Computational Inverse Design of Anatomy-Specific Soft Robot Actuators with Physically-Realizable Material Conditions

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INTRODUCTION

Soft robotics has empowered robots to maneuver, traverse, and complete tasks where traditional rigid robots fall short [1]. Traditionally, soft robots rely on forward design. This limits robots' spectrum of potential actuator shapes [2], but what if a specific patient's anatomy requires a unique actuator shape not realizable by such methods? Inverse design, on the other hand, centers around the desired task and produces a robot geometry for that task. Currently, there exists no technique for the inverse design of a soft robot that can actuate from arbitrary anatomy-specified initial and final shapes. Worse, computational methods may easily result in designs that are physically impossible to realize with real-world materials, e.g., such designs may easily violate volumetric compressibility properties of realistic elastomers.

We introduce a method of computational inverse design of soft robot actuators that simultanesouly achieves required arbitrary actuator shapes dictated by unique patient anatomy and ensures realistic volumetric compressibility throughout the actuator body. We propose a 3 degree of freedom (DOF) parameterization. We test the hypothesis that partitioning 2 DOF to parameterize the required actuator surface shape and 1 DOF for volumetric compressibility provides compuational designs that achieve desired shapes and meet compressibility requirements for typical hyperelastic elastomers.

MATERIALS AND METHODS

The process for creating an anatomy-specific actuator consists of two steps. The first step involves segmenting medical scans to find anatomical centerlines from which the soft robotic actuator shapes are determined. The second step focuses on designing a soft actuator given the parametric representation of the anatomy (e.g. centerline, radius) and selecting the desired actuator shape.

The anatomy-specific soft actuator design method is applied to an abdominal vasculature scan as shown in Fig. 1a. For our demonstration case, we design a spiral actuator in the renal artery as outlined in Fig 1b. The renal artery was selected due to the need for vascular surgeons to anchor guidewires in the renal artery without obstructing blood flow. Moreover, the artery's small size and need for such shapes could not be achieved with the methods in [2].

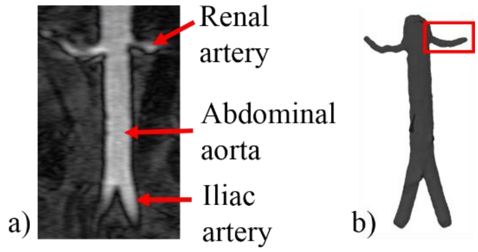

Fig. 1 a) MRI scan data of patient anatomy. b) 3D model, where the actuator is designed for the renal artery (boxed region).

Anatomy Segmentation & Centerline Creation

Our process for creating an anatomy-specific actuator started with processing the abdominal medical image using the Vascular Modeling Toolkit Lab (VMTKLab) [3]. Thresholding and segmentation were performed on the image and a 3D model was created. The 3D model was uploaded into the centerline function in VMTKLab and the centerline data was exported.

The centerline points were loaded into MATLAB and fit with a cubic spline followed by a polynomial fitting as a function of the parametric variable for length, w. This resulted in a 3D parametric centerline with coefficients β and order N=5 as written in eq. 1.

$$\boldsymbol{c_a}(w) = \sum_{i=0}^{N} \beta_i w^i \quad (1)$$

$$r_a = \sum_{i=0}^{N} \gamma_i w^i \quad (2)$$

The cubic spline and polynomial fit were also applied to the anatomical radius data, resulting in eq. 2, with coefficients γ . Eqs. 1-2 yield a continuously differentiable basis for an arbitrary robot actuator shape.

Soft Actuator Design

A helical spiraling actuator was selected for the renal artery as it can anchor in place without obstructing blood flow. The spiral actuator design was created by wrapping a helix about the anatomical centerline and extruding a cylinder about the obtained helical centerline. We start by assigning coordinate frames along the anatomical centerline which is parameterized by w. These frames are defined using Frenet-Serret method by computing the tangent, $\boldsymbol{T}$, normal, $\boldsymbol{N}$, and binormal, $\boldsymbol{B}$, vectors along the centerline $\boldsymbol{c_a}(w)$.

$$\boldsymbol{T} = \frac{\frac{\partial \boldsymbol{c_a}}{\partial w}}{\left|\frac{\partial \boldsymbol{c_a}}{\partial w}\right|},\ \boldsymbol{N} = \frac{\frac{\partial \boldsymbol{T}}{\partial w}}{\left|\frac{\partial \boldsymbol{T}}{\partial w}\right|},\ \boldsymbol{B} = \boldsymbol{T} \times \boldsymbol{N} \quad (3)$$

With eq. 3 defined we move to fit a helix around $\boldsymbol{c_a}(w)$ by applying modifications to the standard helix parameterization resulting in $\boldsymbol{c_f}(w)$ as shown in eq. 4.

$$c_f(w) = c_a(w) + R(w)\left[cos\left(\frac{2\pi kw}{L}\right)\boldsymbol{N} + sin\left(\frac{2\pi kw}{L}\right)\boldsymbol{B}\right] + \rho\boldsymbol{T} \quad (4)$$

$$R(w) = \frac{1}{2}f_{env}(w)r_a(w) \quad (5)$$

$$f_{env}(w) = \frac{1}{\sqrt{2}}tan^{-1}(15w) \quad (6)$$

The radius of the helix is described by $R(w)$, eq. 5, where $f_{env}(w)$, eq. 6, is an envelope function that ensures the ends of the actuator centerline smoothly blend into the anatomical centerline. In eq. 4 the k represents the number of helical turns, ρ is the pitch of the spiral, and L is the length of the centerline. In this specific case, $L = 20.63\ mm$, $k = 3.18$, $\rho = 0.8\ mm$, inner radius $r_i = 0.99\ mm$, and outer radius $r_o = 1.00\ mm$. These can be modified to fit specific design criterions, such as occlusion ratio, actuator contact area with anatomy, or a combination of other objectives.

The next step is to create the actuator body about $\boldsymbol{c_f}(w)$. Since the body is a volume, we must use three parametric variables and choose to use a cylindrical coordinate basis resulting in $X = [u, v, w]$, where $u = [r_i\ r_f]$ represents the radius of the cylindrical actuator, $v = [0\ 2\pi]$ represents the wrap angle, and $w = [0\ L]$ is the length along the centerline. A Frenet-Serret frame is then defined using $\boldsymbol{c_f}(w)$, resulting in $(\boldsymbol{T_f}, \boldsymbol{N_f}, \boldsymbol{B_f})$. The actuator body is then created as a function of parametric variables X as written in eq. 7. The function $f(u)$ is introduced to modify the local thickness of the actuator body at each point as illustrated in Fig. 2.

$$\boldsymbol{\varphi}(X) = \boldsymbol{c_f}(w) + \big(u + f(u)\big)\,\bar{n} \quad (7)$$

$$\bar{n} = \left[\boldsymbol{N_f}\,cos(v) + \boldsymbol{B_f}\,sin(v)\right] \quad (8)$$

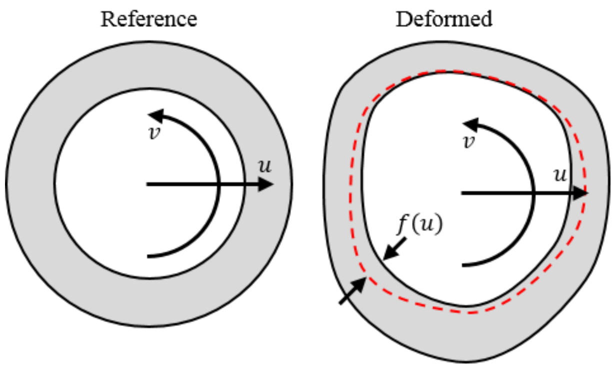

Fig. 2 The cross section of the soft robot is displayed. The initial design focuses on creating a design over [*v*, *w*] and does not directly account for *u*. The function *f(u)* was created to modifiy the thickness to enforce incompressibility.

To enforce material incompressibility of the design, we use the deformation gradient $\boldsymbol{F}$ as calculated by eq. 9 with the reference basis being in cylindrical coordinates, X, and the deformed basis being in Cartesian coordinates, $\boldsymbol{\varphi}(X)$ [4]. If the determinant of $\boldsymbol{F}$ is sufficiently close to one, we consider the design incompressible.

When solving eq. 10 for incompressibility, a first order differential equation is produced in terms of $f(u)$. If we assume that $f(u)$ takes the form in eq. 11, we can solve for α_0 and α_1 at every spatial location X.

$$\boldsymbol{F} = \begin{bmatrix}\frac{\partial\boldsymbol{\varphi}(X)}{\partial u} & \frac{1}{u}\frac{\partial\boldsymbol{\varphi}(X)}{\partial v} & \frac{\partial\boldsymbol{\varphi}(X)}{\partial w}\end{bmatrix} \quad (9)$$

$$\varepsilon = |det(\boldsymbol{F}) - 1| < 10^{-6} \quad (10)$$

$$f(u) = \alpha_0 + \alpha_1 u \quad (11)$$

RESULTS

The renal artery centerline and mesh are shown in Fig. 3. Additionally, Fig. 3 contains the soft robot's centerline and final computational design. The incompressibility constraint was numerically satisfied in eq. 11 with all ε less than 10^6 over the whole range of $X = [u, v, w]$, 2000 spatial locations in all.

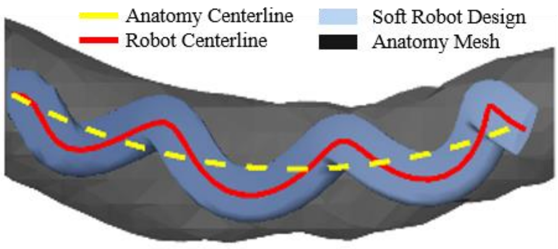

Fig. 3 Spiral actuator design inside the renal artery.

DISCUSSION

The method produced a parametric soft robot actuator design for a specific renal artery while ensuring local volumetric compressibility by partitioning the DOF as hypothesized. This method is highly generalizable for nearly arbitrary shape requirements. It expands the possible design space of soft robots to new patients with unique anatomy. It also alleviates constraints such as those caused by wrap angles in fiber-reinforced actuators, and creates a design based upon the task (specific patient-needs) and not the manufacturing method with which a soft robot designer begins. Future work will incorporate additional and anisotropic material properties into the computational design.

ACKNOWLEDGEMENTS

This work was supported by the MnDRIVE Robotics, Sensors, and Advanced Manufacturing initiative for financial support. Additionally, this material is based upon work supported by the National Science Foundation Graduate Research Fellowship Program under Grant 00039202. We thank VMTKLab for providing the MRI scan.

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