1 Background

In recent years, great strides have been made in the development of computational models for studying fluid-tissue interactions in portions of the mammalian heart. These models tend to be quite sophisticated from a computational fluid dynamics (CFD) perspective, enabling the calculation of fluid velocity profiles, fluid pressures, and wall displacements [1–5]. They incorporate various computational approaches, constitutive relationships, and structural geometries. Taylor et al. modeled a human left ventricle as a deformable sphere with time-dependent fluid boundaries, but the mechanics of the left ventricular wall were not considered [1]. Similarly, Schoephoerster et al. created a two-dimensional model of the left ventricle without addressing ventricular wall mechanics [2]. However, it utilized medical image sequences of an in vivo human heart to produce a more detailed geometry and prescribed wall motions. While Taylor and Schoephoerster treated the ventricular walls merely as fluid boundaries, Peskin modeled the cardiac walls with springs and “contractile elements” capable of interaction with the fluid [3–5]. Peskin first modeled the left atrium and left ventricle in two dimensions [3] and later modeled the entire heart in three dimensions [4,5]. Fluid/structure interactions were analyzed using the immersed boundary method. Combining Peskin’s methods and Patankar’s semi-implicit method for pressure linked equations (SIMPLE), Lemmon constructed a more computationally efficient algorithm to analyze the hemodynamics of the left ventricle and left atrium [6]. Left heart geometries were modeled using combinations of ellipsoids and spheres.

In addition to the CFD models, finite element (FE) models focusing on left ventricular wall mechanics are replete in the recent literature [7–10] and include highly detailed geometries generated directly from anatomical measurements. Perhaps, the most popular constitutive law for ventricular tissue consists of a strain energy equation that sums the contributions of strains in the fiber, cross-fiber, and radial directions [11]. The development of these constitutive laws and FE models has been aided by detailed mapping of ventricular geometry, strains, and muscle fiber orientations [12–17]. While these models aptly simulate the behavior of relatively thick-walled cardiac structures, a three-dimensional stress model may not be necessary for modeling thin-walled cardiac structures, such as the left atrium, which arguably exhibit plane stress characteristics.

A host of studies have been conducted in recent years concerning the mathematical modeling of planar soft tissues [18–24], all

Finite Element Modeling of the Left Atrium to Facilitate the Design of an Endoscopic Atrial Retractor

With the worldwide prevalence of cardiovascular diseases, much attention has been focused on simulating the characteristics of the human heart to better understand and treat cardiac disorders. The purpose of this study is to build a finite element model of the left atrium (LA) that incorporates detailed anatomical features and realistic material characteristics to investigate the interaction of heart tissue and surgical instruments. This model is used to facilitate the design of an endoscopically deployable atrial retractor for use in minimally invasive, robotically assisted mitral valve repair. Magnetic resonance imaging (MRI) scans of a pressurized explanted porcine heart were taken to provide a 3D solid model of the heart geometry, while uniaxial tensile tests of porcine left atrial tissue were conducted to obtain realistic material properties for noncontractile cardiac tissue. A finite element model of the LA was constructed using ANSYS™ Release 9.0 software and the MRI data. The Mooney–Rivlin hyperelastic material model was chosen to characterize the passive left atrial tissue; material constants were derived from tensile test data. Finite element analysis (FEA) models of a CardioVations Port Access™ retractor and a prototype endoscopic retractor were constructed to simulate interaction between each instrument and the LA. These contact simulations were used to compare the quality of retraction between the two instruments and to optimize the design of the prototype retractor. Model accuracy was verified by comparing simulated cardiac wall deflections to those measured by MRI. FEA simulations revealed that peak forces of approximately 2.85 N and 2.46 N were required to retract the LA using the Port Access™ and prototype retractors, respectively. These forces varied nonlinearly with retractor blade displacement. Dilation of the atrial walls and rigid body motion of the chamber were approximately the same for both retractors. Finite element analysis is shown to be an effective tool for analyzing instrument/tissue interactions and for designing surgical instruments. The benefits of this approach to medical device design are significant when compared to the alternatives: constructing prototypes and evaluating them via animal or clinical trials. [DOI: 10.1115/1.2801650]

Keywords: mitral valve repair, finite element analysis, left atrium, retraction

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or most of which exhibit anisotropic material properties. Sacks and Sun have presented an overview of biaxial testing of biological materials and a history of constitutive models formulated to describe their behavior [25]. According to Sacks and Sun, the most popular constitutive model for planar soft tissues is the Fung orthotropic pseudoe lastic model. Sacks and Sun implemented the Fung model into FE code to simulate the biaxial testing of soft tissues (specifically glutaraldehyde treated bovine pericardium (GLBP)) [22]. Commercially available ABAQUS was the software platform; the material model was user defined. Although anisotropic models (e.g., the Fung model) more aptly characterize the behavior of soft tissues in multiaxial loading, Lally et al. have suggested that the Mooney–Rivlin isotropic hyperelastic material model may be sufficient for modeling the general behavior of coronary artery tissue over a wide range of subjects, due to wide variations in stress-strain properties [26]. Similar reasoning suggests that the Mooney–Rivlin isotropic material model may provide an acceptable approximation for the behavior of left atrial tissue in the general case.

Another area of active research involves the construction of real-time surgery simulations as training tools for surgeons. These simulations employ a variety of mechanical model types, including mass-spring, FE, finite difference, mass/tensor, and parametric models with dynamic splines [27–31]. Although FE models are the gold standard for structural analysis, discrete mass-spring systems are the most widely used models in surgery simulations. The mass-spring system’s widespread usage is due to its computational efficiency, which is a necessity for real-time simulations. Use of the FE method for real-time simulations requires reducing the complexity and thus the detail of the FE matrices. For example, the condensation technique, studied by Bro-Nielsen and Cotin [27], reduces the number of calculations by only computing parameters for nodes at the surfaces of biological structures. This practice is considered permissible for surgery simulations since parameters at the surface nodes alone are adequate to produce the wanted force feedback and visual graphics. A German-built surgery simulation package known as KISMET uses the condensation technique for its real-time computations [28].

While CFD models facilitate better understanding and diagnoses of cardiac disorders, they frequently simplify the geometry and material properties of cardiac structures and lack the capability to analyze contact between heart tissue and surgical instruments. FE models of the left ventricle characterize 3D stress behavior, which may not be necessary for modeling left atrial tissue. FE models with planar soft tissues have been implemented in limited studies but have not, to the author’s knowledge, included complex cardiac geometries nor evaluated the interaction of surgical tools with the tissue. Real-time surgery simulations analyze tissue/instrument interactions but sacrifice quantitative mechanical accuracy for computational efficiency. A need exists for computational tools that simulate structural interactions between cardiac anatomy and surgical instruments to accurately quantify stresses and strains in both. Such a tool would facilitate the design of more effective surgical instruments. This need is increased with the advent of minimally invasive procedures, which require that tools be deployed through small incisions but retain or enhance the functionality of conventional tools.

The purpose of this study is to build a FE model of the left atrium that incorporates detailed anatomical features and realistic material characteristics to characterize the interaction of heart tissue and surgical instruments. This model is used to facilitate the design of an endoscopically deployable atrial retractor for use in minimally invasive, robotically assisted (MIRA) mitral valve repair. The left atrial geometry is imported directly from magnetic resonance imaging (MRI) data of an explanted porcine heart, and material properties are derived from experimental testing of noncontractile cardiac tissues. Material stress-strain properties will be an approximation for porcine myocardia rather than a detailed representation of a specific case. Based on this consideration and the availability of the material model in commercial FE software, the Mooney–Rivlin isotropic hyperelastic material model will be used as the constitutive model for the cardiac tissue. Model accuracy is verified by comparing simulated cardiac wall deflections to deflections measured by MRI.

### 2 Method of Approach

Two prerequisites for accurate finite element analysis (FEA) of the mammalian heart include (1) a detailed solid model of the heart geometry and (2) realistic material properties of cardiac tissue.

#### 2.1 Left Atrial Geometry

The three-dimensional solid model required for FEA was obtained from MRI scans of a presurized explanted porcine heart. MRI data (DICOM format) were then converted to a compatible solid model format (IGES).

**2.1.1 Magnetic Resonance Imaging Scans.** An explanted porcine heart (approximately 550 g, 7 cm anteroposterior × 9 cm lateral × 23 cm superior-inferior) was scanned using MRI to generate a 3D solid model of the left atrium (LA) and to measure atrial wall deflections at various internal pressures. The porcine heart was collected from a local abattoir, prepared, and tested within 60 h of mortality. A fluid supply tube (12.7 mm inside diameter (i.d.) vinyl) was inserted into the largest pulmonary vein and affixed using cyanoacrylate adhesive. To prevent leakage, the aorta was sealed with a 15.88 mm vinyl plug, and smaller pulmonary veins were sealed using cyanoacrylate adhesive. The atrium and left ventricle were pressurized using a 60 cm² syringe and viscous solution containing Suave™ body wash, water, and gadolinium contrast agent. Internal fluid pressure was monitored using a 0–300 mm Hg pressure gauge taken from a certified sphygmomanometer, which was kept dry throughout the experiment. Fluid gage pressure was maintained at one of four constant levels for the duration of each MRI scan: 0 mm Hg, 30 mm Hg, 50 mm Hg, and 60 mm Hg. The expansion of the LA with increasing pressure can be seen in Fig. 1.

#### 2.2 Creating the Solid Model From Magnetic Resonance Imaging Data

MRI data obtained from the fluid-filled atrium at zero gage pressure were used to generate the reference geometry, while data obtained at higher gage pressures were used to generate deflected geometries. For each pressure, the MRI scan data were converted to an STL solid model format using MIMICS™ 8.11 software, designed specifically for 3D image processing and editing of data from medical scanners. Easily identified by the contrast agent, the LA, the left ventricle, and surrounding structures were isolated by selecting pixels within the range of grayscale values of the contrast agent. Separation of the atrium from the ventricle at the mitral annulus was achieved manually by comparing the MRI data to the measured porcine heart anatomy. To facilitate better meshing and FEA convergence, the smaller pulmonary vein extending from the LA was eliminated and the solid model was refined using GEOMAGIC™ STUDIO® 7 reverse engineering software. The model was refined with settings at maximum smoothness and minimum strength, which reduced the number of area patches by approximately 75%. The STL model was then converted to IGES format and imported into ANSYS™ Release 9.0.
2.3 Left Atrial Material Properties

2.3.1 Published Data. Since the myocardium remains in a noncontractile state throughout cardiac surgery, this study will only seek to model the passive mechanics of atrial tissue. Common methods of studying the passive stress-strain characteristics of cardiac tissue include uniaxial tensile testing (elongation of tissue in a single direction) and biaxial testing (elongation in two orthogonal directions simultaneously) [26–33]. Since their geometry lends to relatively easy testing, unstimulated papillary muscles are often studied using uniaxial tension tests. Fung [32] has shown that experimental stress-strain measurements can be curve fitted to exponential equations of the form

\[ P = (P^e + \beta)e^{\alpha(\lambda^*-1) - \beta} \]

where \( P \) is the Lagrangian stress (force per unit of the original cross-sectional area), \( \lambda \) is the material stretch, and \( \alpha \) and \( \beta \) are rate-dependent constants, and \( P^e \) and \( \lambda^* \) are a measured set of \( P \) and \( \lambda \) values. Stretch is defined to be the instantaneous tissue length divided by its original length (\( \lambda \))=engineering strain+1). Mirsky et al. [33] generated similar curves for human left ventricular tissue using data from cardiac catheterizations and assuming that the left ventricle was either perfectly spherical or ellipsoidal in shape. The mechanics of cardiac tissue were related by

\[ d\sigma = k\varepsilon + c \]

where \( \sigma \) is the stress, \( \varepsilon \) is the engineering strain, and \( k \) and \( c \) are constants derived from catheterization.

2.3.2 Experimental Tensile Testing. Uniaxial tensile tests of porcine left atrial tissue were conducted to obtain stress-strain characteristics for FEA model development. Samples were taken from ten pig hearts within 24 h of mortality and stored in a Tyrode solution (NaCl, KCl, CaCl₂, 6H₂O, MgCl₂·6H₂O, NaHCO₃, NaH₂PO₄, glucose, and distilled water). Anterior samples were taken from the atrial region adjacent to the aorta, posterior samples were taken from the region surrounding the pulmonary veins, while appendage samples were taken from the relatively flat region of the atrial appendage. The samples were cut to a uniform width (5.5 mm) using a tool with parallel scalpel blades. Each sample was cut to be aligned with or perpendicular to muscle fiber directions based on visual inspection of the exterior surface of the atrial tissue (Fig. 2). The thickness of each sample was measured using a dial caliper at three locations (the midpoint and each end).

These thickness measurements were averaged and multiplied by the sample width (5.5 mm) to calculate the unloaded (original) cross-sectional area of each specimen. An MTS Sintech™ Universal Testing Machine fitted with a 5.0 lb (22.2 N) load cell was used for testing. Sample ends were sandwiched between and adhered to thin stainless steel fixtures using cyanoacrylate. The fixtures were then mounted into the machine grips and cycled to approximately 20% strain for 4 cycles for preconditioning at a rate of 60 mm/min. After preconditioning, each sample was elongated at a rate of 38 mm/min until breakage occurred, with load and elongation data recorded at approximately 10.0 Hz.

2.3.3 Analysis of Experimental Data. Stresses and strains were calculated from the load and elongation data for comparison with published data. Engineering stress was calculated by dividing the load by the average original cross-sectional area of the specimens. Results indicated that the true (Cauchy) stress-strain relationships were not significantly dependent on fiber direction (Fig. 3) and did not vary significantly throughout the LA. Similar data published by Fung [32] and Mirsky et al. [33] were adjusted for comparison purposes, revealing similar stress-strain characteristics (Fig. 3(d)).

2.4 Model Construction. A FE model of the LA was built using ANSYS™ Release 9.0 software, the solid model of Fig. 4, and the material properties of Fig. 3.

2.4.1 Material Model, Meshing, and Boundary Conditions. The ANSYS™ Mooney–Rivlin isotropic hyperelastic material model was chosen to characterize the passive left atrial tissue, as it has been recommended for similar applications [26,34,35]. One notable deficiency of this model is its inability to model creep, which may be significant for extended loading periods [32]. The Mooney–Rivlin assumes the material to be isotropic and incompressible, and determines stress based on the derivative of a strain energy function \( W \), which can include two, three, five, or nine parameters. The governing equations for the five-parameter model are

\[ \sigma_{ij} = \frac{\partial W}{\partial e_{ij}}, \]

\[ W = c_{10}(\bar{I}_1 - 3) + c_{01}(\bar{I}_2 - 3) + c_{20}(\bar{I}_3 - 3)^2 + c_{11}(\bar{I}_3 - 3)^2(\bar{I}_3 - 3) - 3 \]

Parameters \( c_{10}, c_{01}, c_{20}, c_{11}, \) and \( c_{02} \) are material constants, \( \sigma_{ij} \) is the Piola–Kirchhoff stress, and \( \bar{I}_1 \) and \( \bar{I}_2 \) are invariants of strain calculated with respect to the right Cauchy–Green deformation tensor. ANSYS™ calculated these material constants (Table 1) using the uniaxial tensile test data (specifically Sample 4 of the anterior, Table 2) discussed previously. The test data were input in engineering stress-strain form as prescribed by ANSYS™ documentation. Although both the five-parameter and nine-parameter options provided reasonable curve fits to the test data, the five-parameter option (normalized error residual: 0.01833) was chosen since it provided better solution convergence than the nine-parameter option. It should be noted that the ANSYS™ Mooney–Rivlin model is commonly referred to as a polynomial hyperelastic material model. In fact, the ANSYS™ two-, five-, and nine-parameter Mooney–Rivlin models are equivalent to the ANSYS™ polynomial strain energy functions with \( N = 1, 2, \) and 3, respectively.

The LA was meshed using 3D, six degree-of-freedom (DOF) shell elements. The specific element used contains four nodes and is well suited for modeling large strains in nonlinear materials. Meshing resulted in 2664 total elements (2603 nodes) (Fig. 4). Mesh density (convergence) tests were used to confirm that the element sizes were sufficient for the model. Appropriate element thicknesses were specified for each region of the model based on MRI data and measured sample thicknesses (Fig. 5). Boundary conditions were specified to emulate condi-
Fig. 3 Experimental true (Cauchy) stress-strain curves for left atrial tissue: (a) anterolateral, (b) posterolateral, (c) appendage, and (d) average curves for the three regions. Note that graphs on the left represent data for specimens elongated perpendicular to the observed fiber direction; graphs on the right represent data for specimens elongated parallel to observed fiber direction.
tions encountered during MRI scanning of the porcine LA. During these scans, the LA was not only affected by adjacent cardiac structures but also by contact with the container in which it was placed and the supply tubing which entered the largest pulmonary vein. While these constraints did not necessarily resemble those of in vivo tissue, they were necessary to achieve meaningful comparison with MRI data. Additional constraints were added in regions of near-zero deflection as observed from MRI data to simulate the effects of surrounding tissues (ventricular muscles, arteries, veins, connective tissues, etc.), which were not included in the FEA model.

2.4.2 Model Validation Using Magnetic Resonance Imaging Data. Accuracy of the FEA model was then assessed by comparing the computed resultant deflections (the Euclidean norms of the $x$, $y$, and $z$ displacements, Fig. 6(a) and 6(d)) to the experimental deflections (Fig. 6(a) and 6(b)) resulting from internal pressurization. Computer-aided design (CAD) models of the LA at internal pressures of 0 mm Hg, 30 mm Hg, and 50 mm Hg were generated from MRI scan data using MIMICS™ 8.11. Deviations in displacement (between the zero pressure models and the pressurized models) were plotted using GEOMAGIC™ QUALIFY® 7 computer-aided inspection software.

To facilitate a quantitative comparison of computed and measured atrial deflections, a novel correlation algorithm was developed. Spherical coordinate systems were placed at the centroids of each model (Fig. 7(a)). Three-dimensional reference vectors $\hat{u}_i$ were projected from the coordinate system origins at 15 deg increments of $\theta$ and $\phi$, penetrating the model surface (Fig. 7). The radial deflections $\Delta r_i$ associated with each reference vector $\hat{u}_i$ resulting from internal pressurization were then computed. For example, a total of 266 normal deflections $\Delta r_i$ was computed for each model to quantify changes from 0 mm Hg to 30 mm Hg of internal pressurization ($\Delta r_i = r_{30} - r_0$). A correlation coefficient was then computed for all deflections obtained from the FEA and MRI data sets. Because the accuracy of each deflection calculation

### Table 1 Values of Mooney–Rivlin parameters used for the left atrial FEA model

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>$c_{10}$</td>
<td>$-5.84 \times 10^4$</td>
<td>Pa</td>
</tr>
<tr>
<td>$c_{01}$</td>
<td>$6.34 \times 10^4$</td>
<td>Pa</td>
</tr>
<tr>
<td>$c_{20}$</td>
<td>$1.60 \times 10^7$</td>
<td>Pa</td>
</tr>
<tr>
<td>$c_{11}$</td>
<td>$-3.53 \times 10^7$</td>
<td>Pa</td>
</tr>
<tr>
<td>$c_{02}$</td>
<td>$1.97 \times 10^7$</td>
<td>Pa</td>
</tr>
</tbody>
</table>

### Table 2 Measured thickness values for left atrial tissue samples

<table>
<thead>
<tr>
<th>Sample region</th>
<th>Sample orientation</th>
<th>Heart No.</th>
<th>Measured thickness (mm)</th>
<th>Average thickness (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Appendage</td>
<td>a</td>
<td>6</td>
<td>6.58 5.05 5.28</td>
<td>5.64</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>3.99 5.00 4.47</td>
<td>4.49</td>
</tr>
<tr>
<td></td>
<td>b</td>
<td>5</td>
<td>4.60 5.21 5.21</td>
<td>5.00</td>
</tr>
<tr>
<td></td>
<td></td>
<td>All</td>
<td></td>
<td>5.04</td>
</tr>
<tr>
<td>Anterior</td>
<td>a</td>
<td>4</td>
<td>1.50 2.39 2.24</td>
<td>2.04</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>5.21 3.81 4.11</td>
<td>4.38</td>
</tr>
<tr>
<td></td>
<td>b</td>
<td>1</td>
<td>4.62 5.56 6.83</td>
<td>5.67</td>
</tr>
<tr>
<td></td>
<td></td>
<td>5</td>
<td>1.60 1.14 1.32</td>
<td>1.35</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>6.17 4.83 3.45</td>
<td>4.82</td>
</tr>
<tr>
<td></td>
<td></td>
<td>All</td>
<td></td>
<td>3.65</td>
</tr>
<tr>
<td>Posterior</td>
<td>a</td>
<td>2</td>
<td>2.01 2.79 3.63</td>
<td>2.81</td>
</tr>
<tr>
<td></td>
<td></td>
<td>7</td>
<td>2.11 1.98 3.33</td>
<td>2.47</td>
</tr>
<tr>
<td></td>
<td>b</td>
<td>10</td>
<td>3.20 3.45 3.12</td>
<td>3.26</td>
</tr>
<tr>
<td></td>
<td></td>
<td>All</td>
<td></td>
<td>2.85</td>
</tr>
</tbody>
</table>

a Parallel to observed fiber direction.

b Perpendicular to observed fiber direction.
depended on the angle between the surface normal $\hat{u}_n$ and the reference vector $\hat{u}_r$. Values associated with $\alpha_i > 45\%$ deg were eliminated from the correlation computation. The final correlation coefficient, $r=0.72$, confirmed the accuracy of the FEA model in predicting the mechanical characteristics of the ex vivo LA.

Fig. 5 Element thicknesses of the left atrial model as defined per area; (a) posterolateral and (b) anterolateral views.

2.5 Simulating Atrial Retraction: Conventional Blade Retractor

2.5.1 Background. The true value of the validated FEA model lies in its utility as a design tool for surgical devices. Specifically, it was developed to facilitate the design of an atrial retractor for minimally invasive mitral valve repair (MVR). Atrial retraction is commonly performed using a two-piece blade retractor, such as the CardioVations Port Access™ retractor (Fig. 8), that must be assembled intracorporeally. The blade dimensions (up to $45 \times 70$ mm$^2$) necessitate larger access ports, larger atrial incisions, and complicate deployment. Research currently underway at North Carolina State University is focused on the development of an endoscopic atrial retractor that can be deployed through small incisions ($10.0$ mm trocars) yet enhance exposure and surgical access to the mitral valve.

2.5.2 Model Construction. FEA modeling of conventional...
atrial retraction served as a stepping stone for the development of an endoscopic retractor and provided an accurate basis of comparison. A solid model of a 35 × 60 mm² retractor (Fig. 9(a)) was created using ANSYS™ solid-modeling functions and scaled approximately 52% to fit the porcine LA, which in this model was significantly smaller than that of the average human adult. The retractor blade was meshed using 3D ten-noded tetrahedral elements with material properties of stainless steel. To simulate an atrial incision, a small rectangular piece of cardiac tissue was removed from the left atrial model in the proximity of the interatrial groove, the precise location and orientation consistent with standard surgical procedure. The atrium’s translational DOFs were constrained between the anterolateral and posterolateral halves of the LA tracing from the mitral valve opening to the appendage (Fig. 9(b)). The blade retractor was positioned inside

Fig. 8 CardioVations Port Access™ retractor (45 × 50 mm²): (a) outside the patient, (b) application in minimally invasive MVR, and (c) mitral valve exposure provided by the retractor (endoscopic view)
the incision into close proximity with the anterolateral wall of the LA (Fig. 9(b)). To detect collisions between the blade and LA, surface-to-surface contact elements were placed on the upward facing side of the retractor blade and on regions of the anterolateral wall of the LA in close proximity to the blade (Fig. 9(b)). The thin protrusion at the distal end of the retractor was meshed with node-to-surface contact elements in addition to the surface-to-surface contact elements. Initially, the contact elements were “open” (not in contact). The retractor blade was displaced along the path defined by the retractor rod axis by prescribing displacements. To account for the effects of gravity, a global acceleration of 9.81 m/s$^2$ was applied toward the zenith as defined when the human patient is in the supine position. Tables 3 and 4 summarize the FEA parameters used in these analyses. The static friction of 9.81 m/s$^2$ was applied toward the zenith as defined when the human patient is in the supine position. Tables 3 and 4 summarize the FEA parameters used in these analyses. The static friction coefficient $\mu_s$ and cohesion parameter $b$ were experimentally determined for each material pair according to the relation

$$F_f = \mu_s N + b A$$

where $F_f$ is the frictional force, $N$ is the normal force, and $A$ is the contact area. This relationship can be derived by multiplying the ANSYS™ contact sliding resistance equation [36] by the contact area. Tissue samples were placed on a flat metal plate and loaded with variable dead weights. The normal force $N$ was varied by gradually increasing the plate incline angle $\theta$. The normal and frictional forces at the onset of slippage were calculated by summing static forces:

$$N = (m_{\text{tissue}} + m_{\text{dead}}) g \cos \theta$$

$$F_f = (m_{\text{tissue}} + m_{\text{dead}}) g \sin \theta$$

Parameters $\mu_s$ and $b$ were determined using linear regression on a series of $F_f$ and $N$ measurements.

2.5.3 Overcoming Convergence Difficulties. Attaining FEA solution convergence required several minor adjustments in the position of the retractor blade and in contact parameters. Through a trial-and-error process, the retractor was positioned to produce appropriate amounts of retraction near the incision and near the mitral annulus. For mitral valve exposure, the walls of the LA near the incision require less displacement than tissue near the much stiffer annulus. Attempting too much displacement near the annulus led to convergence difficulties due to the high magnitude of force required to move the annulus. These convergence difficulties were overcome by iteratively adjusting the rotation of the retractor as well as its depth into the LA. Adjustments were also made to the contact parameters in accordance with ANSYS™ documentation guidelines. Two parameters, in particular, the “normal penalty stiffness” and “penetration tolerance,” required tuning for optimal contact analyses. ANSYS™ utilizes a normal penalty stiffness to determine the force magnitudes between two contacting surfaces relative to their separation distances, while the penetration tolerance specifies the maximum allowable penetration between two contacting surfaces.

2.6 Simulating Atrial Retraction: Prototype Endoscopic Retractor. The ANSYS™ model was next used to optimize the design of a prototype endoscopic atrial retractor for MVR.

2.6.1 Retractor Geometry. A CAD model of the endoscopic retractor was constructed using ANSYS™ solid-modeling functions (Fig. 10). This prototype contains a narrow central blade with four nitinol wires protruding from the distal end (Fig. 10(a)), whose lengths can be independently adjusted for optimal retraction. The wire tips are capped by stainless steel spheres (1.3 mm diameter) to prevent tissue damage (Fig. 10(b)). The retractor blade, nitinol wires, and stainless steel spheres were all drawn as solid volumetric structures, and the nitinol wires and stainless steel spheres were attached to each other along adjoining edges.

2.6.2 Material Model and Meshing. All components of the endoscopic retractor were meshed using 3D ten-noded tetrahedral elements. Although beam elements are more efficient in modeling

| Element and material properties for atrial blade retractor |
|-----------------|-----------------|-----------------|-----------------|-----------------|
| **Element**     | **Edge length (mm)** | **Material**    | **Type** | **Model** | **Modulus (GPa)** | **Density (kg/m³)** | **Poisson’s ratio $\nu$** |
| Distal protrusion | 1.5             | Stainless steel | Linear isotropic | 200 | 8000 | 0.30 |
| Left atrium     | 2.0             | Cardiac tissue  | Mooney–Rivlin hyperelastic | NA | 1053 | 0.50 |
| Main retractor blade | 1.5 | Stainless steel | Linear isotropic | 200 | 8000 | 0.30 |

| Contact parameters for atrial blade retractor |
|-----------------|-----------------|-----------------|-----------------|-----------------|
| **Target surface** | **Contact surface** | **Normal penalty stiffness (factor)** | **Penetration tolerance** | **Friction coefficient $\mu$** | **Cohesion $b$ (Pa)** | **Contact pair type** |
| LA wall           | Main blade       | 0.01            | $5 \times 10^{-4}$ (absolute) | 0.10 | 116 | STS |
| LA wall           | Distal protrusion | 0.01            | 0.1 (factor)       | 0.10 | 116 | STS |
| LA wall           | Distal protrusion | 0.01            | 0.1 (factor)       | 0.10 | 116 | NTS |

*TARGET 170 elements.
*CONTA 174 elements.
*STS=surface to surface, NTS=node to surface.
the wires structures, they proved to easily penetrate the left atrial tissue in contact simulations. Linear elastic material properties were specified for the central retractor blade (stainless steel) and arms (nitinol). Nitinol is a shape memory alloy with highly nonlinear material characteristics; at temperatures above the austenite finish temperature (in this case, approximately $15.0^\circ$C [37]), its stress-strain curve exhibits bilinear characteristics enabling the material to accommodate very large strains without plastic deformation. For the prototype application, however, the stress-strain relationship remains in the linear range, justifying the use of a simple elastic material model. Table 5 summarizes element and material properties for the endoscopic retractor.

2.6.3 Contact Pair Construction. Interaction between the retractor and surrounding tissue involved three distinct contact groups, each requiring its own friction and cohesion characteristics: (1) contact between the wires and LA, (2) contact between the stainless steel spheres and the LA, and (3) contact between the blade and the LA. Contact element pairs were placed at various locations throughout the model where retractor/tissue interaction was anticipated. Each contact pair contains a moving “contact” surface (on the instrument), which interacts with a “target” surface (on the atrium). Due to the highly nonlinear nature of the wire-to-LA contact, the solution only converged for extremely small values of normal penalty stiffness (between $4 \times 10^{-3}$ and $1 \times 10^{-3}$) and relatively high values of penetration tolerance ($1 \times 10^{-3}$). Table 6 summarizes contact element parameters for the endoscopic retractor.

2.6.4 Investigating Effects of Wire Deployment. FEA simulations were run with the wires at various levels of deployment ranging from no retraction to full retraction (defined as providing optimal exposure of the MV, approximately 12.5 mm extension) to investigate the effects of varying wire deployment.

3 Results

3.1 Conventional Blade Retractor. FEA simulations revealed that a peak force of approximately 2.9 N was required to retract the LA using a conventional blade retractor. To establish a baseline for comparison, forces were monitored during atrial retraction of a cadaver using a conventional blade retractor (Fig. 7(a)) equipped with a miniature load cell (Transducer Techniques MLP-10). The “lifting force” necessary to achieve adequate access and visualization of the MV was determined to remain below 4.5 N. The discrepancy between simulated and measured forces can be explained by the reduced size of the porcine LA and the lack of neighboring cardiac structures in the FEA model. Although the porcine heart was approximately average human size [38], the LA was approximately 52% of average human size.

The forces were seen to vary nonlinearly with retractor blade displacement due to the nonlinear characteristics of cardiac tissue (Fig. 3) and the associated increase in tool/tissue interaction with blade displacement. Near the end of its displacement, the retractor begins to contact the relatively thick-walled mitral annulus. Contact status plots show that retractor/tissue interaction is concentrated near the incision; the maximum contact pressure occurred in this region of contact.

3.2 Endoscopic Prototype Retractor. With the wires fully deployed for optimal exposure of the MV, the peak retraction forces (approximately 2.5 N) were slightly lower than those asso-

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Table 5  Element and material properties for endoscopic atrial retractor

<table>
<thead>
<tr>
<th>Element</th>
<th>Material Information</th>
<th>Edge length (mm)</th>
<th>Type</th>
<th>Model</th>
<th>Modulus (GPa)</th>
<th>Density (kg/m³)</th>
<th>Poisson’s ratio ν</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left atrium</td>
<td>SHELL 181 (3D, four nodes, 6DOFs)</td>
<td>2.0</td>
<td>Cardiac tissue</td>
<td>Mooney–Rivlin hyperelastic</td>
<td>NA</td>
<td>1053</td>
<td>0.50</td>
</tr>
<tr>
<td>Main retractor blade</td>
<td>SOLID 92 (3D, ten nodes, 3DOFs)</td>
<td>1.5</td>
<td>Stainless steel</td>
<td>Linear isotropic</td>
<td>200</td>
<td>8000</td>
<td>0.30</td>
</tr>
<tr>
<td>Spheres</td>
<td>SOLID 92</td>
<td>0.30</td>
<td>Stainless steel</td>
<td>Linear isotropic</td>
<td>200</td>
<td>8000</td>
<td>0.30</td>
</tr>
<tr>
<td>Wires</td>
<td>SOLID 92</td>
<td>2.0</td>
<td>Nitinol</td>
<td>Linear isotropic</td>
<td>41</td>
<td>6500</td>
<td>0.30</td>
</tr>
</tbody>
</table>
associated with the conventional retractor; approximately 1.63 N was exerted by the blade and the remaining 0.90 N by the arms. The two distal arms exerted more force than the proximal arms. Contact with the LA was concentrated near the incision and at the stainless steel spheres, while contact between the wires and tissue was minimal. This illustrates the rigidity of the cardiac tissue, which does not experience excessive sag between contact points. Contact pressures were relatively low on the central blade and arms, but were locally higher on the distal spheres.

As mentioned previously, the arms of the endoscopic retractor can be independently adjusted to optimize retraction. The effects of arm length adjustment were studied by simulating retraction with the arms at each of four positions: 0%, 33%, 66%, and 100% of their fully deployed lengths. Figure 11 shows the fully retracted LA for three of the four cases. From this endoscopic viewpoint, the mitral valve is clearly visible through the atrial incision without the LA wall Main blade 0.01 0.1 LA wall Distal protrusion 0.01 1 × 10⁻⁴ (constant)

Table 6 Contact parameters for endoscopic atrial retractor

<table>
<thead>
<tr>
<th>Target surface</th>
<th>Contact surface</th>
<th>Penalty stiffness factor</th>
<th>Penetration tolerance</th>
<th>Friction coefficient</th>
<th>Cohesion b (Pa)</th>
<th>Contact pair type</th>
</tr>
</thead>
<tbody>
<tr>
<td>LA wall</td>
<td>Stainless steel spheres</td>
<td>0.05</td>
<td>0.001 (absolute)</td>
<td>0.10</td>
<td>116</td>
<td>STS</td>
</tr>
<tr>
<td>LA wall</td>
<td>Nitinol wires</td>
<td>4 × 10⁻⁴– 1 × 10⁻³</td>
<td>0.001 (absolute)</td>
<td>0.35</td>
<td>87</td>
<td>STS</td>
</tr>
<tr>
<td>LA wall</td>
<td>Main blade</td>
<td>0.01</td>
<td>0.1 (factor)</td>
<td>0.10</td>
<td>116</td>
<td>STS</td>
</tr>
<tr>
<td>LA wall</td>
<td>Distal protrusion</td>
<td>0.01</td>
<td>1 × 10⁻⁴ (constant)</td>
<td>0.10</td>
<td>116</td>
<td>NTS</td>
</tr>
<tr>
<td>LA wall</td>
<td>Distal protrusion</td>
<td>0.01</td>
<td>1 × 10⁻⁴ (constant)</td>
<td>0.10</td>
<td>116</td>
<td>NTS</td>
</tr>
</tbody>
</table>

*STGIE 170 elements.
*CONTA 174 elements.
*STS= surface to surface, NTS= node to surface.

**4 Conclusions**

FEA of a porcine LA was validated as being an effective tool for analyzing instrument/tissue interactions and for designing novel surgical instruments. The benefits of this approach to medical device design are significant when compared to the alternatives: constructing prototypes and evaluating them via animal or clinical trials. A number of factors complicate the fabrication of prototypes, most notably (1) acquisition of suitable parts and materials, (2) availability of appropriate manufacturing processes, (3) training and/or recruitment of skilled personnel, and (4) the associated times and costs of each. Although rapid prototyping (RP) techniques have greatly reduced fabrication times, this approach can be costly and often yields products with inadequate material properties. Animal and clinical trials require appropriate IACUC or IRB approvals, extensive planning and preparation, and the involvement of highly skilled and available surgeons and/or other medical personnel. Although extensive effort is needed to develop and validate these FEA models, design modifications to the prototype retractor can be implemented and evaluated with minimal additional effort or cost.

**4.1 Model Limitations.** Despite the demonstrated advantages of this simulation-based design approach, several limitations serve as impediments to its use as the sole benchmark for retractor design.

**4.1.1 Model Geometry.** Differences in size and shape between the porcine and human left atria (the porcine heart was approximately average human size [38], but the atrium was approximately 52% of average human size) made the results not entirely representative of clinical procedures. The diameters of the mitral annuli are approximately the same for the human and porcine left atria; however, chamber dimensions perpendicular to the mitral annulus are much smaller for the porcine LA than for the human LA. The size of the porcine LA mandated that scaling factors be applied when transferring the retractor from the human to the porcine LA and hindered the model’s ability to completely predict correct length-to-width ratios for the retractor. Both the geometry of the model and the lack of adjacent geometries hindered the effectiveness of FEA in calculating the magnitudes of forces exerted between instruments and the LA. Limitations in model geometry could be addressed in future models by incorporating solid models obtained directly from human anatomies.

**4.1.2 Ease of Use.** Using FEs to evaluate retractors required extensive knowledge of the FE software and model construction. Making slight modifications to the models, especially changes in geometry, was tedious and time consuming. Any changes in geometry required importing the modified solid model, meshing it, and positioning it into the proper location. Problems associated with importing volumetric objects necessitated building...
solid models using ANSYS™ functions, even if solid models created with other software programs were already available. The position and angular orientation of the retractor must be defined by specifying displacements and rotation angles with respect to the global Cartesian coordinate system, which is a lengthy trial-and-error process.

4.1.3 Convergence. Attaining convergence required a delicate balance of retractor positioning and contact parameter adjustments. The inability of the solution to converge for certain positions of the retractor limited its use to certain retractor positions.

Positions had to be avoided in which the two contacting surfaces are far from parallel to each other (the angle of contact is much greater than zero). This included placing portions of the retractor near the edges separating the anterolateral and posterolateral halves of the heart. In effect, convergence difficulties limited the positioning of the retractor to the center of the LA. Convergence difficulties were aggravated by nonlinearities in the left atrial material model and nonlinearities in the contact problem. Exploring alternative solution algorithms and FEA packages with more complex contact algorithms might alleviate convergence issues and enhance the utility of this design methodology.

4.1.4 Material Model. Using an isotropic material model for left atrial tissue admittedly limited the accuracy of the FE model. Since biological tissues are known to exhibit anisotropic properties, an anisotropic material model would more aptly represent the characteristics of biological tissue and is recommended for use in
future studies. As FE software continues to mature, more appropriate material models will most likely become commercially available in the near future.

### 4.2 Endoscopic Retractor Design Recommendations

FE modeling of a porcine LA revealed several deficiencies of the current endoscopic retractor design. Contact pressures on the spheres were unacceptably high. This weakness could be alleviated by enlarging the diameter of the spheres to provide increased contact area and enhanced load distribution. Contact pressures could be further reduced by curving the retractor arms away from the LA to increase LA/wire contact. The protrusion at the distal end of the blade received negligible contact with the LA, limiting tissue retraction at the mitral annulus. To address this deficiency, the distal protrusion should be lengthened to project further from the blade.

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### References


