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Characterization of Mechanical Properties of Pericardium Tissue using Planar Biaxial Tension and Flexural Deformation

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Abstract

Flexure is an important mode of deformation for native and bioprosthetic heart valves. However, mechanical characterization of bioprosthetic leaflet materials has been done primarily through planar tensile testing. In this study, an integrated experimental and computational cantilever beam bending test was performed to characterize the flexural properties of glutaraldehyde-treated bovine and porcine pericardium of different thicknesses. A strain-invariant based structural constitutive model was used to model the pericardial mechanical behavior quantified through the bending tests of this study and the planar biaxial tests previously performed. The model parameters were optimized through an inverse finite element (FE) procedure in order to describe both sets of experimental data. The optimized material properties were implemented in FE simulations of transcatheter aortic valve (TAV) deformation. It was observed that porcine pericardium TAV leaflets experienced significantly more flexure than bovine when subjected to opening pressurization, and that the flexure may be overestimated using a constitutive model derived from purely planar tensile experimental data. Thus, modeling of a combination of flexural and biaxial tensile testing data may be necessary to more accurately describe the mechanical properties of pericardium, and to computationally investigate bioprosthetic leaflet function and design.

Graphical Abstract

Keywords

Constitutive modeling; inverse finite element analysis; transcatheter aortic valve replacement

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1. Introduction

Bioprosthetic heart valves (**BHV**) have been widely used to treat valvular disease for over forty years (1) and offer excellent hemodynamic performance and low risk of thrombogenesis. Glutaraldehyde-treated bovine pericardium (**BP**) is recognized for its attractive mechanical properties and has been considered one of the standard sources for BHV leaflets (2). Some recent generations of transcatheter valve designs have elected to use glutaraldehyde-treated porcine pericardium (**PP**). PP tissue is substantially thinner than BP (3), which lends itself to a more low-profile valve design, consequently, improving the safety of transcatheter delivery.

The long-term durability of BHVs is limited by structural deterioration and calcification of the tissue leaflets (4). There have been many studies showing that the regions of BHV leaflet tearing correspond with regions of high tensile and bending stresses acting on the leaflets during opening and closing (5). Therefore, it is important to understand the stresses acting on the leaflets to mitigate BHV failure. Several groups have used computational approaches to investigate structural and hemodynamic valve function (6–18). Numerical simulations allow for rapid, resource-efficient, quantitative analyses compared to experimental methodologies; however, the accuracy of computational modeling hinges on the accuracy of the prescribed material properties (19).

The mechanical properties of glutaraldehyde-treated BP and PP, have been investigated and characterized primarily through uniaxial and biaxial tensile tests (3, 20–27), which are both planar tests. Studies on the out-of-plane, flexural behavior of pericardia are limited (28, 29) despite the critical role leaflet bending plays in bioprosthetic heart valve function and durability (30). In-plane tensile testing data cannot be directly used to derive off-plane properties unless certain material symmetry or isotropy conditions are assumed. Pericardial tissues, along with many other soft tissues, are highly compliant, and offer very little resistance to bending, which complicates tissue bending experiments. Even less has been done to implement both the tensile and flexural behaviors into a single constitutive model to more accurately describe the pericardial tissue mechanical behavior. While data from isolated bending experiments of pericardia can offer insight on BHV function, these experiments cannot replicate the complex in vivo loading conditions of BHVs. Consequently, there is still a jump between understanding the BHV tissue material flexural response and the BHV structural response. Constitutive models describing both the tensile and flexural responses of BHV biomaterials are needed to interpolate and extrapolate the isolated experimental data to different loading states. Simulation-based analyses can then be used to translate tissue-level responses to accurate BHV device-level responses.

In this study, an integrated experimental and inverse finite element (**FE**) procedure was developed to characterize the biaxial tensile and flexural properties of pericardial tissue and implement these properties into a single nonlinear, anisotropic, constitutive model. Biaxial testing data of BP and PP tissues were obtained previously (3). A beam bending experiment was performed to characterize the tissue flexural responses. Thin and thick BP and PP tissues were tested. The material response for each tissue type was described with a structural-based constitutive model. The material parameters were optimized through inverse

FE analysis in order to capture both the biaxial tensile and beam bending experimental data. The optimized material parameters for each tissue type were then implemented in simulations of BHV opening to assess the impacts of tissue type and thickness on performance.

2. Materials and Methods

2.1 Tissue Preparation

Fresh bovine and porcine pericardia stored on ice were obtained from Animal Technologies (Tyler, TX). Details on tissue pericardium preparation have been previously described (3). Briefly, the pericardial sacs were dissected, the fatty nodules were removed cautiously, and large sections were pinned in a trampoline fashion without overextension. Fresh BP and PP sheets were treated with 0.625% glutaraldehyde (GL) solution, and a crosslinking solution (6% Formaldehyde, 2.2% Ethanol, 1.2% Tween 80) before final transfer to a 0.25% GL storage solution at 4° C. Rectangular samples of BP (12 × 5.75 mm) (n=23) and PP (7.5 × 6.5 mm) (n=27) tissue were prepared. Tissue thickness was measured in three areas along the length of the sample using a non-rotating thickness gauge (Mitutoyo, Model 7301) and the mean value was documented. The samples were divided into two distinct thickness groups for each tissue type. The thickness of thick bovine pericardium (**BPK**) (n=12), thin bovine pericardium (**BPN**) (n=11), thick porcine pericardium (**PPK**) (n=15), and thin porcine pericardium (**PPN**) (n=12) were 0.45 ± 0.03 mm, 0.32 ± 0.03 mm, 0.20 ± 0.03 mm, and 0.14 ± 0.02 mm, respectively.

2.2 Experimental Bending

A cantilever beam bending experiment was setup in a 0.9% saline bath by sandwiching a tissue specimen on one end and allowing the rest of the specimen to free float in water as shown in Figure 1A. A small needle with suture was fed through the free end of the tissue such that the weight of the needle deformed the tissue in a beam bending fashion (Fig. 1A). A calibration ruler was fixed in place near the beam for calibration during subsequent image analysis. Two needles of differing weights were applied to each BP (56mg and 125mg) and PP (19mg and 45mg) specimen. Images were taken of each sample with each applied load in order to measure the tissue deformation. The edge of the tissue specimens were darkened to enhance the contrast between the tissue and background to facilitate the image-based measurements.

Image digitization was carried out using custom Matlab (MathWorks Inc., Natick, MA) scripts. A representative example of the digitization process is shown in Figure 1. Images were individually cropped and converted to binary using a background-level threshold as shown in Figure 1B. The sample position along its length was digitized by selecting points along the specimen in the thresholded image, and converted to units of mm using the pixel-to-mm ratio determined from the calibration ruler (Fig. 1C). After digitization of all the experiments, points along the deflection curve were fit to a third-order polynomial. The vertical coordinates at set horizontal intervals along the length of the tissue were averaged between specimens of each group to obtain a mean $\pm$ standard deviation flexural response.

2.3 Inverse finite element analysis

An inverse FE procedure was conducted to obtain the tissue flexural material properties from the tissue cantilever beam experiments. Commercial software for FE analysis, Abaqus/Standard 6.13 and process automation, Isight 5.9 (Dassault Systemes Simulia Corp, Johnston, RI), were used. The 3D tissue geometries were created using the average dimensions of BPK, BPN, PPK and PPN specimens. Tissue finite element models were developed with 6732 solid continuum elements (Abaqus C3D8I) with 51 elements along the length, 33 elements along the width, and 4 layers of elements through the tissue thickness. Cantilever bending was simulated by fixing the nodes along one end of the tissue and applying a concentrated force to the free end mimicking the experimental protocol. In each simulation, two forces were applied: the first corresponding to the lighter weight applied experimentally and the second corresponding to the heavier weight applied experimentally. The tissue deformation due to the application of each weight was exported from the simulations as the position of the nodes along the tissue length midway through the thickness using custom python scripts as shown in Figure 2. The position of the nodes in the region that was fixed were not extracted.

BP and PP were assumed to be anisotropic, incompressible, hyperelastic materials. Therefore, the strain energy function, W , was defined using a modified version of the fiber-reinforced, hyperelastic material model based on the work by Holzapfel et al (32) given by

$$W = C_{10} \{ \exp[C_{01}(\bar{I}_1 - 3)] - 1 \} + \frac{k_1}{2k_2} \sum_{i=1}^2 \{ \exp[k_2(\bar{I}_{4i} - 1)^2] - 1 \} + \frac{1}{D} (J - 1)^2, \quad (1)$$

where C_{10} , C_{01} , k_1 , k_2 , and D are material constants, and $\bar{I}_1$ and $\bar{I}_{4i}$ are deviatoric strain invariants. Parameters C_{10} and C_{01} characterize the material response of the ECM and collagen cross-links, while k_1 and k_2 represent the mechanical response of the collagen fibers. Material constant D controls incompressibility and J is the determinant of the deformation gradient tensor. The fiber orientation was defined by structural tensor, $\mathbf{M}_i = \mathbf{m}_{0i} \otimes \mathbf{m}_{0i}$ with $\mathbf{m}_{01} = [\cos\theta, \sin\theta, 0]$ and $\mathbf{m}_{02} = [\cos\theta, -\sin\theta, 0]$. It has been shown that Eqn. 1 can capture the biaxial response of chemically-treated pericardia to a high degree of accuracy (33) and can describe 3D out-of-plane deformations necessary to model the tissue bending response. Equation 1 was implemented in a user-defined material (UMAT) subroutine in Abaqus.

The material constants of Eq. 1 were obtained through an inverse FE procedure. The initial material constant values were obtained through fitting the stress-strain data from recent biaxial testing of BP and PP (3) in the least-squares sense. Previous work by Mirnajafi et al. (28) has shown that the BP bending response is dominated by the non-collagenous tissue components as well as inter-fiber cross-links as opposed to the stiffness of the collagen fibers themselves. Therefore, in the inverse FE procedure, parameters C_{10} and C_{01} controlling the material response of the ECM and collagen cross-links, the main contributors to the flexural response, were optimized and the remaining parameters related to the

collagen fibers were kept constant at the initial values. Optimization was performed by maximizing the objective functions, R_1^2 and R_2^2 ,

$$R_1^2 = 1 - \frac{\sum_i (y_{i1} - f_{i1})^2}{\sum_i (y_{i1} - \bar{y}_1)^2}, \quad (2)$$

$$R_2^2 = 1 - \frac{\sum_i (y_{i2} - f_{i2})^2}{\sum_i (y_{i2} - \bar{y}_2)^2}, \quad (3)$$

where y_{i1} and y_{i2} are the y coordinates of the tissue measured experimentally at each x increment under weights 1 and 2 respectively, and f_{i1} and f_{i2} are the corresponding y coordinates obtained from simulations. Here R_1^2 and R_2^2 represent the goodness of fit in terms of the coefficient of determination between the computational tissue deformation and experimental tissue deformation under weights 1 and 2 respectively. The Archive-based Micro Genetic Algorithm (**AMGA**), a multi-objective optimization scheme within Isight, was chosen to explore the parameter space. A total of 180 runs were completed for each tissue type. An illustration of the optimization workflow is presented in Figure 3.

After searching the parameter space, the range of C_{10} and C_{01} values to maximize Eqs. 2 and 3 were identified. To ensure that the material parameters could accurately capture the biaxial response in addition to the flexural response, the biaxial data for each tissue group was re-fitted to Eq. 1 in the least-squares sense, with the constraints that the fiber related parameters k_1 , k_2 , and θ , were fixed at their original values obtained from fitting the biaxial test data only, and the ECM-related parameters, C_{10} and C_{01} , were fitted within the optimal bounds identified through AMGA optimization.

2.4 Heart Valve Modeling

A previously described FE model of a generic, 23 mm transcatheter aortic valve (**TAV**) (34, 35) was used in this study. One TAV model was created for each tissue group: BPN, BPK, PPN and PPK. Each leaflet was discretized into 5394 large-strain brick (Abaqus C3D8H) elements with two layers of elements through the thickness using HyperMesh 12.0 (Altair Engineering, Troy, MI). The leaflet thickness was defined based on the average thickness of the tissue samples for each respective tissue group. Deformation of the TAV stent was assumed to be negligible; therefore the stent geometry was excluded from the analysis. Nodes along the stent-attachment line were fixed from displacement to mimic attachment to a rigid stent. The leaflet mechanical properties were defined by Eqn. 1 using both the material parameters defined from the biaxial test data only (**Biax**) and the optimized material parameters (**Opt**) determined through the inverse FE procedure. Quasi-static valve opening was simulated for each valve model using both sets of material properties by applying a uniform pressure from 0 to 10 mmHg to the ventricular side of the leaflets.

Valve opening was assessed by extracting the leaflet deformation at the leaflet midline (Fig. 4A&B) as well as the geometric orifice area (**GOA**). In-house Python scripts were used to export images of the valve from the top view in the x-y plane during opening (Fig. 4C). The images were imported into Matlab and converted to binary using an appropriate threshold (Fig. 4D). The background pixels lying outside of the circular valve profile were removed. The known valve dimensions were used to determine the pixel-to-length ratio. Black and white pixels were totaled and the ratio was used to calculate the GOA. In this way, GOA was computed in a manner similar to aortic valve planimetry, a clinical method used to measure aortic valve area (36).

3 Results

3.1 Pericardium flexural response

The average deflection curves for each tissue group are shown in Figure 5. In each case, application of weight resulted in nonlinear tissue beam deformation. Thinner tissues underwent greater bending deflection than their thicker counterparts. At 56 mg and 125 mg loads, the BPN samples had peak tip deflections of 9.39 mm and 11.5 mm, respectively, while the BPK samples had corresponding deflections of 6.30 mm and 8.38 mm, respectively. Similarly, at 19 mg and 45 mg loads, the PPN samples had tip deflections of 3.46 mm and 5.21 mm, while PPK samples had deflections of 3.23 mm and 4.57 mm, respectively.

3.2 Pericardium tissue material modeling

The tissue cantilever beam deformation was simulated using the material parameters determined through fitting the biaxial data only (**Biax**) as well as the material parameters optimized through inverse FE procedure (**Opt**) and were compared to the experimentally measured mean deformation curves (Fig. 6&7). In all conditions, the simulated deformation using the Opt material parameters was within one standard deviation of the mean experimental deformation, and was more accurate than the simulations using the Biax parameters. The Biax material model coefficients tended to produce a softer flexural response. Corrections to C_{10} and C_{01} , the parameters governing the low-strain, isotropic response, resulted in beam deflections that were more accurate in regards to tip deflection magnitude and deformed tissue shape.

The absolute values of C_{10} and C_{01} were not strongly correlated with the simulated deflection accuracy; however, the product of the two parameters, denoted as $CC = C_{10} \times$

C_{01} , was strongly correlated with the average R^2 , defined as $R^2 = \frac{R_1^2 + R_2^2}{2}$. The relationship between CC and the average R^2 can be well characterized by a second order polynomial as shown in Figure 8. The value of CC at the maximum R^2 was identified for each tissue type, and then to obtain the optimized material parameters for each group, the biaxial test data was re-fit to Eq. 1 given the constraint that the CC was equal to this value. The initial and optimized material parameters for each tissue group are presented in Table 1.

3.3 Simulation of TAV deformation

The initial and optimized material parameters were utilized in FE simulations of TAV opening. The TAV leaflet midlines in the undeformed (0 mmHg) and deformed (4 mmHg) configurations for each tissue type shown in Figures 9A&B, highlight differences in the tissue bending properties. The porcine leaflets deformed more than the bovine leaflets, and the leaflets defined with the initial (Biax) material properties deformed more than those defined with the optimized (Opt) material properties.

Valve deformation developed comparably between tissue types during opening. Leaflets opened rapidly after reaching a threshold opening pressure then the GOA plateaued (Fig. 9C&D). Threshold pressures were 1.64, 2.77, 0.57, and 0.38 mmHg for BPN, BPK, PPN, and PPK tissues, respectively. The porcine leaflets exhibited greater flexure than bovine leaflets at corresponding pressure levels. While both porcine valves reached GOAs approaching 3cm², the GOA for BPN plateaued around 2.6 cm² and stayed below 2.5 cm² for BPK. The differences in GOA between valves using the Biax and Opt material parameters at a transvalvular pressure gradient of approximately 4mmHg, were 10.9% and 6.4% for BPN and PPN, respectively. The optimized material parameters which accounted for flexural deformation consistently predicted smaller GOAs than the models which implemented material coefficients exclusively derived from biaxial testing data.

4 Discussion

Chemically-treated pericardial tissue has low flexural rigidity. Large flexural deflections can occur at low stresses and strains. As such, flexural testing is complementary to planar uniaxial or biaxial tensile testing which typically subjects tissues to a high load and large deformation, and allows for fine determination of tissue stiffness in the low stress-strain region which generally cannot be accurately measured in tensile tests (28). This is particularly true for thin tissues capable of larger bending deflections under very small loads, which may explain why, in this study, the Biax material parameters for the thin tissue groups (BPN and PPN) could not accurately capture the corresponding cantilever beam experimental data, while the Biax material parameters for the thick tissue groups (BPK and PPK) were much closer to the Opt material parameters. Generally, it was observed that deflection optimization resulted in stiffer material parameters than those obtained from a biaxial testing-only approach. The degree of accuracy in which the model parameters of Eq. 1 could capture the cantilever beam experimental data was found to be governed by the product of the matrix-related parameters. This result can be explained by the fact that with this particular constitutive model of Eq. 1, the product of the matrix-related parameters determines the material stiffness at low stresses before the fibers are recruited, i.e. the stress (Second Piola Kirchhoff Stress) in the matrix is defined by $S = 2C_{10}C_{01} \exp(C_{01}(I_1 - 3)) = 2CC \exp(C_{01}(I_1 - 3))$. In physical terms, this indicates that the tissue beam bending responses were dominated by the tissue stiffness at low stresses, as expected. It is clear that multi-modal testing allows for finer characterization of soft tissue mechanics.

Reports of flexural tests of soft tissues in the literature are limited. Flexural test set-ups are complicated by the fact that materials subjected to flexure undergo tensile and compressive loading. Typically groups focus on the tensile properties of soft tissues only. In-plane

compression testing of thin, fibrous soft tissue like pericardium is technically challenging. Mirnajafi et al (28) have presented bending data of BP using a three-point bending experiment specifically designed for soft tissue (37, 38). However, the data from three-point bending tests can be difficult to analyze and incorporate into constitutive law. The Euler-Bernoulli beam theory is often utilized to determine the elastic modulus of the material from the test data with the assumption that the material is linearly elastic (28). To eliminate the linear elasticity assumption, Nicosia (29) developed a theoretical framework to analyze soft tissue bending test data based on finite elasticity theory. Using this approach the material parameters of a general strain energy function can be determined; however, the analytical solution may become complicated when using complex strain energy functions, or when characterizing non-homogeneous or multi-layered materials.

In this study, an integrated experimental and computational inverse FE methodology was developed to analyze the bending experiments of chemically-treated BP and PP. The cantilever beam bending experiment adopted in this study benefits from utilizing simple geometry and loading conditions to cause large deflections. This setup eliminates the uncertainty of frictional force between tissue and tissue support rods that often occurred in a 3-point, or 4-point bending test. More importantly, the inverse FE method developed is more general: it eliminates the assumption of linear elasticity. By taking this approach the set of material parameters to fit both tensile and flexural test data could be determined. This framework is general and could easily be adapted to analyze different datasets as long as the experimental set-up could be accurately modeled using the FE method.

The BP and PP constitutive models were used to define the leaflet mechanical behavior in simulations of TAV opening where flexure is a primary loading condition. While PP tissues are less compliant than BP tissues under biaxial tension (3), the PP TAV leaflets underwent larger flexural deformations than the BP leaflets leading to larger GOAs, which may be due to intrinsic non-collagenous material property differences in addition to the reduced thickness. The leaflet thickness affected the BP TAV opening. The BPN valve, utilizing 30% thinner leaflets than the BPK valve, opened under 40.4% less pressure and achieved a 6.80% larger maximum GOA. This could be an advantage of TAV leaflet materials versus thicker surgical valve leaflet materials allowing for larger GOAs. Interestingly, the PPN leaflets deformed slightly less than the PPK leaflets during valve opening despite the 30% reduced thickness, suggesting that the bending stiffness of PPN tissues is greater than that of PPK tissues.

TAV opening was simulated using both the initial (Biax) parameters and the optimized (Opt) parameters. The same trend of increased flexural stiffness among the Opt tissue responses seen in the cantilever beam simulations, were observed in the TAV opening simulations. The largest differences were observed between the BPN TAVs. A 7.35% increase in the maximum GOA and 50.2% decrease in the opening pressure was observed between the BPN TAV simulated with Biax parameters versus that with Opt parameters, which demonstrates the importance of incorporating flexural experimental data in the leaflet material model in order to accurately simulate TAV deformations.

4.1 Limitations

It is difficult to directly compare the flexural responses of BP and PP from the cantilever beam bending tests given the different loading conditions and specimen sizes used in the experiments. Due to large difference in tissue thicknesses where BP tissues were approximately twice as thick as PP tissues, it was difficult to determine standard tissue dimensions and weights to produce good data for both groups. Differences in the experimental set-up also make it challenging to compare the test data with BP flexural data obtained from three-point bending tests published in the literature (28). In this study, TAV opening simulations utilizing the derived BP and PP constitutive models were run to show the exemplary effects of the inclusion of flexural data in the constitutive models, and to facilitate comparison between the tissue groups. The GOA obtained from purely structural FE simulations of TAV opening as in this study have been shown to be overestimated by our group (39); however the comparison among groups is valid. TAV closure was not simulated under the assumption that the flexural properties occurring at low stresses are negligible at the high stress range tissue responses seen under closing pressure.

5 Conclusions

In this study, an integrated experimental and computational inverse FE methodology was developed to model the planar tensile and flexural mechanical responses of chemically-treated BP and PP with a single, nonlinear, anisotropic constitutive model. The inverse FE approach is general and eliminated many of the assumptions required by analytical methods to characterize bending data including linear elasticity, which is not valid for many soft tissues. It was shown that constitutive model parameters derived from purely planar tensile loading data could not accurately capture the tissue flexural response, particularly for the thinner tissues. Thus, a combination of in-plane and out-of-plane testing data may be necessary to more accurately model tissue behavior. The optimized constitutive model parameters were utilized in FE simulations of TAV opening, which showed that TAVs with PP leaflets can open under lower pressures and to larger GOAs than TAVs with BP leaflets, and GOAs for BP TAVs can be increased by decreasing the leaflet thickness.

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Highlights

- A methodology to model the tensile and flexural properties of tissue is proposed
- This method eliminates the commonly adopted assumption of linear elasticity
- Constitutive models for pericardial tissue were obtained
- Optimized models were used in simulations of transcatheter heart valve deformation

Figure 1.

A) Tissue cantilever bending experiment: the white thread is used to apply load to the specimen, and the ruler at the bottom of the image is for calibration. B) Thresholding result of the image in A) to isolate the tissue specimen. C) The digitized position of the tissue sample along its length.

Figure 2.

A) Simulated tissue cantilever beam experiment in the undeformed (gray) and deformed (green) configurations. B) The deformed tissue geometry defined as the position of the nodes highlighted in red in A).

Figure 3.
Optimization structure for determination of material model parameters.

Figure 4.

Representative valve model with leaflet midline used for extracting the deformation highlighted in red in the A) undeformed and B) deformed (4 mmHg) configurations. C) Top-view snapshot of the deformed valve in the x–y plane exported from the simulation, and D) top-view snapshot of the deformed valve after thresholding to measure the GOA (inner black area) of the deformed valve.

Figure 5.
Mean beam deflection of A) BP and B) PP samples shown with standard deviation bars.

Figure 6.

Average experimental deflection curves (blue) compared to FEA deflection resulting from the Opt material parameters (red) and the Biax material parameters (black) for A) BPN under 56 mg, B) BPN under 125 mg, C) BPK under 56 mg, and BPK under 125 mg.

Figure 7.

Average experimental deflection curves (blue) compared to FEA deflection resulting from the Opt material parameters (red) and the Bi (black) for A) PPN under 19 mg, B) PPN under 45 mg, C) PPK under 19 mg, and PPK under 45 mg.

Figure 8.

Influence of CC on the goodness of fit between the simulated and experimental deflection for A) BPN, B) BPK, C) PPN, and D) PPK samples. The raw data (black circles) were fitted with the quadratic functions displayed (red lines).

Figure 9.

Simulated TAV leaflet midline deformation for A) each valve using the optimized (Opt) material parameters, and B) the thin tissue groups using the initial (Biax) and Opt material parameters. Simulated geometric orifice area (GOA) for A) each valve using the Opt material parameters, and B) the thin tissue groups using the Biax and Opt material parameters.

Table 1

Initial and optimized material model parameters.

Initial (Biax) Material Parameters						
	C₁₀ (kPa)	C₀₁	CC (kPa)	k₁ (kPa)	k₂	Θ (°) R²
BPN	4.814	13.222	63.6507	27.8908	56.3524	8.32 0.9614
BPK	8.385	11.163	93.6018	10.37191	37.1714	0.016 0.9677
PPN	2.044	32.111	65.6349	152.6404	107.272	7.81 0.9872
PPK	2.866	26.486	75.9089	109.6696	91.9787	0.002 0.9788
Optimized (Opt) Material Parameters						
	C₁₀ (kPa)	C₀₁	CC (kPa)	k₁ (kPa)	k₂	Θ (°) R²
BPN	18.5987	7.9345	147.6	27.8908	56.3524	8.32 0.9339
BPK	11.6045	9.8502	114.3	10.37191	37.1714	0.016 0.9658
PPN	15.1414	13.4848	204.2	152.6404	107.272	7.81 0.9252
PPK	2.3196	28.5449	66.21	109.6696	91.9787	0.002 0.9781