DESIGN, FABRICATION, AND APPLICATIONS OF POROUS ELASTOMERS FOR COMPLIANT FLUIDICALLY POWERED MACHINES

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DESIGN, FABRICATION, AND APPLICATIONS OF POROUS ELASTOMERS FOR COMPLIANT FLUIDICALLY POWERED MACHINES

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Open-celled elastomer foams enable many new designs for fluidically inflated, soft machines. These foams allow for simple fabrication of complex shapes containing an embedded fluidic network, without the requirement of laborious molding or assembly necessary for other common fabrication methods.

In the first part of this work, a method is described to produce porous silicone via a lost salt process that allows porosity to be easily tuned. This process also allows the silicone to be molded into a variety of geometries, while retaining an interconnected porous network. Porosity is shown to affect foam mechanical properties and actuator performance. Additionally, the flow rate of air through foam samples is characterized with flow studies and X-ray computed tomography, then compared to a theoretical model. To demonstrate the unique geometries possible with this material system, a heart-inspired, compliant water pump is presented that provides pulsatile flow at physiologically relevant frequencies and pressures.

Next, a buckled polyurethane foam system is described. Actuator characterization reveals that a moderate amount of residual compressive strain augments the applied force or the maximum stroke compared with uncompressed actuators. These actuators are applied in a patient-specific direct cardiac compression (DCC) device design. The compliant DCC device is demonstrated on a porcine heart
and is capable of assisting heart pumping at systolic and diastolic durations and physiological stroke volumes.

Finally, a compliant haptic interface is developed for virtual reality interactions. The interface is demonstrated in the form of an internally porous controller handle that can change stiffness and shape when pressurized, simulating the physical shape and feel of different virtual objects. The elastomeric handle is composed of an internal lattice network which was selected from an array of geometries with corresponding finite element simulations that compare their buckling. As an additional method of interaction, compressions of the handle are demonstrated as a means of user input.
BIOGRAPHICAL SKETCH

Benjamin Mac Murray was born on February 16, 1987 in Exeter, New Hampshire. He grew up in Mankato, Minnesota and attended Iowa State University for his undergraduate education. While at Iowa State, he worked in Professor Michael Kessler’s Polymer Composites lab where he learned several thermal analysis techniques while researching cyanate ester composite materials. He graduated in 2009 with a B.S. in Materials Engineering.

He then worked for several years as a Research Engineer at Cornerstone Research Group in Dayton, Ohio. There he lead and supported several SBIR and STTR materials development programs. He contributed to the development of a variety of novel materials including specialty adhesives, aerospace-grade composites, recyclable resins, shape memory polymers, and self-healing materials. His work led to two patents, covering reversible thermoset adhesives and a fluid absorption and distribution enhancement system.

In 2013, he moved to Ithaca, New York to pursue a PhD in Materials Science and Engineering at Cornell University. Shortly thereafter, he joined Robert Shepherd’s Organic Robotics Lab in the Sibley School of Mechanical and Aerospace Engineering and began his research of foam-based compliant machines. He was aided in this research by his committee members Christopher Ober, an expert in materials and polymer chemistry, Professor Chung-Yuen (Herbert) Hui, an expert of soft matter mechanics, and Professor Jonathan Butcher, an expert in the biomechanics of heart valves, as well as many excellent labmates and support staff. Upon completion of his degree, he will join the research team at 3M in St. Paul, Minnesota.
ACKNOWLEDGEMENTS

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<td>Three Dimensional</td>
</tr>
<tr>
<td>AR</td>
<td>Augmented Reality</td>
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<tr>
<td>BCC</td>
<td>Body Centered Cubic</td>
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<tr>
<td>BS</td>
<td>Bone Surface</td>
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<td>CAD</td>
<td>Computer Aided Design</td>
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<td>Computed Tomography</td>
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<td>DCC</td>
<td>Direct Cardiac Compression</td>
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<td>DMA</td>
<td>Dynamic Mechanical Analysis</td>
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<td>DOF</td>
<td>Degree of Freedom</td>
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<td>ECG</td>
<td>Electrocardiogram</td>
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<td>FCC</td>
<td>Face Centered Cubic</td>
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<td>Finite Element</td>
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<td>FEA</td>
<td>Fluidic Elastomer Actuator</td>
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<tr>
<td>ID</td>
<td>Inner Diameter</td>
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<tr>
<td>KC</td>
<td>Kozeny Carman</td>
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<tr>
<td>MM10</td>
<td>Mold Max 10</td>
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<tr>
<td>MM60</td>
<td>Mold Max 60</td>
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<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
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<tr>
<td>OD</td>
<td>Outer Diameter</td>
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<tr>
<td>PDMS</td>
<td>Polydimethylsiloxane</td>
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<tr>
<td>PU</td>
<td>Polyurethane</td>
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<tr>
<td>Rc</td>
<td>Compression Ratio</td>
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<tr>
<td>SA</td>
<td>Surface Area</td>
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<td>SC</td>
<td>Simple Cubic</td>
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<td>SI</td>
<td>Supplemental Information</td>
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<tr>
<td>STEAM</td>
<td>Science, Technology, Engineering, Arts, and Mathematics</td>
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<tr>
<td>STP</td>
<td>Standard Temperature and Pressure</td>
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<tr>
<td>TGA</td>
<td>Thermogravimetric Analysis</td>
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<tr>
<td>VAD</td>
<td>Ventricular Assist Device</td>
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<td>VR</td>
<td>Virtual Reality</td>
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<thead>
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<tbody>
<tr>
<td>A</td>
<td>cross-sectional area</td>
</tr>
<tr>
<td>d</td>
<td>Thickness</td>
</tr>
<tr>
<td>$D_p$</td>
<td>Distance between pores</td>
</tr>
<tr>
<td>$d_{\text{pore}}$</td>
<td>Mean pore diameter</td>
</tr>
<tr>
<td>E</td>
<td>Young’s modulus</td>
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<tr>
<td>E’</td>
<td>Normal storage modulus</td>
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<tr>
<td>E”</td>
<td>Normal loss modulus</td>
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<tr>
<td>F</td>
<td>Force</td>
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<tr>
<td>G’</td>
<td>Shear storage modulus</td>
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<tr>
<td>G”</td>
<td>Shear loss modulus</td>
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<tr>
<td>L</td>
<td>Length</td>
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<tr>
<td>Q</td>
<td>Flow rate</td>
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<td>r</td>
<td>Radius</td>
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<tr>
<td>$R_c$</td>
<td>Compression ratio</td>
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<tr>
<td>$T_m$</td>
<td>Melting temperature</td>
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<td>$V$</td>
<td>Volume</td>
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<td>$\Phi$</td>
<td>Porosity</td>
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<td>$\rho$</td>
<td>Density</td>
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<tr>
<td>$\mu$</td>
<td>Micro-</td>
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<tr>
<td>$\Delta P$</td>
<td>Pressure differential</td>
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<tr>
<td>$\varepsilon$</td>
<td>Engineering strain</td>
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<tr>
<td>$\dot{\varepsilon}$</td>
<td>Engineering strain rate</td>
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<tr>
<td>$\varepsilon_d$</td>
<td>Densification strain</td>
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<tr>
<td>$\sigma$</td>
<td>Engineering stress</td>
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<tr>
<td>$\eta$</td>
<td>Viscosity</td>
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<tr>
<td>$\Psi$</td>
<td>Sphericity</td>
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<tr>
<td>$\Gamma$</td>
<td>Tear strength</td>
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CHAPTER 1
INTRODUCTION

1.1 Compliant Machine Fabrication

The design of soft robots, defined as robots that are composed of materials with similar moduli to soft biological materials [1], naturally draws much inspiration from biology. The use of low modulus materials allows machine designs that seek to mimic the attributes of soft bodied animals that are often difficult to attain using rigid bodied machines, including gentle interactions with humans and adaptability within unstructured environments [2], [3]. Fabricating compliant machines in the complex biological forms found in nature, however, is no simple task. Compliant machines typically employ a particular class of actuator, Fluidic Elastomer Actuators (FEA), as these provide large strokes and high forces without the need for high electric fields, magnetic fields or temperatures [4] required by other actuators. These actuators consist of an elastomer body and are powered through pressurization of a fluid within an internal chamber. In their simplest form, FEAs are similar to a balloon in which a single elastomeric chamber is able to be pressurized. In their most useful form, machines comprised of FEAs are much more complex; incorporating multiple materials, multiple pressurization chambers, intricate conduit routing, built-in elastomeric valves, and asymmetric chamber geometries in order to program a specific deformed shape. The difficulty in fabricating these actuators stems from the need for multiple and uniquely shaped pressurization chambers necessary to mimic the
functions of soft organisms. Further, when the goal is to match not only the function, but also the form of biological entities, the question arises: How can you fabricate (and eventually manufacture) 3D elastomer geometries with embedded fluidic pathways?

As this question is not easily answered, fabrication has become a strong focus of the field with several recent reviews discussing fabrication methods [1], [2], [4]. In order to form hollow fluidic channels within 3D geometries, researchers have resorted to creative and time-consuming fabrication methods. One common approach is to assemble actuators piecewise from smaller actuator components. Planar actuators are commonly fabricated by first molding an elastomer with an exposed pneumatic network, then sealing that network by adhering additional material to that surface (Figure 1.1). These actuators, often called “Pneu-nets”, have been used to form compliant machines capable of multigait locomotion [5] and gentle gripping [6]. 3D machines fabricated via assembly have used Lego®-like blocks [7], magnetically-aligned modules [8], and interlocking dovetailed tiles [9] as building blocks. While the assembly method of fabrication provides design flexibility, the adhered joints create weak points when pressurized and the time demand for fabrication clearly increases with machine complexity.
**Figure 1.1 Molded pneumatic channel fabrication.** (A) Schematic representation of the soft pneumatic channels, formed by bonding an elastomeric layer (layer 1) to the strain-limiting layer (layer 2). (B) A cross section of a portion of the machine is schematically illustrated at atmospheric pressure (left) and actuated at positive pressure (right). (Inset, left) Top view of the machine and the section removed. (C) An optical micrograph at atmospheric pressure (left) and actuated at 7.0 psi. Scale bar, 3 cm. Adapted from [5].

Another fabrication method is to cast an elastomer around a printed wax form, then remove the wax after the elastomer has cured. This lost wax method enabled hollow chambers within a combustion-powered fluid pump [10], a combustion-powered jumping/rolling machine [11], and a fluidically actuated fish model [12]. Though the lost wax process can form embedded fluidic chambers, the process requires multiple molding and melting/burnout steps that lead to increased fabrication times (as evident in Figure 1.2).
Figure 1.2 Compliant fish tail fabrication process. (A) pour and cure a rubber mold; (B) pour wax cores; (C) combine head constraint, center constraint, and wax cores with tail mold halves; (D) pour rubber into assembled tail mold; (E) melt wax core out of the cured fish tail in an oven; and (F) cook out remaining wax to create desired actuator cavities. Adapted from [13].

Another method to produce FEAs is to 3D print them directly. Examples of this method include actuators printed via Digital Light Projection Stereolithography [14], direct ink writing [15], and multimaterial polyjet printing that forms a functionally-graded machine [16] or bellowed actuators [17]. While 3D printing allows the greatest design flexibility, current commercial printable elastomers typically have dramatically worse mechanical properties relative to commercial castable elastomers and printing processes often require longer durations than casting.

A final fabrication method involves using a custom self-healing elastomer system based on Diels-Alder reversible chemistry that enables origami-like folding of
a flat elastomer sheet to form a hollow, inflatable chamber [18]. This system could repair micro- and macro-scale damage and achieved a 98% healing efficiency of healed storage modulus compared to virgin modulus [19]. While the self-healing material method shows clear promise, its custom chemistry hinders its widespread use.

Elastomer foams (i.e., 3D cellular solids [20]) provide a solution to this fabrication difficulty. Open-celled foams, those composed of an interconnected network of void spaces, intrinsically allow permeation of pressurized air throughout the 3D foam structure. Elastomer foams can be processed in the liquid state, meaning they can be molded into a variety of geometries. Additionally, when the rheology of the liquid is correctly tuned, these liquids can be sculpted in both an additive and a subtractive fashion [21]. Whereas the fabrication methods described above force the machine designer to incorporate embedded pneumatic chambers, foam-based machines allow the designer to mold the chambers directly.

Elastomer foam based machines are also inherently low density as they can exhibit void space volume fractions (i.e., porosity) of 0.98 or higher, enabling lightweight (and buoyant) devices. Porosity (Φ) and density are related according to

**Equation 1.1** [22]

$$\Phi = 1 - \frac{\rho_f}{\rho_s}$$

where $\rho_f$ is the foam density and $\rho_s$ is the density of the solid component of the foam.

The relative density $(\rho_f/\rho_s)$ is also useful to estimate the small strain Young’s modulus, as in **Equation 1.2** [23]

$$\frac{E_f}{E_s} = C_1 \left(\frac{\rho_f}{\rho_s}\right)^2$$

5
where $E_f$ is the foam Young’s modulus, $E_s$ is the Young’s modulus of the solid component of the foam, and $C_1$ is a constant. In many ways high porosity is desirable for foam based actuators as such devices will be light weight, contain a highly open pneumatic network enabling rapid actuation, and have a high compliance allowing large deformations with low inflation pressures. A tradeoff exists, however, in that fracture properties of foams also depend on porosity. Specifically, fracture toughness decreases and fatigue behavior (i.e., crack growth per cycle) increases as porosity increases [23].

Finally, the foam void space can also provide new capabilities to compliant machines. As an example, open cell foams provide an interconnected network which can be impregnated with additional materials imparting new properties to the bicontinuous composite system. These new properties could include shape memory [24] or tunable stiffness behavior [25], as well as increased conductivity or transparency, provided that the second material is able to substantially impregnate the foam. As another example, when considering fluidic inflation, both the porosity and tortuosity of the foam could be used to dictate the rate of inflation and degree of deformation of an actuator. High porosity foams could be molded in contact with low porosity foams. In the resulting actuator, the high porosity foam would deform faster and to a greater extent than the lower porosity regions. As a final example, the void space in foam allows compression with an essentially zero Poisson effect, that is, if the foam is compressed uniaxially it does not significantly deform in the other dimensions. I expand upon this behavior in Chapter 3, showing how it can be useful in augmenting actuator performance.
In this dissertation, I examine the use of porous elastomers as the basis of compliant machines. Each chapter describes a unique material system, porous structure, and potential application. The sum total of this work seeks to answer the question: How can compliant machines benefit from the implementation of porous elastomers? Over the course of the projects presented here, I investigated multiple materials and foam-forming methods, including a polydimethylsiloxane (PDMS) porous elastomer formed through a lost salt process, a liquid cast polyurethane that foams through CO₂ generation, and a printed polyurethane formed through a stereolithography process. I also explored multiple structures and levels of structural order, including a stochastic pore network formed from randomly shaped salt crystals, a network of roughly packed spherical pores, and an entirely designed lattice network. Finally, I also pursued multiple potential applications enabled by these materials systems, including a compliant fluid pump that mimics certain metrics of the human heart, a patient-specific cardiac assist device, and a shape-shifting game controller handle. The sum of these investigations provides insights into the benefits and drawbacks of each material system and fabrication method in terms of both capability and manufacturability.

1.2 Dissertation Scope and Organization

In Chapter 2, I report the first use of elastomer foams as fluidically powered actuators. I adapt a lost salt process to form PDMS foams using ammonium bicarbonate (\((\text{NH}_4)\text{HCO}_3\)) as the thermally decomposable porogen. Using this simple fabrication process, I examine how porosity affects foam stiffness, bending actuator
performance, and flow rate of air through the foam via uniaxial tension tests, bending blocked force tests, and direct air flow measurements, respectively. I compare the airflow measurements with the Kozeny-Carman model describing fluid flow through a packed particulate bed, using X-ray μCT scans to measure porosity parameters, including porosity, pore surface area, and pore sphericity (i.e., shape). Finally, as an example of the utility of this foam actuation system, I report the design, fabrication, and performance of a compliant fluid pump formed in the shape of the human heart.

In Chapter 3, I introduce buckled foam actuators. These actuators are composed of a polyurethane in which I impart residual compressive stresses through a molding process. I then investigate how foam compression affects the pore size distribution and actuated stroke and force. The principle result is that a moderate amount of residual compressive strain within the foam increases the applied force ~1.4x or the stroke ~2x compared to actuators without residual strain. I then demonstrate a potential use for this material system by designing and fabricating a patient-specific cardiac assist device design. I show a process to form a patient-specific device from CT, MRI, or 3D scanning methodologies. The device demonstrates stroke volumes equivalent to an adult human heart in benchtop measurements and physiological pumping frequencies on a cadaveric porcine heart,

In Chapter 4, I report work examining structured porous actuators fabricated using stereolithography. In contrast to the stochastic foams described in Chapters 2 and 3, this work investigated regular lattice structures as the basis for fluidic actuators. I generated printable structures based on Bravais lattice geometries and compared their extension and compression behavior using FEA. Next, I used this information to
design and print a lattice-based fluidic actuator intended to serve as a haptic interface for virtual reality interactions. This specific interface is a compliant controller handle that changes both shape and stiffness when internally pressurized leading to multiple haptic experiences from a single controller. I also present an example of using the compliance of the handle as an interaction input by designing an algorithm to identify rapid compressions of the handle as a control signal.

*In Chapter 5, I summarize the principal findings of this work.*

Additionally, I include *Appendices A* and *B* which convey related work on porous actuators. *Appendix A* describes a composite material of elastomer foam impregnated with a low melting temperature metal. By transforming the metal from a cold solid to a hot liquid, the composite transitions from rigid (metal-dominated) to compliant (elastomer-dominated) mechanical properties. This transition (enabling shape memory behavior) is investigated as well as self-healing behavior. *Appendix B* describes work exploring an alternative porogen to that described in Chapter 2: table salt. This result of this investigation is a less hazardous foam-forming system that can be shaped either through molding or through sculpting.

1.3 Related Work

My research stemmed into several disparate fields. What follows below are the works that either inspired, guided, or supported my research.

1.3.1 Porous Compliant Actuators

While porous materials have been studied as shape memory [25], [26],
ferroelectret [27], and solvent sorption [28] based actuators, there are relatively few examples of fluidically actuated porous materials in the literature.

Subsequent to my work on polyurethane foams, Agarwal et al. [29] evaluated pneumatic foam actuators as components of a human spinal assistance unit. They fabricated actuators from foam discs laser cut from precast polyurethane foam sheet. They sealed the discs with a silicone elastomer and grouped each with two other discs to form a three degree of freedom (DOF) joint (**Figure 1.3**). They applied vacuum to collapse any of the three actuators to contort the spinal assist unit as needed. Later, this same team expanded on the capabilities of these actuators by showing their use in locomotion (both rolling and undulating) and jamming-based stiffening [30]. The authors note the promise of rapid and inexpensive fabrication of foam actuators as well the use of vacuum as a safer actuation mechanism (i.e., vacuum will never overpressurize the actuator).

**Figure 1.3 Foam actuator fabrication.** (a-c) Fabrication of the actuator module using a cutout polyurethane foam chamber which is painted with liquid PDMS. Three such chambers are aligned as shown in (c) to form the 3-DOF module. (d) FE model for a
single module consisting of three chambers with top and bottom end-plates. (e) Fully assembled actuator module with top and bottom fiberglass end-plates. Adapted from [29].

Yang et al. [31] showed that actuators with molded interconnected voids can collapse asymmetrically when the pores are exposed to vacuum (Figure 1.4). They used this controlled transition between uncollapsed and collapsed pores as actuation and show a gripper that opens and closes along with devices for locomotion over solid surfaces and water. They also report that a similar structure (also designed to collapse asymmetrically) can approximate muscle by contracting when exposed to vacuum [32]. Interestingly, the authors also show that this asymmetric collapse can lead to shape memory behavior in specially designed structures [33]. In a similar system, reported by Lazarus and Reis [34], a cylinder containing an array of symmetric pores around its exterior is collapsed by exposure to vacuum. They investigate the mechanics behind different types of collapsing behavior caused by different arrangements of filled (with elastomer plugs) and empty pores. As show in Figure 1.5, different plug arrangements lead to bending and twisting of the cylinder.
Figure 1.4 A soft gripper made of a buckling actuator. Schematics of the buckling gripper. b) The claws of the gripper close upon deflation of the buckling actuator. c) The buckling gripper picks up a piece of chalk. Scale bars, 1 cm. Adapted from [31].
Figure 1.5 Soft actuation of structured cylinders. (a) Flexural and (b) twisting actuation of representative samples, at increasing values of the dimensionless pumped volume, $\eta$. (c) Photograph of the experimental apparatus. Inset: Geometric parameters of the sample and the elastomeric conical plugs (with elliptical cross-section) used for the off (filled void) pores. Adapted from [34].

1.3.2 Cardiac Assist Devices

Compliant machines show promise as new Ventricular Assist Device (VAD)
designs. VADS are mechanical pumps that unload the cardiac muscle and impart increased blood flow [35]. These devices exist primarily because of the imbalance between the large number of patients with end-stage heart failure and the relatively small number of donor hearts available for transplantation [36]. VADS are often an alternative to immediate transplantation, providing mechanical circulatory support either to bridge the patient to a transplant or to serve as a destination therapy.

Many current clinical VADS consist of an electrically powered impeller or rotor within a rigid housing that is sutured to both the left ventricle and the aorta. The benefits of these VADS, including dramatically improved patient quality of life [37] and survivability [38], are often partially offset by a risk of device thrombosis [39], stroke [40], and implantation complications [41]. Direct Cardiac Compression (DCC) devices have emerged as a promising VAD subset in that they apply epicardial compression without direct blood contact, alleviating some of these risks [42]. Pneumatically actuated DCC device examples include designs with an inflatable thermoplastic envelope within either a rigid shell [43] or a shape memory wire scaffold [44], [45]. Recent designs have incorporated compliant actuators based on elastomeric bladders. One device assembles several McKibben actuators arranged helically embedded in an elastomer shell surrounding the heart [46]. The actuators are individually addressed providing many variations of the compressing and twisting motions of the heart. The authors show that the device conforms the heart, synchronizes with native heart motion, displaces physiological volumes of fluid in \textit{vitro}, and provides cardiac assistance in pigs with acute heart failure [47]. To date, this
publication makes the strongest case for the use of compliant machines as VADs.

1.3.3 Haptic Interfaces

The “Shifty“ device study was a conceptually similar approach to dynamic haptic interfaces [48]. This study showed that a fairly simple mechanism, a movable weight within a rigid handheld rod, can provide users with the perception of handling multiple or shape-changing virtual objects. The authors found that shifting the mass during virtual simulations increased the perceived realism and user enjoyment as they observed virtual objects changing length or thickness.

Within my own research group, a project developing a “pneumatic skin“ provided a strong influence on my work. The skin consisted of 12 isolated pneumatic actuators within a thin sheath designed to fit around the exterior of a commercial VR controller. When synced with a virtual reality simulation, the chambers inflated rapidly providing real-time haptic feedback onto the user’s fingers. As each actuator is individually addressed, many patterns of inflation/deflation are possible, providing a plethora of tactile and haptic sensations. While an abstract is the only current publication of this work [49], a follow-up paper describing the full details of the system will be submitted in the near future.

1.4 REFERENCES


N. W. Bartlett, M. T. Tolley, J. T. B. Overvelde, J. C. Weaver, B. Mosadegh, K. Bertoldi,


2.1 Introduction

The field of soft robotics uses compliant structures to reduce machine complexity and approach the mechanical [1,2] and sensing [3] capabilities of biology. Recently, soft (i.e., compliant and extensible) materials and structures have enabled machines capable of elaborate locomotion [1,4-6] as well as analogs for caterpillars [7], fish [8,9], jellyfish [10], and octopus tentacles [11]. These compliant machines often perform favorably when compared with conventional, rigid machines as they reduce control complexity [11], enable natural motion through continuous deformation [12] and interact gently with fragile objects [13].

Fluidic Elastomer Actuators (FEAs) are a class of compliant machines capable of producing large deformations via pressurization of internal bladders [14]. They operate similar to McKibben artificial muscles [15, 16] though often require lower pneumatic pressures [17]. When internally pressurized, FEAs (typically composed of low elastic modulus silicone rubber) can bend [18], extend [19], or twist [20] based on the specific patterning of inextensible fibers within the structure. During inflation, the

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inextensible fibers create a strain gradient resulting in the programmed motion. Although these machines exhibit complex motions with relatively few actuators [21], their fabrication has been largely limited to prismatic structures, which require complex assemblies to approximate the shapes and motions of biological models [17]. Furthermore, these actuators require internal, connecting air chambers that often necessitate complex, costly, and time-consuming mold fabrication. Recently, foam-wax composites have been used to create soft, smart 3D structures that transcend prismatic designs [22], however these porous materials have not yet been used for fluidic actuation.

In this communication, we present a method to easily produce stochastic, open-celled, low density ($\rho \sim 0.47$ g ml$^{-1}$) foams that when hermetically sealed form compliant actuators. Upon internal pressurization, these actuators are capable of large amplitude shape morphing while applying forces that exceed 20 N. Similar to cardiac muscle systems, where the soft tissue defines the function and shape of the organ (e.g. the heart), the poroelastic material forms the entirety of the machine. Unlike other FEAs, these machines have an interconnected open-pore network and require no additional molded air channels. Using existing forming techniques (e.g., casting, sculpting, and sheet cutting) we fabricated machines capable of actuating in simple modes such as bending and extension, as well as complex modes for functional devices such as suction cups and fluid pumps (Figure 2.1a-c).
Figure 2.1 Foam-based pneumatic actuation. (a) Bending actuator fabrication process (from left to right): curing porogen & PDMS mixture, heating to remove porogen, and patterning inextensible fiber into PDMS sealing layer, (b) bending actuation, (c) extending actuation, and (d) µCT foam reconstruction.

Our fluid powered foam actuators are significant because they allow simple production of complex soft machines that cannot easily be formed by other established methods. Though sacrificial 3D printed molds [9] or manual assembly of prismatic structures [23] can, with extra material and labor costs, produce 3D machines—these designs do not intrinsically link the robot’s body with its motion. As our fluidic actuators comprise the entirety of the machine, everywhere there is foam there is also the potential for motion. To demonstrate the potential of the poroelastic foam machines, we fabricated a soft, fluid pump with a complicated internal and external
architecture. The pump we report is shaped like an organ, a human heart, and pushes fluid via pulsatile pressurization of two chambers. It is capable of pumping at flow rates \((\nu \sim 500 \text{ ml min}^{-1})\) faster than any previous soft pump reported [24-26] and at physiologically relevant pressures \((\Delta P \sim 14 \text{ kPa})\).

2.2 Materials and Methods

We fabricated the foams via a lost-salt process adapted from Lin [27] using silicone elastomer (PDMS) as the matrix material and ammonium hydrogen carbonate \((\text{NH}_3\text{CO}_3)\) as the fugitive porogen. An X-ray µCT reconstruction (Figure 2.1d) shows the full foam structure after salt removal. \(\text{NH}_3\text{CO}_3\) is beneficial as a porogen because it (i) thermally decomposes readily and entirely above 50 °C, (ii) yields easily-managed products (water, \(\text{CO}_2\), and \(\text{NH}_3\)) leaving the PDMS foam residue free, and (iii) is inexpensive and readily available. We further describe the porogen thermal decomposition in the supplemental information (SI) text.

We selected Mold Max 10 PDMS (MM10; Smooth-On, Inc.) as the elastomer due to its low tensile modulus \((E \sim 500 \text{ kPa})\) and large ultimate strain \((\varepsilon_{ult} = \frac{L_{ult} - L_0}{L_0} > 3)\), where \(L_{ult}\) is the stretched length at break and \(L_0\) is the initial length, which enables large deformations at low inflation pressures (typically \(\Delta P < 70 \text{ kPa}\)). Additionally, MM10 is beneficial as it exhibits a short room temperature cure (~ 3 hr) and is compatible with the \(\text{NH}_3\text{CO}_3\) porogen since it cures via an organotin-catalyzed condensation reaction. We note that the porogen will significantly inhibit the platinum catalyst in addition-cured PDMS resins [28]. Though we selected a low modulus
PDMS for this work, our process is general to all material systems compatible with the porogen.

To produce actuators, we sealed the foam structure in a ~2 mm thick layer of MM10 which we either brushed onto or molded around the foam. We preferred the brushing method when sealing irregularly shaped foams. Prior to curing the sealing layer, we patterned an inextensible, yet flexible nylon mesh (9318T18, McMaster Carr) to program the actuator motion upon inflation. Though other fiber materials could serve the same function, we selected nylon mesh because it is easily laser-cut into complex shapes and creates an interwoven mechanical bond with the PDMS.

To characterize the mechanical properties of the foam, we performed uniaxial tensile tests on samples of $\phi = 0, 0.5, 0.6,$ and $0.7$ porosity (i.e. pore volume fraction) and measured the forces applied by the sealed actuators via blocked force measurements. Additionally, we characterized the airflow rate through unsealed foam samples, since this parameter dictates actuation speed. We measured the input and output pressures (3847K71, McMaster Carr) across a foam disc (radius, $r = 1.25$ cm and thickness, $d = 1.25$ cm) while measuring the air flow rate (FLDA3422G, Omega) through the system.

2.3 Results and Discussion

Tensile tests on the unsealed foams show a trend of decreasing elastic modulus as porosity increases (Figure 2.2a). The average tensile moduli, measured at $\varepsilon = 1$, of the $\phi = 0, 0.5, 0.6,$ and $0.7$ porous foams were $E \sim 508$ kPa ± 28, 200 kPa ± 31, 83 kPa
± 6, and 48 kPa ± 19, respectively. The average ultimate tensile stresses and strains were $\sigma_{UTS} \sim 2.22 \text{ MPa ± 0.09, 0.40 MPa ± 0.02, 0.24 MPa ± 0.02, and 0.064 MPa ± 0.04}$ and $\varepsilon_{ult} \sim 3.56 ± 0.22, 1.97 ± 0.10, 2.35 ± 0.09, \text{ and 1.58 ± 0.16}$ for the $\phi = 0, 0.5, 0.6, \text{ and 0.7}$ porous foams, respectively. The low variation within these samples (n=3 for each porosity tested) indicates good repeatability even though the foams are stochastically structured. The data also shows that all foams can stretch to $\varepsilon >> 1$ before failure; this elongation is essential to obtain large actuation amplitudes. Within the data, we also note that the pure ($\phi = 0$) PDMS has a higher ultimate strain, $\varepsilon_{ULT} \sim 3.3$, than the foam materials. Due to this difference, the first component to fail during actuator over-inflation is the foam interior and not the encapsulating skin. Fortunately, for a pneumatic actuator in service, this failure mode is preferred over rupture of the external seal. The resulting expansion from foam fracture provides a visual indication of failure (we call it an “aneurysm”).
Figure 2.2 Tensile and blocked force measurements. (a) Tensile stress-strain behavior of unsealed foams, (b) blocked force test setup, and (c) blocked force measurements of bending actuators composed of differing porosity.
To characterize the performance of the bending actuators, we performed blocked force measurements on $\phi = 0.6$ and $\phi = 0.7$ porous samples. This measurement constrains bending actuators to a fixed gap of 3 cm between rigid surfaces (Figure 2.2b) and measures imparted force as a function of inflation pressure (Figure 2.2c). Due to its lower modulus, the $\phi = 0.7$ porous foam actuates at lower pressures; however, it also develops an aneurysm at $\Delta P \sim 50$ kPa. The $\phi = 0.6$ porous actuator inflates to higher pressures ($\Delta P \sim 80$ kPa) without developing an aneurysm, and therefore applies a larger force. The relationship between inflation pressure and actuating force extends to $\phi = 0.5$ porous foams (SI text); however, there was large sample to sample variation likely due to approaching a percolation threshold of pneumatic connectivity. The ability to tailor the onset and magnitude of actuating force by porosity is a new capability for FEAs. We used the same blocked force technique to measure the force applied from an extending actuator and recorded an applied force of 20 N for $\phi = 0.7$ porosity (details of this are in the SI text).

To record airflow measurements through the foam, we used a custom-built acrylic cell to bond a cylindrical foam sample within rigid tubing (Figure 2.3a). We then hermetically sealed the tubing using bolted endplates and rubber o-ring gaskets. We designed this mount to contain the applied pressure and ensure air permeated a constant foam cross-section. We then increased the pressure differential across the foam from 0 to 104 kPa (0 to 15 psi) while measuring the flow rate downstream of the sample. We fabricated both soft and rigid foam samples for this measurement: the soft, elastomeric foam was silicone (MM60, Smooth-On; $E = 2$ MPa, reported) and the
rigid foam an epoxy (System 3000, Fibre Glast; $E = 16$ GPa, reported). Due to the detection limits of our instruments, we required the higher modulus MM60 over MM10 to attain sufficient flow rates for measurement. All foam-fabrication parameters and procedures were unchanged when using MM60 and System 3000 epoxy. As seen in Figure 2.3b, the soft foams exhibited considerably higher flow than the rigid foams. We also observed the expected behavior that for a given material, a higher porosity resulted in a higher flow rate for all applied pressures.
Figure 2.2 Airflow through foam actuators. (a) Airflow measurement sample mount, (b) airflow measurements through soft [red] and rigid [blue] foams with KC model prediction of flow behavior [black], and foam microstructure during 0 L min\(^{-1}\) (c) and 7 L min\(^{-1}\) (d) airflow. Highlighted region shows more open pore network in (d) than in (c), where grey is PDMS and black is air-filled pores.
We next compared the experimental airflow measurements with an adapted Kozeny-Carman (KC) model for fluid flow through a packed particulate bed. The KC model relates fluid flow to key parameters as displayed in \textbf{Equation 2.1},

\[
Q = \frac{\Delta P \Psi D_p^2 \phi^3 A}{180 L \eta (1-\phi)^2}
\]  

(2.1)

where \(Q\) is flow rate, \(\Delta P\) is pressure drop across the sample, \(\Psi\) is PDMS sphericity, \(D_p\) is the spacing between pores, \(\phi\) is sample porosity, \(A\) is the sample cross-sectional area, \(L\) is sample length, and \(\eta\) is viscosity of the fluid. We directly measured \(Q, \Delta P, A,\) and \(L\) from the airflow experiment. We used X-ray \(\mu\)CT to extract \(\Psi, D_p,\) and \(\phi\) for each of the tested foams.

The X-ray \(\mu\)CT scans of both the soft and rigid foam samples yielded cylindrical (radius, \(r = 1.5\) mm and thickness, \(d = 3\) mm) volumetric reconstructions of the foam structure. Using the freely available ImageJ image processing software with the BoneJ [29] plugin (version 1.4.0), we determined the porosity, pore surface area, and average spacing between pores from the scans using the \textit{Volume Fraction}, \textit{Isosurface}, and \textit{Thickness} functions, respectively. Since the foam microstructure is relatively similar to that of cancellous bone, the BoneJ plugin is a particularly useful software package [30]. Further details of this process are provided in SI. Further, we calculated the sphericity (\(\Psi\)) of PDMS using \textbf{Equation 2.2} with measured values of PDMS volume (\(V\)) and surface area (S.A.) from BoneJ.
\[ \psi = \frac{\pi^{1/3} (6V)^{2/3}}{S.A.} \]  

(2.2)

The calculated parameters were similar between soft and rigid foams of equal porosity indicating repeatability of the process independent of matrix composition. When input into the KC model, using no free parameters, the predicted flow rates are in close agreement with the measured rates through rigid foams (Figure 2.3b). The soft foams, however, exhibited much higher flow rates than both the rigid foams and the flow rate predicted using the KC model. To determine the cause of this discrepancy, we compared \(\mu\)CT scans of the sample with and without air flowing through it. Visually, when air flowed through the foam samples, we noted that the foam cylinder deformed to create hemispherical surfaces in the direction of airflow. The \(\mu\)CT scans were consistent with this behavior in that they showed an apparent 12% increase in porosity when permeating at a 7 L min\(^{-1}\) flow rate relative to the porosity at 0 L min\(^{-1}\) (Figure 2.3c,d). We attribute the apparent increase in porosity to the expansion of pores in accommodating the strain of the hemispherical macroscopic deformation. Further verifying this phenomenon, our calculations showed that the porosity was highest in the center of the sample (the region furthest from fixed edges) and became incrementally less porous in regions closer to the perimeter. As the porosity is a heavily weighted parameter in Equation 2.1, we attribute the increase in flow through soft foams to this strain-induced porosity increase.

To demonstrate the utility of foam-based soft actuators, we fabricated a functional fluid pump with a biologically inspired external geometry (Figure 2.4a,b). This pump is actuated pneumatically (Figure 2.4c) and is composed entirely of soft
materials (SI text). Prior to use, we attached the pump to a pressurized air source and fully primed it with water. The pump generates water flow using only a single air source and the timed control of four solenoid valves via a microcontroller (Arduino Uno). We made simple modifications to the microcontroller program to attain a wide range of pumping frequencies. The only electric power required for the pump is that needed to power the controller and valves. Two two-way valves control water flow (W1 and W2 in Figure 2.4d; 08F23O2140A3F4C75, Parker) and two three-way valves control airflow and allow venting (A1 and A2; 912-000001-031, Parker). The pump attains water flow by alternating between 2 states: (State 1) valves A1 and W1 are open, valve A2 is venting, and valve W2 is closed; (State 2) valves A2 and W2 are open, valve A1 is venting, and valve W1 is closed (Figure 2.4d). In State 1, air travels from the source through A1 to inflate the left foam chamber. This inflation pressurizes the adjacent left water chamber generating water flow through W1 into the right water chamber. In this state, valve A2 is venting to accommodate compression of the right foam chamber as the right water chamber inflates. The microcontroller then switches all valves to State 2, causing the right foam chamber to inflate, pressurizing the right water chamber and causing water flow through W2 back to the left water chamber.

When pumping water at frequencies of 2 Hz, 1 Hz, and 0.1 Hz, we observed sustained output flow rates of 190 mL min\(^{-1}\), 370 mL min\(^{-1}\), and 430 mL min\(^{-1}\) (MR3L213NV, Key Instruments) at pressures of 20 kPa, 14 kPa, and 12 kPa, respectively (3846K1, McMaster Carr). The 430 mL min\(^{-1}\) flow rate is 80% faster than the previously highest reported rate from a soft pump [24].
Figure 2.3 The foam-based fluid pump design and principle of operation. (a) Fluid pump as cast, (b) fully assembled fluid pump after applying external inextensible shell, (c) X-ray fluoroscope of pump when left chamber is uninflated (left) vs. inflated (right), (d) schematic of pump operation. Green represents an inflated foam chamber while red represents an uninflated chamber. W1 and W2 represent valves in water transporting tubing; A1 and A2 represent valves in air transporting tubing. A red “X” indicates a closed valve. The pump is composed of a pure MM10 top [A], pneumatic chambers composed of MM10 foam [B], a MM60/woven Kevlar™ barrier separating the left and right halves [C], pneumatic inputs [D], water-transporting tubing [E], and water chambers [F].

While pumping, the water flow is pulsatile due to the alternating inflation and venting of each foam chamber; however, throughout each inflation step, the flow is sustained. Therefore, we observed continuous flow within each half cycle (sustained for 10 s at a 1 Hz frequency). This flow is uniquely different than the rapid bursts of
flow exhibited by previously reported (i.e. combustion-powered) pumps. To the best of our knowledge, this is the only soft pump that offers sustained flow. Similar to other reported soft pumps, our design is a displacement pump that relies on compliant diaphragms. Unlike other pumps, however, our design does not employ combustion for inflation. Although combustion is a powerful energy source [31], the complexity in managing the safety and timing is complicated. As our pump undergoes only mechanical loading (and not mechanical combined with significant thermal loading), we expect the pump to have lifetimes comparable to other soft actuators, which have been reported in excess of a million actuation cycles [32].

This pump provides an alternative to current VADs and artificial hearts, both of which are primarily composed of rigid materials. Consisting entirely of soft materials, our pump could easily deform and conform in response to the forces imposed on internal organs. Additionally, the two primary material components of the pump (i.e. PDMS and Kevlar™) are both biocompatible [33,34]. In further development of the pump, a flow increase and overall volume reduction will be necessary for its utility as a functional, implantable prosthetic. If used as a VAD, a version of this pump could eliminate the need for the complex impeller mechanisms presently used, which show a propensity to collect blood clots [35]. Further, a VAD using a poroelastic foam sleeve would not require boring into the heart itself, reducing complications from surgery [36].

2.4 Conclusions

This work demonstrates elastomeric foams as a new material platform for
designing compliant machines. Unlike previous soft lithography fabrication methods, these foams provide the unique capability to produce truly three-dimensional actuating structures. Further, because the foams are open-celled, we see a significant reduction in design complexity as there is no need for molded air channels. We also demonstrate that the actuating behavior is easily tailored during formulation by tuning porosity. Using this new material system we created an entirely soft fluid pump that generates pulsatile, unidirectional flow at physiologically relevant pressures. Further, the pump produces flow rates higher than any soft pump reported.

In their current state of development, these foams have two primary limitations: i) the porous network limits the inflation rate and ii) the internal structure tears when overinflated. The fibrillar PDMS network within the foam presents a tortuous path for airflow which limits actuation rate. This effect is dependent on foam porosity and is much more pronounced in lower porosity samples. Despite this limitation, however, the $\phi = 0.7$ porosity foam actuators always demonstrate greater flow rate than a comparable pneu-net design (SI Text) [1]. Interestingly, at pressures in excess of $\Delta P \sim 100$ kPa ($\Delta P \sim 15$ psi), the poroelastic actuators show more than double the flow rate of our model pneu-net.

We note that the data Figure 2.2c contain a “toe” region (i.e., there is no measurable actuation force at low pressures). In this region, the internal foam is pressurized but not to a degree sufficient to impart a non-negligible force. While part of this toe region is due to the actuator not yet reaching the blocked curvature, another part is inherent to the actuator composition. That is, there is an inherent resisting force to actuation supplied by both the foam and the sealing elastomer. Pressurization must
overcome the resisting forces in order for strain in the material and, subsequently, displacement of the actuator to occur. This toe region could potentially serve as a benefit to the system. For an actuator system designed to cyclically inflate, a state of zero actuation can be achieved without having to remove 100% of the actuating pressure, potentially enabling higher frequency or more efficient performance. A clear tradeoff exists in this example, however, in that the internal foam would never fully recover to its unloaded state. Any pressurization of the system, whether sufficient to actuate or not, imparts stresses within the foam. Over long durations, these stresses could lead to creep in the foam and, over a high number of cycles, could affect fatigue of the material, each potentially leading to a reduced functional life. As a potential solution, minimizing the toe region would minimize these detrimental effects. A materials approach to this solution would consist of a comparison the blocked force behavior of foam actuators composed of different elastomers. We hypothesize that a lower modulus coating combined with a higher modulus foam (though still a lower modulus than the coating elastomer) would actuate at lower pressures, reducing the toe region, and limit the foam strain at low pressures. Reducing the coating modulus reduces the resisting force to actuation, enabling actuation at lower pressures, as verified in Appendix B. Increasing the foam modulus will reduce its strain within the toe region, limiting creep and fatigue effects. Another approach to minimize the toe region could be accomplished through a controls scheme. A controls system could be developed in which the pressure within the actuator is continually monitored and used as feedback to ensure that the actuator is fully depressurizing between cycles and adjust deflation durations if necessary.
In future work, the use of a tougher elastomer will allow higher porosity foams (with less tortuous airflow pathways) that could withstand higher inflation pressures while avoiding rupture. To further increase the actuation rate, anisotropic and/or oriented fugitive materials could be used to generate pores. We would expect pores with a higher aspect ratio would provide a less tortuous pathway for the inflation fluid and would therefore increase the speed of inflation.

2.5 Experimental Section

2.5.1 Foam Fabrication

We fabricated foams via a lost-salt process using MM10 PDMS unless otherwise noted. To produce the foams, we first mixed the PDMS prepolymer and NH$_3$CO$_3$ (Alfa Aesar, used as-received) at a desired ratio and allowed the mixture to cure at room temperature. The prepolymer-to-porogen volume ratio dictates the final porosity of the foam. We then placed the cured PDMS/porogen composite in a vacuum oven at 150°C until the porogen had entirely decomposed. The porogen typically decomposed in 2-6 hours depending on part size.

2.5.2 Characterization

We performed tensile testing of the foams according to ASTM D412 on a Zwick Roell z010 instrument. We conducted all tests using a 10 kN load cell and a strain rate of 6.25 min$^{-1}$. 

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We imaged the internal structure of the foams using an X-ray µCT (ZEISS Xradia Versa 520) and the inflation of the foam using a Philips EasyDiagnost Eleva. Details of the subsequent µCT image analysis are provided as supporting information.

2.6 Supplemental Information

2.6.1 Thermogravimetric Analysis (TGA)

The collected TGA data reveals several important aspects of the thermal stability of the system. First, we observed that the pure porogen (NH$_4$CO$_3$) begins thermal decomposition at approximately 50°C and that this behavior is unchanged when mixed with the MM10 PDMS (Figure 2.5a). We also note that the pure PDMS is thermally stable through 250°C. At a constant temperature of 150°C, the porogen decomposed entirely within ~10 minutes (Figure 2.5b). Again, we observed similar behavior for the porogen in PDMS. The remaining mass after ~20 min corresponds with the initial mass of PDMS.

![Thermogravimetric analysis of foam components.](image)

**Figure 2.4** Thermogravimetric analysis of foam components. Decomposition curves of (a) PDMS (black), PDMS + porogen (red), and pure porogen (blue) at a ramp rate of 10 °C min$^{-1}$. (b)PDMS + porogen (black) and porogen alone (red) when held at 150 °C
2.6.2 Blocked Force Measurements

Figure 2.6 is an extension of the data presented in Figure 2.2c. We observed the same general trend that actuators withstand greater inflation pressures as porosity decreases. We note that the $\phi = 0.5$ porous actuators show much greater variability than the $\phi = 0.6$ and $\phi = 0.7$ porous samples. We attribute this variation to the $\phi = 0.5$ porous foam being much closer to the percolation porosity required for an open-cell network and therefore local concentration variations have a greater overall effect on the actuating behavior.

![Graph showing blocked force measurements of bending actuators]

2.6.3 Blocked Force of Extending Actuator

In an additional blocked force measurement, our extending actuator provided a force of 20 N when pressurized to 70 kPa with no apparent foam failure. We
performed this measurement by sandwiching the actuator between a balance and a rigid, fixed surface. Because the actuating displacement was minimal (only that necessary to depress the balance), the majority of the inflation energy is transferred as this large measured force.

2.6.4 Image Analysis Process

We processed the raw X-ray µCT image stacks using the following custom Matlab thresholding algorithm.

```matlab
function threshold()

for l=100:900 %picks the 100th to the 900th image
    if l<10
        rawImage=dicomread(['I000','num2str(l),'.dcm']); %reads the images
    else
        if l<100
            rawImage=dicomread(['1000',num2str(l),'.dcm']);
        else
            rawImage=dicomread(['100',num2str(l),'.dcm']);
        end
    end
end

brightImage=immultiply(rawImage,2); %brightens the image by 2x

binaryImage=im2bw(brightImage(:,:,1)); %converts brightened image to a binary image

smoothedImage=bwareaopen(binaryImage,40); %smoothes black region of the image

invertedImage=imcomplement(smoothedImage); %inverts the smoothed binary image

reSmooth=bwareaopen(invertedImage,1); %smooths the inverted image

threshImage=imcomplement(reSmooth); %re-inverts smoothed image

finalImage=rawImage; %defines final output image

[m,n]=size(rawImage);
for i=1:m
    for j=1:n
        if rawImage(i,j)>100
            if threshImage(i,j)==0
                finalImage(i,j)=0; %converts pixel to black in finalImage
            end
        end
    end
end
```
if threshImage(i,j)==1
    finalImage(i,j)=60000; %converts pixel to white in finalImage
else
    finalImage(i,j)=20000; %converts background to arbitrary grey value
end
end
end

imwrite(finalImage,['FinalImage_','num2str(l),'.tiff']); %writes Tiff image
end

We use the *immultiply* and *bwareaopen* parameters to ensure the thresholded images visibly matched the raw image. We next imported the image stack into ImageJ image analysis software and selected the circular field of view as the region of interest. As Figure 2.7a shows, the areas shaded in black represent the foam while the white areas represent the pores.

We measured the average PDMS thickness (i.e. spacing between pores) using the *Thickness* function in the BoneJ plugin. This also returns a graphical representation of the foam thickness (Figure 2.7b).

To analyze foam porosity, we first invert the binary image stack (Figure 2.7c) and run the *Volume Fraction* function in BoneJ. Finally, we calculate the PDMS surface area using the *Isosurface* function with a resampling parameter of 8 and a threshold of 128. This returns a bone surface (BS) value as well as a 3D graphic result (Figure 2.7d).
2.6.5 Pump Assembly

We first cast the foam chamber component of the pump using a 3D-printed mold (Objet24, Stratasys, Ltd.) mirroring the external geometry of the bottom half of the human heart. We filled this mold with a PDMS/porogen mixture (MM10; $\phi = 0.7$ porogen) and allowed it to cure at room temperature overnight (Figure 2.8a, left).

Concurrently, we cast PDMS into a separate mold to form the top half of the pump (Figure 2.8a, right). After demolding, we placed the foam chamber component in a vacuum oven at 150 °C under continuous vacuum for 8 hours to remove the porogen.
At this point, the foam chamber was fully porous (Figure 2.8b). We then sealed the foam chamber by brushing PDMS onto the interior and exterior surfaces. Once sealed, we bonded the top and bottom pump halves together using PDMS. During this step, we also inserted pneumatic lines (1.57 mm ID, 2.08 mm OD; 427446, Becton Dickinson) through the top pump half and into the bottom half, embedding them into the foam. Next, we vertically bisected the pump using a scalpel (Figure 2.8c) and sealed the newly exposed foam creating the two fully isolated foam chambers. We then impregnated Kevlar™ woven fabric (10 x 15 cm; 1065, Fibre Glast) with MM60 and cured the composite at room temperature to create the fluid chamber divider. We then adhered the divider to the two pump halves using MM10 as an adhesive and sealant. The full exterior of the pump was then coated in a mixture of MM60 and chopped carbon fiber (2 inch length; Fibre Glast; 2 wt. % fiber loading) to form an inextensible shell. This shell is essential for efficient pumping as it directs inflation inwards towards the internal fluid chambers. Finally, we completed the assembly by inserting tubing into the fluid chambers and affixing zip ties around the tubing ports to provide a hermetic and watertight seal.
**Figure 2.7 Molding process to form the pump’s foam shell.** (a) Casting molds for lower foam component [left] and PDMS top [right], (b) resulting foam component after porogen removal, and (c) separated pump halves that form the two foam chambers and the two water chambers.

**2.6.6 Flow Rate Comparison of Foams to Pneu-net**

We compared the soft foam samples presented in Figure 2.3 to a model system for pneu-net type actuators. To simulate the limiting air channels within the pneu-net design [1], we used a disc of solid MM10 (radius, $r = 1.25$ cm and thickness, $d = 1.25$ cm).
cm) with a single, 1 mm diameter hole through its thickness. The airflow behavior of this sample is compared to the soft foam samples in Figure 2.9.

![Airflow measurement of soft foams and a pneu-net model.](image)

**Figure 2.8 Airflow measurement of soft foams and a pneu-net model.**

We observed that the $\phi = 0.7$ porous foam showed a higher flow rate at all tested pressures signifying a more rapid inflation of actuators composed of this material. We also observed that the model pneu-net system showed differing flow rate increases throughout the tested pressure range. We attribute the change in slope of this curve to the strain stiffening of MM10. As we increased pressure, we observed the MM10 disc bulging in the direction of flow (similar to what we had observed in the foam samples) increasing the volume of the hollow air channel and reducing flow resistance. The channels, however, eventually stop changing in dimension as the material strain stiffens, reducing the rate of flow increase with pressure.
2.7 Acknowledgements

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2.8 REFERENCES


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CHAPTER 3

COMPLIANT BUCKLED FOAM ACTUATORS AND APPLICATION TO PATIENT-SPECIFIC DIRECT CARDIAC COMPRESSION†

3.1 Abstract

Direct Cardiac Compression (DCC) is a type of implanted mechanical circulatory support that avoids direct blood contact, mitigating risks of clot formation and stroke. This paper describes a first step towards a pneumatically powered, patient-specific DCC design by employing elastomer foam as the mechanism for cardiac compression. To form the device, a mold of a patient’s heart was obtained by 3D printing a digitized x-ray CT or MRI scan into a solid model. From this model, a soft robotic foam DCC device was molded. The DCC device is compliant and uses compressed air to inflate foam chambers that in turn apply compression to the exterior of a heart. The device is demonstrated on a porcine heart and is capable of assisting heart pumping at physiologically relevant durations (~200 ms for systole and ~400 ms for diastole) and stroke volumes (~70 mL). In this paper, the use of buckled foam for pneumatic actuators is also introduced: a moderate amount of residual compressive strain within the foam increases applied force ~ 1.4x or stroke ~2x compared to

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actuators without residual strain. The origin of these improved characteristics are explained analytically. Though further development is necessary to produce a fully implantable device, the material and processing insights presented here are essential to the implementation of a foam based, patient-specific DCC design.

3.2 Introduction

Compliant actuators, relative to their rigid counterparts, uniquely have an increased potential for safe human-robot interaction due to their infinite passive degrees of freedom actuated at low mechanical stresses, a high number of possible degrees of freedom of motion, and an ability to adapt their shape to changing environments [1]. This class of actuators includes piezoelectric polymers [2], electroactive polymers [3], dielectric elastomer actuators [4], thermo- and chemically-responsive gels [5], and fluidically pressurized elastomer actuators [6]. Of these, fluidically pressurized elastomer actuators are a good choice for applying high forces and actuating at high speeds [7] within low density and highly stretchable systems. To transport fluid for actuation, fluidically pressurized elastomer actuators require fluidic channels and chambers [8, 9], the design and fabrication of which limits their geometry, often necessitating complex molding and/or assembly to form fully 3D machines.

Foam actuators simplify this fabrication process by eliminating the need for designed channels and chambers. The elastomer foam itself serves as the inflation chamber while the interconnected porous network serves as fluidic channels. When
sealed within a nonporous elastomer, these foams forms a fluidic actuator enabling 3D soft machines without complex assembly [10, 11] or investment casting [9, 12]. While these actuators can function using air, water, and other fluids as the pressurizing medium, we used compressed air as the actuating fluid for all actuators due to the inviscid nature of air and fast transport through micropores [13]. These foam actuators however experience reduced actuation amplitudes. Because they are composed of ≥ 60% void space, less material participates in stretching making the foams prone to tearing at lower inflation pressures.

Here, we have improved upon our previously published work on pneumatically powered foams [13–15] by buckling them prior to actuation. When the foam is compressed in the direction opposite of our intended actuation mode, the foam elasticity aids in the inflated motion and the resulting device has a thinner form factor. While compression of porous, elastomeric structures (i.e. the controlled buckling of the foam struts) has been shown to produce Negative Poisson Ratio behavior [16], shape memory properties [17, 18], directed deformation [19, 20], and actuation [21, 22], to the best of our knowledge this paper is the first to employ pre-compression to improve pneumatic actuation. Here we show that mechanical foam compression imparts buckling of the microscale struts leading to augmented actuation (i.e., either increased displacement or reduced actuating fluid pressure). These buckled actuators allow the DCC device to displace large volumes of liquid (~ 70 mL) when inflated, and remain thin (device thickness, L ~ 15 mm) when deflated.

To mold these actuators, we used an elastomeric polyurethane (PU) open-celled foam (FlexFoam-It!, Smooth-On, Inc.) to form the inflation chamber. The
foam’s liquid precursors consist of three parts: (i) hydroxyl-functionalized prepolymer, (ii) water, and (iii) aromatic diisocyanate cure agent. The water serves as a chemical blowing agent that, when reacting with an isocyanate at STP, generates carbon dioxide in an exothermic reaction which expands to form the cells of the foam [23]. The use of elastomeric PU foams makes the fabrication of these actuators (SI Figure 3.7) faster than our prior method of dissolving or vaporizing embedded porogen salts [13 - 15]. In addition to rapid fabrication, the PU foam actuators are functional at a much higher porosity ($\phi_{pore} \sim 0.93$, pre-buckled) than our previously reported polydimethylsiloxane (PDMS) foam actuators ($\phi_{pore} \sim 0.70$) allowing for rapid actuation (~86% of maximum strain, $\varepsilon = 0.15$ in 320 ms, SI Figure 3.8) and lightweight devices ($\rho_{actuator} \sim 0.65$ g cm$^{-3}$).

As as potential application for these buckled foam actuators, we have also developed a patient-specific Direct Cardiac Compression (DCC) heart assist design. DCC devices are implanted mechanical pumps that increase blood flow in failing hearts by applying epicardial compression. Promising pneumatically actuated DCC device examples include designs with an inflatable plastic envelope within either a rigid shell [24] or a shape memory wire scaffold [25, 26] surrounding the heart and a fully compliant design that uses an assembly of contracting pneumatic actuators [27–29]. The most recent of these devices demonstrates the strong need for highly complex and compliant shapes; this fluidically powered soft machine demonstrated viability via in vivo testing on a porcine heart [29]. Our design is similar in material properties, but uses an elastomer foam for easy, reliable, and reproducible fabrication. Additionally, the objective of our design is primarily to address the shortcomings of other DCC
devices—they largely neglect variations in patient-to-patient heart geometry and apply
global compression, potentially adversely compressing coronary arteries [30, 31].

In this work, we present a DCC device design with the unique features of a
patient-specific fit and rapid, efficient local actuation. To form the patient-specific
device, we use conventional clinical imaging techniques such as magnetic resonance
imaging (MRI) or computed tomography (CT) to capture the geometry of the patient’s
heart (Figure 3.1A). We then use the digital model of the heart as the basis of a 3D
printed mold for device fabrication. The device employs pneumatically inflated, soft
foam chambers (Figure 3.1B, C) to apply mechanical compression to the heart muscle
adjacent to the ventricles. We have demonstrated rapid DCC device fabrication (< 36
hours) which enables the potential for on-demand device manufacturing. Further, the
portion of the device that contacts the heart is comprised entirely of soft, compliant
materials which exhibit similar mechanical properties to biological tissue [1].
Figure 3.1 The direct cardiac compression device concept. (A) The device fabrication process involves the following steps: collection of a chest MRI or CT of the patient’s heart, reconstructing a digital 3D model of the heart, digitally isolating heart, 3D printing a mold from the digital heart model, casting elastomer foam within the mold, and cropping and assembling two foam chambers to form the final DCC device. Scale bar, 3 cm. (B) A device schematic showing elastomer and strain limiting layers surrounding the foam actuation chambers. Scale bar, 250 μm. (C) The device prior to inflation (left) and during inflation (right) showing volume displacement upon pressurization. Scale bar, 3 cm.

By incorporating buckled foam actuators within the DCC device, it is capable of displacing large volumes of liquid (~ 70 mL) when inflated while remaining thin (device thickness, L ~ 15 mm) when deflated. Additionally, while deflating, the foam passively expels air due to its inherent elasticity. For these reasons, elastomer foams have potential operational advantages over the inflatable plastic envelopes that have been used in other designs [24–26].

Although there are existing long term implantable (i.e. biocompatible) PU
elastomers [32], we chose a commercially available PU foam for this study because it rapidly fills mold cavities, cures in less than one hour, produces a low modulus open-celled foam with high porosity, and is low cost. Though the materials we used in this study have not yet been tested for chronic tissue compatibility, they serve well as initial surrogate materials in that their mechanical properties remain stable over the range of temperatures and strain rates required in the physiological environment (SI Figure 3.9).

We used a commercially available PU elastomer (Vytalex 20, Smooth-On, Inc.) to form the non-porous, impermeable coating around the foam. We selected this elastomer due to its high extensibility, \( \varepsilon_{\text{ult}} = \frac{L_{\text{failure}} - L_0}{L_0} \sim 10 \), and tear strength, \( \Gamma = 10.5 \) kN m\(^{-1}\).

### 3.3 Materials and Methods

#### 3.3.1 Material Preparation:

We used all foams and elastomers as-received and according to the manufacturer’s instructions. We mixed the foam precursors vigorously by hand for 30 seconds prior to casting. All foam samples were FlexFoam-iT! III (Smooth-On, Inc.), except for the extending actuators which were FlexFoam-iT! V (Smooth-On, Inc.). We used a commercially available PU elastomer (Vytalex 20, Smooth-On, Inc.) to form the non-porous, impermeable coating around the foam. We selected this elastomer due to its high extensibility, \( \varepsilon_{\text{ult}} = \frac{L_{\text{failure}} - L_0}{L_0} \sim 10 \), and tear strength, \( \Gamma = 10.5 \) kN m\(^{-1}\). We mixed the PU elastomer parts A and B in a Thinky™ planetary centrifugal mixer at
2,000 rpm for 30 seconds prior to casting onto the foam. For extending actuators, we added a flexibilizing agent (So-flex, Smooth-On, Inc.) to the sealing elastomer precursor in a 1:1 weight ratio with Vytaflex part B in order to lower the elastomer modulus.

### 3.3.2 Actuator Fabrication:

To fabricate our cylindrical extending actuators, we constrained the foam in compression by applying a thin coat of PU elastomer to the exterior of a cast foam prior to applying mechanical compression and holding until the elastomer had cured. To form the final hermetic seal, we cast a second elastomer coating (thickness, t ~ 2 mm) around the entirety of the foam. In the case of the extending actuators, we also wrapped aramid thread (KEV138NATL01B, The Thread Exchange) circumferentially around the actuator to limit radial expansion upon inflation. Additionally, we designed all actuators to actuate parallel to the foam’s rise direction. Further details of this fabrication process are in the SI.

### 3.3.3 Tensile Testing:

We performed all tensile tests according to ASTM D412 on a Zwick Roell z010 instrument pulling parallel to the foam’s rise direction. All tests used a 10 kN load cell at a strain rate of 5 min⁻¹.
3.3.4 Statistical Analysis:

Statistics presented are mean ± standard deviation with N = 3 unless otherwise noted. P values were based on a two-tailed student’s t-test with assumed unequal variance within the sample sets.

3.3.5 μCT Imaging:

We used an X-ray μCT (ZEISS Xradia Versa 520) to image the internal structure of the foams. We used a previously described Matlab program [13] to threshold the μCT image stacks and used ImageJ image analysis package with the BoneJ plugin [33] to determine pore characteristics. Details of the image analysis are provided in the SI text.

3.3.6 Blocked Force Testing:

We performed blocked force tests by first measuring the maximum extension at a given air pressure. We then depressurized the actuator, set it on a balance with the top surface blocked with a weighted plate to ensure zero extension, and subsequently repressurized it while measuring the force exerted. We then repeated these measurements at several values of air pressure.

3.3.7 Actuator Modeling:

The 1D actuator model was implemented in Matlab. We used a large deformation formulation with total stretch on the material taken as the product of the actuation stretch and pre-compression. We extracted foam and coating mechanical
properties from their respective uniaxial tensile tests. The 1D model approximated the full 3D behavior reasonably well because the Kevlar thread provided most of the radial expansion resistance and foams have a relatively small Poisson’s ratio in compression.

3.3.8 DCC Device Fabrication:

The SI contains details of the conversion of a MRI or CT scan into a digital model. From the digital model of the heart, using Autodesk Maya® software and an Object30 (Stratasys) 3D printer, we designed and fabricated two sets of molds: one mold designed for casting a 20 mm thick foam shell surrounding the heart geometry and one for compressing that foam to a thickness of 10 mm. The inset portion of the mold (i.e. the geometry of the heart) was the same for both mold sets. We first cast the PU foam into the 20 mm thick mold. From that foam, we cut out two sections that surrounded the exterior of the ventricles and avoided the coronary artery regions. We then coated the foam sections in PU elastomer and compressed them into the 10 mm thick mold for the duration of its cure. After demolding, we painted on an additional layer of PU elastomer (final thickness ~ 2 mm) and manually applied a coat of 5% (by wt.) chopped carbon fiber (Fibreglast) in PU elastomer to the external surface so that all inflation would be directed inwards.
3.4 Results

3.4.1 Compressed Foam Actuation

The void space within the foam, in addition to providing an inflatble porous network, can be exploited for mechanical advantage (SI Figure 3.10). To achieve this gain, we constrain the foam in a state of mechanical compression using a PU elastomer. The practical limit of uniaxial compression (i.e., complete pore collapse) is the densification strain ($\varepsilon_D$) which we found to be $\varepsilon_D = -0.82$ and $\varepsilon_D = -0.70$ for foams having a density of $\rho = 0.05$ and $\rho = 0.08$ g cm$^{-3}$, respectively (Figure 3.2A). As we require an open porous network for inflation, we operated below $\varepsilon_D$ for all samples.
Figure 3.2 The compressive behavior of the polyurethane foams. (A) Mechanical compression of the foams used in this study, (B) a schematic of constrained compression method, (C) a foam cube before (left) and after constrained triaxial compression to ~40% original volume (right). Scale bar, 5 mm. (D, E) µCT 3D reconstruction (left), a representative 2D slice (right), and pore size distribution (lower) of uncompressed and triaxially compressed foam. Scale bars, 500 µm.

By constraining the foam in a compressed state (Figure 3.2B,C), we observed an increase in the structure’s apparent ultimate strain. Specifically, for foam samples constrained uniaxially to a compression ratio ($R_c = \frac{\text{Initial Volume}}{\text{Compressed Volume}}$) of 1.5, we observed a significant ($p = 0.03, N = 5$) increase in apparent ultimate strain ($\epsilon_{ult,Rc=1} = 1.20 \pm 0.12$ to $\epsilon_{ult,Rc=1.5} = 1.59 \pm 0.26$) in uniaxial tension. This
difference is nearly equal to the strain increase we would expect from the reversible unbuckling of this compression as shown in **Equation 3.1** below:

\[
\varepsilon_{\text{expected}} = \frac{L_{\text{initial}} - L_{\text{compressed}}}{L_{\text{compressed}}} = R_{c,\text{uniaxial}} - 1 = 0.5
\]

We attribute the discrepancy between the measured strain increase \((\varepsilon_{\text{ult}, R_c=1.5} - \varepsilon_{\text{ult}, R_c=1} \approx 0.4)\) and \(\varepsilon_{\text{expected}} = 0.5\) to a slight, macroscopic creasing of the compressed tensile samples that initiates tearing at a lower strain (**SI Figure 3.1**). The improved elongation appears to be from a reversible structural change (i.e., strut buckling), supported by the lack of a significant difference \((p = 0.39, N = 5)\) between the ultimate tensile stress of the samples: \(\sigma_{\text{ult}, R_c=1} = 0.134 \pm 0.024\) MPa and \(\sigma_{\text{ult}, R_c=1.5} = 0.148 \pm 0.024\) MPa.

**Figure 3.2D, E** shows a comparison of the porous structure of the uncompressed (i.e., \(R_c = 1\)) and compressed foam (\(R_c = 2.5\)) obtained from a micro-computed tomography (µCT) scan. Consistent with pore collapse, we noted a decrease in the average pore size of the compressed foam (mean pore diameter, \(d_{\text{pore}} = 98 \pm 65\) µm) compared to the uncompressed foam (\(d_{\text{pore}} = 165 \pm 144\) µm); however, the compressed pore network remained highly porous \((\phi_{R_c=2.5} = 0.85, \text{measured via µCT stack image analysis})\) enabling ample open pathways for inflation. Due to the foam’s high initial porosity \((\phi_{R_c=1} = 0.93)\), we expect an inflatable, open-celled network even after a large volumetric compression. For example, we estimate that a foam compressed to 20% of its initial volume (i.e., \(R_c = 5\)), would have a porosity, \(\phi \sim 0.65\) (details in the **SI Text**).
The actuators can gain a mechanical advantage by starting with the foams in a compressed state. In order to demonstrate the more efficient actuation of actuators fabricated from compressed foams, we fabricated two extending actuators (Figure 3.3); one with $R_c = 1$ (no compression) and one with $R_c = 1.5$ (with compression applied uniaxially). When inflated to equal pressures, the buckled foam ($R_c = 1.5$) actuator showed a much larger displacement than the $R_c = 1$ actuator. We quantified this behavior via a blocked force test measuring both the maximum displacement (under zero load) and the maximum applied force (when constrained to zero displacement) for each actuator as a function of inflation pressure. Representative data from these measurements are shown in Table 3.1 with full results shown in Supplemental Table 3.2. These results indicate that for a given inflation pressure, the compressed foam actuator exhibited ~2 times the displacement and ~1.4 times the applied force as the one without compression. We measured this increased efficiency over uncompressed actuators in terms of displacement or force for all tested pressures in the range of 10 – 90 kPa.

![Figure 3.3 The effect of foam compression on actuation.](image)

Extending actuation of an uncompressed (left) and compressed foam actuator (right). Scale bar, 5 mm.
Table 3.1 Comparison of model and experimental actuation (representative data). Experimental measurements and model predictions for maximum force and actuated strain of uncompressed ($R_c = 1$) and compressed ($R_c = 1.5$) actuators.

<table>
<thead>
<tr>
<th>Inflation Pressure (kPa)</th>
<th>Rc = 1</th>
<th>Rc = 1.5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experiment</td>
<td></td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>1.3 ± 0.4</td>
<td>0.017 ± 0.004</td>
</tr>
<tr>
<td>50</td>
<td>11.8 ± 1.8</td>
<td>0.082 ± 0.009</td>
</tr>
<tr>
<td>90</td>
<td>25.5 ± 4.4</td>
<td>0.155 ± 0.005</td>
</tr>
<tr>
<td>Model</td>
<td></td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>2</td>
<td>0.038</td>
</tr>
<tr>
<td>50</td>
<td>10</td>
<td>0.184</td>
</tr>
<tr>
<td>90</td>
<td>18</td>
<td>0.327</td>
</tr>
</tbody>
</table>

To further understand the actuated motion, we constructed a 1D statics model to predict actuation strain as a function of inflation pressure and externally applied force. The force balance indicated that the force in the foam, the force in the coating, and the externally applied blocking force must sum to the force provided by the inflation pressure as shown in Equation 3.2.

$$\sigma_{\text{foam}}\pi r_{\text{actuator}}^2 + \sigma_{\text{coating}}2\pi r_{\text{actuator}}t_{\text{coating}} + F_{\text{block}} = p_{\text{inflate}}\pi r_{\text{actuator}}^2 \tag{3.2}$$

The stress in the foam ($\sigma_{\text{foam}}$) and coating ($\sigma_{\text{coating}}$) were each related to their strain by the measured uniaxial stress-strain behavior. The foam strain was taken to be the actuation strain minus the pre-compression strain, so the required stress (and consequently required inflation pressure) was much lower to produce a given actuation strain when the pre-compression was large. This equation also showed the direct tradeoff between blocked force and actuation strain at a given pressure. The model
results for uncompressed ($R_c = 1$) and compressed ($R_c = 1.5$) foam actuators for the particular materials and geometry used in the experiment (SI Figure 3.13) show an expected 20% increase in actuation strain from the pre-compression. This model also indicates that the larger the radius to coating thickness ratio, the greater the effect.

We note while the model and experimental data show similar trends, there is a noticeable difference between their values as shown in SI Figure 3.13. We believe that the Kevlar constraining component of the actuator may be a contributor to this discrepancy. The Kevlar thread prevents radial expansion of the foam by imparting hoop stresses which could influence the extension of the foam when inflated. Our 1D model does not include this hoop stress contribution.

### 3.4.2 Direct Cardiac Compression Device Design

As an excellent example of complex machines that could benefit from soft robotics, we applied our buckled foam system to a cardiac assist device. Specifically, we used the more efficient and thin compressed foam actuators to form a prototype DCC device. This device design applies compression to the ventricular walls, aiding the expulsion of blood. Like other pneumatically driven DCC devices [24, 26], this design avoids direct blood contact, mitigating risks associated with thrombosis and anti-coagulation therapies that are present in common VAD designs [30, 34]. The use of buckled foam actuators has the potential to provide the following unique features: (i) application of localized compression on the ventricles avoiding potentially harmful compression of the coronary arteries and atria; (ii) the soft, compliant materials easily deform in response to anatomical forces; (iii) the low tangent modulus ($E < 1$ MPa),
highly porous foams ($\phi > 0.93$) allow for rapid inflation and deflation at physiological rates; (iv) they can be molded to fit each patient, enabling physician-directed digital shape modification prior to surgery to promote reverse remodeling or inhibit ventricular aneurysms.

To demonstrate this potential, we developed a procedure to produce a patient-specific, 3D printed mold from standard medical imaging scans including MRI and CT (SI Figure 3.14), or 3D optical scans from excised hearts (Figure 3.4A), and formed a DCC device (SI Figure 3.15). The process began with isolating the heart as a digital model (details in SI text). From this model, we designed and fabricated (Figure 4B) to cast a foam shell around the heart geometry. From the foam shell, we excised two foam sections in regions corresponding to both ventricles while avoiding areas containing coronary arteries. These two foam chambers formed the active, expanding (when inflated) portions of the DCC device; the area between the chambers (i.e. the coronary artery regions) is passive elastomer. We then constrained the foam into a smaller mold ($R_c = 2$) to buckle the foam struts (Figure 4C). This $R_c$ value was chosen to achieve the needed volumetric displacement based on the ultimate elongation of the foam. Though we did not optimize the $R_c$ parameter, we could tune the parameter to increase or decrease the displacement in future versions of the device. We compressed the foam in a direction radial to the center of the heart and expect the primary expansion of the foam to be in this same direction. We then applied a strain-limiting layer of chopped carbon fiber to the exterior of the foam which is necessary to direct the expansion of the inflated foam towards the device interior. We note that since the carbon fiber forms a transverse isotropic layer, the device applies only radial
compression and does not twist (as the heart does during contraction). Finally, we hermetically seal the device by applying an outer coat of PU elastomer (Figure 4D,E).

Figure 3.4 Fabrication of the DCC device. (A) The porcine heart that served as the basis for the digitally designed mold (B) that was used to form an organ-specific foam shell. Scale bar, 2 cm. (C) The compressed foam inflation chambers that were coated in carbon fiber and sealed with elastomer to form the final DCC device (D) that fit well around the original heart (E).

We inflated the DCC device using pressurized house air run through a regulator (model 900 dispenser, EFD-Nordson). We controlled the airflow using a three-way valve (04F30C2104AF4C05, Parker) which was activated by a microcontroller (Arduino Uno) via a relay (SRD-05VDC-SL-C, Songle). The microcontroller enabled us to dictate the frequency and duration of pulsatile inflation to mimic physiological rates. We oriented the three-way valve within the airline such that it allowed pressurized flow (for inflation) when activated and allowed venting to
the atmosphere (for deflation) when deactivated. An in-line pressure sensor (PSE530-R06, SMC pneumatics) provided the inflation profile within the foam chambers.

We controlled inflation rhythms through either microcontroller code or real-time electrocardiographic (ECG) sensing. While microcontroller coding allowed a stable, static pulsatile inflation frequency and duration for benchtop tests, real-time ECG sensing is more useful within the scope of direct cardiac compression as a therapeutic tool. When applying compression to the exterior of a beating heart, it is key for the device to compress in phase with the heart’s rhythm.

To measure the ECG signals of the heart, we used a prepackaged heart rate monitor (AD8232, Sparkfun) with electrical leads arranged in Einthoven’s triangle (i.e., a constantly negative lead on the right wrist, a switching polarity lead on the left wrist, and a constantly positive lead on the left ankle). This monitor collected an ECG signal which we used to control the solenoid valve, which triggered inflation of the device.

To demonstrate this DCC device, we designed one to fit a porcine heart (SI Figure 3.15) for an ex vivo test. We aimed to mimic the hemodynamics of a young pig including a heart rate of 90 bpm, stroke volume of 70 mL, and systole and diastole durations of 220 and 440 ms, respectively [35]. Using water displacement to measure the inflated foam volume, we observed a ~ 70 mL displacement for each chamber within a 200 ms timeframe (SI Figure 3.16). When operating the device outside of the water, we observed (via the pressure profile) that the chambers inflated and deflated rapidly, at speeds sufficient to mimic a porcine (or human) heart rate. We coordinated the inflation and deflation rhythm with a simulated electrocardiogram showing
sufficiently rapid inflation to align with a native systole and diastole, respectively (Figure 3.5A-C).

![Figure 3.5 DCC device benchtop and ex vivo performance.](image)

**Figure 3.5 DCC device benchtop and ex vivo performance.** (A) Photo sequence of chamber inflation and deflation corresponding to maximum and minimum actuation pressures (B) recorded by an in-line analog sensor. The shaded areas indicate the duration of inflation triggered by a simulated ECG reading (C). Scale bar, 2 cm.

Finally, we performed an *ex vivo* study using the porcine heart that served as the model for the device. This experimental setup ([SI Figure 3.17](#)) included a pressure sensor (model ITV1031-21N2BL4, SMC) on the pneumatic line to the device and a flow sensor (Sonoflow CO.55/120, Sonotec) on the pressurized water lines integrated into and out of the left ventricle, all of which provided real-time measurement of the air and blood surrogate (water) flow. The device’s elasticity secured the heart; however, after observing the heart slightly shifting during pressurization cycles, we added a string harness to inhibit this motion.
Within the *ex vivo* study, manual massage of the heart produced a ~ 0.18 L min\(^{-1}\) peak flow while the DCC device produced ~ 0.12 L min\(^{-1}\) peak flow (Figure 3.6) at a frequency of 60 beats per minute. Though these flows are significantly lower than the native porcine cardiac output (~ 5 L min\(^{-1}\)), this experiment demonstrated that the device is able to produce a repeatable, consistent flow profile and is able to impart ~ 67% the flow of manual massage. While the measured flow rates through the heart were lower than expected, some decrease in performance is attributed to the decreased compliance of the cadaveric heart relative to its living properties. An *in vivo* study, therefore may show increased performance. We also note that the DCC device we fabricated was molded from a contracted heart; this choice ultimately limits the flow rate of our pump. As this paper is focused on the application of buckled foam actuators and not on device efficacy, future work will focus specifically on improving the DCC device design.
Figure 3.6 *Ex vivo demonstration on a porcine heart.* (A) The device setup for the *ex vivo* demonstration on the porcine heart with (B) the collected airline pressure measurements and corresponding water flow rate during *ex vivo* demonstration on the porcine heart.
3.5 Conclusions

Buckled foam actuators exhibit improved force or displacement behavior over previous foam systems; they are more efficient because they either provide a higher force or displacement at a given inflation pressure or require a lower pressure for a given force or displacement. Though we demonstrated this effect using PU foam and elastomer, the fabrication process is general and should be compatible with any variety of other open-celled, elastomer materials and fluidic actuator designs.

We attribute the increased actuation behavior to the foam’s elasticity. Constraining the foam in compression generated residual stresses that provided a restoring force towards the original, unstressed state. Because we compressed the foam in the direction opposite to actuation, the returning force is in the direction of actuation, aiding in this motion. We note that the returning force will only be present up to the state when the foam is fully unbuckled. Another benefit of buckled foam actuators is that the imparted residual stresses limit the effect of foam strain hardening. Because the compressed foam is under a lower strain when actuated, the foam tangent modulus is also lower and provides reduced resistance to expansion upon inflation.

Our demonstration DCC device formed a close fit around the porcine heart. In an \textit{ex vivo} experiment pumping fluid through the heart, the device achieved \(~67\%\) of the peak flow rate obtained through manual massage of the porcine ventricles at a duration of 500 ms for systole and 500 ms for diastole. Though we designed the device to fit a contracted heart, investigating device geometries based on different stages of the cardiac cycle would be an important direction for future studies. For
example, designing the device from a heart at the end of diastole may minimize device effects on ventricle refilling, but may also pose a risk of forming a less conformal fit around the heart.

Fabricating the device to be patient-specific enables not only an optimal fit within the patient’s chest cavity, but is also a step towards two future possibilities. First, because this device is produced from a digital model of the patient’s heart, it is possible for a physician to digitally modify the final geometry prior to device fabrication. This would be particularly important for patients exhibiting dilated cardiomyopathy (an enlargement of the heart, reducing its pumping efficiency). A patient-specific DCC device could be formed to the heart’s healthier, non-enlarged shape, potentially promoting reverse remodeling of the cardiac tissue [36]. Second, this fabrication method does not limit the device size allowing for potential adult and pediatric use alike; further, for pediatric care, its compliant properties could potentially be harnessed to grow with the patient.

The fabrication time (<36 hours) is currently limited by the process of 3D printing the molds. We believe this time could be reduced to <24 hours by implementing new, rapid 3D printing techniques that dramatically reduce printing times [37].

Additionally, the high porosity of these open-celled PU foams allowed the rapid inflation and deflation that is critical for mimicking cardiac dynamics. The repeatable ~70 mL inflation displacement within a 200 ms timeframe closely mimics the heart’s systole in terms of stroke volume and duration.
All demonstrations of the device occurred at constant frequencies (with the exception of the ECG demonstration). While constant frequencies are relevant for feasibility studies, future versions of the device must be able to sync with and function alongside the continuously changing pulse of the human heart. To this end, modifications to the control system of the device should be implemented in future iterations. Currently, the device receives air from a reservoir at a constant pressure. By opening and closing valves in line with that reservoir for specific durations the desired output is achieved (i.e., 80 mL fluid displacement within the systole duration). This works well for a single pumping frequency however when the frequency changes, the output (volume displacement) also changes. To accommodate variable frequencies, a simple modification to the control system could be developed that implements a frequency-dependent duration of inflation and deflation. In general as frequency increases, the durations for inflation and deflation will decrease. Considering this, the peak frequencies of the human heart can help us identify a reservoir pressure that is capable of displacing 80 mL at these highest frequencies. Actuation at lower frequencies, then, can be achieved by reducing the duty cycle of the valve, providing short bursts of air throughout the inflation/deflation duration while still maintaining the high reservoir pressure. To monitor the performance of the device, a two-way flow meter and a pressure sensor could be integrated in line with the actuating chambers. The flow meter would provide feedback to the controller, ensuring that a constant volume of air enters and leaves the chamber each cycle regardless of frequency. The pressure sensor would be helpful in monitoring device performance over time and identifying any component degradation. If any materials within the device begin to
degrade (through component creep, fatigue, or aging), the actuator will change its inflated shape. Since a constant volume of air is introduced into the chamber each cycle, the change in shape will be apparent by a pressure change. In this way, the coupled flow meter and pressure sensor can provide real-time monitoring of device function.

A limitation of this DCC demonstration is that the biocompatibility of its components, to our knowledge, is not yet known. Future studies should integrate known compatible materials with accompanying fatigue tests for measuring durability over 100’s of millions of cycles (a useful lifetime for a VAD). Additionally, we intend to perform the device demonstration using fresh porcine hearts in the future in an attempt to better approximate the mechanics and dynamics of living tissue.

3.6 Acknowledgements:

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1120296). Imaging data were acquired in the Cornell BRC-Imaging Facility using the 
shared, NIH-funded (S10OD012287) ZEISS Xradia Versa 520 µCT.

3.7 Supplemental Information

3.7.1 Actuator Fabrication

We fabricated the extending actuators using the process shown below in 
Figure 3.7. We cast the sealing (i.e. nonporous) polyurethane elastomer around the 
foam within cylindrical mold. We mechanically clamp the foam in compression 
between 2 pieces of acrylic sheet for the duration of the elastomer cure. After this 
cure, we demolded the part and removed the top acrylic block and used a second mold 
to cast a ~2 mm layer of PU elastomer surrounding the foam. This second casting, 
increasing the sealing layer thickness, is necessary to avoid leaks from the overly thin 
initial seal. After the second seal cured, we wound the actuator with an aramid fiber 
with a spacing of 2.5 mm to constrain the actuator from expanding radially.
**Figure 3.7 Fabrication steps to create extending actuators.** A cast foam (A) is inserted into a sealing mold and coated with PU elastomer before it is compressed by a clamp (B) for the duration of its cure. Once demolded (C), the constrained foam is inserted into a second mold (D) to form the final seal thickness. Once cured, the actuator is wrapped circumferentially with aramid thread to restrain radial expansion (E). Scale bar, 5 mm.

### 3.7.2 Actuator Inflation Speed

We performed an inflation speed test on an extending actuator composed of uncompressed foam. We actuator (height = 40 mm, diameter = 25 mm) was inflated and vented through a single input tube (inner diameter = 1.6 mm). We used the image sequence in **Figure 3.8** to determine the actuated strain throughout the inflation. In a separate inflation test, the maximum strain of this actuator was measured to be $\varepsilon = 17.5$.

**Figure 3.8 Inflation of an extending actuator composed of uncompressed foam.**
### 3.7.3 Foam Inflation Frequency and Temperature Characterization

In order for foam actuators to be useful within the human body, they must maintain consistent mechanical properties at a variety of timescales at body temperature (37°C). We characterized the temperature and frequency dependence of the foam using a DMA Q800 (TA Instruments) using a film tension clamp. The machine performed the temperature sweep at a 1 Hz oscillating strain, \( \varepsilon = 0.01 \), and the frequency sweep from 0.1 Hz to 25 Hz at 37°C with an oscillating strain, \( \varepsilon = 0.01 \).

In Figure 3.9 below, both the storage and loss moduli (\( E' \) and \( E'' \), respectively) remain relatively unchanged between -25 to 50°C and throughout the tested frequency range. The temperature sweep shows that both body temperature and room temperature are within the foam rubbery plateau region indicating that the material’s mechanical properties in the lab setting approximate those at 37°C. The frequency sweep also shows relatively flat \( E' \) and \( E'' \) curves throughout the frequency range indicating the foam mechanical properties are relatively frequency independent within this range. Though we were unable to reproduce the high strain rates that we expect within the DCC device (\( \dot{\varepsilon} \sim 3 \text{s}^{-1} \)) in this frequency sweep, the maximum tested strain rate (\( \dot{\varepsilon} = 0.25 \text{s}^{-1} \)) is a first step towards characterizing the foam’s dynamic mechanical properties and is relevant for lower speed actuators.
Figure 3.9 Dynamic mechanical behavior of two foams of different density. Foams of different densities (0.05 and 0.08 g cm\(^{-3}\)) as a function of temperature (left) and frequency (right). In both plots, storage modulus (E’) and loss modulus (E’’) are plotted on the left axis and tan delta (E’’/E’) is plotted on the right axis.

3.7.4 Reversible Foam Buckling

In Figure 3.10 below, we performed a tensile test on both an uncompressed and compressed (R\(_c\) = 1.5) foam. These foams did not have a coating, therefore the compression occurred entirely within the tensile testing machine. The zero strain point of the compressed foam represents its buckled state (indicated by the negative stress values). This curve presents two interesting aspects of the buckled foam behavior. First, the elongation increase is roughly equal to the amount of compressive strain applied initially. Further, the fact that these values are nearly equal indicates that the foam buckling is reversible. Second, the negative (residual) stresses exhibited in the compressed foam (below \(\varepsilon = 0.5\)) show the foam’s spring-like return to its unstressed state. Up to this neutral stress state, this residual stress aids in increasing actuation efficiency as described in the main text.
Figure 3.10 Residual strain-dependent tensile behavior. Tensile behavior of an uncompressed foam (black) and a foam compressed to \( R_c = 1.5 \) prior to the start of the test (red).

3.7.5 Macroscopic Creasing from Compression

Some macroscopic creasing of the constrained foam samples occurred within the molding process. For some tensile samples, these creased areas were the site where foam tearing initiated (Figure 3.11). These regions of concentrated stress caused the foam to begin tearing before the foam had fully unbuckled.
**Figure 3.11 Tensile failure of a compressed foam.** $R_c = 1.5$ sample showing macroscopic creasing present throughout tensile test, eventually leading to tear initiation. Scale bar, 10 cm

### 3.7.6 Compressed Foam Porosity Estimation

By assuming that foam compression reduces only the pore volume (i.e. no PU compression), we predicted the porosity and density of a foam using simple volume calculations. We used the definitions of uncompressed foam porosity (Equation 3.3), compressed foam porosity (Equation 3.4), and compression ratio (Equation 3.5) with our assumption that foam compression reduces only pore volume (Equation 3.6).

\[
\phi_{\text{uncompressed foam}} = \frac{V_{\text{void, uncompressed foam}}}{V_{\text{uncompressed foam}}} \tag{3.3}
\]

\[
\phi_{\text{compressed foam}} = \frac{V_{\text{void, compressed foam}}}{V_{\text{compressed foam}}} \tag{3.4}
\]

\[
R_c = \frac{V_{\text{uncompressed foam}}}{V_{\text{compressed foam}}} \tag{3.5}
\]
\[ V_{uncomp.\ foam} - V_{compressed\ foam} = V_{void,uncomp.\ foam} - V_{void,compressed\ foam} \] (3.6)

where \( V \) signifies the volume of the various foams and components. In combining these, the following relation (Equation 3.7) emerges:

\[ \phi_{compressed\ foam} = 1 + R_c(\phi_{uncompressed\ foam} - 1) \] (3.7)

This estimation does not account for the densification strain limit; however, this simple volume argument predicted the porosity of the \( R_c = 2.5 \) foam to within 3% of calculated porosity from \( \mu \)CT imaging.

### 3.7.7 \( \mu \)CT Scan Image Analysis

We performed all image analysis of \( \mu \)CT image stacks using Matlab® and ImageJ software. After thresholding the images, we used the BoneJ plugin to calculate porosity using the “Volume Fraction” function and to generate a 3D image of the foam using the “Isosurface” function. Because the foam is open-celled, we used the “watershed” function in ImageJ to generate “cell wall” connections between pores (transforming the cells to closed-cells) in the thresholded images in order to measure cell sizes. We then used the “Analyze Particles” ImageJ function to label and measure the area of each pore. Finally, we converted the area calculation to the diameter of an equivalent circular pore and binned these into the histogram in the main text.

### 3.7.8 Model for 1D Actuator

The model for the 1D actuator consists of two elastic materials – the elastomer coating and the elastomeric foam. The state of the actuator under any applied inflation
pressure \((p_{\text{inflate}})\) and blocking force \((F_{\text{block}})\) is given by a force balance (Equation 3.8) with area calculations (Equation 3.9 and Equation 3.10).

\[
p_{\text{inflate}} \times A_{\text{foam}} - F_{\text{block}} = \sigma_{\text{foam}} A_{\text{foam}} + \sigma_{\text{coating}} A_{\text{coating}} \tag{3.8}
\]

\[
A_{\text{foam}} = \pi r_{\text{foam}}^2 \tag{3.9}
\]

\[
A_{\text{coating}} = \pi [(r_{\text{foam}} + t_{\text{coating}})^2 - r_{\text{foam}}^2] \tag{3.10}
\]

where \(A_{\text{foam}}\) and \(A_{\text{coating}}\) are the cross-sectional areas of the foam and coating, respectively. The coating is taken to be linear elastic with an elastic modulus of 0.43 MPa. The stress on the coating is therefore given by Equation 3.11,

\[
\sigma_{\text{coating}} = E_{\text{coating}}(\lambda_{\text{actuation}} - 1) \tag{3.11}
\]

where \(\lambda_{\text{actuation}}\) is the actuation stretch relative to the pre-compressed state. The foam stress at any given stretch is read directly from the experimentally determined stress-strain response of the foam. This stretch experienced by the foam is the product of the pre-compression stretch (0.5 in our case) and the actuation stretch (Equation 3.12).

\[
\lambda_{\text{foam}} = \lambda_{\text{actuation}} \lambda_{\text{pre-compression}} \tag{3.12}
\]

The actuation stretch includes any springback after the actuator is released from the mold. Based on both experimental observations and analytical results, this springback is small for this particular actuator geometry and material properties.

We also used this model to examine the effect of the coating modulus on an actuator inflated to 90 kPa. In Figure 3.12 (left), we examined the effect of the coating elastic modulus on the blocked force properties. The maximum force is independent of the coating modulus as this is the entirely blocked force (i.e. the coating experiences no strain). When the actuator is strained, we observe that a more compliant coating
modulus enables a higher maximum strain. A tradeoff exists, however, because a coating that is exceedingly compliant will be unable to retain the foam compression as shown in Figure 3.12 (right). Here, we observe the behavior of foam pre-compression strains of 0.25, 0.5, and 0.75 (corresponding to $R_c$ values of 1.3, 2, and 4, respectively) as a function of the coating modulus (given the geometric and material parameters measured for our actuators). We see that a coating modulus of less than ~0.2 MPa exhibits noticeable foam springback leading to a higher equilibrium pre-compression strain. A coating with a lower modulus than this is not sufficiently stiff to constrain the residual strain within the foam and the resulting actuator would show reduced benefits of the pre-compression. While we did not attempt to optimize actuator geometry and material properties in this work, the tradeoff would be an important focus for future studies.

**Figure 3.12 Modeled behavior of an extending foam actuator.** The effect of coating stiffness on the blocked force properties (left) and on the equilibrium pre-compression strain (right).
3.7.9 Blocked Force Actuation

The full blocked force data set is shown in Figure 3.13 below.

![Figure 3.13 Blocked force and unloaded strain measurements.](image)

**Figure 3.13** Blocked force and unloaded strain measurements. The maximum force (left) and maximum actuated strain (right) of experimental and modeled uncompressed and compressed extending actuators as a function of inflation pressure.

3.7.10 Conversion of an MRI/CT Image Stack to a Digital Heart Model

We used Mimics (Materialize) medical imaging software to convert MRI/CT image stacks to 3D digital models. We used a heart MRI scan from OsiriX (http://www.osirix-viewer.com/datasets/DATA/AGECANONIX.zip) as the heart model for our device. Raw image stacks (**Figure 3.14A**) were cropped and thresholded (**Figure 3.14B**) to create a mask stack forming a 3D model of the isolated heart (**Figure 3.14C**). The model often contained residual vasculature that we subsequently cropped using the “Edit Mask in 3D” function (**Figure 3.14D,E**). We found it helpful to create a new mask after cropping using the “Calculate Mask from Object” function. This new mask allowed us to apply region growing to remove any areas not attached to the main body of the model. Additionally, this mask allowed us to use the “Smooth Mask” function to reduce the roughness of the resulting digital
model (Figure 3.14F). Finally, we also applied the “Wrap” function to fill any remaining holes in the model (Figure 3.14G). Finally, we exported the model as an .stl file which we either 3D printed directly or used as the basis for the mold built in a CAD program.

Figure 3.14 Conversion of MRI image stack to a 3D digital model. (A) Orthogonal views of raw image stack. (B) Thresholding to isolate the heart (C) which was subsequently cropped (D-E), smoothed (F), and wrapped (G) to form the final model.

3.7.11 Direct Cardiac Compression Device based on Porcine Heart Model
We began fabrication of the porcine-based device by inserting two pieces of PU tubing within a 3D-printed mold (20 mm thickness) before casting self-foaming PU within the mold. After demolding, we visually identified the coronary arteries from the porcine heart and manually duplicated them on the foam surface (Figure 3.15A). We then excised two regions of the foam adjacent to the arteries (Figure 3.15B) and
discarded the foam containing the arteries location. Next, we applied a constraining coat of PU elastomer to the two excised foams and compressed them into a smaller 3D-printed mold (10 mm thickness). Once the elastomer had cured, the foam pieces retained their compressed shape after demolding (Figure 3.15C). We then applied a layer of 5 % (wt.) chopped carbon fiber (1/4” chopped graphite fibers, FiberGlast) in PU elastomer onto the external surface of the foams to direct their inflated expansion towards the center of the device. Finally, we placed the foams back within the 10 mm thick mold and cast PU elastomer to fill the areas between the foams. Figure 3.15D shows the assembled device.

**Figure 3.15 DCC device fabrication and assembly.** (A) Molded foam shell (thickness = 20 mm) with simulated coronary artery location identified, (B) excised foam chambers with input tubes visible, (C) PU elastomer-coated foam chambers after compressing in smaller mold (thickness = 10 mm), (D) final device design after applying PU elastomer + carbon fiber to the exterior foam surface and curing PU elastomer in the coronary artery location. Scale bar, 2 cm.

**3.7.12 Volume Displacement of Device**

We performed a water displacement experiment to determine the volume displacement of an individual foam chamber. We submerged the foam chamber in a beaker of water and while inflating a single foam chamber from a compressed air reservoir at 170 kPa.
The chamber was cyclically inflated for 220 ms and vented for 440 ms. We observed a water displacement of roughly 70 mL within 200 ms of inflation (Figure 3.16) and observed that this displacement was repeatable over several cycles.

**Figure 3.16 Volume displacement of a single foam chamber.** A water displacement setup shows the fluid level at 550 mL at time = 0 (A) and at ~ 620 mL at time = 200 ms (B).

### 3.7.13 Ex Vivo Demonstration Experimental Setup

For this initial demonstration of the DCC device, we compared the water flow produced from manual massage of the porcine heart to the flow generated from the DCC device. To inflate the device, we routed air from a house pressure line through an electro-pneumatic regulator (ITV1031-21N2BL4, SMC) through a solenoid valve (CJV23-C12A1) and into a 3-way valve (912-000001-031, Parker). A house vacuum was also connected to the 3-way valve via an additional solenoid valve. The 3-way valve was then connected to the DCC device through a pressure sensor (015PGAA5, Honeywell). The flow of the water pumped through the porcine heart was measured via an in-line flow meter (Co.55/120, Sonoflow). A schematic of this setup as well as the manual massage and pneumatically pumped fluid flow profile is shown in Figure 3.17 below.
Figure 3.17 Ex vivo demonstration set-up. Schematic of experimental set-up and (A) fluid flow rate profile for a manually pumped (B) and pneumatically pumped (C) porcine heart.
Table 3.2 Comparison of model and experimental actuation. Maximum force and maximum actuated strain for experimental and modeled extending actuators. Experimental values are mean ± standard deviation.

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3.8 REFERENCES


CHAPTER 4

A VARIABLE SHAPE AND VARIABLE STIFFNESS CONTROLLER FOR
HAPTIC VIRTUAL INTERACTIONS

4.1 Abstract

This paper presents an entirely compliant controller handle for use in virtual and augmented reality environments. The controller handle transitions between two static states: a semi-rigid, large diameter state when pneumatically pressurized and a soft, compressible, smaller diameter state when depressurized. We integrated the controller with a modified version of NVIDIA’s VR Funhouse employing the two controller states to simulate the physical feel of two virtual objects. We used finite element modeling to downselect an internal elastomer lattice within the controller that controls deformation upon inflation. Finally, we show an example of using the compliance of the handle as an interaction input by designing an algorithm to identify rapid compressions of the handle as a signal to swap objects in the virtual environment.

4.2 Introduction

Virtual- (VR) and Augmented-Reality (AR) are becoming increasingly
prevailing tools for simulation, training, and entertainment. While current systems excel at immersing the user visually and aurally in a simulated environment, haptic interactions are usually limited to rigid controllers that offer only vibrotactile feedback, or no feedback at all. Many technologies have shown promise towards providing interfaces capable of haptic interactions including ultrasonics [1], voice coils [2], electrovibration [3], and fluidic pressurization [4]. Among these, fluidic actuators are unique in providing both high displacements and high forces [5]. Building upon these benefits, fluidic actuators have been integrated within gloves [6, 7] designed to provide haptic feedback; however, challenges remain in designing a universally-fitting glove that incorporates powerful, yet lightweight and safe actuators. Aside from glove motifs, soft pneumatic devices have also been investigated as shape-changing user interfaces [8–10]. In this project, we aimed to combine the forceful actuation of pneumatically powered soft devices with a universal controller handle that can represent the physical properties of virtual objects by changing shape and stiffness (Figure 4.1). The particular mode of interaction we use may be described as Dynamic Passive Haptic Feedback [11], meaning the shape and stiffness transition between stable states that are maintained for extended durations. We note that the feedback is both tactile and kinesthetic, as the controller’s shape change expands across the user’s fingertips while also pushing against and deforming in response to the user’s grip.
Figure 4.1 Soft haptic controller design, fabrication, and operating principle. The compliant, controller handle is composed of an internal elastomer lattice structure (A) and is printed via stereolithography (B). The handle expands from its unpressurized state (C, left) and increases stiffness when pressurized (C, right). Scale bar, 3 cm.

Our choice to use compliant, pneumatic actuators provides several advantages to this study. Air powered actuators can undergo large deformations in short timespans [12] while having a lightweight structure. Compressed air can also provide forceful actuation from readily available sources (e.g., compressors or storage canisters). Recent studies have shown methods to program the deformation of compliant, pneumatic systems; these methods, however, are limited to 2D deformation from 1D actuators [13] and 3D deformation from 2D actuators [14]. Here we seek to explore a foundation for programming 3D deformation from 3D actuators.

4.3 Materials Selection

To fabricate the haptic controller, we chose to use a commercially available
stereolithography printer (Carbon M1). To print the components, we selected their EPU-40 and RPU-70 materials. EPU-40 is an elastomeric polyurethane resin that exhibits both high compliance and high ultimate elongation; both of these properties are essential for the highly deformable structures designed in this work. After printing, we post-processed all EPU-40 and RPU-70 components using the following procedure: We rinsed all parts briefly with isopropyl alcohol and used pressurized air to remove residual uncured resin from internal chambers, then applied uncured, unprinted resin to close any open vent holes and placed the part in a UV chamber (ECE 5000 Flood, Dymax) to solidify the resin. Finally, we then thermally cured the parts in an oven for 8 hours at 120 °C.

4.4 Design and Fabrication

4.4.1 Internal Lattice

We elected to integrate lattices into the controller’s interior for three reasons. First, the internal lattice provides structural support to the inflation chambers. Though the lattice is composed of an elastomer, it is sufficiently stiff to retain the external chamber shape. Second, internal elastomeric lattices aid in the device’s deflation rate by providing an elastic restoring force towards the controller’s uninflated shape. Finally, the lattices we chose provide an open, interconnected cellular network through the controller that enables resin flow during printing and air flow during actuation. Resin transport is important during printing as it reduces suction forces and trapped resin, both of which could lead to print failure.
To limit the search space for appropriate internal support structures, we used Bravais lattice geometries and modeled their deformation using finite element (FE) analysis. The lattices included simple cubic (SC), body-centered cubic (BCC), face-centered cubic (FCC), the union of BCC and FCC structures (BCC + FCC), and FCC without struts in the yz plane (FCC no yz). In addition to these base lattices, we also investigated a hexagonal honeycomb lattice structure.

4.4.2 Controller Handle

We designed the controller handle to be held with one hand and to have a dumbbell shape that curves around the user’s fingers. A dumbbell shape creates an affordance that guides the user to wrap their fingers around the inner section of the device, where the interaction (and inflation) takes place. This affordance is meant to allow users to understand the shape of the device without seeing it, and to encourage them to readjust their grip during transitions between inflated/deflated states. The handle consists of an exterior skin and a central hollow cylindrical conduit (Figure 4.1A) which are connected internally by the lattice. The cylindrical conduit enables wired connections to reach the button housing at the top of the controller.

The controller is composed of four components: (from the top of the controller, moving down to the handle base) the tracking device, the button housing, the inflatable handle, and the tether to the air source and control box (Figure 4.2). We used a commercial Vive Tracker (HTC) that enables position/orientation tracking and simple integration with the HTC Vive VR setup. The button housing (Figure 4.2A) served to provide both an interface between the Vive tracker and the handle and a rigid
frame for the interactive button. The interface between the Vive tracker and the housing consisted of a 6-pin electrical connection and a threaded screw connection (Figure 4.2A). We designed the housing to be a hollow structure to contain the wires necessary for both the tracker and the button. The modularity of these components allowed for rapid iteration of controller handle designs.

![Figure 4.2](image)

**Figure 4.2 The modular design and assembly of the controller handle.** (A) Our printed button housing connection to the Vive tracker, (B) the connection between the rigid button housing and the compliant handle, (C) the connection between the handle and the tether sleeve, and (D) the assembled controller. Scale bar, 6 cm.

4.5 Experimental Set-up

We built digital models of each lattice in Solidworks 2017. The cubic lattices (simple, BCC, FCC, FCC no yz, and BCC+FCC) consist of 27 unit cells arranged in a 30x30x30 mm cube. These lattices are composed of struts with an octagonal cross-section with an edge to edge thickness of 1 mm. The simple lattice consisted of only vertical and horizontal struts, while the BCC and FCC structures had struts passing through their body diagonal and face diagonal, respectively. The hexagonal lattice is
composed of a tessellated array of hollow hexagonal prisms with an edge to edge
dimension of 12 mm, a thickness of 0.3 mm, and a height of 10 mm as shown in
**Figure 4.1A**. We exported all digital models as .SAT file types to be imported into FE software.

We used Static Structural analysis within Ansys Workbench 18.1 for all FE modeling. We imported the EPU 40 elastomer into the model as a 1st order Ogden material, as the Ogden model provided the best fit to our experimentally-measured uniaxial tensile data (**Figure 4.3**). Ansys provided the fitting parameters of 1.4009E6 Pa, 2.4477, and 0 Pa$^{-1}$ for Material Constant MU1, Material Constant A1, and Incompressibility Parameter D1, respectively, based on our imported tensile data. In the simulations, we meshed the lattices with an element size of 0.1 mm and enabled large deflections under Analysis Settings. We fixed the bottom lattice surface as a fixed support and applied tensile and compressive force (-0.01 N to 0.01 N) to the top surface of each lattice. We measured the directional deformation response (in the direction parallel to loading) as the load was applied.
We performed all tensile tests according to ASTM D412 on a Zwick Roell z010 instrument. All tests used a 10 kN load cell and pulled at a strain rate of 10/min.

In order to integrate our custom controller with the application, we leveraged the HTC Vive tracker’s integration with the SteamVR runtime. A firmware switch caused the tracker to appear as a normal controller to the application with full lighthouse-based tracking. NVIDIA’s VR Funhouse was modified to send commands to a python script connected over a USB cable to an Arduino microcontroller that operated the valves. Furthermore, we added a virtual model of a foam sword using a soft-body physics simulation to the VR Funhouse environment.

4.6 Results and Discussion

We used FE analysis to analyze several lattice geometries in compression and tension. While the relationship between stress and strain is often used to describe a
material’s tensile and compressive behavior, we intended to simulate a user applying a force to the controller’s walls and then feeling a deformation. Because of this, we chose to observe the relationship between force and deformation directly in this study. A representative force vs. deformation plot is shown in Figure 4.4 with accompanying images of the deformed lattice. We observe the deformation response to be linear under tensile loads and nonlinear under compressive loads. This loading (0.7 N) was sufficient to impart buckling of the lattice struts which forms the nonlinear behavior. When under compression, the lattice initially deforms linearly until it reaches a load of approximately 0.5 N. At this load, the struts begin to buckle and the lattice experiences large deformation with small increments in load. This behavior is consistent with compressive collapse of porous elastomers [15, 16].

![Force vs. Deformation Plot](image)

**Figure 4.4 Simulated deformation of an elastomer lattice.** Representative simulated deformation of an elastomer lattice and the tensile and compressive simulated deformation of a simple cubic lattice. The simulated lattice shape (from top to bottom) at high tensile load, no load, and high compressive load.

As shown in Figure 4.5, each lattice deformed differently in compression than
in tension. In tension, all lattices demonstrated a largely linear response between deformation and force. Under compressive forces BCC, FCC no yz, and hexagonal lattices showed nonlinear behavior. This is due to the absence of struts parallel to the applied force in these lattices. The struts easily bend and buckle to allow large deformation under low forces.

![Simulated elastomer lattice deformation](image)

**Figure 4.5 Simulated elastomer lattice deformation.** (A) The lattice models evaluated in this study and (B) their simulated tensile and compressive behavior. Scale bar, 10 mm.

When designing the controller handle prototype, we made the internal structure a hexagonal honeycomb lattice. This structure was unique among those we examined in that it is largely radially symmetric (in the plane of the hexagons) and is relatively rigid in the perpendicular direction. This anisotropic behavior was oriented in the part to aid in printing the handle via stereolithography. That is, we aligned the stiff direction of the lattice with the build direction of the printer to reduce sagging during the print. This orientation of the lattice also allowed resin to drain easily as the part was drawn out of the printing resin.

We integrated the controller handle with a modified version of NVIDIA’s VR
Funhouse software. The demonstration consists of a single virtual environment where the user is surrounded by inflated balloons. The user initially interacts via a virtual metal sword which can pop the balloons. When using this sword, the controller handle is pressurized and increases its diameter and stiffness. With a keyboard or button press on the controller handle, the controller handle deflates to its highly compliant and reduced diameter state and the virtual object changes accordingly to a compliant, foam sword that is unable to pop balloons and instead bends when in contact with the balloons. The user can easily deform the handle by squeezing or bending in this state. The transitions between these states (e.g., inflating and deflating handle) occurs on the timescale of seconds in order to avoid potentially jolting rapid shape change. Inflation occurs over ~2 seconds and deflation occurs over ~7 seconds. These durations are dependent on the air source flow rate and the handle’s air input diameter.

Inflation/deflation could be increased and decreased by modifying these parameters.

The handle is controlled by a microcontroller (Arduino Uno) that integrates pneumatic valves (912-000001-031, Parker), a pneumatic pressure sensor (PSE530-R06, SMC Pneumatics), a relay (SRD-05VDC-SL-C, Songle), and the serial connection to a VR-ready computer (Figure 4.6). The valves control the handle’s internal air pressure by inflating from a pressure source (air compressor, model TC-848, TCP Global) or venting to atmosphere. The relay allowed an external button to trigger the transition between virtual objects.
We also demonstrated handle deformation as an input mechanism. Using real-time readings of the inline pressure sensor, we identified short duration pressure spikes (caused by the user squeezing the handle) as a signal to change objects in the virtual environment. Because we intended occasional deformation of the handle as part of normal handling, we wrote an algorithm to avoid false positives. This algorithm ignores isolated pressure spikes and identifies only two pressure spikes in quick succession as a signal to change the virtual object. A typical pressure profile showing this interaction is shown in Figure 4.7.
Figure 4.7 Interaction with the soft haptic controller. (A) Representative pressure profile of user-triggered inflation and deflation. (B) Sequence of demonstration interactions: [from left to right] interaction with foam sword, compression of the unpressurized handle triggering inflation, interaction with the metal sword, and compression of the pressurized handle triggering deflation. Scale bars, 3 cm.

4.7 Conclusions and Future Work

In this work, we investigated a foundation for designing and evaluating lattice-based pneumatic devices. We used FE analysis to evaluate several representative lattice structures in both tension and compression. Each lattice demonstrated unique behavior relative to the other structures. Lattices that contained struts which were
easily bent and buckled contributed to more compliant behavior in compression. An expanded deformation map, including additional lattices and/or additional loading directions, could be used to design an internal lattice structure in which each unit cell dictates the local deformation at that point in 3D space, allowing fully programmable force/deformation properties.

We also designed and fabricated a controller handle for interaction with different virtual objects. The controller handle switched between two states, pressurized and unpressurized, to simulate both a rigid and a compressible virtual object. We used the compliance of the controller as a means of interaction. By squeezing in quick succession, the user was able to change the form of objects in the virtual environment and the stiffness and shape of the pneumatic controller.

While this handle design contained a single pneumatic chamber, future versions could contain multiple individually addressable chambers to simulate objects with more complex geometry and finer mapping to the human hand. With further refinement of the design, it may be possible to simulate changes in texture as well as changes in shape and stiffness.
4.8 REFERENCES


CHAPTER 5
CONCLUSIONS

5.1 Summary of Contributions

This work describes my contributions in the areas of compliant machine fabrication, design, and application using porous elastomers. These contributions are summarized below.

An elastomer foam system that enabled simple fabrication of intricately shaped compliant machines. The PDMS/ammonium bicarbonate system described in Chapter 2 generated an open-celled foam allowing simple molding of soft actuators in several shapes.

A compliant fluid pump inspired by the human heart that provided a higher flow rate of than any other entirely compliant pump. The fluid pump described in Chapter 2 attained a maximum flow rate of 430 mL min\(^{-1}\). At the time of publication, this was greater than any other reported values for compliant fluid pumps. In addition, this pump mimicked the human heart is several ways including providing unidirectional and pulsatile fluid flow at physiological frequencies and fluid pressures.

A buckled elastomer foam fabrication method that enhanced actuation while reducing machine volume. The material system described in Chapter 3 was the first to use foam in a state of residual compression to aid actuation. In particular, I found that a foam compressed to half of its length exhibited ~ 1.4x the force and ~2x the stroke compared to an uncompressed foam actuator.
A unique patient-specific DCC device design. Though several other DCC devices existed previously, the design described in Chapter 3 was the first to be formed as a patient-specific device, through the use of CT, MRI, or 3D scanning processes.

A haptic controller capable of changing stiffness and shape. The work presented in Chapter 4 showed a process to compare the tensile and compressive behavior of elastomer lattices using FEA. I used this information to design a lattice-based, pneumatic controller handle. We also imparted a sensing capability into the handle, where squeezes of the controller were detected as a user input.
A.1 Introduction

Synthetic composites usually have fixed internal structures and mechanical properties tuned for a particular application. Natural composite materials, however, can alter their structures and mechanical properties in response to changing environmental conditions. Bone, for example, remolds itself upon induced mechanical stresses [1,2]. When touched, the sea cucumber can rapidly and reversibly increase the stiffness of its skin for protection [3,4]. Here we present a synthetic composite material that demonstrates some of these abilities (i.e. stiffness variation and shape morphing) and incorporate this material into a soft robotic tentacle and morphing wing.

Recent work on variable stiffness composites based on inducing phase changes in polymers [5-9] and metals [10,11] has begun to show promise in overcoming the

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usual tradeoff between shape adaptability and load-bearing capability. These synthetic materials are often layered composites [9] or contain soft microchannels filled with a low melting temperature material [8,11]; when this phase change material is molten, the composite can be easily deformed. Such structures are highly sensitive to cracks in the stiff material and their need for layered assembly results in anisotropic mechanical properties. In this communication, we describe the material design and simple processing of a bicontinuous [12] network of two foams—elastomeric and metallic.

The metal (an alloy of indium, tin, and bismuth) melts at $T_{m}^{\text{alloy}} = 62^\circ C$. When the composite’s temperature is below $T_{m}^{\text{alloy}}$, the mechanical properties of the solid metal foam dominate and the material is stiff; above $T_{m}^{\text{alloy}}$, the metal is molten and the elastomer foam dictates the composite’s mechanical properties, making the composite rubbery. In this paper, we demonstrate six capabilities of this bicontinuous foam composite: (i) reversible stiffness, (ii) an ability to change shape by stretching to over 200% strain when heated, (iii) shape memory actuation [13-16], (iv) non-autonomic, heat-triggered self-healing [17,18], (v) assembly into larger continuous structures from smaller sub-components, and (vi) strengthening along the primary axis of deformation after morphing.

We chose eutectic Sn-Bi-In (Field’s Metal, RotoMetals) as the metal for its low melting temperature, low toxicity, and high elastic modulus ($E_{m}^{\text{alloy}} \approx 9.25 \text{ GPa}$) [11] at room temperature. To allow a large range of shape morphing with little resistance to deformation, we used a silicone elastomer (Elastosil® M4600, Wacker) for our matrix material due to its large ultimate strain, $\varepsilon_{\text{ult}}^{\text{silicone}} = \frac{L-L_0}{L_0} \sim 8$, and low...
elastic modulus, $E_{silicone} = 0.54 \text{ MPa}$ [19]; additionally, silicone is stable over the temperature range required to melt the metal alloy [20,21].

To fabricate this bicontinuous structure, we formed an open-cell network of pores in silicone (average diameter of 2 mm) (Figure A.1A) via dissolution of a densely packed fugitive salt [22]. We then impregnated the silicone foam’s open-cell network with molten metal (Figure A.1B) using a microfluidic channel outgas technique (Figure A.2) [23]. This process produces uniformly distributed, interpenetrating stochastic networks of Field’s metal and silicone, resulting in an isotropic composite that can be cut or molded into a variety of 3-dimensional geometries. See Supplementary Information (SI) for more details about this procedure.
Figure A.1 The elastomer foam/Field’s metal composite system. (A) Silicone foam, (B) imbibed with Field’s metal. Composite under 500 g bending stress at (C) \( \sim 23 \) °C and (D) > \( T_{m}^{\text{alloy}} \). Mechanical testing of foam and composite in (E) tension and (F) compression. See Supporting Information Figure S4 for complete curves of E and F.
At above $T_m^{alloy}$, the molten metal remains within the elastomeric foam even during deformation and contact with other composite surfaces. This strong impregnation is likely due to surface oxidation of the liquid metal, which adheres strongly to the silicone surface [24-26]. As the molten metal is now in contact with its own oxide, strong surface wetting results in high capillary forces. Table A.1 shows the results of an experiment conducted to demonstrate the strong capillary forces (see SI text for more details).
Table A.1 Data from repeated welding experiment

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Although some commercially available elastomer foams exist, the selection is limited and tends to contain primarily small-pore or closed cell foams that have lower ultimate strains and higher elastic moduli than our selected silicone material. As we found that large pore sizes ease infiltration of the metal and we require an open cell foam network, we chose to form our own foams using the lost salt method with large grained salt (i.e., “Himalayan”, Pure Himalayan Salt). While directly 3D printing the pore network could enable precision control of the composite architecture, our stochastic process is simple, easily duplicated, and results in an isotropic structure. Controlling pore size and porosity allows us to tune the properties of our composite. For example, increasing the silicone porosity would result in a lower average cross-sectional area of the silicone and a higher average cross-sectional area of the metal. In this case, the effective elastic modulus of the silicone foam would decrease and the overall elastic modulus of the composite foam would increase. Presently, we have chosen an intermediate elastomer foam porosity \( \phi_{EilFoam} = \frac{v_{sil}}{v_{sil} + v_{metal}} = 0.5 \); however, 3D printing strategies could increase this porosity even further and result in...
stronger metal foams that would likely improve the self-healing and welding capabilities of the composite. If the pores become too large, however, the capillary forces retaining the molten metal may become insufficient. Alternatively, increasing the volume fraction of elastomer in the composite will increase the composite’s restoring force for shape memory actuation.

For mechanical testing, we molded dumbbell and cylindrical specimens using ASTM D412 - 06a (2013) and ASTM E9-09 standards respectively. We imbibed half of the samples and measured the mechanical properties of the resultant pure elastomer foams and composites by conducting uniaxial, monotonic tension and compression tests on a Zwick & Roell Z010 testing system according to the ASTM standards (See SI for more details). For reference, we tested sealed samples of empty elastomer foam and the foam imbibed with molten metal and found that the mechanical properties were nearly identical (Figure A.3); therefore, we present here only the empty elastomer foam. We note here that the mechanical properties of the sealed samples do not match those in Figure 1E; this difference is likely due to silicone degradation as a result of salt exposure over the extended periods between testing [27].
Figure A.3 Mechanical testing. Data for sealed samples of silicone foam vs. the composite (A) at room temperature and (B) above $T_m^{alloy}$.

Bars (2.5 cm x 11.7 cm x 1.1 cm) of the composite sustained bending loads of 4.9 N without noticeable deflection (Figure A.1C); upon melting of the metal, the bars immediately yielded (Figure A.1D). Tensile testing data (Figure A.1E; SI Text) show that at room temperature, the composite has an elastic modulus of $E_{tensile}^{BiFoam} \sim 1.8 \text{ MPa}$, which decreases to $E_{tensile}^{ElFoam} \sim 0.1 \text{ MPa}$ at $T > T_m^{alloy}$, measured at
strains $0.01 < \varepsilon < 0.03$ and $0.20 < \varepsilon < 0.40$ respectively. The compression testing data (Figure A.1F) show similar results, with $E_{\text{BiFoam}}^{\text{compr}} \sim 3.1 \text{ MPa}$ at room temperature and $E_{\text{ElFoam}}^{\text{compr}} \sim 0.1 \text{ MPa}$ at $T > T_m^{\text{alloy}}$, measured at strains $0.01 < \varepsilon < 0.03$ and $0.20 < \varepsilon < 0.40$ respectively.

The composite has two functional mechanical states: stiff and compliant. Exploiting the dramatic variance in mechanical properties, we morphed composite structures by heating them to $\sim 70$ °C, deforming them into bent, twisted, and elongated shapes, then allowing them to cool below $T_m^{\text{alloy}}$ to freeze the material into these configurations (Figure A.4A-D). Above $T_m^{\text{alloy}}$, the molten metal deforms and reorients with the foam structure under tensile loading (Figure A.5A-B). We note, however, that during significant compression, the pressure component of the stress forces some of the molten metal out of the elastomer matrix (Figure A.5C-D). In situations where significant compression was expected to occur, we sealed the composite with a thin ($\sim 2$-3 mm) silicone skin (Figure A.3).
Figure A.4 Shape fixing of the composite. Morphed configurations that demonstrate the composite’s ability to hold (A) bent, (B) twisted, (C) relaxed, and (D) elongated positions at room temperature. (E) Stress/Strain plot of “pre-strained” tensile samples.
Figure A.5 Temperature dependent stiffness demonstrations. Composite bar in tension (A) at room temperature with 0.95 kg and (B) above $T_m^{\text{alloy}}$ with 2.00 kg. Composite bar in compression under a 2.00 kg mass (C) at room temperature and (D) above $T_m^{\text{alloy}}$. 
When liquid, the cells of the metal foam align and lengthen in the direction of tensile loading; upon crystallizing, the metal foam increases the composite’s strength along the axis of deformation. We characterized this capability by heating dumbbell samples to 70 °C, stretching them to three different strains ($\varepsilon \sim 0.27, 0.50, 0.71$), and allowing the metal to cool below $T_m^{alloy}$ (Figure A.4). We found that increasing pre-strain causes the ultimate strength to increase. By pre-straining the composite to $\varepsilon \sim 0.7$, the yield stress increases from $\sigma_y^{0} \sim 0.08$ MPa to $\sigma_y^{0.7} \sim 0.27$ MPa. Commensurately, the residual extensibility of the composite is reduced from $\varepsilon_{ult}^{0} \sim 2.9$ to $\varepsilon_{ult}^{0.7} \sim 1.6$. Though not previously reported in a dynamic system, aluminum foams have shown similar dependence on pore structure [28]. Nieh et al. report that when a cell experiences load along the longitudinal direction, the effective elastic modulus is proportional to the length of the cell’s primary axis and inversely related to the cross-sectional area. Therefore, as the cells elongate and their cross-sectional areas shrink, the effective elastic modulus of the composite increases.

The elastomer foam acts as an entropic spring [29] that stores energy when deformed, allowing it to be used as a shape memory actuator when a shape is locked in using the frozen metal foam. We took advantage of this resilience to enable shape memory actuation. As a demonstration, we morphed a rectangular cuboid (27 mm x 27 mm x 9 mm) into a cylinder (diameter, d= 25 mm; Figure A.6A, left). By heating the composite to above $T_m^{alloy}$, the metal melts and the elastomer foam relaxes and returns to its cuboid form (Figure A.6A, right). When the strain of the elastomer foam remains below $\varepsilon \sim 2.5$ (Figure A.7A), the force at a given strain is linearly
proportional to $E_{\text{compr}}^{\text{ElFoam}}$. Exposed to a stream of hot (500 °C) air, the compressed cube took 78 seconds to recover 89% of its original shape. By encapsulating the composite in an elastomer membrane, we prevented leaking during compression and, within the limits of our imaging system, recovered 100% of the original shape.

Figure A.6 **Shape memory behavior and pneumatic actuation.** (A, left to right) A compressed cylinder of metal-elastomer composite expanding into a cuboid after melting the metal foam. (B) Pneumatic tentacle with (left) solid endoskeleton being (middle) actuated after melting the metal foam and then (right) frozen in position by
the solidification of the composite. (C) Pneumatic wing with (left) solid exoskeleton that is (right) liquefied and caused to bend by pneumatic actuation.

Figure A.7 Mechanical testing data. Complete curves for mechanical testing of foam and composite in (A) tension and (B) compression.

In addition to characterizing the composite’s mechanical properties and intrinsic shape memory actuation, we incorporated this material into soft machines for
“on-demand” skeletal networks. We demonstrate two soft machines using pneumatically actuated structures: a tentacle and a morphing wing. The tentacle design was inspired by a previously reported soft actuator [30] with three degrees of freedom, allowing it to bend in any direction. At the core of this tentacle, we inserted the metal elastomer composite to act as a variable stiffness endoskeleton and enable stiffness even when unpressurized. **Figure A.6B** (left) shows the tentacle relaxed, (middle) heated and bent through actuation of two of the chambers, (right) then frozen in this position while the chambers are deflated. The wing used a sealed pneumatic foam chamber [31] with a nylon strain-limiting layer to direct the mode of actuation to achieve bending (**Figure A.6C**). When unpressurized, the wing has a retracted and curved motif, yet remains stiff due to the exoskeleton of crystalline metal foam. To extend and unfurl the wing, we melt the metal and pressurize the foam actuator; in this mode, the wing is stiff due solely to pneumatic inflation.

Since the molten metal can recrystallize with itself, it is possible for cracks in the metal foam to self-heal. We used this ability to heal damaged foam composites. We quantified the results by cutting tensile testing bars in half and rejoining them by melting and refreezing the metal foams (SI Text). Though the silicone foam does not re-bond, **Figure A.8A** shows that the self-healed composite has an elastic modulus that is very close to that of a virgin sample. Self-healed samples also achieve 78% of the undamaged samples’ inelastic limit (measured at $\varepsilon=0.04$). The decrease in toughness and ultimate tensile strength comes from the irreversible damage done to the elastomer. The severed elastomer foam cannot heal and therefore provides no resistance to deformation; once the metal foam in a healed sample is damaged, the
elastomer foam cannot provide structural support as it does in virgin samples. **Figure A.8B-D** shows the cutting and healing of a composite bar, where the cut becomes difficult to see after repair—visually demonstrating the remarkable degree of healing achieved.
Figure A.8 Self-healing behavior. (A) Mechanical testing of self-healed, welded, and virgin composite specimens. (B) Composite bar being cut with scissors, (C) damaged composite bar, (D) self-healed composite bar. (E) Two independent composite samples (left) before welding and (right) after.
We conducted mechanical tests on self-healed samples with the worst possible damage: complete severing of the metal and elastomer foams. There are, however, more minor types of damage from which this composite may heal more completely. Most impacts will damage only the metal foam. In such cases, the elastomer foam remains intact and the metal foam can be melted and recrystallized. This material could also suffer puncture damage, where some of the elastomer foam remains intact and able to support load. In this case, self-healing could restore more of the original ultimate toughness.

Using the same method as for self-healing, we also welded two independent composite pieces into a monolithic structure (Figure A.8E). Each separate composite surface has both silicone and metal present (Figure A.1B); thus, when they touch, there is likely metal-metal contact. By heating the two composites while touching, the metallic surfaces mix and, upon freezing, the Field’s metal recrystallizes into a single form. We measured the mechanical properties of welded samples and compared them to virgin and self-healed ones (Figure A.8A). The elastic modulus and inelastic limit of the welded samples were nearly identical to those of the self-healed ones. Although cell alignment is not guaranteed in the welded samples, the similar moduli are likely due to the molten metal layer that forms along the cut surface during re-bonding. Remarkably, self-healed and welded surfaces can support loads almost as great as virgin samples before plastically deforming. As in the self-healed tests, the welded samples exhibited decreased toughness due to a plane of discontinuity in the elastomer phase.
The composite material we designed has a stiffness that can be dramatically varied through changes in temperature. This material can be stretched and reshaped into different rigid structures and can use stored energy in the elastomer to enable shape memory actuation. Under the application of heat, the same composite can self-heal after sustaining damage and can be assembled from a few subcomponents into myriad super-structures with a continuous metal skeletal system. These capabilities could be used to create multifunctional tools by reforming rigid structures into new shapes, (e.g., a hook into a spear). Like bone, this shape changing capability could also be used to make adjustments to the geometry and strength of a structure in response to its environment. Finally, we demonstrated that these composites can be used for skeletal networks in soft machines without sacrificing their shape adaptability.

The strength and shape adaptability of this class of composites can be improved by employing existing techniques. By using different porogens [32] or a 3D printed lost-wax mold [33,34], we could directly control the foam structure and optimize it for various applications. Designing the microstructure such that the metal grains have larger contact areas and therefore fewer areas of stress-concentration would increase the strength. We could also spatially vary the pore structure to form gradient mechanical properties that intentionally induce anisotropic behavior. Additionally, it is possible that the interpenetration of the metal is not total—future X-ray μCT studies will allow analysis and improvement of the metal foam’s interconnectivity[35]. The melting and freezing of the metal in this paper was generally achieved via external heating; however, the composite’s stiffness can be controlled via Joule heating the metal directly [10] or embedding soft, stretchable heaters [11], which will enable
shorter melting times. While we report variable stiffness caused by melting and freezing of low melting temperature metal alloys, similar mechanical behavior could be achieved by replacing the phase-changing material—the metal foam—with a thermoplastic one [5,6,8].

A.2 Acknowledgements

This work was supported in part by the Air Force Office of Scientific Research under award number FA9550-15-1-0160, by the National Science Foundation Graduate Research Fellowship under Grant No DGE-1144153, and by a grant from the Alfred P. Sloan Foundation.

We also thank Bryan Peele, who aided with the electronics for the embedded heating elements in our soft robotic devices.
A.3 Supplemental Information

A.3.1 Materials and Methods

Silicone Foam Fabrication

We formed the metal-elastomer composite by first mixing equal portions of parts A and B of Elastosil® M4600 together to initiate curing. We immediately added an equal volume of 1-3mm Himalayan salt crystals (Red Himalayan Crystal Salt, Pure Himalayan Salt). We spread this mixture into a laser-cut acrylic mold and cured it in an oven for 20 minutes at ~85°C. We then de-molded the samples and placed them into a warm water bath until all the salt was dissolved. This required changing the water bath several times to speed up the dissolution process. Once all the salt appeared to have dissolved, we rung the samples dry by hand and then placed them in the oven to allow all excess water to evaporate. To verify that all the salt had indeed dissolved, we inspected the samples by hand to feel for hard crystals, then measured their mass and compared that number to our recorded silicone mass, which was measured when initially mixing the two-part silicone together.

Composite Fabrication

After fabricating the silicone foam, we heated the samples to ~85°C, then submerged them in a molten bath of Field’s metal also at ~85°C. The metal bath was contained in an acrylic box and the foam was held beneath the surface of the metal by an acrylic grill that was adhered to the walls of the acrylic box (Figure A.2). With the
silicone foam submerged, we placed the metal bath back into the oven at ~85°C and allowed the system to sit for 10 minutes to ensure thermal equilibrium. At this point, we removed the system from the oven, placed it in a vacuum chamber, and reduced the pressure inside the chamber to -25 inHg for 1-2 minutes. Once air bubbles stopped emerging from the molten metal bath, we removed the impregnated foam samples from the molten metal bath, laid them on a flat acrylic surface, and allowed them to cool to room temperature. We then removed the thin metal skin that remained on the surface of the sample and measured the weight of the composite. Given the silicone and Field’s metal densities and volume fractions, we could calculate the expected total mass of the composite—which we did—and compared that with the measured mass. If the measured mass was too low, we returned the composite sample to the molten metal bath, waited for thermal equilibrium, and repeated the vacuum process.

A.3.2 Experimental Set-up

Tension Tests

All tension tests were uniaxial, performed on a Z010 Zwick Roell equipped with a BW91272 thermal chamber at a strain rate of 0.36 min⁻¹. This rate was chosen as a compromise between the testing standards for rubber (ASTM D412 - 06a(2013)) and those for metal (ASTM E8/E8M – 15a). All tensile samples were fabricated according to the dumbbell dimension standard Die A in ASTM D412 - 06a(2013).

Compression Tests

All compression tests were uniaxial, performed on a Z010 Zwick Roell at a
strain rate of 0.36 min⁻¹. This rate was chosen to match that of the tension tests. All compression samples were fabricated according to the “short” cylinder dimension standard in ASTM E9-09.

Tension Tests of Empty Elastomer Foam and the Composite

These tests were conducted using the sample fabrication and testing method described in “Tension Tests”. We tested 2 empty elastomer samples and 2 composite samples.

Tension Tests of Sealed Empty Elastomer Foam and Sealed Composite

We fabricated dumbbell samples for tension tests using the fabrication method described in “Tension Tests”. We then sealed those samples in a 3 mm skin of a lower elastic modulus silicone (Ecoflex® 0030, Smooth-On). We tested one empty elastomer sample at room temperature, one composite sample at room temperature, two empty elastomer samples at 80°C and four composite samples at 80°C. These tests were conducted to failure of the composite, before the skin ruptured. Each test at 80°C was conducted within the thermal chamber after heating for 30 minutes to attain thermal equilibrium.

Tension Tests of Self-Healed, Welded, and Virgin Composite Samples

We fabricated dumbbell samples for these tests using the fabrication method described in “Tension Tests”. To create self-healed samples, we cut the dumbbells at their vertical middle using a sharp razor blade. We then heated the samples to ~85°C,
placed the two cut surfaces in contact, and allowed the samples to cool back to room temperature. To fabricate welded samples, we cut three samples at their vertical middle and re-paired the segments so that no dumbbell half was paired with its original partner. We then heated the samples to ~85°C, placed the two cut surfaces in contact, and allowed the samples to cool back to room temperature. We tested three self-healed, three welded, and three virgin samples using the tensile testing method described above.

Compression Tests of Empty Elastomer Foam and Composite

These tests were conducted using the sample fabrication and testing method described in “Compression Tests”. We tested four empty elastomer samples and four composite samples.

Repeated Self-Healing Tests

Figure A.9 shows that this material can self-heal at least 3 times with a slight decrease in elastic modulus each round. Tensile tests were conducted using the sample fabrication and testing method described in “Tension Tests of Self-Healed, Welded, and Virgin Composite Samples”. Two virgin samples were tested and their average taken. One healed sample was tested and re-healed three times in total. The reduction of elastic modulus is possibly due to the elastomer skin delaminating from the composite surface and enabling molten metal to flow to that surface. Due to testing constraints (e.g., sample geometry and mechanical testing configuration), slight mechanical stimulation at the healing site was needed to promote flow of the molten
metal. Improved fabrication methods that directly bond the elastomer skin to composite should mitigate the issues of mechanical testing.

![Mechanical behavior over damage-heal cycles.](image)

**Figure A.9 Mechanical behavior over damage-heal cycles.** Mechanical testing data for a virgin sample and for one sample being repeatedly damaged and healed.

Repeated Welding Tests

**Table A.1** shows the results of an experiment conducted to demonstrate the strong capillary forces. Two independent composite pieces were welded and separated repeatedly and their masses measured at each step. The masses of the individual pieces changed by less than 0.7% at each step. The small changes in mass could be due to handling error when pulling the samples apart.
A.4 REFERENCES


B.1 Abstract

This paper describes an easily accessed manufacturing process for soft actuators. It does not require molds and uses safe, readily available materials: table salt and rubber molding compounds. This process involves sculpting or casting uncured rubber compounds and results in soft, open-cell foam structures, which can be sealed to form actuators. The foams have low elastic moduli ranging from 20 to 30 kPa, large ultimate strains over 3.5, and rapid fluid transport rates up to 30 L min\(^{-1}\) cm\(^{-2}\). To demonstrate the capabilities of this process, we sculpted a simple bending actuator, a gripper, and many other 3D shapes. Blocked-force measurements demonstrated that the simple bending actuator can exert up to 5 N of force at its tip, and the gripper picked up a 200 g object. This technique could enable engineers of all ages and skill levels to engage in soft robot fabrication, contributing to K-12 STEM education. In addition, the proposed manufacturing technique could be also interesting for the STEAM (Science, Technology, Engineering, Arts and Mathematics) community, thus combining Science and Arts. Additionally, this work has the potential to inspire a new, more inventive form of engineering by combining the artistic practice of free-form sculpting with robot fabrication.

** Published as:
B.2 Introduction

In the past few years, the field of soft robotics has become a well-defined discipline with practical uses [1,2]. Research groups have developed actuators, sensors, and robots that have potential use in biomedical devices for surgery and rehabilitation [3-7], in mobility and manipulation for autonomous walking and swimming [8-10], and as wearable robots [11-14]. Due to their compliant materials and manufacturing techniques, soft robots present a number of advantages over conventional ones: safe human-robot interaction, low-cost production [10,14-17], and simple fabrication with minimal assembly, all of which allow their potential fabrication and use in classrooms or small laboratories [18-20].

Current state-of-the-art soft robots are often composed of prismatic actuators, an artifact of the replica molding process used to make many of them [8, 21]. To fabricate 3D robots, without complex molding or multiple assembly steps, our lab developed poroelastic foam actuators [16]. We fabricated these foams by combining silicone elastomer as the base material and ammonium hydrogen carbonate (NH₅CO₃) as a fugitive porogen. The resulting foams have intrinsic pathways for fluid transport, which enable actuation when sealed and inflated. While useful in a laboratory or industrial environment, one of the product gases—ammonia—is toxic, making the fabrication process unsuitable for classrooms, small laboratories, or artist studios not equipped with exhaust hoods for toxic gases.

In order to increase participation in soft robotics by non-experts (e.g., artists, K-12 students and STEAM members), we present a manufacturing process that builds
on our prior work in foams that employs sculpting rather than casting and uses benign table salt (NaCl) as the porogen, allowing fabrication outside of specialized laboratories. We describe the viscoelastic properties of this silicone-salt compound and its ability to be sculpted rather than molded, eliminating the need for expensive tools such as 3D printed molds or milling machines or computer aided design (CAD) software. We report the resulting mechanical properties of the foams (i.e., elastic modulus, ultimate strain, and fluid flux through the pores) along with the capabilities of the assembled actuators (i.e. block force measurements and weight-lifting demonstrations).

B.3 Materials and Methods

B.3.1 Foam manufacturing

Using the lost salt method described previously [22], we fabricated the foams by mixing together a widely available silicone pre-polymer (Ecoflex 00-10; Smooth-On, Inc.) and table salt (Morton Salt, Inc.). We chose Ecoflex 00-10 because of its low elastic modulus, \( E = 50 \, \text{kPa} \), and its very high ultimate strain, \( \varepsilon_{\text{ult}} \sim 8 \) (manufacturer's data sheet). These properties allow large and reversible deformation. We chose common table salt because of its low health risk, broad availability, and ability to dissolve aqueously. After mixing Ecoflex 00-10 Part A and Part B (1:1 ratio, by weight), we added the predetermined amount of salt to the uncured elastomer and manually mixed the silicone and salt. We based the amount of salt added on the desired porosity. After mixing Ecoflex 00-10 and salt together, we directly sculpted
(or sometimes molded) the mixture into a variety of 3D shapes (Figure B.1a-g). To facilitate sculpting, we increased the viscosity of the mixture by either allowing the silicone to partially cure for 20 minutes at 25 °C or by adding THI-VEX thickener (Smooth-On, Inc.) (4% wt. of Part A), which is able to immediately increase elastomers viscosity and reduce the curing time. We noted that when the mixture cured fully at room temperature for four hours as prescribed in the Ecoflex 00-10 technical datasheet, a two-layer structure in the foam formed, in which the top layer was composed only of Ecoflex, and the bottom of both Ecoflex and salt (Figure B.1h). To achieve a homogeneous pore distribution, we sped up the curing process by placing the molded/sculpted mixture into an oven at 80 °C. We note, however, that the two-layer foam could be useful in creating a built-in actuation direction or for sealing.
Figure B.1 Fabrication of foam forms. (a-d) Sculpting process. 3D Sculpted examples: (e) hand, (f) Y-shaped gripper and (g) geometric shapes. (h) Ecoflex 00-10 and table salt foam double layer when curing at room temperature.
Once the mixture cured fully, we dissolved the salt in warm water (~50°C) overnight. To speed up dissolution, we used an ultrasonic bath (VWR Model 75T) and periodically stretched (just few millimeters, $\varepsilon<10\%$) the foams under water to increase salt-to-water contact. After drying the foam, we used compressed air (at flow rates accessible by computer keyboard air dusters) to remove residual salt crystals, which ensured an interconnected fluid pathway by breaking thin walls between closed cells. Figure B.2 shows the foam porous structure. We could further shape the foams with scissors before sealing them with another elastomer.

![Porous structure of elastomer foam.](image)

**Figure B.2** Porous structure of elastomer foam.

**B.3.2 Tensile Tests and Air Flow Rate Measurements**

To characterize the mechanical properties of the cured foams, without adding THI-VEX thickener, we molded samples with three different porosities: $\phi_{pore} =$
\[
\frac{V_{NaCl}}{V_{silicone + V_{NaCl}}} = 0.5, 0.6, 0.7 \text{ and performed uniaxial, monotonic tensile tests (3 samples per porosity) according to the ASTM D412 standard (Figure B.3a). We performed the tests on a Zwick & Roell Z010 at an elongation rate of 50 mm min}^{-1}\text{ using a 10 kN load cell. As a reference, we also performed uniaxial tensile tests on pure elastomer (} \phi_{\text{pore}} = 0.0)\].
Figure B.3 Mechanical testing and airflow measurements. (a) Ecoflex 00-10 and manufactured foams uniaxial tensile tests; (b) flow rate measurements.

Towards predicting actuator inflation rates, we performed flow rate measurements through foams with three different porosities. We molded 150 mm long
cylindrical samples of diameter 25 mm and cut them into 15 mm lengths. We then glued the samples to the inner surface of rigid acrylic tubes with length and inner diameter of 25 mm using Sil-Poxy silicone rubber adhesive (Smooth-On, Inc.), leaving 5 mm of space above and below the sample. Next, we mounted each tube between two 3D printed caps and rubber O-rings to make the tube air-tight (Figure B.4a-b). The mount connected the encased sample to a pressurized air source with controllable flow rate. We measured the output flow rate as well as the pressures upstream and downstream of the foam to calculate the pressure drop (using H271A-005, Hedland Inc. and FLDA3422G, Omega Inc., respectively). Our tests covered the range of pressures: 0<ΔP<100 kPa (Figure B.3b).
After conducting mechanical tests and flow rate measurements, we chose to fabricate foams with $\phi_{\text{pore}} = 0.7$ porosity because of its higher flow rate and greater ease of sculpting (due to its larger salt-to-elastomer ratio).

**B.3.3 Rheological tests**

We quantified the flow properties of our pre-foam mixture using oscillatory rheology on a DHR-3 rheometer (TA Instruments) with a parallel plate tool. Using a
salt volume fraction of $\phi_{pore} = 0.7$, we measured the elastic, $G'$, and loss, $G''$, moduli as well as complex viscosity, $\eta^*$. We applied increasing stress, $0.0 \ kPa < \sigma < 3.0 \ kPa$, at a constant frequency of $\omega = 1.0 \ Hz$ to simulate the estimated stress range and rate experienced during sculpting (Figure B.5a). Finally, we measured the evolution of $G'$, $G''$, and $\eta^*$ as the silicone cured (Figure B.5b). The constancy of $G'$ over the two-hour test, while the elastomer was curing, indicates that the salt dominates the elasticity, and the viscosity evolution demonstrates the large window of time for sculpting (about 60 minutes). All rheological tests were performed at room temperature.
Figure B.5 Rheological behavior of liquid foam precursor. (a) Stress sweep and (b) Time sweep
B.3.4 Soft Actuator Fabrication

To fabricate these foam actuators, we added both an external seal to trap the inflating fluid and an inextensible strain-limiting layer to direct motion [1,15,23]. We fabricated multiple actuators sealed with Ecoflex 00-10, Ecoflex 00-30, Ecoflex 00-50, Mold Max 10, OOMOO 30 (All from Smooth-On), or Elastosil M4601 (Wacker Chemie AG). Table B.1 displays some key mechanical properties of these elastomers.

Table B.1 Mechanical properties according to manufacturers’ datasheets.

<table>
<thead>
<tr>
<th></th>
<th>Ecoflex 00-10</th>
<th>Ecoflex 00-30</th>
<th>Mold Max 10</th>
<th>OOMOO 30</th>
<th>Elastosil M4601</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shore hardness</td>
<td>00-10</td>
<td>00-30</td>
<td>10A</td>
<td>30A</td>
<td>28A</td>
</tr>
<tr>
<td>Modulus at $\varepsilon = 1$ (MPa)</td>
<td>0.06</td>
<td>0.07</td>
<td>0.24</td>
<td>0.75</td>
<td>1</td>
</tr>
<tr>
<td>Elongation at break (%)</td>
<td>800</td>
<td>900</td>
<td>529</td>
<td>250</td>
<td>700</td>
</tr>
<tr>
<td>Tensile strength (MPa)</td>
<td>0.83</td>
<td>1.38</td>
<td>3.26</td>
<td>1.65</td>
<td>6.5</td>
</tr>
</tbody>
</table>

We used paper (40 lb Kraft Paper Roll; Staples) for the inextensible layer (E~GPa), as it is easily accessible, low cost, and can be cut to any shape using a laser cutter or scissors. As paper is porous, this layer merges completely with the elastomer while curing. When sealing with lower hardness elastomers (Ecoflex), we reinforced the paper with a thin (2 mm) layer of OOMOO 30 (Figure B.6).
Figure B.6 Actuators composed of different sealing materials. Casted actuators (a), Ecoflex 00-10 actuator section (b), and painted actuators (c).
We used two different methods to form the external seal: painting and molding. A thin sealing layer (about 1-2 mm) can be painted, and this is faster (especially for complex shapes) because it eliminates the need to design a mold; however, it can result in an uneven seal. Molding yields reproducible seals and should be used where reproducibility is required. In moldless fabrication, however, we apply the sealing layer with paint, which results in more variation as can be seen in blocking force tests (Figure B.7). The one-step seal cure (by molding or painting) avoids bonding issues between asynchronously cured seal layers, as opposed to the multi-step cure often used in traditional approaches [24], as shown in Figure B.8 and Figure B.9. Though the foam and seal cure at different times, the bond is sufficiently strong for inflation due to the large bonding area available on the foam's porous surface.
Figure B.7 Bending blocked force measurements. Ecoflex 00-10 \( \phi_{\text{pore}} = 0.7 \) foam with MoldMax 10, M4601 and OOMOO 30 seal for casted (a) and painted (b) actuators.
Figure B.8 Casting process to seal foam actuators. (a) first layer sealing elastomer casting; (b) strain limiting layer placement; (c) foam bonding; (d) second layer sealing elastomer casting; (e) developed block actuator after curing.
Figure B.9 Painting process for foam actuators.
**B.3.5 Bending blocked Force Measurements**

We performed blocked force measurements to characterize the actuators’s applied force as a function of inflation pressure. We tested $\phi_{pore} = 0.7$ foams with three sealing elastomers, (Mold Max 10, M4601, and OOMOO 30) each of which has a higher hardness than Ecoflex 00-10. We used an acrylic sample mount to constrain the curvature of the bending actuator (30x105x8 mm size) to a constant gap and a balance to record applied force (Figure B.10). We increased inflation pressure from 0 kPa until the actuator failed (e.g., sealing rupture or bursting). Figure B.7a shows actuation force measurement results for molded actuators, while Figure B.7b shows results for painted actuators.
Figure B.10 Bending blocked force measurement experimental set-up. (a) lateral view and (b) top view.

B.3.6 Load-Bearing Actuation

Using the actuator sealed with Ecoflex 00-10, we measured the curvature as a function of the inflation pressure. By using a very soft foam and seal, we obtained high bending (curvature radius $r = 25$ mm) with low pressures (27.5 kPa) (Figure B.11a). Using a different sealant elastomer (Elastosil M4601) we were able to lift different weights by inflating a similar actuator (Figure B.11b) in accordance with the actuation force measurements for the selected elastomer (Figure B.11c).
Figure B.11 Bending actuation demonstrations. (a) Bending example of soft block actuator. Foam is sealed with Ecoflex 00-10 and the strain-limiting layer is paper and OOMOO 30; (b) M4601 Actuator test with 100g/450g/500g weight in accordance with actuation force measurement for M4601 sealing (c), as reported in Figure B.7.
**B.4 Results and Discussion**

As expected, the foams' elastic moduli decreased as porosity increased (Figure B.3a). Table 2 displays, for each porosity, the average tensile modulus $E$, ultimate tensile stress $\sigma_{ult}$ and ultimate strain $\varepsilon_{ult}$. All the foam samples showed an ultimate breaking strain $\varepsilon_{ult} > 3.5$. Though this value is less than half the nonporous Ecoflex ultimate breaking strain ($\varepsilon_{ult} \approx 8$), it is adequate for fluidic actuation.

The flow rate measurements (Figure B.3b) demonstrate that $\phi_{pore} = 0.7$ foam has the highest flow rate, which we attribute to its higher porosity. The figure shows a nearly linear relationship between airflow rate and differential pressure. The sample-to-sample variability is likely due to the stochastic nature of the pore network and foam fabrication variables (salt crystal size, degree of foam stretching, and amount of air pumping to remove residual table salt). All porosities tested had a suitable flow rate for fluidic actuation.

Rheological tests, confirmed that we can manually sculpt the foams (Figure B.5a) within the mixture sculpting time (Figure B.5b), according to the material rheological properties.

Through our blocked force tests (Figure B.7), we observed that actuation force is related to the mechanical properties of the sealing elastomer, as expected. Specifically, higher hardness seals required greater inflation pressures to initiate actuation and to attain a given endpoint force. For molded actuators, we recorded actuation initiation pressures of 45 kPa, 55 kPa, and 65 kPa and maximum actuation forces of 4.8 N, 4.9 N, and 4.3 N for Mold Max 10, M4601, and OOMOO 30,
respectively. Sculpted and painted actuators inflate at lower pressures, $\Delta P \sim 25$ kPa. We also measured maximum actuation forces of 3.1 N, 3.6 N, and 4.8 N for M4601, OOMOO 30, and Mold Max 10. In the case of M4601 and OOMOO 30, the lower maximum actuation force is likely due to the longer cure time of these elastomers, resulting in longer flow times and dripping. For painting uniform layers, Mold Max 10 is the best choice.

Figure B.11b shows an actuator with 0.7 porosity foam and M4601 seal raising different weights in accordance to the force-pressure relation in Figure B.7. The inflation pressure necessary to raise the different weights was consistent with the blocked force data in Figure B.11c. We applied internal pressures of $\Delta P = 100$ kPa, 130 kPa, and 145 kPa in an attempt to raise 100 g, 450 g, and 500 g, respectively. The actuator completely lifted the 100 and 450 g weights, but only succeeded in dragging the 500 g about 5 cm along the resting surface. We were not able to increase the inflation pressure in order to raise 500 g because of actuators' failure (rupture, bursting or irreversible deformation) at higher pressures (about 150 kPa).

We fabricated a functional Y-shaped gripper by first sculpting a Y-shaped Ecoflex 00-10 foam with $\phi = 0.7$ porosity. We sealed the foam with Ecoflex 00-30 (cured for 30 minutes at 80°C). Each of the three fingers had dimensions of 200 x 10 x 7 mm, which sufficed to wrap around an apple. The strain-limiting layer was comprised of paper (40 lb Kraft Paper Roll; Staples) and OOMOO 30 reinforcement. After curing, we demonstrated grasping (via pneumatic inflation) and movement of the apple (~200 g; Figure B.12).
Figure B.12 A sculpted, three-pronged gripper lifting an apple.
The gripper successfully grasped the object at low inflation pressures ($\Delta P \approx 45$ kPa). We deliberately used only the foam and a strain-limiting layer to demonstrate the foam's capability as an internal inflation chamber without other mechanical constraints (e.g. external fibers or molded air chambers, as in PneuNet devices). With this approach, we sculpted a functional, monolithically sealed gripper in about 60 minutes that did not require 3D printed molds and used only low cost, readily available materials.

**B.5 Conclusions**

In this work, we present a new method for moldless manufacturing of soft FEAs and, in general, soft robots. We report mechanical properties and actuation measurements for the proposed foams and actuators. Our approach presents some unique advantages over traditional soft actuator fabrication.

A primary advantage of this manufacturing technique is that our method enables anyone to fabricate soft robots. This process is safe, low cost, and uses easily available materials making it applicable in K-12 classrooms, small laboratories, and art studios. Moreover, this technique enables fully functional 3D machines without a mold. The soft, stretchable, and sculptable porous materials eliminate the need for both molds to form an actuator’s external shape and sacrificial molds to form an internal air chamber. This aspect is notable because it significantly reduces the time and number of fabrication steps necessary to manufacture complex 3D actuators. Additionally, since our process forms the sealing layer in a single step, our seal does not have the interface (created between asynchronously cured layers) that can be prone
to delamination and leaking.

**B.6 Acknowledgments**

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**B.7 REFERENCES**


