shape and size, including atypical anatomy known to increase TP difficulty. The simulated ICE subsystem does not currently display the TP catheter assembly until tenting begins, preventing the user from using imaging to troubleshoot issues that arise during navigation of the right atrium. In a training environment, the system is intended to be used with the overhead camera enabled to assist the user with pre-tenting navigation. However, visualization could be improved by using the overhead camera to simulate an ICE view during

the whole procedure. The SATPS also lacks simulations of other common feedback tools such as fluoroscopic imaging, pressure feedback, and contrast injection.

The system validation study was limited by the effects of using a small convenience sample, which was determined by the availability of expert TP specialists in the local area. A larger sample size would improve the accuracy of the results. This study’s participants were predominantly electrophysiologists who operate on adults. Testing with a broader range of cardiac specialists, who perform TP on a variety of patients and conditions, could provide more diverse perspectives on the simulator’s design.

4.5.2 Future Work

Design improvements to the SATPS are driven by participant feedback during system validation. We will develop an algorithm using data from the overhead depth camera to track the catheter assembly as it travels through the right atrium, providing the necessary input for expanded simulated ICE and simulated fluoroscopy environments. Development of a physical controller for adjusting the simulated ICE view is underway. Additional soft robotic right atria should be developed to represent different heart shapes, and new fossa inserts should be designed to represent more challenging puncture conditions such as redundant and calcified atrial septa. Realism of fossa ovalis inserts could be further improved with thinner, tougher materials enabled by revised manufacturing methods, including composites or 3D-printed materials with tunable material properties [81], [82]. Further reduction of friction inside the atrium would also improve realism. Expanding the scope of the SATPS to include guidewire placement would give trainees practice in another critical step of TP. Finally, incorporating an adjustable femoral vein insertion angle, a parameter dependent on patient body fat, would provide an opportunity to practice TP with and without the ability to rest one’s hands on the patient’s leg during atrium navigation and puncture.

Following design changes, the SATPS should be validated with expanded face and content validation. Additional, more objective testing should include construct validity to demonstrate

that the SATPS can be used to correctly distinguish between expert and novice TP operators, and concurrent validity to determine if the SATPS improves real-world performance metrics for TP.

4.6 Conclusion

We designed the SATPS to simulate the right atrium for the purpose of training in three critical steps of transseptal puncture: locating the fossa ovalis, verifying the position of the TP catheter during tenting, and puncturing the FO with appropriate force. The SATPS is novel because it accurately simulates atrial dynamics, FO force feedback, and ICE imaging during tenting. System validation showed that an active TP simulator is not necessarily more realistic than a passive one, but that it is still advantageous to allow for both in order to simulate a variety of cardiac conditions. The SATPS was considered a mostly realistic representation of the right atrium during TP, and a good training tool for beginners before they operate on a real patient for the first time. The realism and training utility of the SATPS will be improved with more advanced imaging simulation and a wider variety of physical atrium and fossa ovalis models.

Chapter 5

Curved Fiber Reinforced Soft Actuators

Abstract

Fiber-reinforced elastomeric enclosures (FREEs) are an established class of soft pneumatic actuators with kinematics determined by the orientation of fibers wrapped around an elastomeric bladder. FREEs have primarily been manufactured as cylinders with straight axes to be used in point-to-point force generation applications. There have been recent applications that use straight FREEs in curved or wrapped configurations, but curvatures for these bent actuator paths are limited by the tendency of straight FREEs to buckle under bending loads. The motivation of this work is to expand the capabilities of FREEs to include curvilinear actuation. This project presents a new type of pre-curved FREE actuator, and seeks to explore curved FREE kinematics using finite element analysis (FEA). A custom program was developed in Python to procedurally generate curved FREEs based on user-specified input parameters, automatically define FEA models for kinematic analysis, and post-process FEA results. Preliminary observations from an iterative FEA study include similarity in axial deformation of straight and curved FREEs, and the dependence of curvature change on

averaged fiber angles and local gradients of fiber angles across the cross section of a curved FREE.

5.1 Introduction

Fiber-reinforced elastomeric enclosures (FREEs) have primarily been designed for linear actuation or twisting and bending from an initially straight configuration [1]. There is potential for expanded utility of FREEs in curvilinear configurations, but linear actuators tend to buckle when tightly wound. Some researchers have used contracting FREEs, also known as pneumatic artificial muscles, in curved applications. Roche et al. cast straight FREEs in a curved configuration in an elastomer matrix for their ventricular assist device [26] and Singh et al. wrapped a long straight FREE around a rigid cuff for a powered arm orthosis [33]. However, in both of these applications the radius of the FREE actuator was small compared to the radius of curvature of the bent actuator path, eliminating the threat of buckling. As a result, both groups of researchers could assume negligible bending forces in their analysis. Applications demanding uniform contraction along a tightly curved surface, such as the Soft Active Transseptal Puncture Simulator (SATPS) in Chapter 4, require FREE actuators with an outside radius on the same order of magnitude as the radius of curvature of the actuation path. Our early SATPS prototypes used straight FREEs wrapped around a cast silicone atrium, but these actuators had a tendency to collapse and pull during installation (Fig. 5.1a). Thus we identified that there was a need for a new type of FREE that can be applied to curved surfaces without buckling (Fig. 5.1b–c).

Curved FREEs have seen little development due to the limitations of existing manufacturing methods. FREE bladders are typically made from preformed elastomeric tubing or from elastomer castings formed around a straight rod for ease of removal [4], [13], [24], [26]. Fibers can be wrapped manually, but maintaining a consistent fiber angle on a curved FREE is difficult without preformed grooves. Instead of a wrapped fiber network, the strain-limiting

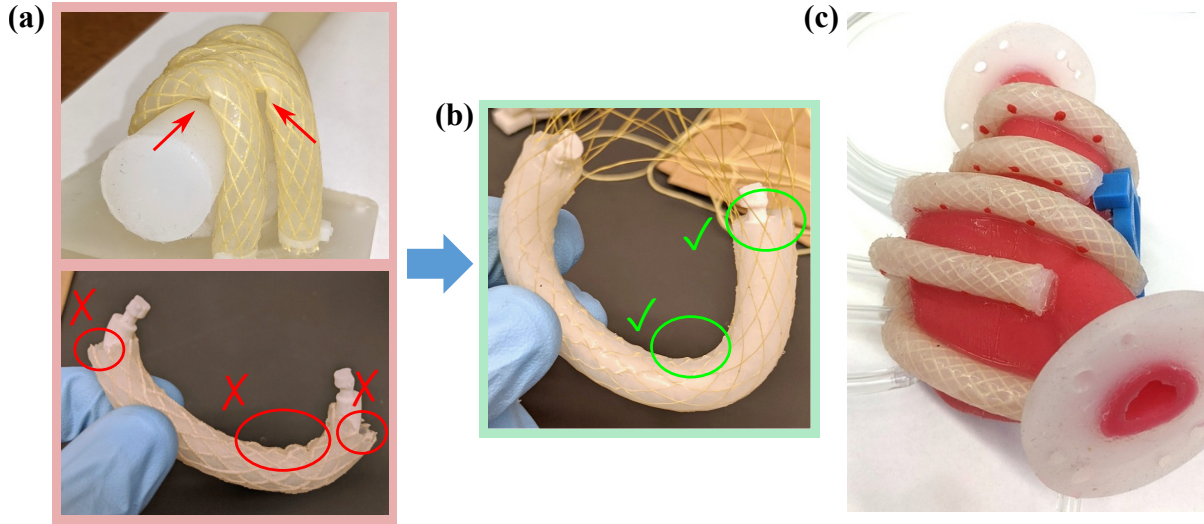


Figure 5.1: (a) Buckling and stress buildup in straight FREEs wrapped around tight curvatures. (b) Casting FREEs in preformed shapes eliminates buckling. (c) Curved FREEs on a soft robotic cardiac simulator.

layer can be made from braided plastic mesh, but premanufactured meshes cannot bend around tight curves without buckling [22], [26]. Hu et al. recently developed and modeled segmented pre-curved FREEs to mimic biological actuation systems using 3D-printed molds and multi-material bending segments [17]. The authors wrapped their fibers manually, but the curved sections had fiber angles designed for extension, which are easier to wrap than fibers designed for contraction. The curved FREEs used in our work are constructed from a single soft silicone material, meaning their kinematics are defined only by their curved actuation paths and fiber angles. The FREEs in this study were formed by using 3D printed molds developed from digital models with accurate fiber paths to define grooves for simple and accurate assembly.

The goal of this chapter was to explore the curved FREE design space and fundamentally understand their kinematics through finite-element analysis (FEA). A parametric modeling software was developed to interface with FEA software (e.g., Abaqus/CAE, Dassault Systèmes) to model curved FREEs of any shape, size and fiber angle along any predefined straight or curved path. This work is the first step in identifying a generalized design methodology for curved FREEs.

This chapter describes the fabrication of curved FREEs, the methods for modeling and analysis of curved FREEs in the design software, preliminary validation of the software, and kinematic observations from iterative curved FREE FEA. The chapter concludes with recommendations for future work.

5.2 Curved FREE Fabrication

The anatomy of a curved FREE is the same as that of a straight FREE, but the fabrication method is complicated by curved geometry. To make either type of FREE, an elastomeric tube-shaped bladder is wrapped with two helical networks of inextensible fibers, and the ends are plugged with stiff end caps, one of which includes a pneumatic inlet. While straight FREE bladders can be fabricated with pre-manufactured tubing or cast from universal molds and cut to length, curved FREE bladders must be cast from custom 3D-printed molds with mold cores that follow the curved centerline path of the FREE.

5.2.1 Mold Development

Curved FREEs are first defined by their pre-actuation path. Circular arcs and Bézier curves are well-suited for curved FREEs because they are inherently smooth and easy to parametrize, but any smooth path can be used as long as the radius of curvature of the actuator path remains larger than the outer radius of the bladder at all points along the curve to avoid self-intersecting geometry. Winding paths such as spirals or helices are theoretically possible, but practically challenging when manufactured with traditional molds due to the need for mold core removal. 3D printing mold cores from water-soluble materials such as polyvinyl alcohol (PVA) can allow for more complex geometries than traditional cores. 3D printing with soft elastomers will vastly expand the geometries available to curved FREE designers, but this technology is still commercially unavailable [32], [83].

FREE centerline paths can be defined with any 3D modeling software and then used to

create a two-part outer mold and a matching inner core (Fig. 5.2). Mold cores are constrained at the ends by the outer mold halves. Complex geometries require additional constraints along the length of the core to maintain uniform wall thickness in the cast bladder. Helical paths for each fiber are modeled along the actuator path and used to create protrusions in the mold, creating grooves in the cast bladder (Fig. 5.2b). Curved FREEs with tight curvatures and low fiber angles are especially difficult to wrap manually, making the preformed grooves essential for maintaining correct fiber angle. Centerline paths are used to create swept outer mold and mold core features. After the addition of an inlet and an air vent, the mold is ready for 3D printing (Fig. 5.2c).

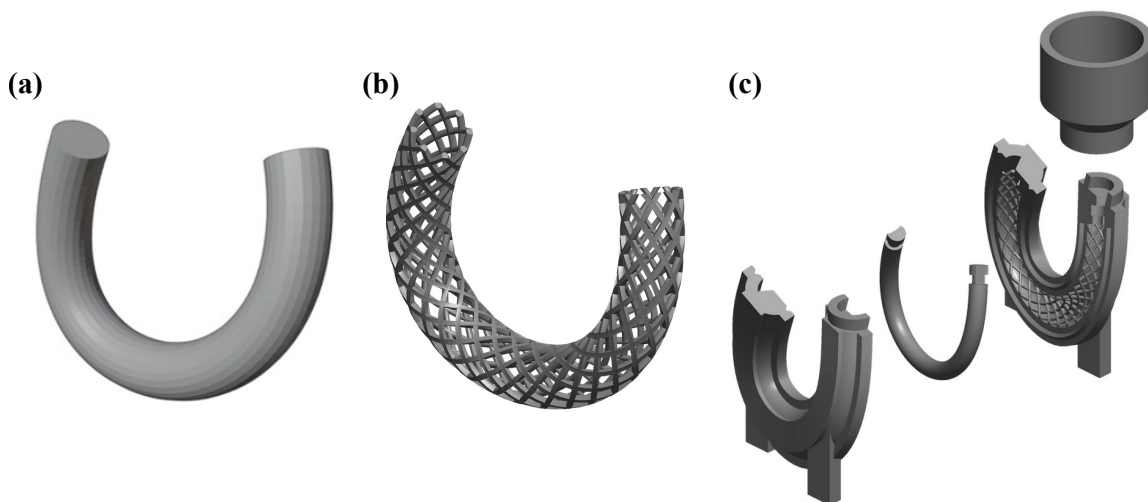


Figure 5.2: Curved FREE mold development. (a) Actuator paths are defined in a 3D modeling software and (b) used to define helical paths for fibers. (c) 3D printed curved FREE mold assembly including two outer mold halves, a mold core, and a filling reservoir.

5.2.2 Casting and Assembly

Curved FREE bladders can be cast from any soft elastomer that has sufficiently low viscosity to flow into the complex molds before curing. The required pressure for actuation increases with elastomer stiffness. The curved FREEs produced for this work were cast from platinum cure silicone rubber (Dragon Skin 10, Smooth-On Inc.) with 10 wt% silicone thinning fluid to reduce viscosity (Silicone Thinner, Smooth-On Inc.). Curved FREE molds are assembled and

clamped together for filling (Fig. 5.3a). After the elastomer cures, the bladder is removed from its mold but kept mounted on the mold core for fiber wrapping to ensure that the desired curvature is maintained in the final assembly. Alternatively, bladders can be transferred to a wrapping tool with the same shape as the mold core (Fig. 5.3b). Fibers are wrapped around the bladder and secured with a final coat of elastomer. The fiber material should bond well to the chosen elastomer to ensure they do not shift out of place. FREEs in this work used cotton string or Kevlar thread as fiber materials. Placing the wrapped and coated FREE under vacuum for several minutes before the silicone cures can improve penetration of the elastomer into the fibers. After wrapping, the FREE is removed from the mold core and the ends are trimmed. The final step is fitting the FREE with end caps and pneumatic fittings (Fig. 5.3c). The FREEs in this work used end caps cast from stiff silicone (Dragon Skin 30, Smooth-On Inc.) and 3D printed barbed fittings, both secured to the FREE with silicone adhesive (Sil-Poxy, Smooth-On Inc.).

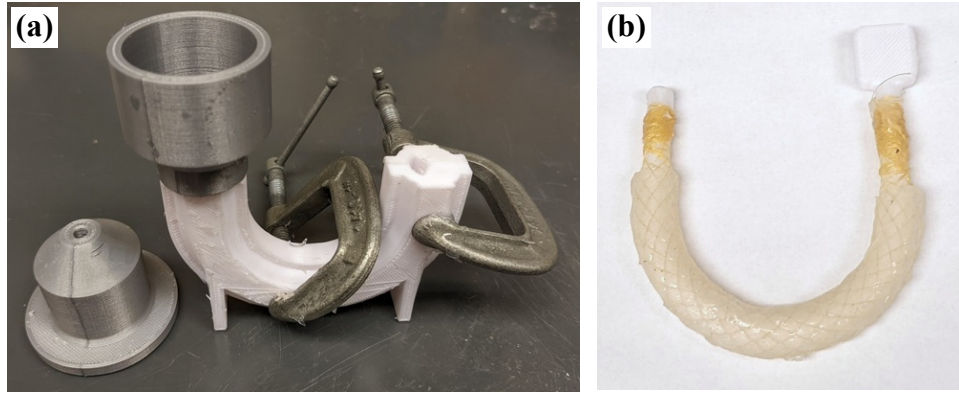


Figure 5.3: Curved FREE manufacturing process. (a) Molds are assembled, clamped together, and filled with elastomer. A 3D-printed plunger (*left*) helps speed up casting and prevent air gaps inside the mold. (b) Curved FREE bladders installed on curved wrapping tool with the same shape as the bladder mold core. A finished curved FREE is shown in Fig. 5.6a.

5.3 Programming and FEA of Curved FREEs

A custom program developed to build curved FREE models was written in Python, which interfaced with Abaqus/CAE for finite-element analysis (FEA). Models are constructed and

analyzed in three basic steps: (i) geometry definition and mesh creation, (ii) finite-element analysis, and (iii) post-processing using predefined output nodes. A link to the repository for the Python code is included in Appendix B.1.

5.3.1 FREE Geometry Definition

Curved FREEs can be programmatically constructed from user-defined input parameters. Basic bladder parameters are outer radius r_o and wall thickness t . The centerline path, which define the FREE's initial trajectory, can be any discretized path with n_n nodes along its length. Fiber paths that twist around the centerline are determined from the fiber angle of each fiber group, α and β , and number of fibers n_f (Fig. 5.4). The remaining input parameters control the final pneumatic pressure p_f , the bladder material, positions of output nodes for post processing, and the finite-element mesh for the bladder, fibers, and end caps.

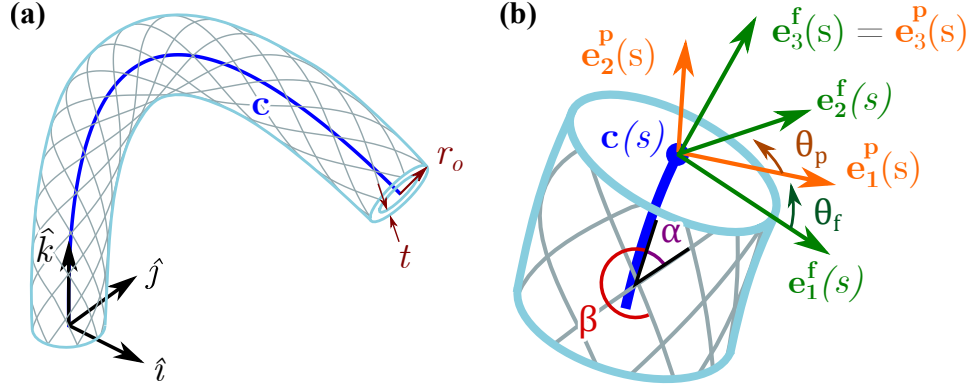


Figure 5.4: (a) Global coordinate system and curved FREE centerline path definition. (b) Local Frenet frame $\{\mathbf{e}^f\}$ and parallel frame $\{\mathbf{e}^p\}$ bases. In $\{\mathbf{e}^f\}$, the normal vector $\mathbf{e}_1^f$ always points to the inside of the path's curve. Circumferential position can be measured in either frame (θ_f for $\{\mathbf{e}^f\}$ and θ_p for $\{\mathbf{e}^p\}$). Fiber angles are measured relative to the centerline tangent vector $\mathbf{e}_3^f = \mathbf{e}_3^p$.

For an arbitrary curved centerline path, the corresponding helical fiber paths can be positioned such that the fiber angles, and therefore the kinematics of the FREE can be programmed during setup. The following equations use the symbol α to represent fiber angle, but all equations also hold true for the second fiber network, defined by fiber angle β . This chapter limits its scope to contracting FREEs with symmetric fiber angles, i.e., with $\alpha = -\beta$.

Local Coordinate Systems

Positioning of fibers along a curved path begins with the establishment of a local basis at every point along the path. The FREE centerline path $\mathbf{c}(s)$ is defined as

$$\mathbf{c}(s) = x(s)\hat{\mathbf{i}} + y(s)\hat{\mathbf{j}} + z(s)\hat{\mathbf{k}}, \quad (5.1)$$

where s is the position along the centerline ranging from 0 to length L . The start of the path $\mathbf{c}(0)$ is positioned at the global origin tangent to the $\hat{\mathbf{k}}$ unit vector and with the initial radius of curvature, i.e., the normal vector at $\mathbf{c}(0)$, pointing along the $\hat{\mathbf{i}}$ unit vector (Fig. 5.4b).

Our model uses two types of moving reference frames to define FREE geometry during setup and to calculate output parameters after FEA. A Frenet frame is calculated at every position s along $\mathbf{c}$ to form an orthonormal basis $\{\mathbf{e}^f(s)\}$, defined by the set of unit vectors $\{\mathbf{e}_1^f(s), \mathbf{e}_2^f(s), \mathbf{e}_3^f(s)\}$, corresponding respectively to the normalized normal, binormal, and tangent vectors for $\mathbf{c}(s)$ (Fig. 5.4b). The Frenet frame is useful for determining the orientation of the curve at any position, since the normal vector always points toward the path's local center of curvature. This property of Frenet frames is useful for curvature-dependent equations such as Eqn. (5.7). However, Frenet frames become undefined in straight sections of a path and can produce abrupt basis rotations anywhere that the path changes direction. To keep track of fiber positions, it is more convenient use a parallel transport frame [84]. At every point along $\mathbf{c}$ we use a parallel transport frame to define a another local orthonormal basis $\{\mathbf{e}^p(s)\}$, which is made up of unit vectors $\{\mathbf{e}_1^p(s), \mathbf{e}_2^p(s), \mathbf{e}_3^p(s)\}$. Normalized tangent $\mathbf{e}_3^p$ is equivalent to $\mathbf{e}_3^f$, while $\mathbf{e}_1^p$ and $\mathbf{e}_2^p$ are orthogonal vectors in the cross-sectional plane positioned to minimize point-to-point twist, as defined by Hanson et al. [84]. The twist-minimizing property of parallel transport frames makes them well-suited for positioning fiber nodes along $\mathbf{c}$. For planar convex curves such as those presented in Section 5.5, $\{\mathbf{e}^p(s)\} = \{\mathbf{e}^f(s)\}$ everywhere along $\mathbf{c}$.

Defining Fiber Paths

A fiber path $\mathbf{f}_i(s)$ for fiber i is defined by points in the $\mathbf{e}_1^p \mathbf{e}_2^p$ cross-sectional plane at position s :

$$[\mathbf{f}_i(s)]_{\mathbf{e}^p} = (r_o \cos \theta_p) \mathbf{e}_1^p + (r_o \sin \theta_p) \mathbf{e}_2^p, \quad (5.2)$$

where $[\mathbf{f}_i(s)]_{\mathbf{e}^p}$ is the path point expressed in the local basis $\{\mathbf{e}^p(s)\}$ and θ_p is the in-plane circumferential angular position with respect to the $\mathbf{e}_1^p$ axis. The circumferential position with respect to the $\mathbf{e}_1^f$ axis is θ_f . The change of basis for a point on the fiber path between the local and global reference frames are:

$$\mathbf{f}_i(s) = \mathbf{R}(s)[\mathbf{f}_i(s)]_{\mathbf{e}^p} + \mathbf{c}(s) \quad (5.3)$$

$$\text{and } [\mathbf{f}_i(s)]_{\mathbf{e}^p} = \mathbf{R}^T(s)(\mathbf{f}_i(s) - \mathbf{c}(s)) \quad (5.4)$$

$$\text{where } \mathbf{R}(s) = \begin{bmatrix} \mathbf{e}_1^p(s) & \mathbf{e}_2^p(s) & \mathbf{e}_3^p(s) \end{bmatrix} \quad (5.5)$$

is a 3×3 rotation matrix.

There are two methods that can be used to plot fiber paths along $\mathbf{c}$. The first is a constant fiber angle approach that yields a more idealized FREE with performance similar to that of straight FREEs. The second is a constant pitch approach that is better suited to current manufacturing methods. 3D modeling software typically uses a constant pitch approach when constructing swept helices around a predefined path. For the current modeling and manufacturing process for curved FREEs, this swept helix operation is necessary to create fiber grooves in the curved FREE bladder molds, so it is practical to use the constant pitch approach. Therefore the experimental FREEs in Section 5.4 use the constant pitch method. To minimize the kinematic effects of variable fiber angles around the circumference of the FREEs, the constant fiber angle method was used for most of the analysis in Section 5.5.

Constant Fiber Angle Method

In the constant fiber angle (CFA) method presented by Hu et al., the fiber angle is held constant from point to point along the discretized fiber path and is independent of circumferential position, resulting in variable helix pitch if the path is transformed using Eqn. (5.3) [17]. Rotation about the $\mathbf{e}_3^p$ axis must be calculated for each discretized position i along $\mathbf{c}$ as follows:

$$\theta_p(i+1) = \theta_p(i) + (r_o^{-1} - \kappa(i) \cos \theta_f(i)) \|\mathbf{c}(i+1) - \mathbf{c}(i)\| \tan \alpha \quad (5.6)$$

where $\kappa(i)$ is the local curvature of $\mathbf{c}(i)$, and α is the chosen constant fiber angle.

Constant Pitch Method

In the constant pitch (CP) approach a straight-axis helix with radius r_o , fiber angle α_m , and constant pitch is “bent” into a curved-axis helix following path $\mathbf{c}(s)$ using Eqn. (5.4). This transformation results in a constant fiber angle α_m at the midline $\theta_f = \pm 90^\circ$, but curvature-dependant fiber angles around the rest of the circumference that vary from $\alpha_i(s)$ on the inside of the FREE’s curvature to $\alpha_o(s)$ on the outside of the FREE’s curvature. In the constant pitch method, a fiber angle of the undeformed FREE can be expressed anywhere on the FREE as:

$$\alpha(s, \theta_f) = \arctan \left(\frac{\tan \alpha_m}{1 - \kappa(s) r_o \cos \theta_f} \right). \quad (5.7)$$

Curvature κ is calculated as the length of the derivative of the tangent vector:

$$\kappa(s) = \left\| \frac{d}{ds} \mathbf{e}_3^p(s) \right\| \quad (5.8)$$

By holding pitch constant in the straight-axis “undeformed” helix, we assert that for any point on the helix, the change in height dL with respect to change in angular position $d\theta$ is

constant and related to fiber angle as follows:

$$\tan \alpha_m = \frac{r_o d\theta_p}{dL} \quad (5.9)$$

We can rearrange Eqn. (5.9) and express it for a discrete curve $\mathbf{c}(i)$ as:

$$\theta_p(i+1) = \theta_p(i) + \|\mathbf{c}(i+1) - \mathbf{c}(i)\| \left(\frac{\tan \alpha_m}{r_o} \right) \quad (5.10)$$

for index $i = 0, 1, \dots, n_n - 1$, where $s = Li/(n_n - 1)$.

To define the fiber paths for either method, starting points $\mathbf{f}_i(0) = [\mathbf{f}_i(0)]_{\mathbf{e}^p}$ are first placed at arbitrary and equally spaced angular positions in the xy plane around $\mathbf{c}(0)$. Point-wise changes in θ_p for each fiber are calculated with the user-defined α and Eqn. 5.6, or α_m and Eqn. (5.10). The resulting points are then transformed to the global coordinate system with Eqn. (5.3).

5.3.2 Finite-Element Model Construction

A Python script run in the Abaqus/CAE Python environment uses the FREE geometry described in Section 5.3.1 to build a mesh for FEA. The FREE path is built using wire spline features defined by discrete control points from $\mathbf{c}(i)$. The FREE bladder is a swept solid feature along the path with outer radius r_o and wall thickness t . Fibers are modeled with wire splines defined by points from $\mathbf{f}_i$. End caps are solid cylinders with radius r_o and an arbitrary thickness of 1 mm. Mesh seed sizes for all components are driven by the bladder mesh seed size, which should be small enough to fit at least three elements across the thickness of the bladder. Fiber seed size and end cap seed size must be smaller and larger, respectively, than the bladder seed size in accordance with guidelines for tie constraints, i.e., fixed constraints with no relative motion [85]. Additional independent output nodes are placed on the bladder and along the fibers at regular intervals to be used for measuring FREE displacement during and after loading (Fig. 5.5).

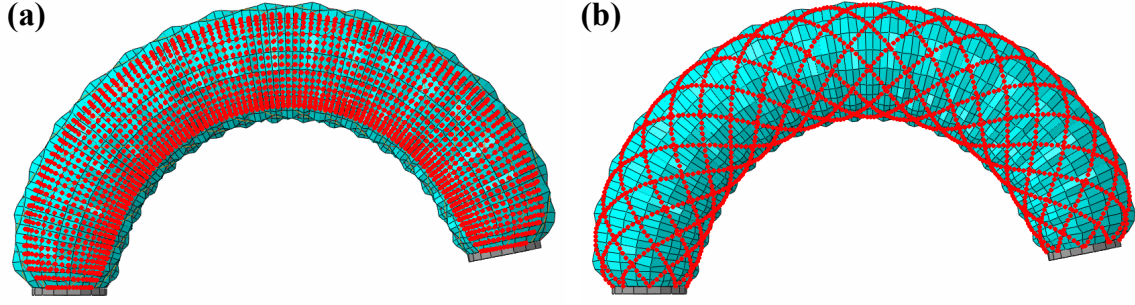


Figure 5.5: FREE FEA setup in Abaqus. Output nodes are positioned at regular intervals (a) on the interior of the FREE bladder and (b) along each fiber.

Elements and materials for the FEA model were chosen to suitably represent the materials used in FREE construction while keeping computation time as low as possible. For this study, the FREE bladder material is modeled as Dragon Skin 10 silicone using a Neo-Hookean hyperelastic model with $C_1 = 42.5$ kPa as determined by Connolly, et. al [10], but the software allows for the use of a variety of common elastomers with hyperelastic model constants from literature [25]. The bladder mesh uses the solid element type C3D8RH, a reduced integration hybrid eight-node linear brick. The mesh is sized to accommodate two to three elements across the thickness of the bladder. 20-node quadratic elements are more accurate than linear elements but they increase solve time by an order of magnitude, making them impractical for iterative analysis. Reduced integration hybrid elements are recommended for hyperelastic materials that are assumed to be incompressible, such as those used to make FREE bladders [85].

Fibers are modeled in one of two ways, depending on the design of the FREE. Typically, fibers are built from the T3D2 element type, which is a two-node linear truss that only transmits axial loads. Truss elements are given the material properties of steel, representing the inextensibility of the fibers compared to the bladder. For some FREEs, including extending FREEs with high fiber angles, the FEA model tends to converge more successfully when fibers are modeled as Kevlar threads using material properties published in [10]. These Kevlar fiber meshes use the element type B32H, which is a hybrid quadratic shear flexible beam element.

End caps are modeled as steel to represent their relative rigidity compared to the bladder material, and they use C3D8RH elements simply because these were found to reduce computation time compared to other linear elements. Due to their high stiffness compared to the bladder, end caps undergo virtually zero strain during analysis, making the element choice inconsequential in terms of kinematic fidelity.

The last step in FEA model setup is assembly. Part instances are created for all parts. End caps are positioned at each end of the FREE bladder and attached with tie constraints. A fixed boundary condition is applied to the end cap closest to the global origin. Fibers are attached to the bladder with a tie constraint. Bladder and fiber output nodes are attached to their respective parts with tie constraints. Pressure p_f is applied linearly and uniformly to the interior surfaces of the bladder and end caps. Lastly, the FEA job is initialized, making the model ready for analysis.

5.3.3 Post-Processing

A second script executed in the Abaqus/CAE Python environment calculates displacement data for all bladder and fiber output nodes for all frames in the completed simulation. The data are saved locally to be processed by a third Python script, executed outside of Abaqus in a Python integrated development environment (IDE) or directly from the terminal or command line. This step is necessary to make use of Python modules that are only available for the current version of Python (3) and therefore cannot be used within the Abaqus Python 2 environment.

FREE centerlines are calculated for each load increment using the bladder output nodes. The output nodes are defined in circumferential sets, providing approximately planar boundary nodes for perpendicular cross sections along the full length of the FREE (Fig. 5.5b). Positions of each circumferential node set are averaged to define a point along the deformed FREE's centerline. Deformed centerline and fiber paths are smoothed for analysis using a Savitzky–Golay filter and spline interpolation [86], [87]. Fiber angles with respect to axial

and circumferential position are calculated from fiber path and centerline tangent vectors. Local curvature is calculated along filtered centerlines for each FEA frame using Eqn. 5.8.

5.3.4 Model Limitations

The FEA model is limited by convergence failure at high strains. An ideal kinematic model would deform FREEs with symmetric fiber angles completely to their “locked” configuration, with $\alpha = -\beta = 54.7^\circ$ [3]. The FEA model presented here typically fails to converge as it approaches this configuration. The rate of a FREE’s deformation with respect to pressure approaches zero as the fiber angle approaches the locked configuration. As pressure climbs higher in real FREEs, eventually it exceeds the level that can be contained by the elastomer bladder in between the fibers, terminating in a “bulging” effect and bladder rupture. In our FEA model, the bladder mesh starts to bulge in between the fibers and elements undergo high levels of strain, resulting in convergence failure. Contracting curved FREE analysis in our software tends to converge up to $\alpha = -\beta \approx 50^\circ$. Low fiber angle FREEs ($\alpha < 25^\circ$) tend to fail earlier in the simulation than FREEs with higher fiber angles, likely due to high strain and large areas of unconstrained bladder between individual fibers, both of which result from the large radial expansion these FREEs undergo.

Accuracy of the FEA model could be improved with higher-order mesh elements, finer meshes, and hyperelastic model validation. The program in this work was created primarily for the study of curved FREE kinematics. Precise force analysis was not a priority, so linear elements were chosen for the bladder mesh instead of more accurate but much slower quadratic elements. For the same reason, meshes were sized to accommodate just two to three elements across the bladder thickness to speed convergence. Additionally, hyperelastic model constants were taken from literature rather than determined experimentally. There is high variance among hyperelastic constants for identical materials studied by different authors, suggesting that any pressure- and force-dependant output from FEA should be validated experimentally [25]. For more accurate force or displacement output with non-normalized pressure input, the

FEA model could be adjusted to use quadratic elements for the FREE bladder with material constants experimentally calibrated across the expected range of strains. Polynomial-type hyperelastic models such as the Yeoh or Ogden model would be more accurate for high-strain FREEs, i.e., those with low fiber angles, than the Neo-Hookean model used in this study [88].

5.4 FEA Model Validation

5.4.1 Methods

Straight and curved FREEs modeled with FEA were compared to physical FREEs with the same parameters (Fig. 5.6). Both tested FREEs had identical lengths, radii, and fiber angles, but one was straight and the other had an initial radius of curvature R_0 (Table 5.1). Physical FREEs were cast from Dragon Skin 10 silicone. Colored silicone markers were added along the centerline of each FREE. The FREEs were pressurized to 80 kPa in 10 kPa increments for three trials, and depressurized from 80 to 0 kPa for three trials. Each trial was recorded by a video camera mounted perpendicular to the FREE's plane of motion. Marker positions were extracted from each video with a video tracking software (Tracker 5.1.5, Open Source Physics) and used to plot and analyze the centerline path of the FREEs at each pressure.

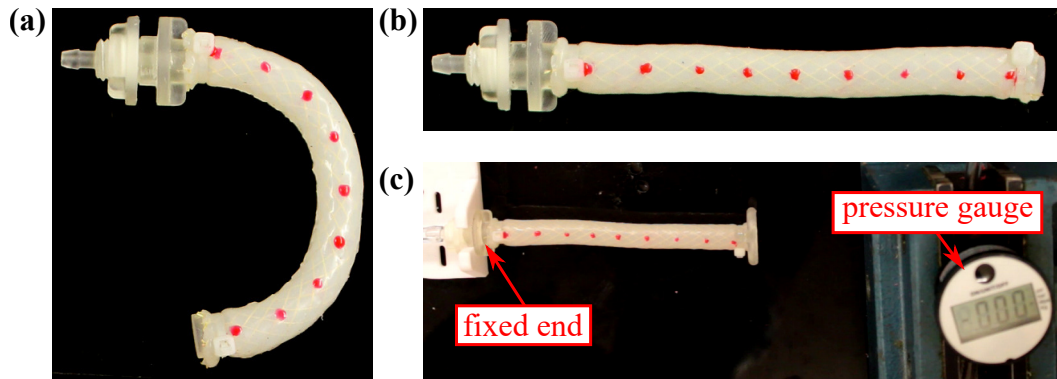


Figure 5.6: (a) Curved and (b) straight FREEs used for experimental validation of curved FREE FEA model. (c) Straight FREE in test setup for experimental validation. FREE displacement was recorded using static marker positions during pressurization in 10 kPa increments. 3D printed fittings allowed the unconstrained ends of the FREEs to slide with low friction.

5.4.2 Results

The FEA model was generally accurate in predicting the the kinematics of both experimental FREEs. FEA predictions for curved FREE axial stretch ratio were within 3% of the experimental measurements for all tested pressures (Fig. 5.7a). For the straight FREE, the FEA model overpredicted axial stretch ratio by a maximum of 6% at 50 kPa. Centerline displacement of the FEA and experimental curved FREEs matched within 2 mm along the entire length of the FREE for all pressures (Fig. 5.7b). Mean curvature for the experimental curved FREE was evaluated using a circular arc fitted to the marker positions at each pressure. Curvatures were normalized by dividing them by the initial curvature. Normalized curvature for the experimental FREE decreased more linearly with pressure than the FEA FREE, but the two curvatures matched within 4% at the maximum pressure (Fig. 5.7c). Overall, the experimental results indicate that the FEA model is a reasonably accurate representation of the tested straight and curved FREEs.

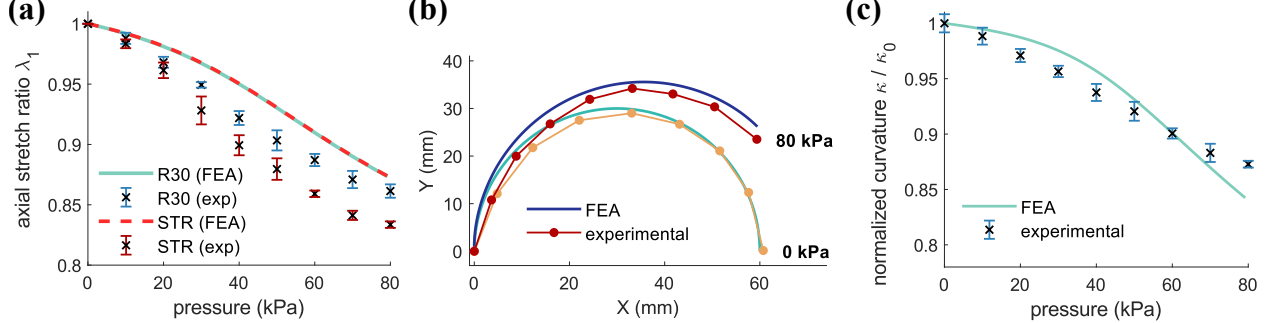


Figure 5.7: (a) Axial stretch ratio with increasing pressure for curved and straight FEA and experimental FREEs (mean $\pm$ SD, $n = 6$). (b) Displacement and (c) curvature with increasing pressure for a FEA and experimental curved FREE (mean $\pm$ SD, $n = 6$).

Table 5.1: Input parameters for FREEs for experimental validation of FEA

Label	FREE Type	Path Type	l (mm)	t (mm)	r_0 (mm)	R_0 (mm)	α ($^\circ$)	β ($^\circ$)
STR	Contracting	Straight	100	2	5	-	30	30
R30	Contracting	Arc	100	2	5	30	30	30

5.4.3 Validation Limitations

Experimental validation in this study was limited by the small sample size and error that was most likely introduced by manufacturing inconsistencies. Measurement of a wider range of starting curvatures, fiber angles, and outer radii would better validate the curved FREE FEA model. It is apparent from the 0 kPa centerline path in Fig. 5.7b that the experimental curved FREE did not follow a precise circular arc path. Small deviations in marker position from the FREE centerline also decreased accuracy of centerline path and curvature estimates for the curved FREE. Inconsistencies in the final coat of silicone can result in asymmetries that affect changes in curvature and overall stiffness. A viscosity-reducing additive (Silicone Thinner, Smooth-On, Inc.) was mixed into the silicone for casting, which may have reduced the stiffness of the experimental FREE.

Experimental evaluation of two- and three-dimensional FREE geometries is limited by the method of measuring physical FREE displacement. Straight FREE displacement can be evaluated from the single-axis motion of a single point at the end of the FREE, which can easily be measured using a ruler, a linear encoder, or motion tracking. Displacement of planar curved FREEs, such as the R30 FREE from this study, require two-dimensional tracking along the whole length of the FREE. Motion capture of markers along the FREE are suitable for this task, but expansion of the FREE bladder during pressurization can offset marker positions relative to the FREE centerline. Three-dimensional curved FREEs would require three-dimensional motion tracking, using one or more 3D sensors such as depth cameras.

5.5 Kinematic Observations of Curved FREEs

5.5.1 Methods

Contracting FREEs with varying fiber angles, radii, and curvatures were modeled with FEA for the purpose of observing curved FREE kinematics. FREEs in this study were modeled

with the constant fiber angle (CFA) method unless otherwise noted. All FREEs had a length of 115 mm. Three FREEs shaped like circular arcs with varied radius of curvature values ($R = 20$ mm - R20, $R = 30$ mm - R30, $R = 60$ mm - R60) were modeled with two different FREE radii ($r = 5$ mm - r5, $r = 8$ mm - r8). The R20 r8 combination was excluded due to convergence issues, which are common when r/R approaches unity. An additional r5 set was modeled for all curvatures using the constant pitch (CP) method. One straight r5 geometry (STR) was also modeled. All FREEs were modeled with fiber angles of $\alpha = -\beta = 25^\circ$, 30° , and 40° (α_m and β_m for CP FREEs). In FEA, FREEs were pressurized to 150 kPa with one end of the FREE fixed. Output data was processed as described in Section 5.3.3. Representative images of deformed and undeformed FREEs are shown in Appendix B.1.1.

5.5.2 Kinematic Similarity to Straight FREEs

FEA results suggested that curved FREEs share similarities with straight FREEs when they deform under pressure. Axial and radial stretch ratios of curved FREEs were nearly identical to stretch ratios of straight FREEs with equivalent fiber angles and radii (Fig. 5.8a–b). All axial and radial stretch ratios for FREEs sharing the same radius and fiber angle were within 2% of each other across all simulated pressures. Like straight FREEs, curved FREE fiber angles asymptotically approach a “locked” fiber angle of approximately 54° as pressure increases [3]. There are many well-established kinematic models for straight FREEs [18]–[21], [24]. These findings suggest that designers can rely on existing models for curved FREE stretch ratios and fiber deformation.

5.5.3 Dependence of Curvature on Fiber Angle

Results indicate that change in curvature for curved FREEs is dependent primarily on fiber angle and FREE radius. We evaluated mean normalized curvature as the mean curvature along the centerline, normalized by the initial curvature of the FREE. All modeled curved FREEs showed decreasing curvature with increasing pressure (Fig. 5.9a). Contracting curved

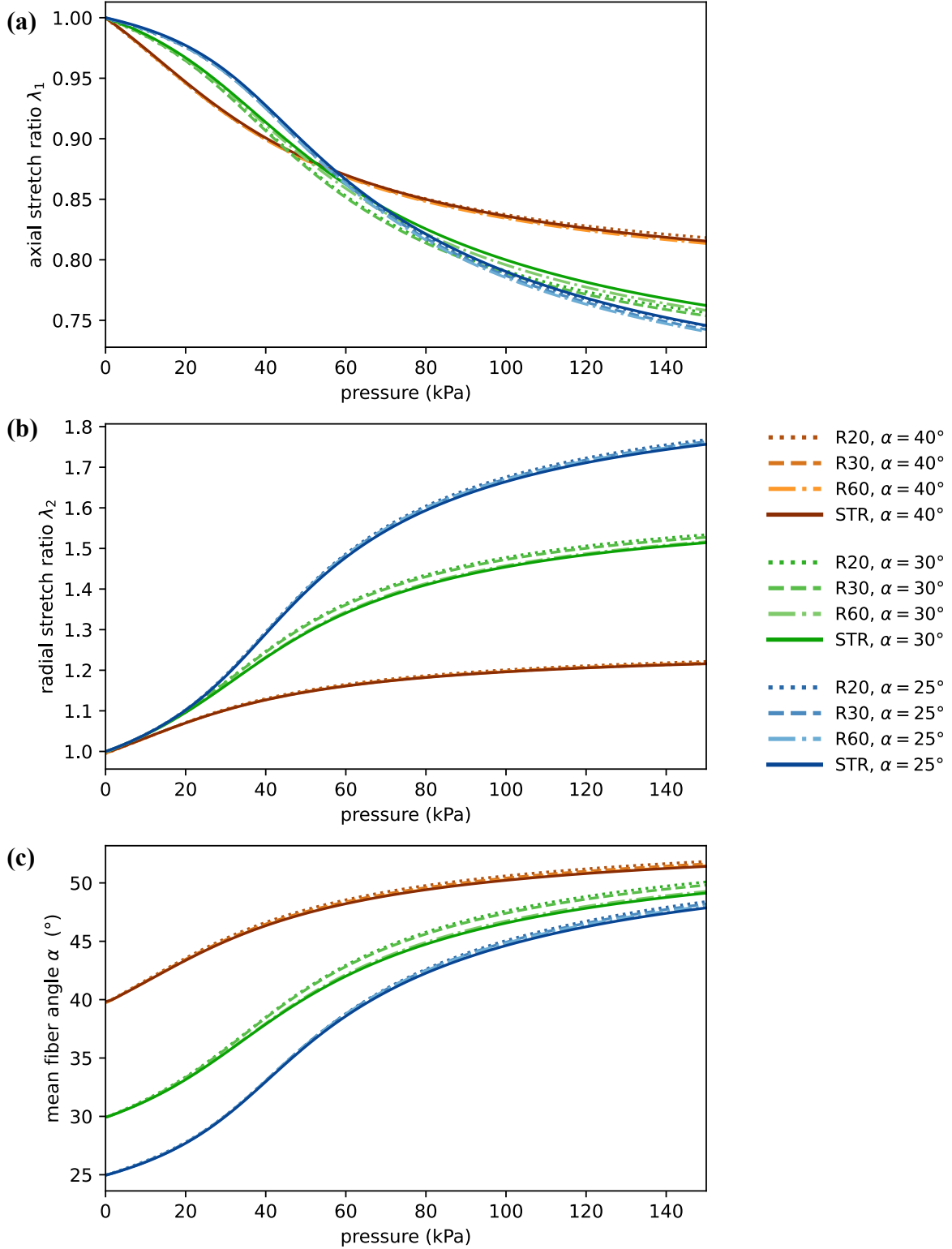


Figure 5.8: Comparison of r5 straight and CFA curved FREEs. (a) Straight and curved FREEs with the same outer radius and fiber angle had nearly equal axial stretch ratios λ_1 and (b) radial stretch ratios λ_2 . (c) Like straight FREEs, curved FREE fiber angles approached a “locked” configuration with $\alpha \rightarrow 54.7^\circ$ as pressure increased.

FREEs appear to decrease in curvature rapidly across the same pressure range during which they undergo large changes in fiber angle (Fig. 5.9a and 5.8c). FREEs with equal fiber angles and radii showed nearly identical mean normalized curvature values across all simulated pressures, indicating that initial pre-curvature has little effect on the relative change in curvature. Contracting FREEs with larger radii tended to decrease in curvature more rapidly than FREEs with the same fiber angle but smaller radii. Increasing FREE radius appears to have a positive scaling effect on curvature change for a given pressure.

Kinematically, normalized curvature was observed to be primarily dependent on fiber

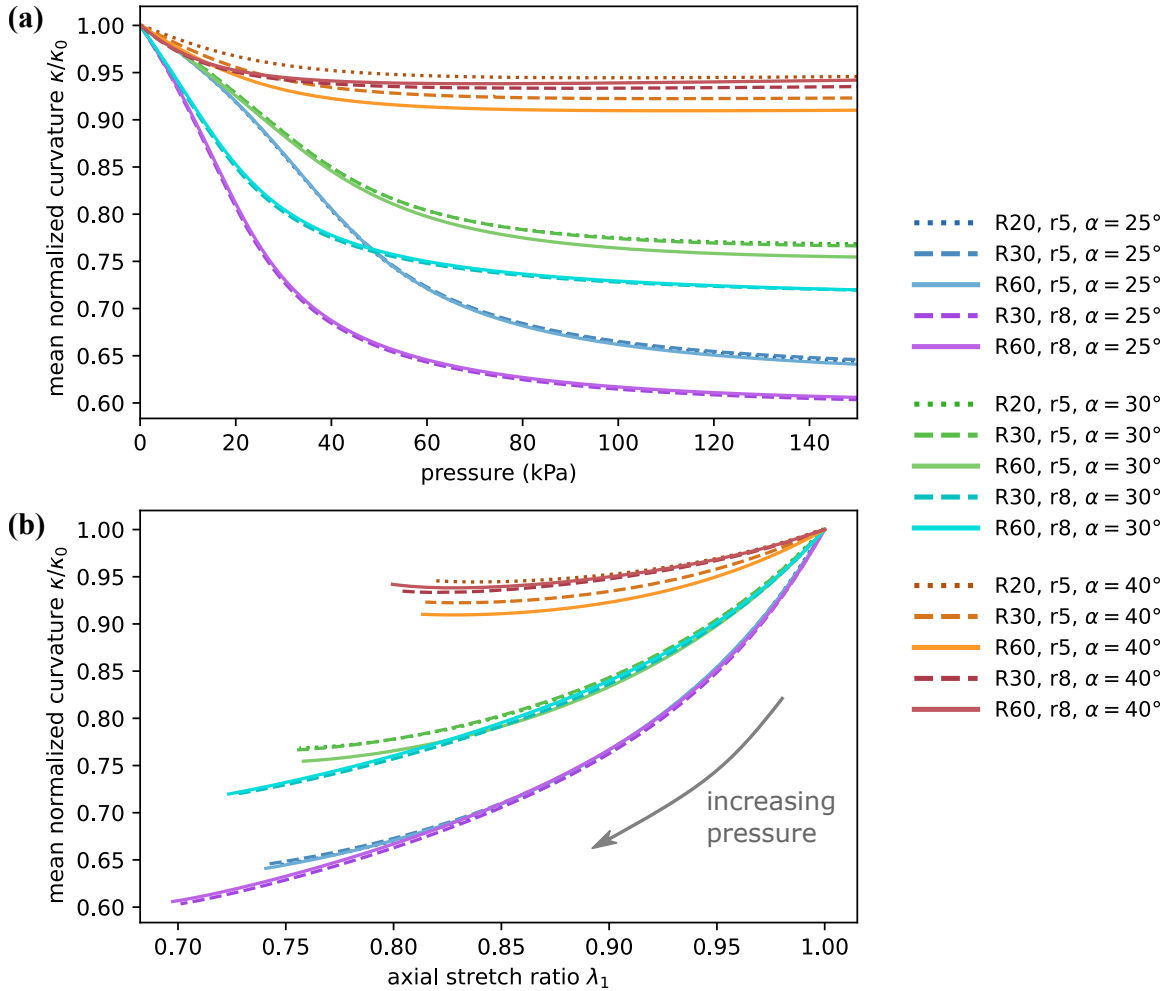


Figure 5.9: (a) Normalized curvature of CFA curved FREEs as pressure increases. FREE radius and fiber angle appeared to be the main factors in curvature change. (b) When plotted as a function of axial stretch ratio, normalized curvature appeared to primarily depend on fiber angle.

angle. All arc-shaped CFA curved FREEs with shared fiber angles experienced the same relative curvature change with respect to normalized contraction, i.e., axial stretch ratio (Fig. 5.9b). Normalizing contraction eliminated the scaling effect from FREE radius, revealing very similar normalized curvature trajectories for all FREEs with shared fiber angles. These clusters of overlapping lines for FREEs with equal fiber angles and differing radii indicate that the kinematic relationship between axial stretch and curvature is the same for any FREEs that share the same starting fiber angle and fiber plotting method. Within the simulated test group, normalized curvature tended to decrease with decreasing fiber angle.

5.5.4 Effects of Fiber Angle Variation

FEA simulations indicated that curvature change is strongly affected by local variations in fiber angle around the circumference of the FREE, even among FREEs with the same midline fiber angle. CP FREEs had the same axial stretch ratios as their CFA counterparts, meaning FREEs made with both methods have stretch ratios that can be approximated by straight FREE kinematic models (Fig. 5.10a). Like CFA FREEs, CP FREEs with shared fiber angles and radii also shared normalized curvature at a given pressure (Fig. 5.10b). However, normalized curvature change followed a different trajectory for CP FREEs than for CFA equivalents with respect to pressure. CP FREEs tended to continue decreasing in curvature as pressure increased, while CFA FREEs asymptotically approached an equilibrium curvature. CP FREEs also showed different normalized curvature trajectories with respect to axial stretch ratio compared to CFA FREEs (Fig. 5.10c). Since all FREE parameters were identical between CFA and CP FREEs aside from the fiber definition method, the large discrepancies in normalized curvature values were likely the result of local fiber angle variations from the CP method of fiber definition, as described by Eqn. 5.7. Unlike for CFA FREEs, normalized curvature did not consistently decrease with decreasing fiber angle. Normalized curvatures for CP FREEs with $\alpha_m = 30^\circ$ were greater than those with $\alpha_m = 20^\circ$, which were greater than those with $\alpha_m = 40^\circ$. This inconsistency suggests an interplay

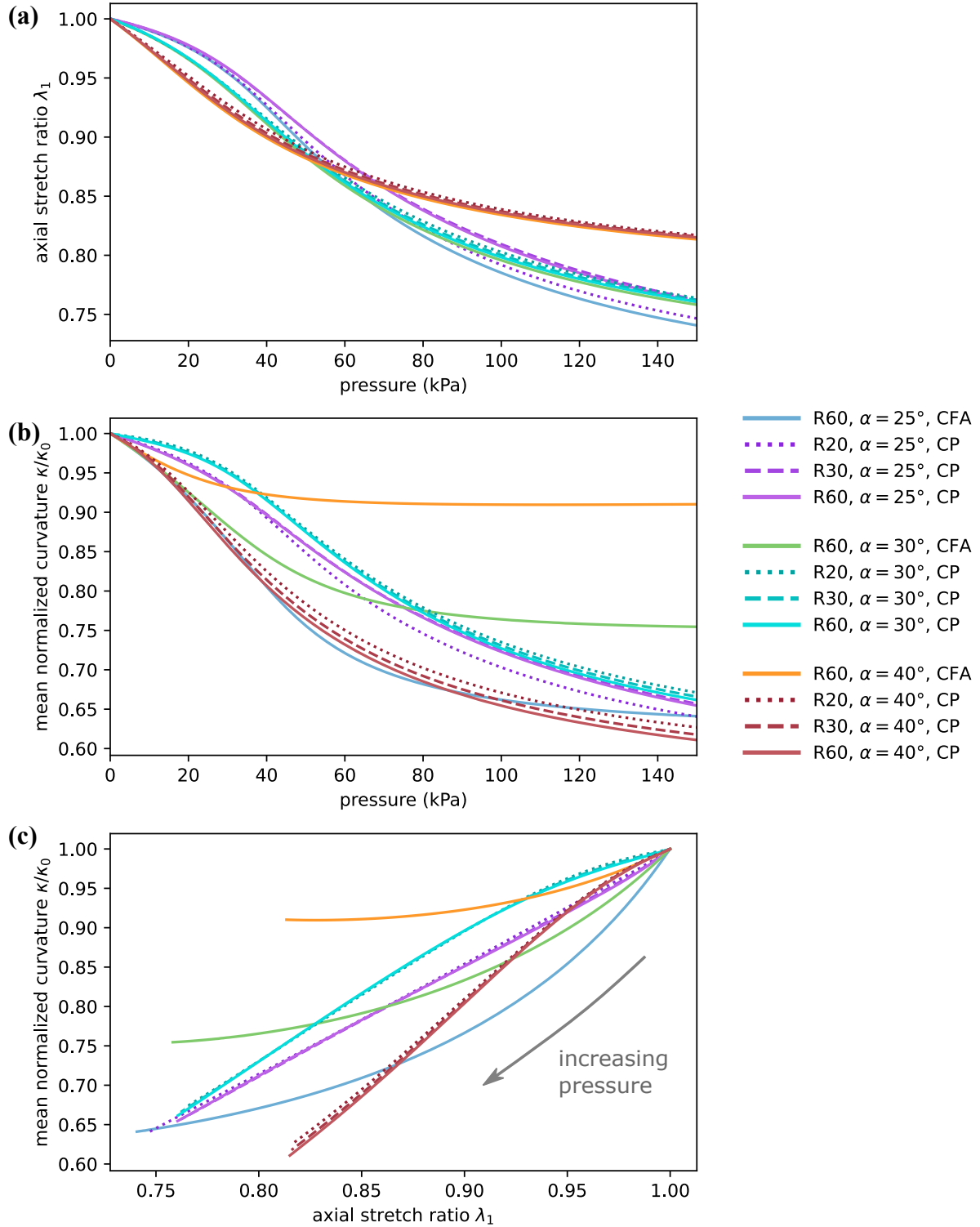


Figure 5.10: (a) Axial stretch ratio of FREEs defined with the constant pitch (CP) method were consistent with representative constant fiber angle (CFA) FREEs. (b) CP FREEs had different changes in normalized curvature than CFA FREEs. CP FREEs that shared fiber angles had the same normalized curvature. (c) Normalized curvature for CP FREEs was dependent on fiber angle when plotted against axial stretch.

between opposing forces resulting from average or midline fiber angle, and additional local fiber-angle dependent forces from varying fiber angles around the circumference of the FREE.

5.6 Future Work

5.6.1 Controlling Curvature with Variable Fiber Angles

The observations in Section 5.5.4 suggest that it is possible to modulate the change in curvature of a FREE by adjusting its local fiber angle in different circumferential positions. FREEs have been designed previously with multiple fiber angles to produce bending, but only with discrete regions of fibers [89]–[91]. Point-wise fiber angle definition could offer much finer control of FREE bending kinematics. Future work should use the point-wise fiber angle definition from the CFM method in Section 5.3.1 to explore variable fiber angles as a means of curvature control. Fiber angles can be adjusted by replacing the constant α in Eqn. 5.6 with a variable fiber angle that is a function of Frenet circumferential position $\alpha(\theta_f)$.

5.6.2 Additional Validation

To verify accuracy of the FEA model, additional experimental validation trials should be conducted. The samples should include FREEs with different radii, curvatures, and fiber angles. Improved accuracy over the methods in Section 5.4.1 could be achieved with the use of a 3D depth camera to capture experimental FREE displacement. Computer vision techniques could also be leveraged to track experimental FREE fiber angle and curvature.

5.6.3 Code and Process Improvements

The program developed in this work has many potential areas for expansion. The code and the FEA settings should be optimized to reduce computation time and maximize convergence success. Implementing one-step modeling, analysis, and post-processing with a user-friendly

graphical user interface would improve accessibility and efficiency. Parametrizing the curved FREE design process has provided a platform for design optimization algorithms. Curved FREE bladder and fiber geometry are mathematically defined in the program. This capability should be used to interface with Python environments in digital modeling software such as Blender for automatic 3D model generation. The procedurally generated 3D models could then be used for mold modeling or 3D printing.

5.7 Conclusions

In this study, a custom program was developed to procedurally generate curved FREE geometries, use the geometries to define FEA models, and post-process the results for the purpose of exploration of curved FREE kinematics. Accuracy of the deformation of the FEA model was validated against experimental results. Through iterative simulation of curved FREEs, it was determined that averaged kinematic metrics like axial stretch ratio and radial stretch ratio were the same for curved FREEs and straight FREEs. Normalized curvature was found to be primarily dependent on FREE fiber angles, including average (CFA) or midline (CP) fiber angles and local variations in fiber angle with respect to circumferential position. Future work should include improvements to the custom program, additional experimental validation studies, and further exploration of curved FREE kinematics with circumferentially-variable fiber angles.

Chapter 6

A 3D Printer for Fabrication of Soft Actuators

Abstract

Soft robot design possibilities can be expanded with the introduction of multi-material 3D printing of soft materials. Soft pneumatic actuators rely on strain-limiting features embedded in highly elastic matrices to define their kinematics, and they require the inclusion of empty chambers for pressurization. Current soft actuator manufacturing methods are time-consuming and limited to simple geometries. 3D printed elastomers could enable new soft robot design features such as complex networks of pneumatic chambers, and active materials with embedded actuators. Advancing the field of soft actuator design requires a 3D printer that is capable of printing (i) highly elastic materials, (ii) stiff but flexible materials, and (iii) a sacrificial support material in a single part. There have been recent advances in the field of soft material 3D printing, but these printers either fail to meet all three criteria or are prohibitively expensive. This project seeks to expand the possibilities for soft robot design through the development of a low-cost multi-material direct-ink-writing (DIW) 3D printer.

6.1 Introduction

Biological actuation systems often serve as inspiration for soft robots, but it is challenging to replicate the complex geometry and material compositions found in nature using traditional manufacturing methods [92]. Soft pneumatic actuators commonly employ soft and stiff elastomers in tandem to control strain during pressurization [2]. Actuators embedded in or bonded to soft matrices can provide continuous movement simulating contracting biological forms, but design iterations are costly due to multi-step fabrication processes with manual assembly [26] (e.g., Chapter 4). Manual techniques such as casting are labor intensive and impose design constraints to accommodate mold removal. Soft material 3D printing has the potential to enable far more complex soft actuator designs than those made with traditional manufacturing methods.

Commercial soft material 3D printing is in its infancy. Commercially available thermoplastic elastomer filaments for fused deposition modeling (FDM) printing, elastomer resins for stereolithography (SLA) printing, and most soft photopolymers, such as those for Stratasys’s polyjet printers, are too stiff for use in pneumatic soft actuators [93]–[97]. SLA technology does not allow for multi-material printing, i.e., printing multiple materials in a single part, which is necessary for useful features such as selective strain-limiting structures and complex internal chambers. The softest commercially available 3D printing material, Stratasys’s TissueMatrix, shows some promise for high strain applications like soft robotic actuators, but it can only be used with a printer that currently costs hundreds of thousands of dollars [98], [99].

There have been many recent advancements in soft 3D printing technology with a focus on soft robotics [100]. A popular method of printing soft actuators is direct-ink-writing (DIW), in which elastomer “inks” or “resins” are deposited directly onto the printer’s build plate. The method of curing varies between printer designs. Schaffner et al. have developed photosensitive silicone inks for DIW printing, but the inks were only validated on expensive bioprinters, costing in excess of \$100,000 [32]. Walker et al. developed a method for printing

with thermoset silicone by controlling reaction kinetics and transient rheology, but printed structures were limited to a single material without support structures [83]. Joseph et al. developed silicone / epoxy hybrid resins for DIW printing of custom-tuned elastomers, and demonstrated their printer’s capabilities with soft robotic components [82]. However, these resins were printed one at a time in each assembly, and there was no support ink to enable large overhangs or pneumatic chambers in their prints.

Developing a multi-material elastomer 3D printer while keeping costs low and operation simple is a challenge. Pusch et al. developed a low-cost stepper motor-driven syringe pump extruder for desktop 3D printers, but our prior attempts indicate the Bowden-style extrusion of their printer is not well-suited for highly viscous DIW inks [101]. A previous iteration of our 3D printer used a Bowden-style extrusion system with a high pressure syringe pump (Fusion 6000, Chemyx Inc.), but back pressure generated by the viscous silicone ink made precise flow control impossible (Appendix C.2.1). Muthusamy et al. presented a promising and simple solution for supporting silicone DIW inks with their hybrid DIW / FDM printer that used soluble polyvinyl alcohol (PVA) filament in a commercial FDM extruder coupled with a pneumatic air syringe for extruding silicone DIW inks [102]. The photosensitive silicone inks developed by Schaffner et al. were used to produce effective multi-material soft actuators, but our adaptation of these inks in an earlier version of our printer came with high material costs, inconvenient ink preparation protocols, and expensive equipment requirements [32]. For details about the previous iteration of our printer, see Appendix C.2.2.

The aim of this project is to develop a low-cost multi-material 3D printer for elastomers, primarily for the purpose of soft actuator research. Printing soft and stiff silicone inks with a water-soluble support material will allow for the design and fabrication of soft robot components with strain-limiting features and internal pneumatic chambers. The printer in development is a commercial FDM model converted for DIW extrusion of custom-tailored silicone inks and a water-soluble support material from commercial air syringes. The inks make use of readily-available commercial elastomer products to simplify ink production and

printer operation. The ink curing mechanism is room temperature vulcanization (RTV) with an extended pot life to allow time for printing. A key area of focus during ink development was rheological characterization and tuning, which are necessary steps for all DIW inks to ensure they flow during extrusion, but remain static until cured [32], [82], [83], [103].

6.2 Printer Design

A commercial FDM 3D printer (Lulzbot TAZ 6, Fargo Additive Manufacturing Equipment 3D, LLC) was converted into a DIW printer for silicone inks through hardware and software modifications (Fig. 6.1). Mechanical hardware modifications included replacement of the extrusion system to accommodate DIW inks. Accompanying electronic hardware modifications made use of unused digital input/output (I/O) pins throughout the printer’s motherboard (RAMBo v1.4, Ultimachine), which consists of a microcontroller (ATmega2560, Arduino) and auxiliary circuits for the printer’s stepper motors, heaters, fans, limit switches, and LCD screen. Software modifications were necessary to set printer parameters and control the modified extrusion system.

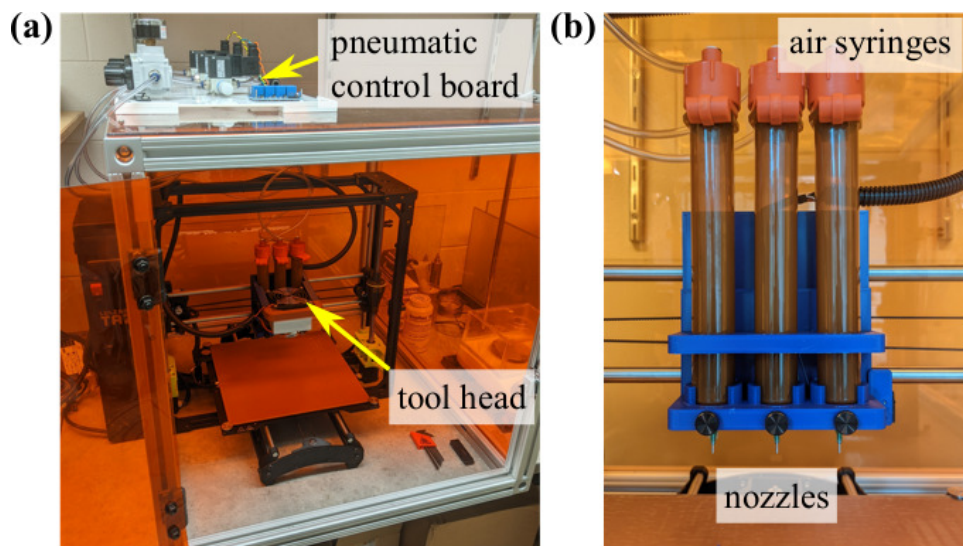


Figure 6.1: (a) Silicone DIW 3D printer. (b) Custom DIW tool head with air syringes.

6.2.1 Hardware

The FDM 3D printer's tool head was replaced with a 3D printed bracket to accommodate up to three 50 cc air syringes (JG50A-15-UVA-WK, Jensen Global) (Fig. 6.1b). Each syringe is fitted with a stainless steel dispensing tip with an inner diameter of 0.84 mm (JG18-0.25HPX, Jensen Global). Air is supplied to the syringes by a pneumatic control system consisting of a small compressor and reservoir (CAT-1P1060SP, California Air Tools), three pneumatic regulators (PPR2-N02BG-2, PneumaticPlus), three solenoid valves and a MOSFET power switch module connected to the printer's Arduino control board. The solenoid valves control extrusion from each needle with pneumatic pressure, and the regulators control maximum flow rate via pressure adjustment (Fig. 6.1a). Solenoid valves are modulated to control extrusion flow rate using pulse-width-modulation (PWM) signals from previously unused output pins on the printer's motherboard.

6.2.2 Software

The 3D printing sequence is managed primarily by two pieces of software. Model slicing software (Slic3r 1.3.0) on a PC converts digital parts into sequential control commands (G-code). The printer's firmware (Marlin 1.0.x), which is installed on the printer's motherboard, interprets and executes the G-code commands to perform printing tasks such as motor movement. Software modifications for the soft material DIW printer included insertion of custom G-code commands with Slic3r and extrusion protocol changes in Marlin. A third software (Pronterface 2.0) is used to directly interface with the printer from a PC and transfer G-code for printing. Slic3r, Marlin, and Pronterface are all open-source [104]–[106].

Slicing Software

Slic3r allows users to insert custom G-code commands at various stages of printing. For the silicone DIW 3D printer, this function is used to store nozzle diameters of each air syringe for dynamic flow rate adjustment. Slic3r is also used to set up standard printer parameters

such as layer height, number of extruders, i.e. air syringes, offsets between extrusion nozzles, and nozzle diameters.

Printer Firmware

Marlin is written in C++ and can be easily modified to include custom protocols. The two files used to modify the DIW printer’s movement control were the “planner”, where incoming G-code movement commands inform algorithmic decisions about motor velocity and speed, and the “stepper”, where the decisions from the planner are converted into electrical signals to the printer’s motors (Fig. 6.2). The original extrusion system used a stepper motor to feed filament into a heated nozzle. Since the stock system was replaced by a pneumatic extrusion system in the modified printer, the stepper file was modified to deactivate signals to the extrusion motor pins. The planner file was modified to respond to incoming G-code extrusion commands by controlling PWM output pins connected to extrusion solenoid valves. The original firmware adjusts the extrusion motor speed based on the X-Y movement of the gantry. The modified system adjusts rate of extrusion by calculating the corresponding flow rate and sending an appropriate PWM signal to the extrusion solenoid valves, modulating pneumatic pressure and therefore flow rate from the corresponding air syringe. Pressure-flow rate relationships were determined experimentally. Additional modifications to Marlin were made to enable G-code commands to multiple extruders and to avoid errors caused by the modified extrusion system. All Marlin code modifications are included in Appendix C.1.2.

6.3 Ink Development

The silicone DIW 3D printer uses soft and stiff silicone-based inks and a water-soluble support ink. All components are commercially-available, low-cost, and safe to handle with basic personal protective equipment.

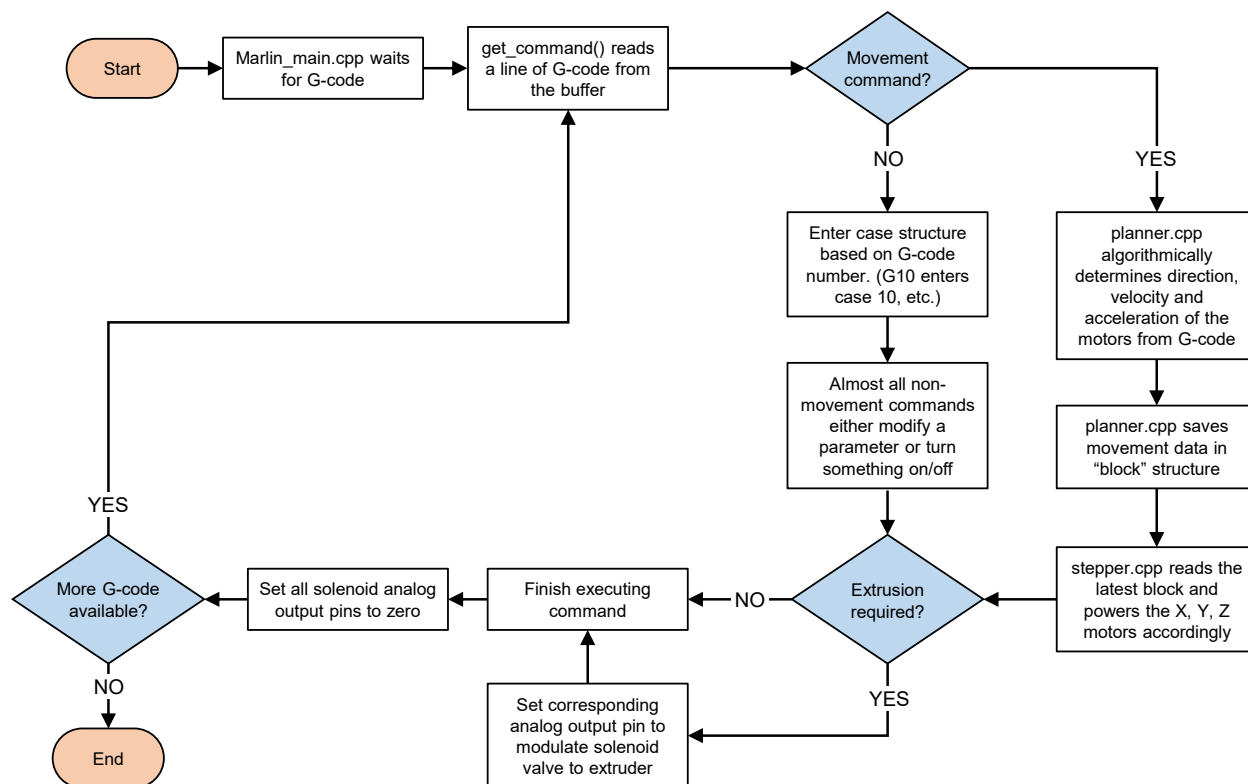


Figure 6.2: Flowchart of 3D printer firmware logic when interpreting G-code, including modified extrusion commands for pneumatic extrusion.

6.3.1 Ink Composition

Soft and stiff inks are primarily composed of two-part room-temperature-vulcanizing (RTV) silicone products, Ecoflex 00-30 (Smooth-On, Inc.) and Sylgard 184 (Dow Inc.), respectively. A commercial cure retardant (SLO-JO, Smooth-On, Inc.) is used to increase the soft ink's pot life, which is the amount time the silicone can be molded or extruded before it begins to polymerize. This extended pot life serves as the available time for printing after the inks are mixed. Joseph et al. used the same approach for an Ecoflex-based ink for their soft materials DIW printer [82]. Both inks have a pot life of 180 minutes, so each batch of ink must finish extruding within that time frame. A rheological modifier is added to both inks to introduce shear thinning and yield stress, as discussed in detail in Section 6.3.2.

The support ink is modeled after the work of Schaffner et al., who developed a stiff water-soluble gel to support their photosensitive DIW silicone inks using kappa-carrageenan,

hyaluronic acid, and hydrophilic fumed silica in distilled water [32]. Support ink preparation methods and component quantities can be found in [32].

6.3.2 Silicone Ink Rheology

DIW 3D printing inks require specific rheological properties to ensure they will flow during extrusion but remain stable on the build plate after extrusion. Viscosity η measures the ink's resistance to flow at a given shear rate $\dot{\gamma}$. It is advantageous for DIW inks to have viscosity that decreases with increasing shear rate. This behavior, known as shear thinning, allows the inks to flow when forced through an extrusion nozzle, but thicken when they exit the nozzle onto the build plate. Additionally, DIW inks should have sufficiently high yield stress σ_y to ensure good shape retention before curing. In a rheological context, yield stress is the required stress for the inks to transition between solid and liquid behavior. During 3D printing, the inks must resist the forces of gravity and forces generated by acceleration of the build plate to maintain part accuracy.

Rheological Modifiers

Two rheological modifiers were considered for the silicone inks in this study: hydrophobic fumed silica and a commercial silicone thickener (Thi-Vex, Smooth-On Inc.). Both of these materials have been used previously to introduce shear thinning and yield stress in silicone-based DIW inks [32], [82], [83]. For rheological testing in this study, base soft inks were mixed with several concentrations of fumed silica and Thi-Vex (Table 6.1). Stiff inks were mixed only with Thi-Vex. Unmodified soft and stiff control inks were used for baseline measurements. The testing in this section was conducted on inks intended for photopolymerization (Appendix C.2.2). However, the photosensitive inks and the inks proposed in Section 6.3.1 are primarily composed of the same RTV silicone products (Ecoflex 00-30 and Sylgard 184 for soft and stiff inks, respectively). Therefore, the effects from the rheological modifiers discussed in this section are expected to be consistent for both types of ink.

Table 6.1: Ink sample compositions for rheological testing

Ink Type	Rheological Modifier	Concentrations
Soft	control	-
	fumed silica	6 wt%, 10 wt%, 14 wt%
	Thi-Vex	1.5 wt%, 3 wt%, 6 wt%, 10 wt%
Stiff	control	-
	Thi-Vex	2 wt%, 5 wt%

Methodology for Assessing Ink Rheology

Steady shear flow tests were conducted on each ink sample to measure viscosity and stress across a range of shear rates. Tests were conducted with a rotational rheometer (Discovery HR-3, TA Instruments) fitted with 20 mm diameter stainless steel parallel plates with a 1000 μm gap. All tests were also conducted with 800 and 1200 μm gaps to verify the absence of slip; measurements were consistent for all tested gaps. All steady shear flow tests were performed from high-to-low shear rates. All tests were conducted at 25 C.

Measurements from the steady shear flow tests were used to identify shear thinning and dynamic yield stress. Viscosity and stress measurements were each plotted on a log-log scale for clear visualization of the data. Materials were considered to exhibit shear thinning if their viscosity decreased with increasing shear rate. For each sample, dynamic yield stress was indicated by a plateau at the low shear-rate regime (leftmost end) of the stress curve. Reported yield stresses values were found by fitting the stress data to the Herschel-Bulkley model [107].

Results and Discussion of Rheology Assessment

Soft ink samples with Thi-Vex and fumed silica both exhibited shear thinning and increased yield stress compared to a control ink with no rheological modifiers (Fig. 6.3a). Low-shear viscosity was 3.2 Pa for the soft control ink, 3300 Pa for the 1.5 wt% Thi-Vex ink, and 5400 Pa for the 14 wt% fumed silica ink. Dynamic yield stress was negligible for the control ink, 40 Pa for the 1.5 wt% Thi-Vex ink, and 53 Pa for the 14 wt% fumed silica ink. Neither

rheological modifier had an appreciable effect on viscosity and stress at high shear rates.

Thi-Vex and fumed silica produced different rheological effects when concentrations were increased in soft inks. Increasing the concentration of fumed silica in the soft ink resulted in increased viscosity and yield stress (Fig. 6.3b). Conversely, concentrations of Thi-Vex exceeding 1.5 wt% appeared only to dilute the ink, with no further improvements to shear thinning or yield stress. This result suggests that there is a saturation point for Thi-Vex, likely related to the proprietary chemical reaction the thickener uses to modify the viscosity of silicone resins (Fig. 6.3c).

Stiff ink testing with Thi-Vex yielded similar results to soft ink testing, with the notable difference that the saturation point appeared to be greater for the stiff ink than for the soft ink (Fig. 6.3d). Adding Thi-Vex to stiff ink introduced shear thinning and increased yield stress. Low-shear viscosity was 10 Pa for the stiff control ink and 2200 Pa for the 5 wt% Thi-Vex ink. Dynamic yield stress was negligible for the stiff control ink and 22 Pa for the 5 wt% Thi-Vex ink.

The shear thinning and yield stress introduced by Thi-Vex was comparable to that of fumed silica in the concentrations we tested. The results suggest that increasing the concentration of fumed silica beyond 14 wt% would further increase low-shear viscosity and yield stress, which could improve printing performance compared to Thi-Vex inks. However, achieving a homogeneous mixture in inks with large quantities of fumed silica is challenging without specialized paste mixing equipment. This practical limitation of fumed silica and the satisfactory performance of Thi-Vex led us to choose Thi-Vex for our inks.

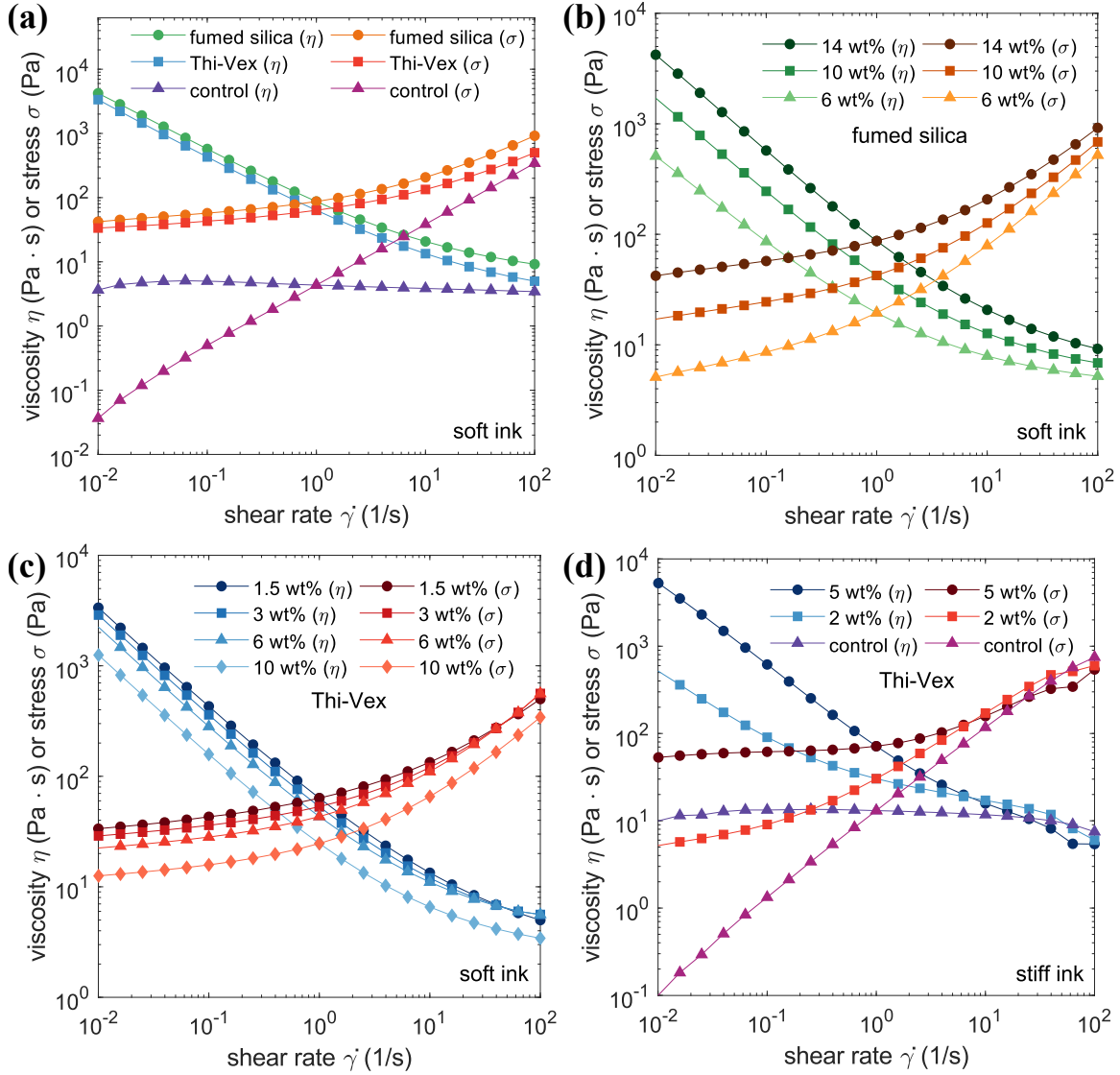


Figure 6.3: (a) Viscosity and stress for soft ink with 1.5 wt% Thi-Vex, 14 wt% fumed silica, and no rheological modifiers. Inks with rheological modifiers exhibited shear-thinning and increased yield stress compared to the control ink. (b) Increasing fumed silica concentration increased viscosity and yield stress in soft inks. (c) Increasing Thi-Vex concentration beyond 1.5 wt% does not increase viscosity or yield stress. (d) Viscosity and stress for stiff ink with 2 and 5 wt% Thi-Vex and an unmodified control. Inks with Thi-Vex exhibited shear-thinning and increased yield stress compared to the control ink.

6.4 Limitations and Future Work

6.4.1 Limitations

Ink Limitations

The primary limitation to the proposed printing method is the time constraint imposed by the RTV silicones' pot life. Both the soft ink and the stiff ink have a pot life of 180 minutes. For a 0.820 mm nozzle with 0.50 mm layer height and a typical DIW print speed of 10 mm/s, these pot life values allow for continuous printing of up to a maximum volume of 45 mL. However, this estimate does not account for support ink printing time or tool head movement time. Large prints will require new syringes with freshly mixed ink to be loaded into the printer after the pot life of each ink has expired during printing.

DIW printed parts have an inherent height limitation for inks that are not cured immediately after being extruded onto the build plate, such as the RTV inks used in our printer. Even with rheological modification as presented in Section 6.3.2, the silicone inks presented in this chapter will eventually yield if too much weight is placed on top of them. The maximum number of layers that can be printed on uncured inks must be determined experimentally. It is possible to mitigate this limitation by supporting the silicone inks on all sides with a stiff support ink. Tall prints may require an intermediate curing step during which the print is paused and the inks are allowed to polymerize. The time limitation imposed by the curing method for these inks offers natural breaks during which curing can be allowed to finish. Curing time can feasibly be sped up by enclosing the printer in a heated chamber.

Equipment Limitations

Manual preparation of viscous inks introduces several potential limitations that can be avoided when using a degassing planetary mixer. Poor homogenization can result in incomplete curing and inconsistent rheology, both of which can be detrimental to the final part's structure and accuracy. Inks that use fumed silica are especially challenging to homogenize by hand.

Manual mixing takes longer than mechanical mixing, wasting valuable silicone ink pot life and therefore decreasing print capacity. Lastly, viscous materials cannot easily be degassed in standard vacuum chambers, increasing the likelihood of air bubbles in the extrusion syringes. Degassing planetary mixers are designed to eliminate this problem and achieve excellent homogeneity using high-speed rotation.

6.4.2 Future Work

Several hardware and software modifications must be completed for the silicone DIW printer to be operational. The current system's solenoid valves and power switching module have insufficient operating frequencies for precise PWM control, so they should be exchanged for faster valves and compatible MOSFET switches. Extrusion commands in the firmware code should be updated to send the correct PWM duty cycles to each solenoid valve to achieve the desired extrusion flow rate.

The relationship between pneumatic supply pressure, solenoid duty cycle, and nozzle flow rate should be determined experimentally. The following is a possible experimental protocol. Suspend an air syringe with each ink over a graduated cylinder placed on a precision scale. Set the duty cycle to the syringe's solenoid valve to 100% and increase the supply pressure until the flow rate reaches $4.2 \text{ mm}^3/\text{s}$. This maximum flow rate corresponds to a 0.84 mm nozzle printing a 0.5 mm thick layer at a linear speed of 10 mm/s, which is the upper end of the typical range of printing speeds for DIW printing. Calculate the flow rate using the time under pressure, the extruded mass, and the estimated ink density. Decrease the duty cycle in 10% increments and hold for 30 seconds at a time to collect flow rate data. Apply a least squares fit to the data to obtain duty cycle as a function of flow rate. Use each ink's equation in the printer's firmware for open-loop control of the solenoid valves.

Once the printer is operational, it should first be evaluated in terms of ink properties and print capabilities. Mechanical tests of printed silicone samples will validate elastic modulus, ultimate tensile strength, and elongation at break against equivalent cast samples. Overhang

and bridging tests can be used to determine print settings that can improve printing speed while maintaining part accuracy. Layer stacking tests will establish the maximum part height that can be printed before failure due to layer compression.

Future upgrades to the 3D printer should focus on resolving the limitations discussed in Section 6.4.1, as well as improving printer capabilities through advanced materials. Combining DIW and FDM extrusion methods as demonstrated by Muthusamy et al. would introduce a much more rigid polyvinyl alcohol (PVA) support material [102]. PVA supports could be used to contain uncured RTV inks more securely, thereby enabling taller prints. Heating the printer’s enclosure would speed up silicone curing times. Joseph et al. developed silicone / epoxy hybrid inks to achieve a broad range of material stiffnesses in their DIW 3D printed parts [82]. Inks such as these should be used with the 3D printer in this study to increase the design possibilities for printed soft actuators.

3D printing of architectures of soft pneumatic actuators will demonstrate the utility of the printer for creation of novel soft robot components with complex geometry. Inclusion of stiff silicone features to act as “fibers” in soft pneumatic actuator designs will allow for a broad range of actuator kinematics [3]. Actuator networks embedded in curved structures [26] and actuators with customized three-dimensional curved paths (Chapter 5) can be among the planned areas of exploration.

Chapter 7

Conclusions

Linear fiber-reinforced elastomeric enclosures (FREEs) have been well-documented and used in many innovative applications, but there is a vast and relatively unexplored design space accessible when one deviates from point-to-point force generation with single actuators. Output performance and kinematics of FREEs can be tailored using novel architectures and curvilinear geometries. These areas of exploration offer exciting new pathways for soft robot design, with the aim of extending the applicability of FREE actuators to new solutions for unique mechanical challenges. To understand and use new FREE architectures and geometries, new tools must also be developed in the form of predictive models and fabrication methods.

7.1 Contributions

The work in this dissertation advanced the field of soft robotics by exploring new FREE design possibilities, enabling their use with modeling and new fabrication techniques, and demonstrating their utility in novel applications. Chapter 1 provides a brief history of FREEs and a summary of the state of the art, organized into the categories of design, modeling, fabrication, and applications. This framework spans the design engineering solution development process from concept formation through preliminary implementation.

Chapter 2 presented the characterization of novel helical and nested linear contracting FREEs in the context of a soft exoskeleton for revolute elbow joint actuation and stiffening [14]. The helical architecture offered an effective method of increasing joint stiffness by leveraging the compliance of FREEs to form a nonlinear actuation path that is uniquely attainable by a soft contracting actuator. The nested linear architecture demonstrated improved actuator displacement compared to single or parallel FREEs of equal length. When used in tandem, the two architectures improved joint actuation and stiffening beyond the capabilities of either architecture on its own.

Chapter 3 explored pennate and cable-driven nested linear architectures of contracting FREEs to augment shoulder flexion in an exoskeleton application [16]. The pennate architecture was shown to be a means to tailor the “gear ratio” of the pneumatic actuator system without changing manufacturing parameters of the individual FREEs [15], [16]. Linking the FREEs serially via Bowden cables in the nested linear architecture reduced the relative size and rigidity of the assembly compared to an earlier version of the same architecture presented in Chapter 2. The nested linear architecture proved to be more suitable for the shoulder exoskeleton application thanks to its larger maximum displacement. Revolute joint actuation from the FREE architecture was validated on a passive mannequin with anatomically accurate joint center locations, arm segment masses, and arm segment center of mass locations.

Chapter 4 focused on the design and validation of the Soft Active Transseptal Puncture Simulator (SATPS), an application that leverages curved FREEs to simulate cardiac atrial musculature. The SATPS motivated the development of the first curved FREEs, which were necessary to follow the contours of an anatomically accurate soft right atrium. The curved FREEs were able to displace physiological volumes from the right atrium during pressurization, and produced atrial dynamics mimicking a beating heart. Additional subsystems were integrated into the SATPS to produce a realistic training environment for transseptal puncture. Face and content validity of the SATPS were established by experienced clinicians in a pilot validation study. Haptics generated by the pulsing FREEs and the static response of carefully

selected elastomers were considered to be realistic by validation participants. This project demonstrated targeted application of novel FREE geometries to meet an established real-world need.

Chapter 5 introduced a versatile program that interfaces with FEA software to explore two- and three-dimensional curved FREE paths with custom fiber angles and dimensions. Parametrization of curved FREEs during model setup makes it easy to iterate designs and study effects of each parameter. The FEA model accurately predicts FREE kinematics, providing a simple and quick method for assessing curved FREE designs without time-consuming fabrication. Automated post-processing extracts useful output parameters including stretch ratios and centerline curvature. This tool is being used to identify kinematic trends in curved FREEs, building toward predictive analytical models. Prototypes designed with this tool use a mold development process using traditional 3D printing.

Chapter 6 sought to enable the fabrication of complex geometries explored in Chapter 5 through low-cost multi-material elastomer 3D printing. Commercial room-temperature-vulcanizing (RTV) silicone products were modified for use as inks in a hobby-grade 3D printer customized for direct-ink-writing (DIW). The printer is designed to print a soft silicone ink and a stiff silicone ink in a single part, allowing for embedded strain-limiting features, similar to the fibers in FREEs. Rheological modification introduced shear thinning and yield stress to the silicone inks, allowing them to flow from a printing nozzle but hold their shape on the printer's build plate. A water-soluble support ink from literature will enable networks of interior pneumatic chambers [32].

7.2 Future Work

Future work stemming from this dissertation should include additional exploration of FREE architectures such as those in Chapters 2 and 3, continued development of the Soft Active Transseptal Puncture Simulator (SATPS) in Chapter 4, and further development of the

FREE design and fabrication tools in Chapters 5 and 6.

There should be continued investigation of architectures of FREEs for the purpose of tailored force and displacement output. Chapters 2 and 3 demonstrated the ability of FREE architectures to deliver application-tailored performance using basic contracting FREEs as building blocks. Development of the curved FREEs from Chapter 5 can compliment this effort by introducing new building blocks to the architecture design space.

Future improvements to the SATPS in Chapter 4 should be driven by the feedback from experienced validation study participants. The visualization features of the SATPS should be upgraded to include live simulated fluoroscopy and intracardiac echocardiography (ICE) views during all stages of transseptal puncture (TP). The simulated ICE view should be adjustable using a handheld controller representative of a real ICE catheter controller. Additional soft right atria and fossa ovalis inserts should be designed and produced to represent different heart geometries and TP-complicating heart conditions such as redundant and calcified atrial septa. The soft right atrium would also benefit from further reduction of friction to improve haptic realism. The important task of placing a guidewire in the superior vena cava for TP catheter positioning should be possible on the SATPS to increase training utility. A larger SATPS validation study should be conducted after the proposed design changes. In addition to expanded face and content validity assessments, the new study should include construct validation to demonstrate measurable differences between expert and novice TP operators when they use the SATPS, and concurrent validation to test the ability of the SATPS to improve real-world performance metrics for TP.

The curved FREE software presented in Chapter 5 should serve as a foundation for larger-scale design and analysis of curved FREEs. The software should be made more user-friendly with a graphical user interface and one-step mesh generation, FEA, and post-processing. The parametrized FREE model and the automatic post-processing could be built into a design optimization algorithm for curved FREEs. Non-orthogonal and nonlinear geometries such as those that make up curved FREEs are time-consuming to individually model. Digital

modeling software such as Blender can interface with Python, so the geometry definition code in the curved FREEs software could be used for procedural generation of actuator meshes for 3D printing or for conversion into casting molds. Streamlining the digital modelling step of the design and fabrication process would further reduce the barrier to entry for complex FREE design. Iterative exploration and analysis of curved FREEs based on observations from the design tool in Chapter 5 could lead to analytical models that circumvent computationally expensive FEA.

The primary bottleneck in FREE design and application is fabrication, so future work should include continued development of the low-cost soft material 3D printer presented in Chapter 6. The printer should be made operational through implementation of the pneumatic extrusion system and experimental determination of correct flow control signals for the extruder solenoid valves. Future improvements should first focus on mitigating the primary limitations of the 3D printer, including the height limit imposed by slow-curing inks. Combining DIW and FDM extrusion methods as demonstrated by [102] would introduce a rigid polyvinyl (PVA) support material to the printing process. Rather than only using PVA for overhang support, the material could be printed as an exterior mold for the extruded DIW inks, reducing the risk of failure by compression for tall prints. The introduction of a broader range of materials, such as the versatile silicone / epoxy hybrid inks developed by Joseph et al., would introduce more diverse design possibilities [82]

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Appendix A

Soft Active Transseptal Puncture

Simulator: IRB Approval Letters

Notice of Approval: New Submission

April 13, 2021

Principal Investigator	Girish Krishnan
CC	Elizabeth Hsiao-Wecksler, Nicholas Thompson
Protocol Title	<i>Face and Content Validation of a Soft Active Heart Simulator (SAHS) for Transseptal Puncture Training</i>
Protocol Number	21731
Funding Source	UIUC – Jump Arches (1-636952-239024-191100)
Review Type	Expedited 6, 7
Status	Active
Risk Determination	No more than minimal risk
Approval Date	April 13, 2021
Closure Date	April 12, 2026

This letter authorizes the use of human subjects in the above protocol. The University of Illinois at Urbana-Champaign Institutional Review Board (IRB) has reviewed and approved the research study as described.

The Principal Investigator of this study is responsible for:

- Conducting research in a manner consistent with the requirements of the University and federal regulations found at 45 CFR 46.
- Using the approved consent documents, with the footer, from this approved package.
- Requesting approval from the IRB prior to implementing modifications.
- Notifying OPRS of any problems involving human subjects, including unanticipated events, participant complaints, or protocol deviations.
- Notifying OPRS of the completion of the study.

Notice of Approval: Amendment #01

August 13, 2021

Principal Investigator	Girish Krishnan
CC	Elizabeth Hsiao-Weckler, Nicholas Thompson, Sanghyun Shin
Protocol Title	<i>Face and Content Validation of a Soft Active Heart Simulator (SAHS) for Transseptal Puncture Training</i>
Protocol Number	21731
Funding Source	UIUC – Jump Arches (1-636952-239024-191100)
Review Type	Expedited 6, 7
Amendment Requested	• Research team modification
Status	Active
Risk Determination	No more than minimal risk
Approval Date	August 13, 2021 (Amendment Approval Date)
Closure Date	April 12, 2026

This letter authorizes the use of human subjects in the above protocol. The University of Illinois at Urbana-Champaign Institutional Review Board (IRB) has reviewed and approved the research study as described.

The Principal Investigator of this study is responsible for:

- Conducting research in a manner consistent with the requirements of the University and federal regulations found at 45 CFR 46.
- Using the approved consent documents, with the footer, from this approved package.
- Requesting approval from the IRB prior to implementing modifications.
- Notifying OPRS of any problems involving human subjects, including unanticipated events, participant complaints, or protocol deviations.
- Notifying OPRS of the completion of the study.



Office of the Vice Chancellor for Research & Innovation

Office for the Protection of Research Subjects
805 W. Pennsylvania Ave., MC-095
Urbana, IL 61801-4822

Notice of Approval: Amendment #02

January 13, 2022

Principal Investigator	Girish Krishnan
CC	Elizabeth Hsiao-Wecksler, Nicholas Thompson, Sanghyun Shin
Protocol Title	<i>Face and Content Validation of a Soft Active Heart Simulator (SAHS) for Transseptal Puncture Training</i>
Protocol Number	21731
Funding Source	UIUC – Jump Arches (1-636952-239024-191100)
Review Type	Expedited 6, 7
Amendment Requested	Adding a new study site – OSF St. Anthony in Rockford, IL (permission provided)
Status	Active
Risk Determination	No more than minimal risk
Amendment Approval Date	January 13, 2022
Closure Date	April 12, 2026

This letter authorizes the use of human subjects in the above protocol. The University of Illinois at Urbana-Champaign Institutional Review Board (IRB) has reviewed and approved the research study as described.

The Principal Investigator of this study is responsible for:

- Conducting research in a manner consistent with the requirements of the University and federal regulations found at 45 CFR 46.
- Using the approved consent documents, with the footer, from this approved package.
- Requesting approval from the IRB prior to implementing modifications.
- Notifying OPRS of any problems involving human subjects, including unanticipated events, participant complaints, or protocol deviations.
- Notifying OPRS of the completion of the study.

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Appendix B

Curved FREEs Supplemental Information

B.1 Curved FREEs FEA Program

The custom program developed for curved FREE modeling and FEA is open source can be accessed using the link below:

<https://github.com/Nicholas-A-Thompson/Curved-FREEs-FEA>

B.1.1 All Modeled FREEs

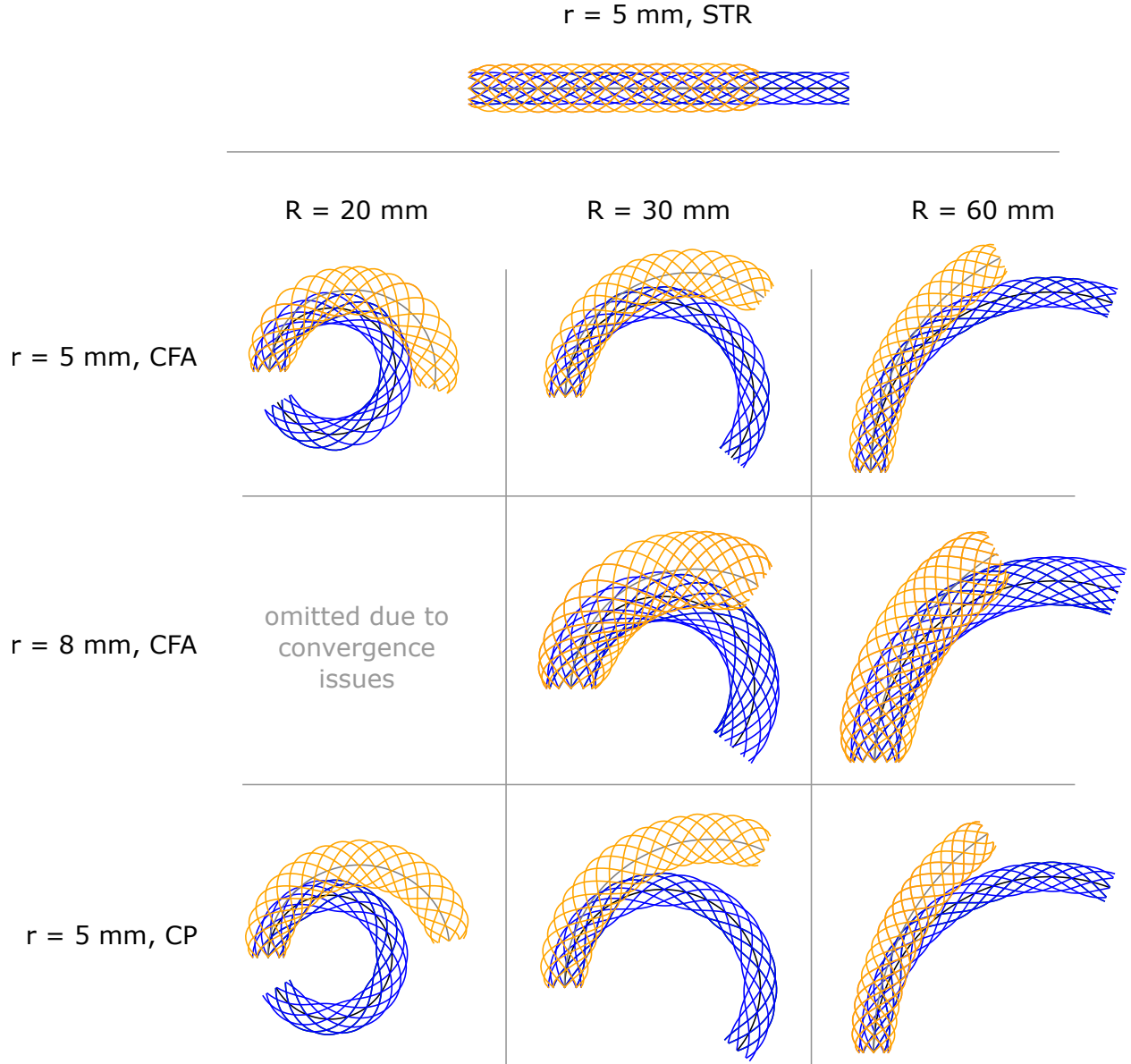


Figure B.1: All FREEs modeled in Chapter 5, Section 5.5. Each FREE was modeled with $\alpha = -\beta = 25^\circ, 30^\circ$, and 40° . FREEs shown here have a fiber angle of 30° .

Appendix C

Soft 3D Printer Supplemental Information

C.1 Software Modification Details

C.1.1 Slic3r Settings

Table C.1: Slic3r custom G-code command in Settings → Printer Settings → Custom G-code

Command Location	Command	Description
Start	M200 Dx.xx Ty	Store nozzle diameter (x.xx mm) and syringe number (y = 0, 1, or 2). Used for flow rate adjustment.

C.1.2 Marlin 1.0.x Code Changes

Configuration.h

- Line 58: Changed number of extruders to 3.

Configuration_adv.h

- Commented line 516 to disable removal of HEATER_2_MAXTEMP to prevent errors. Inconsequential for DIW printing.

Marlin_main.cpp

- Public variable nozzle_area added to store nozzle areas for each extruder.
- G-code command M200 modified to calculate and store the nozzle area for flow rate calculations.

pins.h

- Added dummy HEATER_2_PIN and TEMP_2_PIN assignments to enable third extruder commands.
- Removed line disabling HEATER_2_PIN. Prevents errors when three extruders are used.
- Added dummy E2_STEP_PIN, E2_DIR_PIN, E2_ENABLE_PIN, E2_MS1_PIN, and E2_MS2_PIN assignments to third extruder commands.

planner.cpp

- Declared external public variables that stores extruder nozzle area.
- Added macro definitions to respond to G-code extrusion commands by adjusting corresponding PWM output pins connected to solenoid valves.

stepper.cpp

- Commented line 29 to remove inclusion of temperature.h.
- Commented lines 731 and 735 to disable commands to extrusion motor DIR pins.

- Commented lines 883 and 886 to disable commands to extrusion motor STEP pins.
- Commented line 1235 to disable heater management.

temperature.cpp

- Commented lines 1711, 1726, and 1742 to prevent minimum temperature errors caused by extrusion system conversion.

C.2 Previous Printer Design

C.2.1 Syringe Pump Extrusion

The first iteration of the silicone DIW 3D printer used high-pressure syringe pumps (Fusion 6000, Chemyx Inc.) for each extrusion line, but this approach was found to be incompatible with the viscous silicone inks. Each syringe pump was connected to a dispensing nozzle on the printer’s tool head via semi-flexible perfluoroalkoxy alkane (PFA) tubing. Commands were sent to the syringe pumps via serial communication from the printer’s motherboard using customized firmware. The syringe pumps were able to precisely control flow rate from the syringe, but the 1 m length of tubing and the high viscosity of the ink resulted in back pressure during extrusion. Incomplete degassing and possible compressibility of the ink may also have contributed to this issue. The back pressure caused the flow rate from the nozzle to be less than the specified flow rate from the syringe pump, and resulted in continued extrusion from the nozzle after the syringe pumps were stopped. Serial communication to the syringe pump could not be completed fast enough to withdraw the pump and offset the back pressure during printing. Yirmibesoglu et al. and Walker et. al successfully used a syringe pump for their silicone DIW printer, but their single-material design could be operated with continuous flow from the syringe pump, and a mixing unit attached to the tool head helped contain the back pressure and regulate flow [83], [108]. The pneumatic air syringe extrusion

system presented in Section 6.2.1 can be toggled almost instantaneously using pulse width modulation (PWM) output pins on the printer’s motherboard, eliminating overextrusion and dripping during printing.

C.2.2 Photosensitive Silicone Inks

The original design for this silicone 3D printer used photosensitive silicone inks based on the work of Schaffner et al. [32]. The authors’ soft and stiff silicone inks were composed primarily of Ecoflex 00-30 Part A (Smooth-On, Inc.) and SYLGARD 184 Base (Dow, Inc.), respectively. Both of the original inks used a thiol-based crosslinker and a photoinitiator to polymerize the silicone, along with an “enhancer” to improve ultraviolet (UV) light transmission for inter-layer curing and hydrophobic fumed silica as a rheological modifier. The stiff ink from [32] also included flax fibers for additional stiffening. Schaffner et al. cured their inks with a mercury vapor curing lamp with a broad emission spectrum.

Adapting the inks from [32] to a lower-cost printer proved to be infeasible due to equipment limitations, and incongruous with the low-cost objective for our printer. Our DIW printer was equipped with a custom tool head that included a commercially-available UV LED curing lamp with a narrow light emission peak centered on a wavelength of 365 nm (1DGL-JC-100W-365, Chanzon). To adapt the silicone inks from [32] to our DIW printer, we exchanged the photoinitiator for one with an absorption peak near 365 nm (Irgacure 819, Ciba, Inc.). We also used an alternative rheological modifier (Thi-Vex, Smooth-On, Inc.) based on the results of rheological testing in Section 6.3.2. Flax fibers were omitted from the modified stiff ink due to unavailability. Our printer was designed to extrude a layer of silicone ink, and then shine the UV lamp onto the printed layer to cure it before printing the next layer. Test prints exhibited incomplete curing characterized by a thin uncured exterior layer and poor inter-layer bonding that resulted in delamination. The uncured ink located on the exterior of each printed layer suggested oxygen inhibition from the surrounding air as a possible source of error. It is also possible that the photoinitiator did not create enough

free radicals for crosslinking due to insufficient light emission from the LED curing lamp or incomplete dissolution into the elastomer matrix. Potential solutions to these problems, such as nitrogen submersion during printing for oxygen inhibition, an industrial-strength UV lamp for more effective curing, and a centrifugal paste mixer for better ink mixing, were prohibitively expensive. The photoinitiated inks were abandoned in favor of far the more accessible silicone inks presented in [Section 6.3](#).