Investigating the Effect of Leaflet Quantity on Aortic Valve Stress-Strain Distribution

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As one of the greatest challenges faced by modern medicine, cardiovascular disease is the world's leading cause of death. Complications including regurgitation, stenosis, and atresia continue to perplex researchers seeking to develop new treatments. In particularly challenging forms of illness, patients are born with an abnormal number of aortic valve leaflets, deviating from the standard tricuspid valve. Current treatments for this include implantation of bioprosthetic and mechanical valves, but these options are lacking because they often require frequent reimplantation and continual monitoring. The goal of this investigation is to aid in the development of implantable heart valves by addressing the effect of varying the number of leaflets, which may lead to more effective tissue engineered aortic valves. Importing models generated in Solidworks, simulations were run using COMSOL Multiphysics in order to analyze the stress and strain distributions of heart valve models with varying leaflet numbers. Our analysis shows that stresses and strains tend to be localized along the leaflet interfaces, and maximum stress and strain values tend to decrease as the leaflet number increases. Some possible explanations for these findings include the impact of leaflet contact areas and degree of displacement in the aortic wall. Future studies may improve upon the models and simulations by developing more accurate models of aortic leaflets by incorporating more true-to-nature geometric parameters.

Aortic Valve | Leaflet Geometries | Stress-Strain Distributions | ...

Introduction

Cardiovascular disease is the number one cause of death in the world (1). Consequently, heart valve disease and valve replacement failure present a growing problem. The four heart valves (aortic, mitral, pulmonary, and tricuspid) play a crucial role in regulating blood flow during each heartbeat. Complications with heart valves can be classified into three main categories: regurgitation, the backflow resulting from an incompletely shut valve; stenosis, the thickening or fusion of valve leaflets; and atresia, the lack of an opening through which blood may pass. Annually, there are approximately 90,000 valve replacement implants in the United States and 280,000 worldwide with half of the replacements being mechanical valves and the other half being bioprosthetic valves (2). These implants are essential medical procedures, but have several constraints. Mechanical valves induce a high risk of thromboembolism and require the patient to take anticoagulant medication, and bioprosthetic valves tend to degrade rapidly in the body (3). Patient-prosthesis mismatch often occurs when patients are given suboptimal valve replacements given their current needs and requirements (2). Furthermore, neither valve replacement is capable of growing over a patient's lifetime, and surgical intervention is often required, especially for patients with congenital heart valve complications.

Heart valves can be anatomically categorized into two geometries: semilunar and atrioventricular. The semilunar valves (aortic and pulmonary) prevent backflow into the ventricles and have three, similarly sized leaflets. The atrioventricular valves (mitral and tricuspid) prevent backflow from the ventricles into the atria and have more complex geometries. The mitral valve has two leaflets, and both the mitral and tricuspid valves are asymmetrical with chord-like structures called chordae tendineae that connect the leaflets to the ventricular inner walls. The mitral and aortic valves tend to require more surgical intervention and replacement than the do the tricuspid and pulmonary valves (4). Furthermore, about 1% of the population is born with a bicuspid aortic valve, which is associated with higher risks of regurgitation, aortic aneurysm, and stenosis (5). This presents the interesting question of how differing the number of leaflets may affect valve performance. Previous research has shown that various geometric factors like valve diameter, stent height, and nonuniform leaflet thickness play a significant role in overall stress distributions (6), but the primary purpose of our investigation was to confirm the validity of the claim that tricuspid valves contain the optimal leaflet number and to investigate the consequences which varying leaflet number has on stress distributions across the valves.

The basic biomechanics of heart valve leaflets can be broken into three physical loading models: flexure, shear, and tension. Flexure occurs when the valve opens, shear occurs when blood flows past the leaflets, and tension occurs when the valve shuts to prevent retrograde flow (4). Shear and flexure have proven

Significance Statement

Heart valves play a crucial role in controlling unidirectional blood flow for heartbeats. The aortic valve is especially important considering that it is the last valve before oxygen-rich blood reaches the body. Aortic valves are normally tricuspid with three, roughly symmetrical leaflets. However, about 1% of the population is born with a bicuspid aortic valve (BAV) and about 0.1% with a quadricuspid aortic valve (QAV). BAV and QAV patients often experience higher rates of heart valve disease and other associated complications. This raises the question on the effect number of leaflets affects valve on performance. Traditional mechanical or bio-prosthetic valve replacements are bicuspid or tricuspid, but perhaps exploring quadricuspid geometries may be valuable. Our research hopes to explore how varying the number of leaflets affects the stress-strain distributions of aortic heart valves under diastole, with hopes of introducing new options and suggestions for improving aortic valve replacements.

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Table 1. Parameters for Model Geometries

Leaflet radius	13.5 mm(7)
Leaflet thickness	1.5mm(8, 9)
Leaflet height	15 mm(10)
Aortic annulus radius	12.2 mm (11)
Vessel wall thickness	1.87mm(8)
Poisson's ratio	1.87mm(12)
Young's modulus (radial)	1.87mm(13)
Transvalvular pressure	30 mmHg (14)

Parameters used in geometric and computational modeling of heart valve.

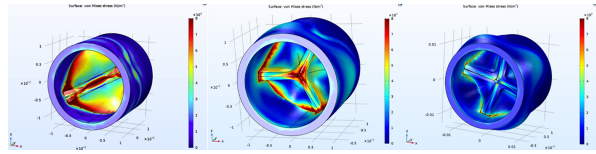

Fig. 2. Distribution of Stress in Aortic Valve COMSOL Models (left to right: bicuspid, tricuspid, quadricuspid)

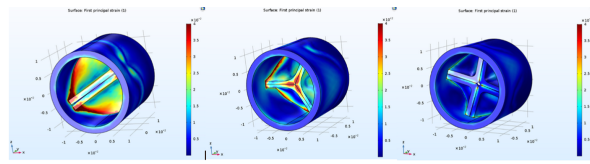

Fig. 3. Distribution of Strain in Aortic Valve COMSOL Models (left to right: bicuspid, tricuspid, quadricuspid)

difficult to model due to the introduction of fluid mechanics, but when the leaflets are closed, the system becomes easier to model. Realistically, human valve leaflets are nonlinear and anisotropic due to an intricate network of collagen fibers, but we pursued a LEHI (linear, elastic, homogeneous, isotropic) model given the scope of the BioE 102 curriculum and the complexity of nonlinear, anisotropic models. Given these considerations and the frequency of aortic valve failure, our investigation varies the number of leaflets to understand its effect on aortic heart valve performance in terms of stress and strain distributions.

Methods. Each heart valve was generated in Solidworks before being imported into COMSOL Multiphysics. First a guide curve, roughly cup-shaped to mimic leaflet curvature, was created. Planes were then generated along the guide curve, and boomerang shapes were sketched in each plane. The loft feature was used to connect each of these boomerang cross-sections and thus create a single leaflet, which was then replicated in a circular array around a central axis. This circular array of leaflets was cut to the proper size before a vessel wall was extruded around this to form the completed heart valve. To change the number of leaflets, the angle of the boomerang shape was changed (120° for tricuspid, 90° for quadricuspid, etc.) in addition to the number replicated in the circular array. To match the geometry of heart valves, values from literature were used to determine the size of the overall heart valve, thickness of each leaflet, and other similar details as summarized in Table 1.

In COMSOL, a stationary study in solid mechanics was initiated, and the Solidworks part file for each heart valve (as shown in Figure 1) was imported into the geometry node. Constants for loads and material behavior were found from literature and are summarized in Table 1. Using a simplified

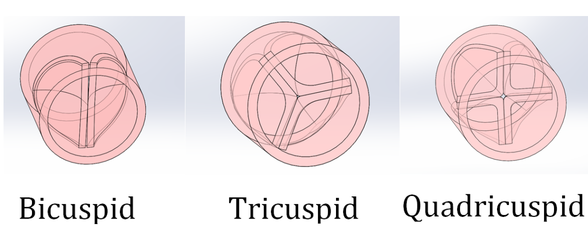

Fig. 1. Aortic Valve Solidworks Models

Using dimensions from literature, heart valves with varying numbers of leaflets were generated in Solidworks and imported into COMSOL for solid mechanics analysis.

isotropic assumption about the heart valve's behavior, Young's modulus and Poisson's ratio from literature were input in the materials node. While a typical heart valve has separate circumferential and radial Young's moduli, the radial Young's modulus was input because the heart leaflets are primarily under radial stress. To keep the valve in place for the applied load, the vessel openings parallel to the annulus plane were designated as the fixed constraints. A transvalvular pressure was input on the interior of the leaflets' cup shape and adjacent vessel wall to mimic diastolic conditions. Finally, a free tetrahedral mesh was formed and refined once before the stationary study was computed.

Results and Discussion. The stress and strain distributions resulting from applying the COMSOL study under the conditions described are shown across our models in Figures 2 and 3, respectively. Shown consistently across each study is an apparent localization of stress on the leaflet edges and on the interface between leaflets. The magnitude of the stresses applied onto each model are found to be decreasing with increasing leaflet number, as shown in Figure 2. These qualities hold true across our strain distributions (Figure 3), as expected due to the dependence of strain on stress.

As described in Figures 2 and 4, the maximal stress, as well as the general magnitude of the stress distribution, decreases as the number of leaflets increases. We posit some explanations for our model behavior: the central intersection of the leaflets has a greater volume with increasing leaflets; the total length of the leaflet interface increases with the number of leaflets; and the smaller distance between leaflets in valves with a high number of leaflets allows for more exaggerated bulging.

The thickness of the leaflets in each of our models is fixed at 1.5mm. However, because of this, the volume at the central intersection is significantly larger as the number of leaflets

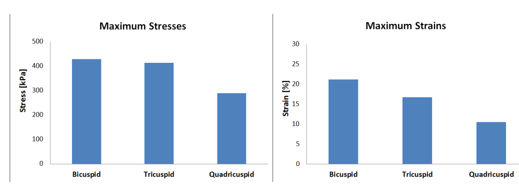

Fig. 4. Plots of Max Stresses and Strains in Aortic Valve Models

increases. This may allow stress at the center of the valves to be more widely distributed, resulting in lower maximal and overall stresses with increasing number of leaflets.

The length of contact interfaces (the interface of the leaflets and the aortic wall) similarly increases as the number of leaflets increases. Noticing again that the highest stress values occur at interfaces of the model, and that large stress values propagate from there, it is plausible that having more intersection area leads to a larger distribution of force and thus a decrease in the magnitude of the stress distribution. The length of contact interfaces (the interface of the leaflets and the aortic wall) similarly increases with the number of leaflets. It is plausible that having more intersection area distributes the forces throughout a larger area, thereby decreasing the stress levels.

Probably most explanatory of our results is the change to the nature of the interface of the leaflet with the aortic wall as the number of leaflets increases. Observing the model, the degree of "bulging" (displacement in the aortic wall) increases as the number of leaflets increases, since the number of bulges increases. This causes the interface of the leaflet and the aortic wall to occur as a shallower angle. Because much of the high stress occurs as a result of the abrupt changes, it is not unexpected that a shallower angle results in lower overall stress.

It is unclear how descriptive our models are of physical aortic valves, especially considering the fact that the maximum stresses in the bicuspid and tricuspid valve models, but not the quadricuspid model, were above the maximum level that heart tissue can withstand (400 kPa) (13). Our geometric parameters are based on an approximation of the aortic valve, and has leaflets which are thicker than the generally reported value in order to generate simulations that compute without errors (10, 15). Furthermore, we made LEHI assumptions about the physical properties of the valve for model simplicity, but it is well known that the radial and circumferential Young's Moduli have significant differences, and stress-strain relationships are nonlinear. It is also likely that the interfaces of the valves are quite complex, possessing different material and geometric properties than the simplified, abrupt interface that we modeled.

The combination of these differences indicates that the particular values that we obtained for maximal strain, which is greater than the values determined for maximal strain before rupture (15), could be subject to differences in the physical body. In addition, our models did not incorporate physiological features of the aortic valve whose functions are to decrease transvalvular pressure, such as the sinus of valsalva (9). Future models which include these features will help to develop more accurate analyses of the aortic valve geometry.

Conclusion. Our COMSOL models and studies suggest that varying the number of leaflets does affect the stress and strain distributions in the heart valve: specifically, the maximal stress and the overall magnitude of the stress-strain distributions decrease as the number of leaflets increases. This result proposes that quadricuspid valves might be more effective compared to tricuspid and bicuspid geometries, but this in turn contrasts with the overwhelming presence of tricuspid aortic valves compared to the rarity of bicuspid and quadricuspid valves.

In addition to discrepancy between our models' geometries compared to physical aortic valves as discussed earlier, this investigation's oversight of the body's response to stress may reveal the logic of the apparent favoring of tricuspid geometries. In recorded cases of bicuspid aortic valve anomalies, the incidence of aortic stenosis (death resulting from thickening of the heart valves) occurs with much higher incidence (16). Our COMSOL study suggests that the bicuspid valve geometry experiences higher stresses compared to the tricuspid and quadricuspid geometries. The body likely reacts to these higher stresses with a geometric response: namely, reacting to the higher levels by thickening the leaflets and ultimately leading to aortic stenosis. Thickening the leaflets increases the volume at the central intersection, allowing force to be more widely distributed to decrease the stress levels. Even our COMSOL simulations reflect this; the quadricuspid valve model had higher volume at the central intersection, thereby distributing stress more widely.

Patients with quadricuspid aortic valves only make up about .01% of the population, but in the limited cases, it has been found that these individuals often experience aortic regurgitation (17). It is posited that this aortic regurgitation is a result of prolapse, which is often a symptom of floppy heart valves. As reflected in our study, stress values are lower for quadricuspid valves in comparison with standard tricuspid ones, which likely leads to reduced development of the heart valves through dystrophy, the opposite of the thickening that bicuspid valves experience. These thinner, underdeveloped leaflets result in heart valves that hold together less than optimally.

No model is without its faults and simplifications, and our investigation is no exception. Nonetheless, our results indicate that the geometries of aortic valves can be explored to better understand the form versus function relationship for improving the design of engineered heart valves.

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1. Kochanek, K. D. MSLXJTVB (2016) Deaths: final data for 2014. national vital statistics reports: from the centers for disease control and prevention, national center for health statistics. *National Vital Statistics System* 65(4):1.
2. Pibarot, P. DJG (2009) Prosthetic heart valves selection of the optimal prosthesis and long-term management. *Circulation* 119(7):1034–1048.
3. Kidane, A.G. BGCPSSBPEMSA (2009) Current developments and future prospects for heart valve replacement therapy. *Journal of Biomedical Materials Research Part B Applied Biomaterials* 88:290–303.
4. Sacks, M. S. MWDSDE (2009) On the biomechanics of heart valve function. *Journal of Biomechanics* 42(12):1804–1824.
5. Lewin, M. B. OCM (2005) The bicuspid aortic valve adverse outcomes from infancy to old age. *Circulation* 111(7):832–834.
6. Luo, X. Y. LWGLJ (2003) Geometrical stress-reducing factors in the anisotropic porcine heart valves. *Journal of Biomechanical Engineering* 125(5):735–744.
7. Thubrikar M (1990) *The Aortic Valve.* (CRC-Press).
8. Sahasakul, Y. EWDNJMTAJ (1988) Age-related changes in aortic and mitral valve thickness: Implications for two-dimensional echocardiography based on an autopsy study of 200 normal human hearts. *The American Journal of Cardiology* 7(1):424–430.
9. Beck, A. TMJRF (2001) Stress analysis of the aortic valve with and without the sinuses of valsalva. *The Journal of Heart Valve Disease* 1(1):1–11.
10. Loerakker, S. AGOCWBFP (2013) Effects of valve geometry and tissue anisotropy on the radial stretch and coaptation area of tissue-engineered heart valve. *Journal of biomechanics* 46(11):1792–1800.
11. Book GA (2010) Modeling of aortic valve anatomic geometry from clinical multi detector-row computed tomography images.
12. Chen, E. J. NJJWKOWD (1993) Structure and anatomy of the aortic root. *IEEE Transactions on ultrasonics, ferroelectrics, and frequency control* 43(1):191–194.
13. Balguid, A. DNJMASJPVFBCVBFP (2008) Stress related collagen ultrastructure in human aortic valves—implications for tissue engineering. *Journal of biomechanics* 41(12):2612–2617.
14. Brogan, W. C. GPALRAHLD (1993) Structure and anatomy of the aortic root. *European Heart Journal-Cardiovascular Imaging* 21(7):1657–1660.