Biomechanical analysis of novel leaflet geometries for bioprosthetic valves

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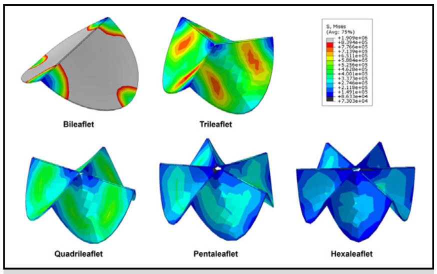
ABSTRACT

Objectives: Although bioprosthetic valves have excellent hemodynamic properties and can eliminate the need for lifelong anticoagulation therapy, these devices are associated with high rates of reoperation and limited durability. Although there are many distinct bioprosthesis designs, all bioprosthetic valves have historically featured a trileaflet pattern. This in silico study examines the biomechanical effect of modulating the number of leaflets in a bioprosthetic valve.

Methods: Bioprosthetic valves with 2 to 6 leaflets were designed in Fusion 360 using quadratic spline geometry. Leaflets were modeled with standard mechanical parameters for fixed bovine pericardial tissue. A mesh of each design was structurally evaluated using finite element analysis software Abaqus CAE. Maximum von Mises stresses during valve closure were assessed for each leaflet geometry in both the aortic and mitral position.

Results: Computational analysis demonstrated that increasing the number of leaflets is associated with reduction in leaflet stresses. Compared with the standard trileaflet design, a quadrileaflet pattern reduces leaflet maximum von Mises stresses by 36% in the aortic position and 38% in the mitral position. Maximum stress was inversely proportional to the square of the leaflet quantity. Surface area increased linearly and central leakage increased quadratically with leaflet quantity.

Conclusions: A quadrileaflet pattern was found to reduce leaflet stresses while limiting increases in central leakage and surface area. These findings suggest that modulating the number of leaflets can allow for optimization of the current bioprosthetic valve design, which may translate to more durable valve replacement bioprostheses. (JTCVS Open 2023;14:77-86)



Increasing the number of leaflets in a bioprosthetic valve decreases von Mises stresses.

CENTRAL MESSAGE

Computational analysis demonstrated that increasing the number of leaflets in a bioprosthetic valve is associated with reductions in leaflet stresses.

PERSPECTIVE

Modulating the number of leaflets can allow for optimization of current bioprosthetic valve designs, which may translate to more durable valve replacement bioprostheses.

More than 200,000 heart valve replacement surgeries are performed annually worldwide, with a predicted increase to 850,000 by 2050.[1] In patients with significant valve insufficiency or stenosis in whom successful surgical repair is not feasible, valve replacement is offered with either mechanical or bioprosthetic valves.[2] Xenografts account for most of bioprosthetic heart valves and are sourced from bovine or porcine pericardia or porcine aortic valves that have been processed through fixative treatments. Bioprosthetic valves are either stented or stentless and are designed with 3 leaflets to mimic the anatomy of the native aortic valve.[3] These devices largely eliminate the need for lifelong anticoagulation therapy, due to their lower thrombotic risk compared with mechanical valves.[4] As a result, patients with bioprosthetic valves have a significantly reduced risk of bleeding. In addition, bioprosthetic valves demonstrate excellent hemodynamic properties similar to those of native valves.[1]

Along with these advantages, however, bioprosthetic valves are associated with greater reoperation rates due to their limited durability.[5] Structural valve degeneration (SVD) is a major cause of limited durability and consists of any intrinsic permanent damage to the bioprosthesis, such as calcification, leaflet fibrosis, and tears.[5,6] Although the mechanisms underlying SVD are not completely understood, studies have shown that calcification mainly develops in areas with high mechanical stress.[1]

Even with the advent of transcatheter valve replacements, the risks of SVD still prevail due to additional mechanical

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Abbreviations and Acronyms
3D = 3-dimensional
FEA = finite element analysis
SVD = structural valve degeneration

stresses from valve crimping, balloon expansion, and leaflet-to-frame attachment.[6] Newer generations of bioprosthetic valves have largely iterated on the delivery system, tissue anti-calcification fixation process, and anchoring for the mitral position, with little to no advancement in leaflet geometry to mitigate SVD.[7] Computational studies have previously been conducted to optimize leaflet design, but the recommended geometric adjustments have been minor. Outside of bioprosthetic valve development, studies have reported substantial biomechanical effects of congenital diseases, such as unicuspid, bicuspid, and quadricuspid aortic valves,[8] as well as with mechanical valve designs, such as monoleaflet, bileaflet, and trileaflet mechanical valves.[9] Despite this diversity, the only commercially available bioprosthetic valve geometry is the standard trileaflet design. Furthermore, this design feature has not changed since the first aortic valve heterograft replacement in 1965.[10] To our knowledge, there is no study reported in the literature that explores the impact of leaflet quantity variation in bioprosthetic valves. This study computationally examines the biomechanical effects of modulating the total leaflets in bioprosthetic valves implanted in the aortic and mitral positions.

METHODS

No human or animal studies were carried out by the authors for this article. To create the geometric models, valves with 2 to 6 leaflets were modeled using 3-dimensional (3D) computer-aided design software Fusion 360 (Autodesk). For each leaflet geometry, 2 valves were designed: one aortic bioprosthesis with a circular diameter of 23 mm and one mitral bioprosthesis with a circular diameter of 29 mm. For a design with *n* number of leaflets, an extruded cylindrical body was divided into *n* sections. A surface patch was then projected onto the segmented cylindrical body using quadratic spline geometry and then patterned around the central axis to form a valve with *n* leaflets (Figure 1, *A*). The nadir to coaptation height was kept constant. The leaflets were assumed to have a uniform thickness of 0.25 mm. Total leaflet surface area and central leakage area were recorded.

A mesh of each 3D computer-aided design was structurally evaluated using finite element analysis (FEA) software Abaqus CAE (Abaqus) (Appendix E1). A static analysis was conducted to simulate valve closure with encastre boundary conditions and applied uniform pressures on the leaflets (Figure 1, *B*). To evaluate the designed aortic bioprostheses, a pressure differential of 95 mmHg was applied to the aortic side of the valve.[11] To evaluate the designed mitral bioprostheses, a pressure differential of 120 mmHg was applied to the ventricular side. These values were derived from previous ex vivo testing of bioprosthetic valves in our left heart simulator flow loop system. As our model used static pressurization with encastre boundary conditions, the method is equally extensible to simulations in both the mitral and aortic positions by modulating the hemodynamic parameters to either systolic or diastolic pressures, respectively, which would effectively invert the hemodynamic orientation relative to the left ventricle. The leaflets were modeled as fixed bovine pericardial tissue, with elastic modulus of 8 MPa, density of 1100 kg/m^3, and Poisson ratio of 0.45.[11-13] Maximum von Mises stresses were assessed for each valve model, and the maximum stresses, surface area, and central leakage data were all plotted against leaflet quantities and regression models were fit to the data.

RESULTS

As the number of leaflets in our bioprosthetic valve model increased, we observed a decrease in both principal and von Mises stresses (Figures 2 and 3). The quadrileaflet valve provided a reduction in maximum principal stresses by 34% in the aortic position and 42% in the mitral position. Adding more leaflets further decreased the maximum stresses and regions of high stress concentrations. Compared with the trileaflet aortic valve, adding leaflets decreased the maximum von Mises stresses: 36% reduction with the quadrileaflet pattern, 48% reduction with pentaleaflet, and 55% reduction with hexaleaflet. When compared with the trileaflet mitral valve, the reduction in

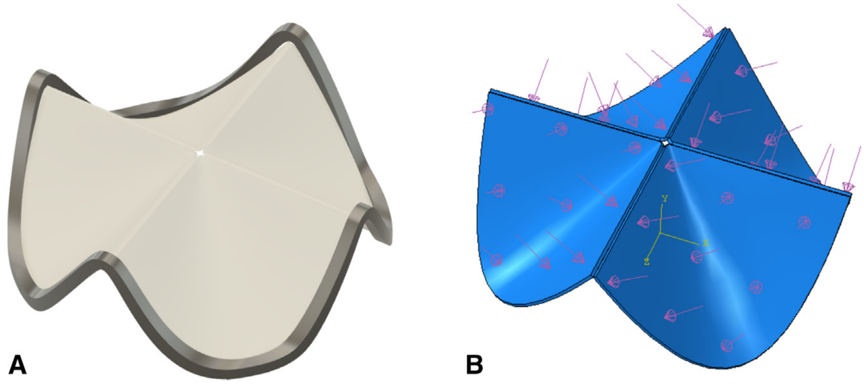

FIGURE 1. Computational model of quadrileaflet (4 leaflets) bioprosthetic valve. A, Each valve was designed in Fusion 360 using quadratic spline geometry. The diameter was set to 23 mm for aortic valves and 29 mm for mitral valves. B, Finite element analysis was conducted for each configuration in Abaqus CAE. To structurally evaluate the designed aortic bioprostheses, a uniform pressure differential of 95 mmHg was applied to the "aortic side" of the valve. To evaluate the designed mitral bioprostheses, a uniform pressure differential of 120 mmHg was applied to the "ventricular side" of the valve.

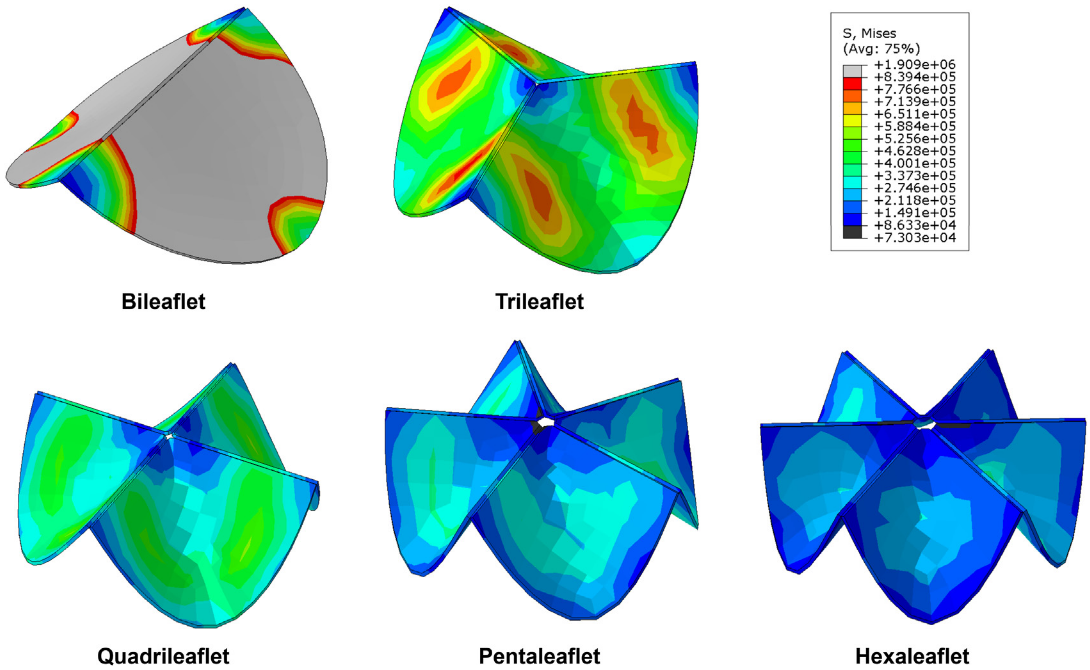

FIGURE 2. Increasing the number of leaflets of a bioprosthetic valve in an aortic position decreases von Mises stresses (in Pascals) on each leaflet during mid-diastole.

maximum von Mises stresses was 38% with the quadrileaflet pattern, 50% with pentaleaflet, and 60% with hexaleaflet. In contrast, a bileaflet valve substantially increased the maximum von Mises stresses on the leaflet: 127% increase in the aortic position and 87% increase in the mitral position.

With data captured from the 3D designs and FEA, regression models were created to characterize the design outputs

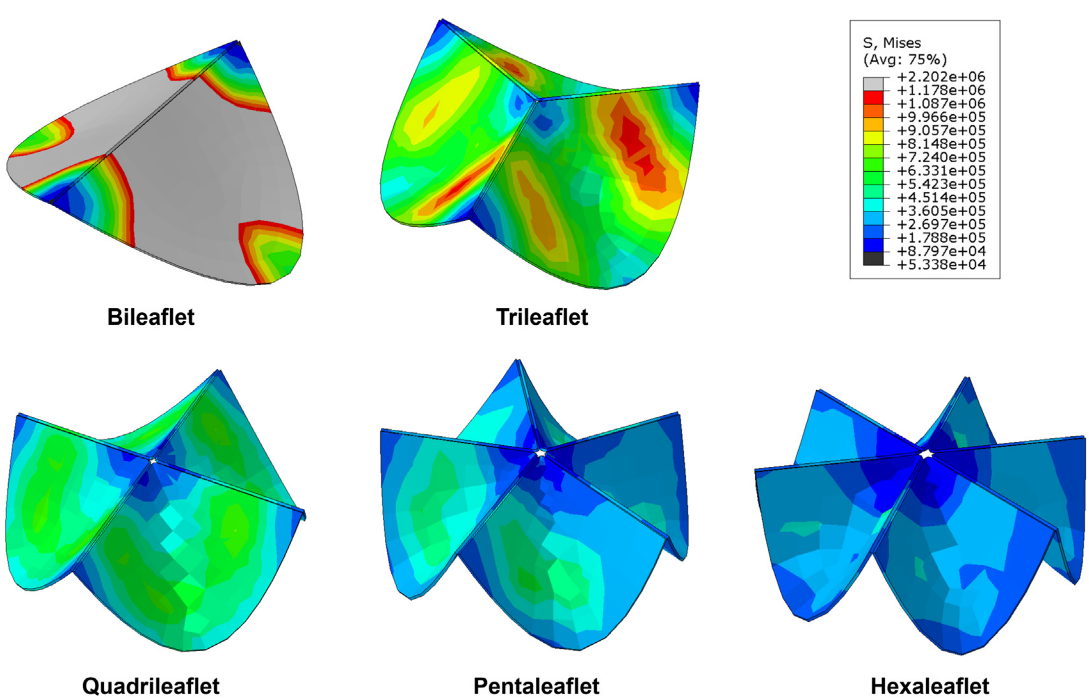

FIGURE 3. Increasing the number of leaflets of a bioprosthetic valve in a mitral position decreases von Mises stresses (in Pascals) on each leaflet during mid-systole.

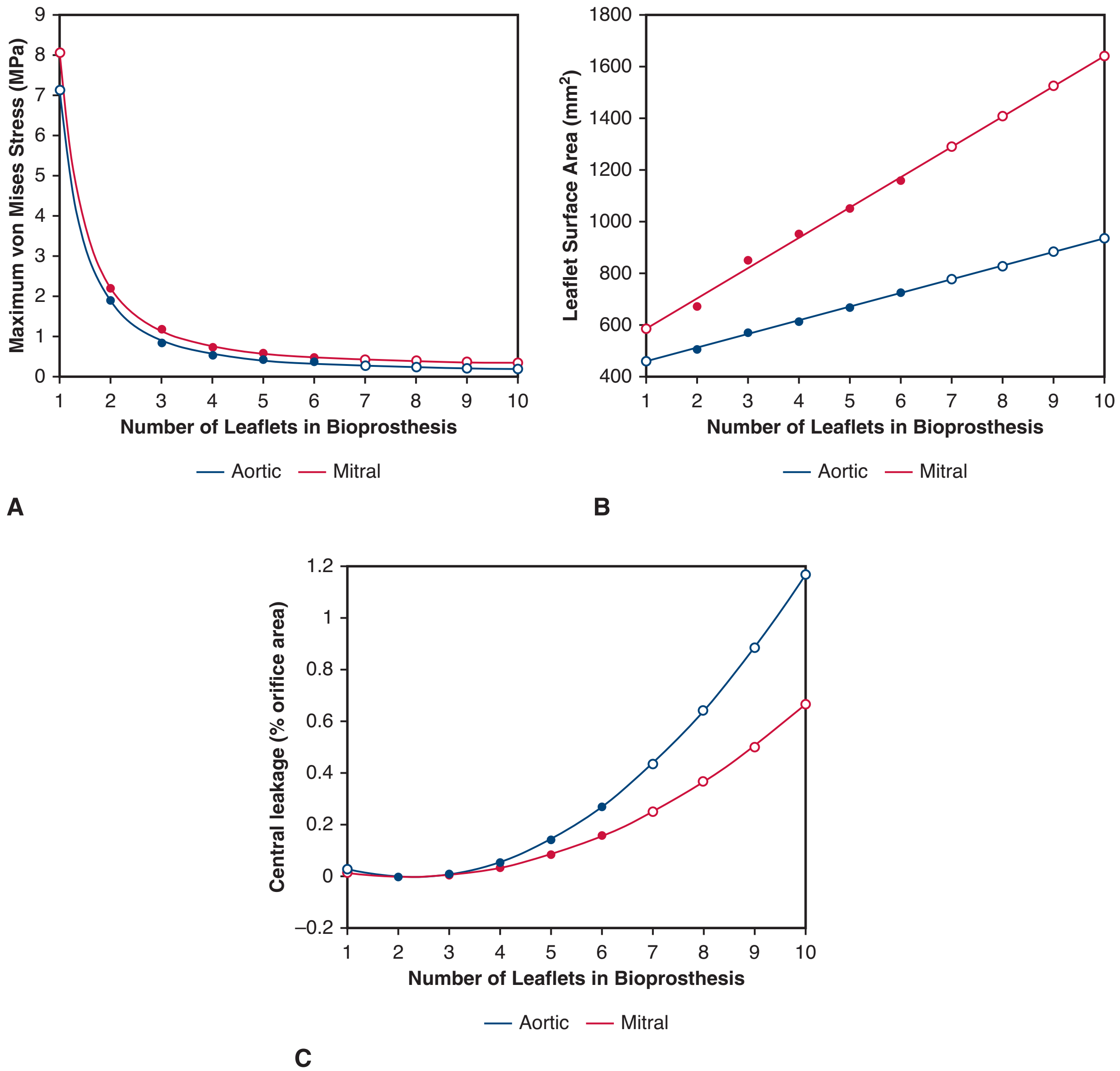

FIGURE 4. Valves with 2 to 6 leaflets were structurally evaluated, and a regression model was created to derive biomechanical metrics for valves ranging from 1 to 10 leaflets. A, For both the aortic and mitral positions, the maximum von Mises stresses of leaflets during valve closure decrease as leaflet quantity increases ($R^2 = 0.99$). B, The leaflet surface area, of both aortic and mitral bioprostheses, increases linearly as leaflet quantity increases ($R^2 = 0.98$). C, As the number of leaflets increases, uniform coaptation can become more complex, causing an increase in central leakage, which is quantified as a percentage of the orifice area ($R^2 = 0.99$).

of bioprostheses with total leaflets ranging from 1 to 10. Figure 4 shows the effect of modulating the leaflet quantity on maximum stress, surface area, and central leakage, fit to regression models. As seen in Figure 4, *A*, maximum stress was inversely proportional to the square of the leaflet quantity ($R^2 = 0.99$). In Figure 4, *B*, surface area was seen to increase linearly with leaflet quantity ($R^2 = 0.98$). Increasing the number of leaflets from 3 to 4 increased the total leaflet surface area by 7% in the aortic position and 12% in the mitral position.

The 3D models demonstrated that increasing the number of leaflets results in a small area of leakage in the valve center. Central leakage is reported as a ratio of the central orifice area and the total projected orifice area of the bioprosthetic valve. In Figure 4, *C*, a quadratic relationship was observed between central leakage and leaflet quantity ($R^2 = 0.99$). With the trileaflet model, 0.01% of the projected leaflet area may leak. The bileaflet valve theoretically caused no leakage due to the plane of coaptation. In contrast, increasing the number of leaflets results in an increase in central leakage. For aortic

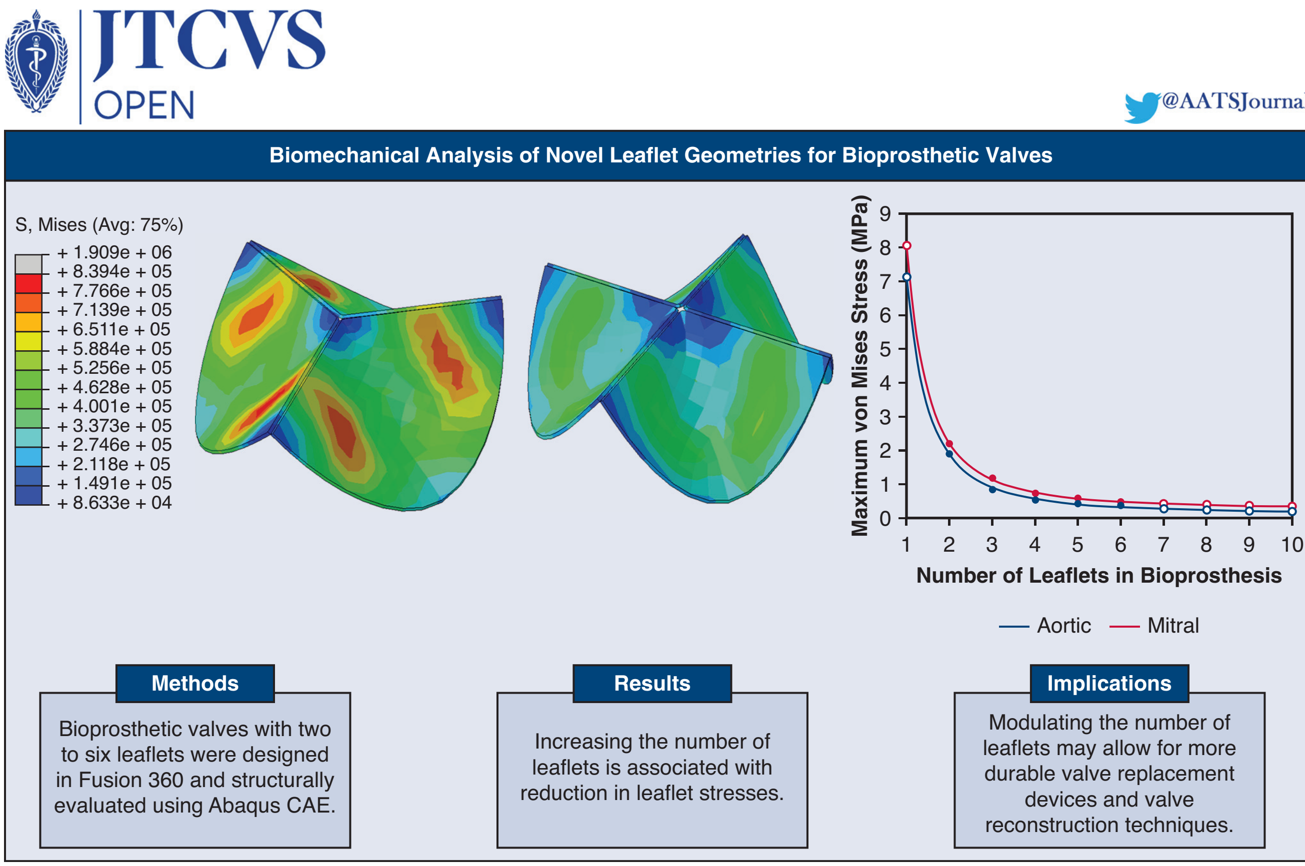

FIGURE 5. Bioprosthetic valves with 2 to 6 leaflets were designed in Fusion 360 and structurally evaluated using finite element analysis software Abaqus CAE. Computational analysis demonstrated that increasing the number of leaflets in a bioprosthetic valve is associated with reduction in leaflet stresses.

bioprostheses, central leakage was 0.06% with the quadrileaflet pattern, 0.14% with pentaleaflet, and 0.27% with hexaleaflet. For mitral bioprostheses, central leakage was 0.04% with the quadrileaflet pattern, 0.09% with pentaleaflet, and 0.16% with hexaleaflet. A visual summary of the results is provided in Figure 5.

DISCUSSION

Increasing the number of leaflets in a bioprosthetic valve was shown to substantially reduce the magnitude and distribution of von Mises stresses. The reduction in stresses may clinically translate to increased durability and improved mitigation of SVD. This is because a higher stress amplitude can cause cumulative fatigue damage over each cardiac cycle, resulting in a decrease in cycles to failure. As seen in Figures 2 and 3, modulating the leaflet quantity not only affected the maximum stress but also regions of stress concentrations, as seen in the red zones of the "belly" region of the leaflet. It is commonly accepted that any reduction in stress concentrations is advantageous, as stress concentrations within the leaflet can cause tearing or initiate calcification. However, one single time point may not be sufficient to conclude improvements in durability, due to stress softening under cyclic loading.[14] Therefore, additional studies are necessary.

Although increasing the number of leaflets may be correlated with improvements in device durability, it was also associated with a linear increase in surface area. This may correlate to increases in manufacturing costs due to the increase in xenograft raw material needed to assemble the valve, and the relationship between surface area and leaflet quantity may be useful in informing the manufacturing of novel bioprosthesis designs as well as the development of novel valve reconstruction techniques. As such, increases in leaflet quantity and surface area are subject to the supply chain availability of animal pericardial tissues as well as the increased costs associated with the added complexities of assembling multileaflet designs. These costs could be reconciled by using alternative biocompatible polymers or even bioprinted leaflets with advances in bioprinting technologies. Given the significant current infrastructure surrounding bioprosthesis manufacturing, these added marginal costs may be inconsequential when expanding multileaflet designs to be manufactured at scale. Although we are not aware of the current financial constraints of industrial bioprosthesis manufacturing systems, we would assume that the increased costs of raw material may be negligible relative to the overall costs involved in creating the production line of a new valvular product. However, bioprosthetic valves as a category of devices retain substantial before regulatory approval, indicating reduced regulatory risks and decreased costs related to development and testing.

In addition, our findings on the relationship between leaflet surface area and leaflet quantity can elucidate the relationship between stress and leaflet surface area: the maximum von Mises stress would also be inversely proportional to the square of the leaflet surface area. As the leaflet quantity increased, the pressure was distributed to a larger surface area, while the surface area per leaflet diminished. In other words, while total aggregate leaflet surface area increased, the area per leaflet decreased. Therefore, a reduction of leaflet surface area means that there is less force applied to each leaflet for a given constant pressure. Since each leaflet is smaller, we observe reduced patches of loading and improved support between adjacent leaflets, which results in reduced elastic deformation. This nonlinear relationship may be mechanically intuitive because increased contact support from the coaptation boundaries reduces the relative proportions of leaflet surface contributing to leaflet "belly" forces, as both edges of each leaflet are supported via coaptation. However, the specific quantification and even the nature of the squared inverse proportionality may not be clear without performing the FEA because leaflet surface area increases while stresses decrease, which is not intuitively true as surface area is directly proportional to increased forces given constant pressures. This exact squared inverse relationship is uniquely calculated based on our model and maximum von Mises stresses, and this phenomenon is further explained from the simulation results, which identify that the ratio of the leaflet geometries (ie, coaptation surfaces) to the relative additional forces from increased surface area plays an important role in stress reduction.

It is also worth noting that the rates of SVD are generally greater in bioprosthetic valves in the mitral position compared with those in the aortic position. This phenomenon is mechanically intuitive, as the mitral system is under significantly greater systolic pressure loads compared with the lower diastolic pressure gradients experienced in the aortic position. Likewise, our different hemodynamic parameters manifest these discrepancies in the resulting stress analyses, as these differences are visualized in Figures 2 and 3 where the Von Mises stress scales in the mitral position (Figure 3) at approximately an order of magnitude higher compared with that of the aortic position (Figure 2), which aligns well with the clinical observations of increased SVD rates in the mitral position. Regarding the percent reduction of stresses on the leaflets when comparing the quadrileaflet and trileaflet conditions for the mitral versus aortic positions, it was our finding that maximum principal stresses were reduced by 34% for quadrileaflet valves in the aortic position versus 42% for quadrileaflet valves in the mitral position. Therefore, our results indicate that the greater systolic pressure parameters exposed to the mitral valve actually amplify the stress reduction effects of additional leaflets. Although this may not be intuitive, this finding aligns well with mechanical theory, which would indicate that stress reduction is proportional to loading magnitudes.

Increasing the leaflet quantity was also found to increase the "leakage" through the center of the bioprosthetic valve. These results indicate that uniform coaptation can become increasingly complex as the number of leaflets increases. This is because central regurgitation, as defined in our model, can be functionally interpreted as a proxy for the added complexity of multileaflet coaptation that results from the additional geometric constraints of leaflet conformity to the leaflet attachment apex and nadir (ie, the leaflet attachment spanning the sinotubular junction and aortic annulus) over a reduced arc length with the same coaptation height (ie, the sinotubular junction to annulus height). This structure necessitates a greater amount of leaflet bending or deformation upon increasing leaflet number, as seen in Figures 2 and 3, which, when factoring in leaflet stiffness properties, reveals a central, aberrant orifice through which regurgitation proceeds. However, our model does not account for the more complex additional leaflet tissues that contribute to the coaptation surface, which would allow for a greater ability of the leaflet tissues to cover the central leakage of the valve upon pressurization. It is our assumption that such a leakage medium may behave differently within a fully implemented prototype, yet the issue remains that the leakage jet adds significant coaptation complexity, increasing the risk of regurgitation, for which this metric serves as proxy, and which must be accounted for during bioprosthetic design. Despite this disclaimer, unloaded valve models as well as commercial tricuspid bioprosthetic valves display a similar central leakage orifice, which would practically indicate that such an issue persists regardless of increased coaptation, creating a design challenge for multileaflet bioprosthetic development. As such, although increased durability is clinically advantageous, there is a marginal complexity cost to increasing the leaflet quantity beyond the standard trileaflet design. Therefore, a quadrileaflet pattern may translate to improved mitigation of SVD, as this design decreases leaflet stresses, while also limiting increases in central leakage.

Our work presents numerous clinical applications of interpreting multileaflet valvular biomechanics in addition to suggesting novel bioprosthetic designs. Currently, there is much interest around both aortic and mitral valve leaflet reconstruction in the operating room via the Ross procedure, pericardial patch reconstruction, and the Ozaki procedure, to name a few. For all these advanced valvular reconstruction operations, our study provides valuable initial evidence, exploring possible new factors for improved leaflet designs, that can directly inform these procedures. Specifically, our stress visualizations imply that material reinforcement of the belly region of the tricuspid valve could help better support the leaflets, potentially increasing durability of the valve. Moreover, bileaflet valves display significantly greater forces, invoking many new questions as to the optimal bicuspid aortic valve repair procedure. These new insights can provide valuable intuition to surgeons for better understanding the biomechanical underpinnings of leaflet design and valve behavior in the operating room for improved patient care.

Limitations

The 2 primary limitations of this computational study are the exclusion of leaflet coaptation and material modeling of the leaflets. Coaptation was disregarded to reduce FEA computational time correlated with generating contact surfaces.[11,13] This design decision also allowed us to exclude the variability of contact forces. Contact forces can vary because of both the leaflet number and the coaptation height, so by setting the coaptation height to zero, we can isolate it as a confounding variable. Furthermore, we chose to exclude the integrated simulation of fluid dynamics, as doing so would add significant complexity to this initial biomechanical analysis of a novel structural valve concept. As our study is far more concerned with the loaded, closed valve conditions at systolic and diastolic pressures, using combined fluid structure interaction solvers would not provide us significant additional benefit in analyzing the static structural stress distributions of closed valves.

Since our main goal was to study the effect of different leaflet geometries, rather than assess a specific prototype, we chose material properties within Abaqus that would simplify computations.[12,15] Although standard leaflet parameters were used for the FEA, the leaflets were modeled as a linearly elastic, isotropic material. Such an assumption neglects the nonlinear, viscoelastic, constitutive behavior typical of biological materials. However, there is evidence that after the fixation process of xenografts, leaflets act more like a homogenous, isotropic material with mechanical characteristics different from those of a native heart valve.[1] In addition, a previous computational analysis showed that an isotropic model had similar FEA results as an anisotropic pericardial valve, both of which were compared with a porcine valve.[16] The porcine aortic valve leaflets resulted in significant heterogeneities in the deformation patterns, with the belly region of the leaflets caving in and with increased maximum in-plane Green-Lagrange strain, while the anisotropic pericardial and isotropic valves depicted significantly more homogeneous and reduced deformation patterns, showing quite similar behavior to each other.[16] In addition, the isotropic case had greater stresses near the commissure regions of the valve, which is not seen in our study due to the encastre boundary conditions. Taken together, these results would indicate that the use of anisotropic porcine leaflet valves could accentuate the deformation characteristics in the leaflet regions of greater stress concentration, leading to a less durable outcome, which aligns well with literature favoring pericardial valves when considering clinical complications and hemodynamic profiles.[17] Despite the material conditions applied, the computed stresses of our study were comparable to results reported in other papers. The values computed for our trileaflet aortic valve closely resembled studies that used a nonlinear, orthotropic Fung-elastic constitutive model.[11,14]

In addition, as our study was limited to computational analysis of the valve, it carries with it the limitations of in silico modeling. However, computational simulation has been frequently used to study the heart in ways that would otherwise be very difficult or nearly impossible to do experimentally, particularly for structural analyses, and is often provided as critical data for regulatory approval. The benefits of these analyses are that they offer an unparalleled ability to generate unique comparisons, visualizations, and quantifications that would otherwise be very difficult to do. Specifically in our study, this technology allowed us to compare multileaflet designs while removing significant confounding variables accrued in other non-computational biomechanical analyses. However, in silico models require many unifying, homogenous assumptions, such as cellular response, parameter identification of tissue properties, and tissue interactions, which can deviate from the heterogeneous, patient-specific, in vivo physiology.[18] These limitations were accounted for by simplifying the models to study very specific mechanical aspects of the system, maximizing their benefits, yet we acknowledge that SVD is a very complex entity involving many other factors apart from leaflet stresses, and by simplifying our models, we aim future work towards better analyzing the holistic biomechanical effects of modulating leaflet quantity.

CONCLUSIONS

Increasing the number of leaflets may be associated with improved bioprosthetic valve performance, as our results show that maximum von Mises stresses are inversely proportional to the square of the leaflet quantity, whereas surface area increases linearly and central leakage increased quadratically with leaflet quantity. These findings suggest that modulating the number of leaflets can allow for optimization of current bioprosthetic valve design, which may translate to more durable valve replacement devices. Compared with the standard trileaflet designs, a quadrileaflet pattern was computationally shown to reduce leaflet maximum von Mises stresses by 36% in the aortic position and 38% in the mitral position. Combining these results, this modeling study found that a quadrileaflet pattern reduces leaflet stresses while simultaneously limiting increases in central leakage and surface area. Despite this promising set of initial data, it is probable that many other mechanical and biological factors, such as the effects of leaflet calcification and the stenotic potential of multileaflet systems, may influence valve longevity adversely in the setting of pulsatile flow. Although it is difficult to determine what factors will predominate, we strongly believe that further research focused on the holistic outcomes of novel valve designs is crucial for translating our insights to the clinic. As such, sustained further research has been focused on developing novel multileaflet valve designs, analyzing the effects of increased leaflet quantities under pulsatile flow, both ex vivo and in vivo, with particular attention focused on analyzing the leaflet dynamics and stenotic potential of these novel bioprosthesis designs. Overall, our computational study not only informs better clinical understanding of leaflet biomechanics, particularly for leaflet reconstruction or remodeling repair techniques, but also provides the basis for further advancement and optimization of bioprosthesis design, which may translate to improved durability, reduced rates of reoperation, and expanded indications of bioprostheses.

Conflict of Interest Statement

The authors reported no conflicts of interest.

The *Journal* policy requires editors and reviewers to disclose conflicts of interest and to decline handling or reviewing manuscripts for which they may have a conflict of interest. The editors and reviewers of this article have no conflicts of interest.

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Key Words: heart valve replacement, leaflet biomechanics, leaflet stresses, finite element analysis, bioprosthesis

APPENDIX E1

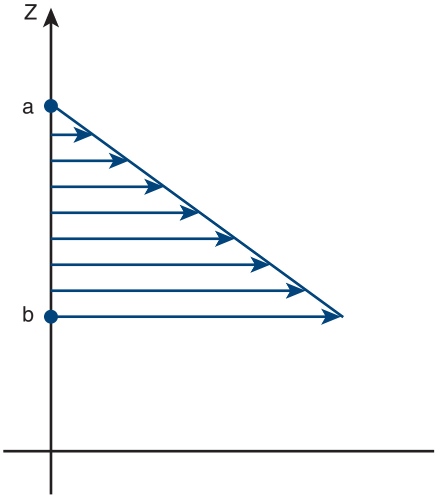
Finite Element Analysis

To create the geometric models, valves with 2 to 6 leaflets were modeled using 3-dimensional (3D) computer-aided design software Fusion 360 (Autodesk). For each leaflet geometry, 2 valves were designed: one aortic bioprosthesis with circular diameter of 23 mm and one mitral bioprosthesis with circular diameter 29 mm. For a design with n number of leaflets, an extruded cylindrical body was divided into n sections. A surface patch was then projected onto the segmented cylindrical body using quadratic spline geometry and then patterned around the central axis to form a valve with n leaflets (Figure 1, *A*). The nadir-to-coaptation height was kept constant. The leaflets were assumed to have a uniform thickness of 0.25 mm. Total leaflet surface area and central leakage area were recorded.

A mesh of each 3D computer-aided design was structurally evaluated using finite element analysis software Abaqus CAE (Abaqus). A static analysis was conducted to simulate valve closure with encastre boundary conditions and applied uniform pressures on the leaflets (Figure 1, *B*). To evaluate the designed aortic bioprostheses, a pressure differential of 95 mm Hg was applied to the aortic side of the valve.[11] To evaluate the designed mitral bioprostheses, a pressure differential of 120 mm Hg was applied to the ventricular side. These values were derived from previous ex vivo testing of bioprosthetic valves in our left heart simulator. As our model used static pressurization with encastre boundary conditions, the method is equally extensible to simulations in both the mitral and aortic positions by modulating the hemodynamic parameters to either systolic or diastolic pressures, respectively, which would effectively invert then hemodynamic orientation relative to the left ventricle. The leaflets were modeled as fixed bovine pericardial tissue, with elastic modulus of 8 MPa, density of 1100 kg/m^3, and Poisson ratio of 0.45.11 to 13 Maximum von Mises stresses were assessed for each valve model, and the maximum stresses, surface area, and central leakage data were all plotted against leaflet quantities and regression models were fit to the data.

The following description depicts the basic hydrostatic pressure analysis implemented for our finite element analysis by the Abaqus software package. Detailed information is derived directly from the dedicated Abaqus documentation and modified with basic finite element modeling theory.

To define hydrostatic pressure in Abaqus, one must provide the Z-coordinates of the zero pressure level and the level at which the hydrostatic pressure is defined in an element-based or surface-based distributed load definition. For levels above the zero pressure level, the hydrostatic pressure is zero. In planar elements the hydrostatic head is in the Y-direction and for axisymmetric elements the Z-direction is the second coordinat.



One can specify external pressure, internal pressure, external hydrostatic pressure, or internal hydrostatic pressure on pipe elements. When pressure loads are applied, the effective outer or inner diameter must be specified in the element-based distributed load definition. By default, the loads resulting from the pressure on the ends of the element are included. Open-end loading can be specified in the element-based distributed load definition. Closed-end conditions correctly model the loading at pipe intersections, tight bends, corners, and cross-section changes, whereas open-end conditions require application of additional loads at such points. In straight sections and smooth bends, the end loads of adjacent elements cancel each other precisely. The only case in which closed-end conditions yield an incorrect end load occurs if the pressure at the end of a pipe is supported by an independent structure (such as a piston), which is rather unusual.

Viscous pressure loads are defined by the following equation:

$$p = -c_v \mathbf{v} \cdot \mathbf{n},$$

where p is the pressure applied to the body; c_v is the viscosity, given as the magnitude of the load; $\mathbf{v}$ is the velocity of the point on the surface where the pressure is being applied; and $\mathbf{n}$ is the unit outward normal to the element at the same point. Viscous pressure loading is most commonly applied in structural problems when you wish to damp out dynamic effects and, thus, reach static equilibrium in a minimal number of increments. An appropriate choice for the value of c_v is important for using this technique effectively.

To compute c_v, consider the infinite continuum elements. In explicit dynamics those elements achieve an infinite boundary condition by applying a viscous normal pressure where the coefficient c_v is given by ρc_d; ρ is the density of the material at the surface, and c_d is the value of the dilatational wave speed in the material (the infinite continuum elements also apply a viscous shear traction). For an isotropic, linear elastic material

$$c_d = \sqrt{\frac{\lambda + 2\mu}{\rho}} = \sqrt{\frac{E(1-\nu)}{\rho(1+\nu)(1-2\nu)}},$$

where λ and μ are Lamé's constants, E is Young's modulus, and ν is Poisson's ratio. This choice of the viscous pressure coefficient represents a level of damping in which pressure waves crossing the free surface are absorbed with no reflection of energy back into the interior of the finite element mesh. For typical structural problems it is not desirable to absorb all of the energy (as is the case in the infinite elements). Typically, c_v is set equal to a small percentage (perhaps 1 or 2 percent) of ρc_d as an effective way of minimizing ongoing dynamic effects. The c_v coefficient should have a positive value. For more information on the mathematical analysis and classical mechanics of hydrostatic pressure loading for finite element analysis via Abaqus CAE, please visit the following reference:

https://classes.engineering.wustl.edu/2009/spring/mase5513/abaqus/docs/v6.5/books/usb/default.htm?startat=pt06ch19s04aus92.html#usb-prc-ploaddistributed-hp.