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Effect of Geometry on the Leaflet Stresses in Simulated Models of Congenital Bicuspid Aortic Valves

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Abstract

The aim of the study was to assess the effect of geometric variations on the stresses developed in the leaflets of congenital bicuspid aortic valves (CBAV). We developed a model for the human tri-leaflet aortic valve based on the geometry and dimensions published in the literature. We also developed simulated CBAV geometry based on the most common geometry present in patients with CBAV that is published in the literature. We employed a constitutive relationship for the leaflet material from the previously published experimental data of fresh porcine aortic valve leaflet specimens for the analysis. We performed dynamic finite element (FE) structural analysis of the valves in the aortic position in order to compute the strain and stress distribution on the leaflets of the tri-leaflet valve and the CBAV models. Our results showed that large changes in the computed in-plane leaflet strain and stress occurred with variations in the geometry of the simulated CBAV whereas changes due to alterations in material constants were correspondingly less. The valve orifice area in the fully open position was significantly reduced in CBAV compared to that for the tri-leaflet valve. The changes in geometry of CBAV resulted in large changes in in-plane strain and stress and our results suggest that geometrical variations may be a potential risk factor inducing calcific aortic stenosis frequently present in patients with CBAV.

Keywords

Dynamic simulation; Finite element analysis; In-plane leaflet stress and strain; Calcific aortic stenosis

INTRODUCTION

While the normal aortic valve usually consists of three cusps, and their corresponding sinuses, in 1–2% of the population the aortic valve develops with two cusps, a condition which is referred to as the congenital bicuspid aortic valve (CBAV).⁵ The true mechanism behind development of a CBAV is unknown but recent studies point to an inheritable genetic trait with a slight familial and male predominance.⁵ The resulting valve geometry

varies but the most common morphology has been identified as a fusion of two leaflets and a single raphe along the site of fusion where raphe refers to a small strip of non-valvular tissue.¹⁵ Approximately 33% of the CBAV population develop major cardiovascular complications including aortic stenosis, regurgitation, and endocarditis.¹⁸ There also exists an increased risk of developing other complications in the ascending aorta including aortic dissection and coarctation which may be the result of abnormal blood flow out of the CBAV.⁵ Factors engendering calcification include abnormal geometry and tissue structure.⁵ There have been several studies investigating the reasons behind calcification in CBAVs, with primary theories correlating regions of calcification to relatively high leaflet in-plane stresses.^{14,19} The reasons complications develop in the ascending aorta accompanying a CBAV are not well understood however, it has been proposed that the altered blood flow pattern from the valve may have an effect.^{14,19}

Deposition of calcium on the leaflet tissue is a known biological cascade present on the normal aortic valve in patients of advanced age.¹⁴ It is also known that patients with a CBAV develop more prominent calcific regions on the leaflets earlier than with the tri-leaflet aortic valve. It has been suggested that abnormal mechanical stresses on the leaflets quickens the process of calcification.¹⁴ Due to the short duration of the cardiac cycle, and large deformations that the leaflets are subjected to, it is impractical to measure the leaflet deformation and stresses accurately *in vivo*.

Numerous studies have employed the finite element (FE) method for the structural analysis of the aortic valves, particularly with quasi-static analyses of the stress distribution on the leaflets with the valve in the fully closed position.¹³ One of the first FE heart valve simulations was conducted by Cataloglu *et al.*² who employed a linear isotropic material model for the leaflets. More recently, several studies have been reported on the dynamic analysis of the valvular function during the cardiac cycle. Gnyaneshwar *et al.*⁷ developed a FE analysis of a normal tricuspid aortic valve, which included an aortic root. Simulation condition for the study included physiologically realistic pressure applied uniformly to the ventricular surface of the leaflet. This study pointed out the important role played by the aortic ring and the root on the valve function. A fluid structure interaction analysis of the dynamics of the aortic valve was reported by De Hart *et al.*⁴ who employed a neo-Hookean elastic material property model for the leaflet tissue and reported the presence of regions of stress concentration in the commissural region. Sun *et al.*¹⁶ have reported on a quasi-static FE analysis of the bioprosthetic heart valve in the fully closed position in order to determine the leaflet stress distribution with the pressure load. This work emphasized the importance of the leaflet tissue properties and structure on the resulting stress distribution. Kim *et al.*^{11–13} have reported on the dynamic analysis of a bioprosthetic valve in the aortic position with experimentally determined non-linear anisotropic material property specification for the leaflets. These studies underscored the importance of the use of physiologically realistic constitutive models and tissue structure on the quasi-static and dynamic analyses for the computation of the stress distribution on the leaflets. Yet, few studies have been reported on the structural analysis of CBAVs and the effect of geometry and material property alteration on leaflet calcification with these valves. Robicsek *et al.*¹⁴ reported on an *in vitro* experimental study of a CBAV mold obtained from a cadaveric valve specimen and pointed out the importance of the CBAV geometry as a factor resulting in structural failure. More recently, Weinberg and Kaazempur-Mofrad¹⁹ compared the mechanics of the normal tri-leaflet geometry of the aortic valve to a purely bicuspid geometry of a CBAV. Since the geometry of the CBAV varies widely among patients, the effect of geometric variation on the dynamics of the valvular function and on the stress distribution on the leaflets has not been analyzed in detail. We present a dynamic FE structural analysis of a simulated tri-leaflet aortic valve and four models of simulated CBAV representing commonly observed

geometry of the malformations observed in patients in order to delineate the effect of geometric alterations on the stress distribution.

FINITE ELEMENT (FE) DYNAMIC ANALYSIS

Challenges associated with this FE analysis include leaflet contact, anisotropic tissue properties, and the overall large leaflet deformation over a short duration. The FE analysis software package used was ABAQUS/Explicit (HKS Inc., Pawtucket, RI) for its efficacy in solving dynamic contact simulations involving large deformation. Aortic valve tissue possesses a complex microstructure composed of several layers and components allowing the valve to function properly. The three basic layers are the fibrosa, spongiosa, and ventricularis. Realistic analysis of this structure requires the specification of an anisotropic non-linear constitutive model incorporating the preferential fiber directions for accurate computational simulations.¹⁹ Currently, *in vivo* and *ex vivo* experimentation to quantify CBAV tissue has been limited. Hence, this study employed material properties measured from fresh porcine aortic valve leaflet tissue specimens and the details of the experimental data acquisition are included in Huang *et al.*¹⁰ and Huang.⁹ The material property was assumed to be constant throughout the leaflet excluding the raphe which was assumed to be made of stiffer material as discussed below. In order to computationally model the aortic valve leaflet tissue, this study employed an anisotropic, Fung-type hyper-elastic strain energy function as available in ABAQUS/Explicit, defined in Eqs. (1) and (2).^{6,12,16} The stiffer axis was assumed to be in the circumferentially direction with the cross fiber axis specified to be along the radial direction of the leaflets in all the simulations.

$$U = \frac{C}{2} (e^Q - 1) + \frac{1}{D} \left(\frac{(J^{el})^2 - 1}{2} - \ln J^{el} \right) \quad (1)$$

$$Q = \bar{\epsilon}_{ij}^G b_{ijkl} \bar{\epsilon}_{kl}^G \quad (2)$$

In Eq. (1), U is the strain energy per unit of reference volume; C and D are material parameters. Q is defined in Eq. (2) where b_{ijkl} is a dimensionless symmetric fourth-order tensor with material constants corresponding to the components of the modified Green strain tensor $\bar{\epsilon}_{ij}^G$. J^{el} is the elastic volume ratio; and D is two divided by the bulk modulus. The modified Green tensor is defined by $\bar{\epsilon}_{ij}^G = 1/2(J^{-2/3}c_{ij} - \delta_{ij})$ where c_{ij} are the components of the standard Green tensor. Biaxial testing of porcine aortic valve tissue, similar to human valve tissue, yields in-plane material constants exclusively, or only 6 of the 21 constants needed for the fully three-dimensional implementation of the Fung constitutive model. To facilitate the analysis, the values of material constants in the thickness direction have to be assumed. Since the leaflet is thin and the thickness direction is not load-bearing, we expect the material to be even softer than the weaker in-plane direction and have minimal influence on the in-plane response. Based on this consideration, the material parameters are assumed to be equal to half of the radial direction in-plane material constants. The parameters determined by nonlinear regression of the biaxial test data are shown in Table 1. The constant c was experimentally determined to be 9700 Pa, and assuming near incompressibility, D was assigned a value of $1e^{-6}$. Uniaxial stress-strain curves in the circumferential and radial directions derived from the material model employed in this simulation are included in Fig. 1. To account for viscous force dissipation created by blood interacting with the surface of the leaflet, damping has been introduced in ABAQUS/Explicit through the introduction of a thin skin layer of homogeneous linear elastic material

properties $E = 1000 \text{ Pa}$, $\nu = 0.4$, $\rho = 1100 \text{ kg/m}^3$ thickness 0.0386 mm , and damping coefficient $\alpha = 90000 \text{ g/s}$ which was applied to the diagonalized damping matrix.

Raphe presence was simulated by specifying arbitrarily stiffer material with $E = 2 \text{ MPa}$, $\nu = 0.15$, $\rho = 2200 \text{ kg/m}^3$ since the actual material behavior of the raphe have not been measured. Triangular shell elements were employed for the CBAV models in order to overcome excessive elemental distortion errors in ABAQUS, and square shell elements were employed for the normal tricuspid valve model. For mesh generation, the edges of the valve leaflets were seeded with nodes having a distribution of 0.8 node/mm , and this density was used for all of the models.

MODEL DESCRIPTION

A simulated normal tricuspid aortic valve model, with the geometry as shown in Fig. 2a, was used to validate the material model and the dynamic FE analysis with previously published results.¹² The results from the normal tri-leaflet aortic valve model were used for qualitative and quantitative comparison for the results from the CBAV simulations employed in this study. The radius, height, and thickness of the normal tricuspid valve are taken to be similar to average physiological values, 11.5 , 10.9 , and 0.386 mm respectively¹³ (Fig. 2b). Previous studies have documented the types of CBAVs present in the population along with corresponding number of occurrences.¹⁵ From those studies, four types of CBAV geometries have been created as shown in Fig. 2a. To create the CBAV models, the normal tricuspid aortic valve model was modified to maintain dimensions similar to physiologic average values of normal tricuspid valves. The geometrical features unique to the type of CBAV being modeled are approximate and idealized based qualitatively on surgical images. A majority of CBAV patients have a small strip of non-valvular tissue forming along the site of leaflet fusion at what would have been a commissure thus creating a false commissure.¹⁵ This study also includes an analysis of the effect of this raphe on the mechanical stresses of the valve by adding a section of stiffer material to the area of leaflet fusion to simulate its presence.

The outer boundary of the leaflet has all translational degrees freedom fixed; the inner edges are allowed to move freely. The valve annulus *in vivo* is also flexible and will deform during the opening and closing phases of the valve function. However, in this simulation to analyze the effect of valve geometry on the leaflet stress distribution, we assumed the annulus to be rigid. The leaflet motion was driven by a pressure load on the ventricular surface of the leaflet equal to the difference between experimentally measured ventricular and aortic pressures for a normal healthy human. The applied pressure magnitude as a function of time is shown in Fig. 3 and the instants for the fully open and closed positions of the leaflets for the comparison of the leaflet stress distribution among the various models are also indicated in this figure. Leaflet contact was simulated with three master/slave pairs and enforced by the penalty constraint. A mesh density test was conducted by doubling the number of nodes on each edge, increasing the number of nodes per millimeter on each edge from 0.8 to 1.6 . This increase in mesh density resulted in a 1.7% change in maximum in-plane stress.

RESULTS

The dynamic analysis of the in-plane principal stress distribution on the leaflets from the simulated tri-leaflet aortic valve in the fully open and fully closed positions is shown on the top row of Figs. 4 and 5, respectively. The stress distribution for the various CBAV valve geometries is also included in these figures. The magnitudes of the in-plane stress magnitudes with the valve in the closed position are much larger than the corresponding magnitudes in the open position. The maximum stress magnitudes in the normal tricuspid

valve were present at the leaflet attachment boundary, and the same was true for the CBAVs as well. The presence of raphe resulted in two local stress concentration spots on the fusion site and increased the region of high stress in the central portion of the free edge, however the global maximum located at the commissures was not affected. Further increases in the elastic modulus for the raphe ($E = 4$ or 8 MPa) resulted in increased magnitudes of the local stress concentration regions on the fusion site, but the overall stress distribution on the leaflets including the maximum in-plane stress magnitudes near the cuspal attachment regions were unaffected. These results also show that the maximum in-plane strain and stress occur with Type I (where two leaflets are smaller than normal and the third is large) and Type IV (two of the three leaflets are fused with a raphe along the fusion) compared to Types II and III. In order to assess the variations in the stress distribution due to alterations in the material property of the leaflets, the analysis was repeated with increasing and decreasing the constant c of the Fung constitutive model by 25% and those results are included in the left and right columns respectively of Figs. 4 and 5.

The computed maximum in-plane principal stress and strain results of the simulated tricuspid valve were then compared with the simulated CBAV geometries as shown in Tables 2 and 3. The data included in the tables are with the valves in the fully closed position where the stresses on the leaflets are the largest. The percentage change in the in-plane principal stress and the in-plane principal strain for the CBAV models in comparison to that of the tri-leaflet valve model with the mean material constant c are also included in the tables. The largest change with the geometrical variations in the CBAV models was 161% in the maximum in-plane principal stress, and 64% in the maximum in-plane strain. The largest change when varying the material constants was 42% in the maximum in-plane principal stress and 13% in the maximum in-plane principal strain as can be observed from these tables.

Previous studies reporting on dynamic FE analysis with tri-leaflet valves have reported the results for von Mises stresses rather than in-plane principal stresses. The maximum in-plane von Mises stress computed in the tri-leaflet valve in our simulation was about 3.36 MPa in the commissural region with the leaflets in the fully closed position. These results are in the same range of maximum stress values reported by previous studies: Gnyaneshwar *et al.*,⁷ -0.8 MPa; Kim *et al.*,¹¹⁻¹³ -2.09 MPa; and Balguid *et al.*,¹ -0.6 MPa also in the same anatomic region of the aortic valve. Gnyaneshwar *et al.* employed an isotropic linear elastic property for the leaflet with a 0.40 mm thickness for the leaflet simulating a human aortic valve. Balguid *et al.* employed a neo-Hookean elastic property with a leaflet thickness of 0.46 mm for a fiber-reinforced valve leaflet. The studies by Kim *et al.* and the present work employed an anisotropic Fung model based on bi-axial experimental data for pericardial and porcine aortic leaflet tissue respectively with a leaflet thickness of 0.386 mm. Hence the maximum magnitude of the maximum Von Mises stress in the present work is slightly higher than those for Gnyaneshwar *et al.* and Balguid *et al.* but closer to those reported by Kim *et al.*

The aortic valve orifice area with the valve in the fully open position for the tri-leaflet valve is shown in Fig. 6. The triangular open area shown in this figure is similar to that reported in Kim *et al.*¹¹⁻¹³ The corresponding orifice area in the fully open position for the CBAV is also shown in this figure for comparison. A CBAV of Type I reduced the open orifice by 33%, Type II by 81% and Type III and IV by 51%. These results suggest that, as the blood is forced to flow through the significantly reduced orifice with the CBAV, the flow in the ascending aorta will be altered and may play a role with problems in the ascending aorta reported in patients with CBAV. We assumed that the valve annulus is rigid in this analysis in order to compare the stress distribution and valve orifice area in the fully open position as

a function of the geometry of the CBAV. However, *in vivo*, the deformation of the flexible valve annulus may result in alterations in the predicted valve orifice area.

DISCUSSION

Patients with CBAV are at increased risk for leaflet calcification as well as aortic dissection and coarctation. It can be anticipated that the induced mechanical stresses on the leaflets with CBAV will be different from those with the normal tri-leaflet valve geometry. Moreover, the geometry of the CBAV varies widely and the propensity for valvular diseases in these patients may be associated with specific geometry. The present work was intended to determine the effect of CBAV geometry on the stresses developed on the leaflets, since these abnormal stresses may potentially be a precursor for valvular diseases in these patients. We developed a simulated model of the normal tri-leaflet aortic valve based on the nominal dimensions and geometry published in the literature. We also employed the Fung non-linear anisotropic model for the valve leaflets and obtained the material constants for the constitutive relationship from previously published experimental data for normal porcine aortic valve tissue. We performed a dynamic FE structural analysis to compute the leaflet deformation and the in-plane stress distribution on the leaflets during the opening and closing phases of the cardiac cycle. The results presented for the normal aortic valve leaflets agreed with those from prior publications. We also developed four simulated CBAVs based on the most common geometry of these malformations reported in the literature and performed dynamic analyses with these simulated models. The maximum magnitudes of the computed in-plane stresses on the leaflets with CBAV varied by as much as about 161% compared to the maximum stress magnitudes developed in the simulated tri-leaflet valve geometry. In order to assess the relative importance of geometric variations with the CBAV in comparison with the alterations in leaflet material property, we also performed the simulations with material constants increased or decreased by 25% from the magnitudes from the experimental data. The maximum variation in stresses with the variation in material property was about 42%. These results suggest that variations in the geometry of CBAV have a more dominant effect on the stresses developed in the leaflets. Thubrikar *et al.*¹⁷ reported on the calcific patterns on surgically excised end-stage calcified valves. Their results show the calcium deposits in the areas of cuspal flexion and commissural area with the calcification spreading radially into the belly area of the leaflets. Our results show regions of stress concentration near the cuspal attachment area for the normal tri-leaflet valves with more extensive regions of stress concentration with some geometries of CBAV. In as much as that abnormal stresses may lead to structural damage on the leaflet microstructure leading to calcification, our study suggests that patients with certain types of CBAV geometry may be at increased risk to develop leaflet calcification and ascending aortic wall problems and this method can be potentially employed to stratify patients with increased risk among those with CBAV.

In these simulations, we employed a normal tri-leaflet valve model without any leaflet calcifications and hence employed clinical pressure data for normal patients in the dynamic FE analysis. We then compared the stress pattern on the leaflets with simulated bi-cuspid valves with the same pressure load with the assumption that the analysis is being performed on bi-cuspid valves prior to calcification. As calcifications develop on these leaflets, the hemodynamic conditions in the patients will be altered and appropriate pressure load data from patients need to be employed in analyzing the dynamics of calcified bi-cuspid valves.

Our simulations also showed that the orifice area of the valves in the fully open position for the CBAV were significantly decreased compared to that for the tri-leaflet aortic valve. Even though our dynamic FE analysis neglected the effect of the local flow dynamics on the leaflet deformation, this study suggests that the orifice area in the fully open position with

CBAV may be smaller than that for the healthy tri-leaflet valves. As the blood is forced through such potentially restricted orifices during peak forward flow, abnormal jet-like flow past the orifice can also be anticipated with potential impact of the jet with the aortic wall in the ascending aorta. Hence it can be anticipated that the stresses that the segments of the ascending aortic wall is subjected to due to these abnormal flow pattern may result in aortic dissection and coarctation observed with these patients. Further study on the ascending aortic fluid dynamics in CBAV patients may prove to be very useful in understanding any potential relationship between fluid induced stresses and the problems in the ascending aortic wall that these patients develop. The orifice open area predicted in our analysis for the truly bicuspid valve (Type II in Fig. 6) is similar to that shown by Robicsek *et al.*¹⁴ from *in vitro* experiments of an excised bi-cuspid valve. Nevertheless, we assumed a rigid valve root in our analysis and introduction of flexibility of the root can be anticipated to alter the orifice opening even though the area of opening may still be significantly different from the normal tri-leaflet valve. With the availability of appropriate valve root material property experimental data, the simulations can incorporate the flexibility of the root on the dynamic analysis as well.

LIMITATIONS

There are several limitations in the study presented that must be pointed out. We developed the models for the normal tri-leaflet valve and the most often occurring CBAV based on idealized geometry. Several studies have been reported in the literature on obtaining data with ultrasound and MR imaging to obtain morphologically realistic three-dimensional geometry of the aortic valve.^{3,8} Performing the dynamic analysis on morphologically realistic patient-specific geometry with the boundary conditions specified based on the hemodynamic data from those patients will result in more realistic simulation for the computation of the stress distribution on the leaflets. If these patients can be followed up for the development of aortic valve disease, the proposed correlation between the presence of regions of abnormal stresses in the leaflets and the leaflet calcification can be established. Such studies may prove to be important in our ability to stratify CBAV patients with increased risk to develop calcification. We also assumed the mechanical behavior of the leaflets for the CBAV to be the same as that of the healthy tri-leaflet valve leaflets and specified the material constants for the analysis from previously reported experimental data for fresh porcine aortic tissue. We also arbitrarily assumed that the elastic constant in the region of raphe will be twice as stiff as in the other regions of the leaflets. There are limited numbers of publications that have reported on the force–deformation behavior of normal human aortic valve leaflet tissue. As far as we are aware of, there are no published data on such experimental data with CBAV leaflets. Experimental studies performed with tissue specimens from patients with CBAV can also prove to be of great interest in the studies relating induced mechanical stresses and valvular disease in patients with CBAV.

Irrespective of these limitations, this study provides an insight on the effect of geometrical variations of the CBAV affects the valve performance. Our studies on simulated idealized geometry of the CBAV indicate that specific geometrical types induce regions of relatively high stress concentrations in the leaflets. We believe this to be an important initial step in our understanding of the mechanical effects of valve geometry on leaflet calcification with CBAV. Further studies on the structural analysis of patient-specific CBAV geometry and clinical follow up with the patients to correlate the regions of stress concentration with calcification in these patients will be valuable in stratifying patients with risk of valve failure.

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REFERENCES

1. Balguid A, Driessen NJ, Mol A, Schmitz JP, Verheyen F, Bouten CV, Baaijens FP. Stress related collagen ultrastructure in human aortic valves—implications for tissue engineering. *J. Biomech.* 2008; 41(12):2612–2617. [PubMed: 18701107]
2. Cataloglu A, Clark RE, Gould PL. Stress analysis of aortic valve leaflets with smoothed geometrical data. *J. Biomech.* 1977; 10(3):153–158. [PubMed: 858719]
3. Chandran K. Role of computational simulations in heart valve dynamics and design of valvular prostheses. *Cardiovasc. Eng. Technol.* 2010; 1(1):18–38. [PubMed: 20606715]
4. De Hart J, Peters GW, Schreurs PJ, Baaijens FP. A three-dimensional computational analysis of fluid-structure interaction in the aortic valve. *J. Biomech.* 2003; 36(1):103–112. [PubMed: 12485644]
5. Fedak PW, Verma S, David TE, Leask RL, Weisel RD, Butany J. Clinical and pathophysiological implications of a bicuspid aortic valve. *Circulation.* 2002; 106(8):900–904. [PubMed: 12186790]
6. Fung, YC. *Biomechanics: Mechanical Properties of Living Tissues.* New York: Springer-Verlag; 1993.
7. Gnyaneshwar R, Kumar RK, Balakrishnan KR. Dynamic analysis of the aortic valve using a finite element model. *Ann. Thorac. Surg.* 2002; 73(4):1122–1129. [PubMed: 11996252]
8. Gorman JH III, Gupta KB, Streicher JT, Gorman RC, Jackson BM, Ratcliffe MB, Bogen DK, Edmunds LH Jr. Dynamic three-dimensional imaging of the mitral valve and left ventricle by rapid sonomicrometry array localization. *J. Thorac. Cardiovasc. Surg.* 1996; 112(3):712–726. [PubMed: 8800160]
9. Huang, HY. *Micromechanical Simulations of Heart Valve Tissues.* Pittsburgh: University of Pittsburgh; 2005.
10. Huang HY, Liao J, Sacks MS. In situ deformation of the aortic valve interstitial cell nucleus under diastolic loading. *J. Biomech. Eng.* 2007; 129(6):880–889. [PubMed: 18067392]
11. Kim H, Chandran KB, Sacks MS, Lu J. An experimentally derived stress resultant shell model for heart valve dynamic simulations. *Ann. Biomed. Eng.* 2007; 35(1):30–44. [PubMed: 17089074]
12. Kim H, Lu J, Sacks MS, Chandran KB. Dynamic simulation pericardial bioprosthetic heart valve function. *J. Biomech. Eng.* 2006; 128(5):717–724. [PubMed: 16995758]
13. Kim H, Lu J, Sacks MS, Chandran KB. Dynamic simulation of bioprosthetic heart valves using a stress resultant shell model. *Ann. Biomed. Eng.* 2008; 36(2):262–275. [PubMed: 18046648]
14. Robicsek F, Thubrikar MJ, Cook JW, Fowler B. The congenitally bicuspid aortic valve: how does it function? Why does it fail? *Ann. Thorac. Surg.* 2004; 77(1):177–185. [PubMed: 14726058]
15. Sievers HH, Schmidtke C. A classification system for the bicuspid aortic valve from 304 surgical specimens. *J. Thorac. Cardiovasc. Surg.* 2007; 133(5):1226–1233. [PubMed: 17467434]
16. Sun W, Abad A, Sacks MS. Simulated bioprosthetic heart valve deformation under quasi-static loading. *J. Biomech. Eng.* 2005; 127(6):905–914. [PubMed: 16438226]
17. Thubrikar MJ, Aouad J, Nolan SP. Patterns of calcific deposits in operatively excised stenotic or purely regurgitant aortic valves and their relation to mechanical stress. *Am. J. Cardiol.* 1986; 58(3):304–308. [PubMed: 3739919]
18. Ward C. Clinical significance of the bicuspid aortic valve. *Heart.* 2000; 83(1):81–85. [PubMed: 10618341]
19. Weinberg EJ, Kaazempur Mofrad MR. A multiscale computational comparison of the bicuspid and tricuspid aortic valves in relation to calcific aortic stenosis. *J. Biomech.* 2008; 41(16):3482–3487. [PubMed: 18996528]

FIGURE 1. Uniaxial stress–strain curves in the circumferential and radial directions derived from the material model employed in the simulation.

FIGURE 2.

(a) Aortic valve models created and used in this study: tri-leaflet valve and CBAV: Type I, three malformed leaflets, half of a purely bicuspid valve and two leaflets smaller in size than normal; Type II, purely bicuspid; Type III, single fusion of leaflets lacking raphe; and Type IV, most common CBAV morphology with raphe included through the addition of stiffer material to the section highlighted in red; (b) leaflet height and radius for the leaflets are shown on the left. Three-dimensional representation of the model used for normal tri-leaflet geometry is shown on the right.

FIGURE 3.

Pressure load specified on the ventricular surface of the leaflet. Values represent the difference between ventricular and aortic pressure throughout the complete cardiac cycle of normal healthy humans. The leaflet stress distribution is compared for the various simulated geometries with the valve in the fully open position and the fully closed position as indicated in the plot.

FIGURE 4.

In-plane principal stress contours for the tri-leaflet aortic valve and simulated CBAVs with the leaflets in the fully open configuration approximately 90 ms into the cardiac cycle. Units for stress are in Pascal (N/m^2). The CBAV leaflets shown were chosen for having the largest stress magnitude.

FIGURE 5.

In-plane principal stress contours for the tri-leaflet aortic valve and simulated CBAVs with the leaflets in the fully closed configuration approximately 300 ms into the cardiac cycle. Units for stress are in Pascal (N/m^2). The CBAV leaflets shown were chosen for having the largest stress magnitude.

FIGURE 6.

The comparison of the valve orifice area in the fully open position for the tri-leaflet valve and CBAV: Type I; Type II; and Type III. The addition of a raphe did not affect the fully open geometry and hence valve Type IV has the same open orifice as that of Type III.

TABLE 1

Material constants employed in the fully anisotropic Fung constitutive model to the components of the modified Green strain tensor.

$b_{1111} = 49.558$	$b_{1313} = 24.779$	$b_{3313} = 24.779$
$b_{2222} = 5.2871$	$b_{2323} = 8.02$	$b_{1213} = 8.016$
$b_{3333} = 24.779$	$b_{1112} = -0.004$	$b_{1123} = -0.002$
$b_{1122} = -3.124$	$b_{2212} = -0.02$	$b_{2223} = -0.01$
$b_{1133} = 24.779$	$b_{3312} = -0.002$	$b_{3323} = -0.002$
$b_{2233} = -1.56$	$b_{1113} = 24.779$	$b_{1223} = 8.016$
$b_{1212} = 16.031$	$b_{2213} = -1.562$	$b_{1323} = -0.002$

TABLE 2

Maximum in-plane principal stress (MPa) at the instant when the valve is fully closed.

Material constants	Tri-leaflet	Type I	Type II	Type III	Type IV	$\Delta\%$
Normal	0.199	0.408	0.281	0.323	0.519	161
Increase	0.197	0.404	0.289	0.332	0.329	105
Decrease	0.202	0.368	0.272	0.307	0.302	82
$\Delta\%$	2	10	3	5	42	

$\Delta\%$: Maximum percent change in principal stress with change in material constants ($\pm 25\%$) or valve geometry compared to that of the normal tri-leaflet valve.

TABLE 3

Maximum in-plane logarithmic strain at the instant when the valve is fully closed.

Material constants	Tri-leaflet	Type I	Type II	Type III	Type IV	$\Delta\%$
Normal	0.157	0.185	0.181	0.232	0.255	62
Increase	0.154	0.174	0.171	0.244	0.242	58
Decrease	0.16	0.197	0.193	0.262	0.259	64
$\Delta\%$	2	7	7	13	6	

$\Delta\%$: Maximum percent change strain with change in material constants ($\pm 25\%$) or valve geometry compared to that of the normal tri-leaflet valve.