Fluid–Structure Interaction Models of Bicuspid Aortic Valves: The Effects of Nonfused Cusp Angles

Bicuspid aortic valve (BAV) is the most common type of congenital heart disease, occurring in 0.5–2% of the population, where the valve has only two rather than the three normal cusps. Valvular pathologies, such as aortic regurgitation and aortic stenosis, are associated with BAVs, thereby increasing the need for a better understanding of BAV kinematics and geometrical characteristics. The aim of this study is to investigate the influence of the nonfused cusp (NFC) angle in BAV type-I configuration on the valve’s structural and hemodynamic performance. Toward that goal, a parametric fluid–structure interaction (FSI) modeling approach of BAVs is presented. Four FSI models were generated with varying NFC angles between 120 deg and 180 deg. The FSI simulations were based on fully coupled structural and fluid dynamic solvers and corresponded to physiologic values, including the anisotropic hyper-elastic behavior of the tissue. The simulated angles led to different mechanical behavior, such as eccentric jet flow direction with a wider opening shape that was found for the smaller NFC angles, while a narrower opening orifice followed by increased jet flow velocity was observed for the larger NFC angles. Smaller NFC angles led to increased stresses on the NFC. Jermihov [18] generated different geometric variations of BAV; four models with NFC angles of 120 deg and 180 deg, with and without a thickened raphe. During diastole, for both BAV models, they found higher stresses on the fused leaflet than on the NFC, particularly at the attachment edge of the raphe, while the thickened raphe model had lower stresses on the fused cusp, but increased stresses on the NFC. Jermihov [18] generated different geometric variations of BAV; four models with NFC angles of 120 deg and 180 deg, with and without a thickened raphe, and a symmetric TAV. When comparing between the type I model of 120 deg and 180 deg, higher stresses and an increased orifice area were found located between the right and left cusps. The most common nonfused cusp (NFC) angle is 150 deg (92%) [13].

Both experimental and numerical studies have been previously performed on the BAV. Experimental studies characterized the BAV with a limited valve opening orifice, increased turbulent flow, and asymmetric jet flow toward the NFC [14–17]. Numerical models were used to analyze the structural and fluid dynamic aspects of the BAV, which cannot be captured in experimental studies. Some of the previously published models are limited to structural finite element (FE) analysis, so-called dry models, where a predefined transvalvular pressure difference is applied as an input to the FE-model, thereby not considering the interaction of the fluid and the structure. The proposed biomechanical models could explain the early failure of BAVs with decreased NFC angles, and suggest that a larger NFC angle is preferable in suture annuloplasty BAV repair surgery. [DOI: 10.1115/1.4038329]
and their sinuses are identical. The parametric geometry, metric aortic valve, assuming that two portions of the fused cusp generated based on a 3D parametric representation of an asymmetrical anisotropy and nonlinearity [27,28]; Chandra et al. [27] created simplified two-dimensional simulations only, disregarding the few FSI studies have investigated the BAV. Some studies are limited both the structure and flow mechanics simultaneously, while only previously introduced for TAVs [22–26], involving computations of the structure and flow mechanics simultaneously, while only.

Many advanced fluid-structure interaction (FSI) models were previously introduced for TAVs [22–26], involving computations of both the structure and flow mechanics simultaneously, while only a few FSI studies have investigated the BAV. Some studies are limited to simplified two-dimensional simulations only, disregarding the leaflet anisotropy and nonlinearity [27,28]; Chandra et al. [27] created models of TAV, and two asymmetric BAVs, with and without calcification. In the BAV models, during peak systole, they noticed a strong jet toward the NFL, with higher shear stress than in the TAV. Kuan and Espino [28] aimed to evaluate the requirements for future BAV models by creating a simplified asymmetric BAV model without raphe and concluded that the BAV asymmetry must be taken into account. Other studies employed more accurate and complex three-dimensional (3D) FSI models of the BAV; Katayama et al. [29] compared rare configurations of a symmetric BAV type 1 to a healthy and stenotic TAV. They observed an increased pressure gradient in BAV compared with TAV. Our group [30] previously compared TAV to symmetric BAV without raphe, and to asymmetric type 1 BAV with and without raphe, at peak systole. An increased presence of vortices was found in the asymmetric BAV sinuses, and higher flow shear stress (FSS) magnitudes were found in the BAV cusps. Weinberg and Kaazempur Mofrad [31] modeled symmetric BAV and TAV in three length scales: the cell, tissue, and organ. They assumed isotropic tissue, while the fibers were aligned in the radial and circular directions. Cao and Sucosky [32] employed FSI models of the aorta of asymmetric BAV type 1, with three variations of fusing locations, between the right-left, non-crownary-left and non-crownary-right cusps. The jet flow direction was toward the NFC side in all the models, and resulted in increased wall shear stress on the aorta at that side.

From the above literature review, it is clear that the BAV has been studied extensively, due to its high prevalence and continuing failure rates. Recent studies have indicated that there is a strong correlation between the BAV geometry and its long-term durability [33,34], which raises the need for a better understanding of the BAV kinematics and geometrical characteristics [30]. Yet, most of the current studies have focused on the comparison between TAVs and BAVs with rare configurations [13,18,29,30] while they do not take into account the various morphologies that accompany BAVs [35]. This study aims to investigate the influence of the NFC angle on the valve’s functionality in the most common BAV configuration (type 1). To that end, a parametric FSI modeling approach of BAVs is presented and employed for four geometries of BAV with select NFC angles between 120 deg and 180 deg. Response variables, such as orifice area, hemodynamic metrics, and stress magnitudes on the tissues, are examined. These response metrics can help test the hypothesis of early failure in BAVs with NFC angles smaller than 160 deg [33], and to optimize the BAV geometry toward repair surgery in BAV patients.

2 Methods

2.1 Geometry. The geometries of the cusps and roots were generated based on a 3D parametric representation of an asymmetric aortic valve, assuming that two portions of the fused cusp and their sinuses are identical. The parametric geometry, previously suggested by our group for TAVs, was constructed from mathematically formulated 3D curves and surfaces and was validated against BAV echocardiography scans [36]. With several modifications, including angular sizing of the cusps and their root portions, as well as fusion of the two identical cusps, the geometry can represent a BAV with nonfused and fused cusps. The morphology of BAV with one fused cusp was chosen since it is the most common BAV configuration [1,11–13]. The dimensions of the parametric BAV model are function of the annulus diameter, which was chosen to be 24 mm (Fig. 1). Four geometries of BAV type 1 were generated, with NFC angles that vary from 120 deg to 180 deg (Fig. 2).

2.2 Structural Models. The cusps were assumed to have a heterogeneous structure composed of collagen fibers embedded in the elastic matrix. A collagen fiber network model material was employed in accord with the realistic bundle orientations within the layers [37,38] (Fig. 3(a)). The fiber bundles were modeled as beam elements, with radii varying from 0.1 to 0.4 mm. Assuming a thick structural behavior, the cusps and the root were modeled using shell elements with a thickness of 0.3 and 2 mm, respectively [39]. The hyper-elastic mechanical behavior of the sinuses was calibrated using the stress–strain curve published by Gundiah.
The collagen fiber alignment along the cusp. (b) True stress–strain curves for the hyper-elastic elastin and collagen in the cusp and the root. (c) The material constants suitable for employing the Ogden model with a first-order for the elastin and collagen, and a third-order for the sinuses. (d) The performed pressure in the dry model, calculated by the pressure difference in the aorta and left ventricle as a function of time.

et al. [39], after they experimentally tested 6 porcine sinuses. The stress–strain curves of the elastin and collagen were calibrated to the experimental study by Missirlis and Chong [40]. These material properties were used within prior calibrated and validated simulations of the native TAV [30,38,41,42]. The hyper-elastic Ogden model was chosen to best fit the above material properties, and was employed with a first-order form for the elastin and collagen, and a third-order form for the root (Figs. 3(b) and 3(c)). It was assumed that for simulating the coaptation area, linear elastic material properties would be appropriate [43] only for this limited part of the cusp. This assumption was needed for improving the convergence due to contact discontinuities. Nonlinear coaptation material was also examined to demonstrate the relative local and small error due to this linear material simplification. Poisson’s ratio of 0.45, Young’s modulus of 0.45 MPa, and a density of 1100 kg/m³ were used for the coaptation area. As for the root and the cusps (including the fibers), we used a density of 2000 and 1100 kg/m³, respectively [43,44].

Predominantly, dry models were employed in order to examine the structural behavior of the models, especially during diastole where the fluid is minimal. In these models, a physiologic time-dependent transvalvular pressure load was applied on the cusps alone, which constitutes the gradient between the aorta and the left ventricle as a function of time (Fig. 3(d)). The annulus and the sinotubular junction were constrained, while the cusps were tied to the root. A master–slave contact algorithm was employed between the cusps assuming a nonfrictional contact. The dry structural model was previously validated against in vitro tests [45]. The structural solver was ABAQUS 2016 (Dassault Systemes, Simulia Corp., Providence, RI) FE software, where force and momentum equilibrium equations were used [46] and an implicit nonlinear dynamic analysis was employed.

2.3 Fluid–Structure Interaction Models. The FSI simulations were employed in order to confirm the dry models and compare the hemodynamics between the four BAV models. In order to keep the boundary conditions away from the cusps, two straight circular and rigid tubes (1 and 4 cm in length) were added upstream and downstream, respectively. These lengths were chosen to be physiologically relevant [47]. The annulus and sinotubular junction were constrained to the rigid tubes. The flow was assumed to be laminar [48] and the blood to be Newtonian and isothermal at a temperature of 37 °C [46]. The blood was assumed to be slightly compressible, with a realistic and physiologic compressibility of $3.75 \times 10^{-6}$ m²/N [22,49]. Navier–Stokes equations were solved to model the flow. Physiologic time-dependent pressure waveforms were employed in both the upstream and downstream boundaries, presenting healthy left ventricle and aortic pressures [47]. A partitioned solver was used, where a sequential two-way coupling process exchanged the node displacements from the structure solver, and the calculated traction loads (including the fluid pressure and shear) from the flow solver. Since the flow solver calculates different pressure values on both sides of the cusps, virtual surfaces that represent the thickness of the cusps were added to both sides of the cusps [30,38,47]. Nonconformal meshes were used; the structure equations were solved using the Lagrangian approach, and the flow equations were solved with a finite volume method and Eulerian approach. The mesh was dynamically adapted and refined near the boundaries of the wall and the cusps using the cut-cell method with a Cartesian mesh. Mesh refinement study was previously performed for both structural and flow meshes [47]. The nodes on the moving bodies were meshed with polyhedral cells and followed the motion of the valve surface. Since the boundary nodes are part of the fluid mesh, the flow-related parameters, such as shear stress, can be directly calculated and no-slip condition can be directly imposed on the moving valve. The maximum time steps were 1 and 5 ms for the flow solver and structure solver, respectively. A spatial-temporal second-order upwind discretization technique was employed by the implicit flow solver. The initial spacing of the Cartesian mesh was 1 mm in the x and y directions and 1.2 mm in the z direction. The flow domain was discretized with a mesh of approximately 700,000 elements [47]. The structural problem was solved by ABAQUS 6.16 (Simulia, Providence, RI) FE software. FLOWVISION HPC 3.09 (Capvidia, Leuven, Belgium) was used as the flow solver and FLOWVISION MULTI-PHYSICS manager (Capvidia NV, Leuven, Belgium) managed the coupling between the two solvers.

3 Results

3.1 Dry Models

3.1.1 Peak Systole. Figure 4 presents the maximum principal stress distribution acting on the BAV cusps in all four models.
during peak systole, when the valve reaches its maximal opening at 0.1 s from the beginning of the systolic phase. For all the models, the highest maximal principal stress values were concentrated in the raphe region; a gradual change in the stress values was observed in the four models, while in the 180 deg model, the highest stress value of 146 kPa was found. In the 160 deg and 140 deg models, the same stress values of 123 kPa were spread across the raphe, and in the 120 deg model, the lowest stresses were found with value of 100 kPa.

The opening morphologies and orifice areas of the four BAV models during peak systole are also shown in Fig. 4, where the view is from the aortic side. The opening shapes change gradually as the NFC angle increases; the opening shape for the 120 deg angle is the most circular, yet eccentric in its location, closer to the NFC. As the NFC angle increases, the opening shape becomes more elliptic (a fish-mouth shape) and more centered. In the 180 deg model, the orifice is fully centered, which is closer to an ideal jet flow. As for the orifice area values, no significant differences were found among the four models; the orifice areas of the 120 deg and 140 deg models were almost identical, about 2.25 cm²; and in the 160 deg and 180 deg models, they were 2.26 cm².

3.1.2 Diastole. The maximum principal stress distribution in the BAV cusps during diastole, at 0.33 s, is presented in Fig. 5. This time point was chosen since it represents the typical stress distribution during diastole, 0.13 s after the cusp begins to close. For all the models, high stress concentrations were found in the raphe region, accompanied by a stress distribution pattern; as the NFC angle decreased, the stresses on the raphe region increased while decreasing in the NFC belly (Fig. 5). The values of the average maximum principal stress of the nonfused and fused cusps for all the models are also presented in Fig. 5. The average maximum principal stress of each cusp was calculated as a weighted average of the stress value based on the local element volume. Most of the stresses in the fused cusp were concentrated in the raphe region; the highest averaged maximal principal stresses were found in the 120 deg model, with a magnitude of 327 kPa. As the NFC angle increased, so the averaged stress magnitudes decreased to 244 kPa for the 180 deg model. The opposite phenomenon occurred in the NFC, particularly in the belly region; the lowest averaged maximal principal stresses were found in the NFC angle of 120 deg, with a magnitude of 269 kPa, while as the NFC angle increased, the averaged stress magnitudes increased to 372 kPa for the 180 deg model.

3.2 Fluid–Structure Interaction Models

3.2.1 Comparison of the Flow Patterns. Figure 6 presents the blood flow velocity distribution in all four models, during peak systole. As was predicted from the opening pattern calculated from the dry models (Fig. 4), the NFC angle affects the opening shape and eccentricity, and therefore the jet flow pattern. An eccentric oriifice area, as in the 120 deg model, led to an asymmetric jet flow directed toward the NFC side of the aortic wall. None of the jets of the other models directly hit the aorta because they were relatively centered. The maximum jet flow velocity increased with the increase in the NFC angle from 1.47 to 1.93 m/s for 120 deg and 180 deg, respectively. Nevertheless, similar maximum flow velocity magnitudes were found in the 180 deg and 160 deg models (1.89 m/s in the latter). The increased jet flow velocity magnitudes can be related to the narrower elliptical orifice area shape associated with the models with increased NFC angle. It is important to note that the same pressure gradients were used for all the models.

3.2.2 Comparison of Flow Shear Stress Distribution on the Cusps’ Surfaces. Figure 7 presents the FSS contours on the ventricular side of the nonfused and fused cusps from the results of the four FSI models at peak systole. The highest shear stress magnitudes were found in the coaptation area, while the shear stress values were significantly lower in the belly region. The same maximal FSS magnitudes were found for all the models, the difference lies in the high FSS distribution. In the NFC of all the models, concentrated high FSS contours were located along the free edge. As the NFC angle decreased, the high FSS zone spread to a larger portion of the cusp, whereas in the 120 deg model the higher FSS zone spread along most of the coaptation region. All of the examined fused cusps were also exposed to concentrated higher FSS along the free edge. However, the fused cusps of the 160 deg and 180 deg models also had regions of high FSS concentration, albeit only slightly, in the belly region. We observed a high FSS distribution pattern; in the 120 deg model, the high FSS contours were concentrated and covered most of the coaptation area of the NFC, and as the NFC angle increased, particularly in the 160 deg and 180 deg models, the high FSS distribution spread along the...
free-edge area and the belly of the fused cusp, and was divided more equally between the two cusps.

4 Discussion

This study compared four BAV geometries with NFC angles between 120 deg and 180 deg by performing both dry (structural only) and FSI simulations. A 3D parametric geometrical representation of the most common BAV configuration, with a raphe and asymmetric cusps, was employed to generate these anatomies. The predicted jet flow patterns (hemodynamics) and stresses were presented. One of the goals of this study was to investigate the reasons that underlie early failure of BAVs with NFC angles smaller than 160 deg that often requires reoperation [33].

It should be noted that prior mesh refinement and convergence studies showed, in part, that when the local geometry of the raphe is smoother along with geometrical discontinuities, the concentration of the local deformation is reduced. However, both crude and smoother models yield the same maximum deformation levels and did not severely affect their critical values. Therefore, the moderate mesh model was selected to be used for the parametric FSI models due to the associated efficiency and fast convergence. It is interesting to note that the stresses in the collagen fibers were not affected from the mesh refinement.

The calculated orifice areas from the dry models (Fig. 4) and the jet flow results from the FSI models (Fig. 6) suggest a tradeoff between these two parameters. For smaller NFC angles, the orifice shape is wider and closer to circular, leading to lower jet flow velocity magnitudes. Yet, the orifice opening is located eccentrically, causing the jet flow to be asymmetric and directed toward the aortic wall on the NFC side, leading to increased FSS in the aorta [50,51], which can spur medial elastin degradation, and contribute to the development of aortopathy [52,53]. On the other hand, as the NFC angle increases, the orifice shape becomes narrower and is followed by higher jet flow velocity magnitudes. This could result in an increased peak systolic pressure gradient [54–56] in order to compensate for the narrowed orifice area. The jet flow of the increased NFC angles is centered, which is the preferred jet flow direction. The same jet flow pattern [51] and trend for the peak gradient [56], which increases in accordance with the maximum flow velocity, were also observed in clinical studies of flow in BAVs. It should be emphasized that in order to make this study comparable, the same pressure gradient was applied in all the dry and FSI models; therefore, the differences in the velocity magnitudes resulted only from the BAV geometry. Furthermore, the eccentric orifice area of the smaller NFC angle models can be explained by the increased fused cusp angle, which is very restricted and cannot be adequately opened. As for the opening shape patterns, our models correlated with those of Jermihov [18] even though they compared only two BAV models with NFC angles of 120 deg and 180 deg.

The maximal principal stress during diastole, as calculated from the dry models (Fig. 5), was found to have two contrasting distribution patterns in the fused and nonfused cusps. As the NFC angle increased, the stress magnitudes decreased in the raphe region of the fused cusp while increasing in the NFC belly. This outcome can be explained by the size of the cusp; the larger the cusp the more load it has to carry, thereby increasing its stress distribution. A very good correlation was found with the numerical study by Conti et al. [19]. They presented a BAV with small NFC angle, where the stress magnitudes and distribution during diastole were similar to our findings: around 375 and 500 kPa in the belly and raphe regions, respectively. High FSS distribution pattern was also observed in the fused cusps during peak systole (Fig. 7). While this pattern was spread equally between the two cusps in the 160 deg and the 180 deg models, it was very concentrated along the coaptation area of the NFC, and was unequally spread between the two cusps in the 120 deg model. This type of concentrated high FSS may be related to the initiation of calcification [57,58]. Nevertheless, the maximal FSS magnitudes (Fig. 7) were much smaller (~50 Pa) than the maximum principal stresses in the tissues during both peak systole (~100–150 kPa) (Fig. 4), and diastole (~250–350 kPa) (Fig. 5).

A performance summary of the four examined geometries that account for the above parameters is presented in Table 1. Three plus signs (+++) mark excellent performance, representing the relatively ideal outcomes for the tested parameters, while one plus sign (+) represents a poor performance. The scoring decision is based on the model results and previous knowledge in the literature on factors leading to early failure of the valve. A tradeoff exists between the jet flow direction and the orifice shape, which leads to change in the jet flow velocity. The ideal jet flow direction is axial (180 deg model), similar to TAVs, and the optimal jet flow velocity is low (120 deg model) as in TAVs. On the other hand, the flow velocities are not yet considered to be pathological (they are all lower than 2 m/s) [59,60]; therefore, the velocity magnitude parameter will not be considered in the summary table. The connection between concentrated high FSS to the initiation of calcification has been proven [57,58]. Therefore, it would be preferred to avoid exposing large portion of the cusp to high FSS as in the NFC of the 120 deg model, and to strive for the high FSS to spread equally between the cusps and along smaller portion of the NFC as in the pattern found in the 160 deg and 180 deg models. The desired diastolic stress distribution should have the lowest magnitudes and maintain equal distributions on both cusps, as found in the 140 deg dry model. The 140 deg and 160 deg geometries were found to best represent the ideal combination between maximum principal stresses, shear stresses, and jet flow direction, and therefore they were chosen as the optimal configurations regarding the parameters we examined. These results can also explain why BAVs with NFC angles less than 160 deg tend to fail early [33]. The 120 deg model demonstrates this phenomenon; it has a very low performance compared with the other geometries because of the asymmetric jet flow, which may impair the aorta, as well as concentrated shear stress, which could lead to initiation of calcification and high diastolic stresses in the raphe region, which can also expedite calcification. These findings, together with the clinical results of the long-term durability of BAV repair [10,33], may have important clinical implications, which can contribute to improve techniques of surgical BAV repair, thereby enhancing BAV long-term durability.

4.1 Limitations. One of the limitations of the current study is the lack of information in the literature about the material properties of healthy BAVs, particularly regarding the raphe region. Therefore, in our models, we assumed material properties of TAVs, while the raphe properties were identical to the hyperelastic properties of the belly region. Other limitations are related to the physical and numerical assumptions of the models; the current BAV geometry was based on parametric representation of TAV geometry, which may not be suitable for BAVs. In order to make this study comparable, we used similar dimensions for all the models, except for the angle of the cusps and sinuses. Another assumption is that the initial simulation conditions were of an almost closed valve with no residual stresses in the tissue. This assumption has been used in prior studies where it showed good approximation for the overall kinematics [30,47]. Another limitation is the use of laminar flow, which is widely used in BAV.

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simulations [20,61,62]. In the current study, the Reynolds number was lower than 3000, which could have led to transitional flow, yet it is not a fully developed turbulent flow. Moreover, a study that was carried out in our research group [59] on calcified aortic valve disease (which is more prone to develop turbulent flow than BAVs) compared FSI models with laminar assumption and turbulence models. It was found that a negligible difference exists between these two cases, which gives further support for the current laminar flow assumption. Also, the use of shell-based structural elements over higher order shell or brick-solid elements may have limitations in a reduced accuracy to compute and resolve the deformations. It is well known that higher order shell finite elements can better capture the bending of thick plates and shells. The current shell triangular elements (S3R) is a general purpose conventional shell element capable of capturing the transverse shear deformation in the case of relatively higher shell thickness and enforces the Kirchhoff assumptions in case of a relatively thin shell. This special treatment of this type of element makes it well suited for the aortic valve modeling. It also accounts for finite membrane strains and arbitrarily large rotations; suitable for large-strain analysis, however, it is limited to a linear spatial case within the element. Finally, although no experiments or clinical measurements were performed in this study, the results were compared to available clinical data and to other numerical models, while the conclusions were based on trend comparisons rather than on absolute values.

5 Conclusions

Parametric FSI modeling approach for BAVs is presented. The proposed parametric modeling approach was employed for the biomechanical analyses of BAVs in order to examine the influence of the NFC angle. It was found that BAVs with smaller NFC angle have the least effective mechanical performance, expressed in asymmetric jet flow and unequal stress distribution between the cusps that can lead to aortopathy and calcification initiation, respectively. This can explain the reasons behind the poor durability of BAVs with NFC angles smaller than 160 deg. The proposed computational model can pave the way towards simulations of patient specific BAV repair surgery, such as suture annuloplasty, and improvement of BAV surgical repair techniques for better long-term durability.

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