27th INTERNATIONAL CONGRESS OF MECHANICAL ENGENEERING
COBEM 2023
Brazilian Society of Mechanical Sciences and Engineering
27th ABCM International Congress of Mechanical Engineering
December 04-08, 2023, Florianópolis-SC, Brazil

COB-2023-0372
INFLUENCE OF THICKNESS ON FLUTTER IN PROSTHETIC BIOLOGICAL HEART VALVES

Matheus Carvalho Barbosa Costa
Graduate Program in Mechanical Engineering – Universidade Federal de Minas Gerais, Avenida Antônio Carlos, 6627, Belo Horizonte, Minas Gerais, Brazil
mt94_carvalho@hotmail.com

João Victor Curado Fleury
Universidade Federal de Minas Gerais, Avenida Antônio Carlos, 6627, Belo Horizonte, Minas Gerais, Brazil
joaovictorfleury@gmail.com

Saulo de Freitas Gonçalves
Graduate Program in Mechanical Engineering – Universidade Federal de Minas Gearis, Avenida Antônio Carlos, 6627, Belo Horizonte, Minas Gerais, Brazil
saulodfg@gmail.com

Mário Luis Ferreira da Silva
Graduate Program in Mechanical Engineering – Universidade Federal de Minas Gearis, Avenida Antônio Carlos, 6627, Belo Horizonte, Minas Gerais, Brazil
marioluisfs@gmail.com

Artur Henrique de Freitas Avelar
Universidade Federal de São João Del-Rei, Praça Frei Orlando, 170, São João del Rei, Minas Gerais, Brazil
arturavelar@ufsj.edu.br

Rudolf Huebner
Universidade Federal de Minas Gerais, Avenida Antônio Carlos, 6627, Belo Horizonte, Minas Gerais, Brazil
rudolf@demec.ufmg.br

Abstract. *Despite being associated with an increase in calcification, fatigue, hemolysis and thrombus formation of biological aortic valve prostheses, flutter has barely been studied in the dynamics of these valves. The present study aims to evaluate the influence of leaflets thickness on the dynamics behavior of heart valve bioprotheses using the Finite Element technique. For this purpose, the geometry of the aortic valve prostheses was developed, based on geometrical models available in the literature, and only the thickness was changed for each simulation. Transvavular pressure was applied on the side facing the left ventricle, and a fixed support was considered on the sides of the leaflets. The mesh was constituted by quadrilateral shell elements. The first order Ogden hyperelastic constitutive model was used to calculate the deformation energy function of the valve material. The results indicate that the oscillatory behavior of the leaflets depends on their thickness. It was observed that thicker valves had higher amplitudes and lower frequencies compared to thinner valves. Furthermore, other quantities, such as stress and strain, are influenced by this oscillatory behavior. With the present work, it is possible to observe that numerical studies can be a viable alternative to evaluate and quantify the processes responsible for the deterioration of prosthetic heart valves.*

Keywords: *Aortic Valve, Bioprotheses, Finite element method, Flutter, Thickness, Shell elements.*

1. INTRODUCTION

Aortic valves are structures formed by flexible leaflets constituted of fibrous tissue with the function of ensuring that the flow between the left ventricle and the aorta is predominantly unidirectional (Johnson et al., 2020, Xu et al., 2020). These valves are characterized by low flexural stiffness, allowing a rapid opening and closing movement, and also have a complex structural organization, so that these elements can tolerate the high cyclic loads during the cardiac cycle (Chen and Luo, 2020).

Dysfunction of the aortic valves, such as congenital, rheumatic, and degenerative heart diseases, is among the main causes of death among the group of adversities that affect the cardiovascular system (Hasan et al., 2014). In addition, in economically developed societies, it is estimated that more than 30 million people live with valvulopathies, and the incidence increases with age (Oveissi et al., 2020).

Replacement of damaged native valves with mechanical or biological prostheses is the main strategy for treating patients affected by heart valve dysfunction (Borazjani, 2013, Bhagra et al., 2016). Worldwide, more than 290,000 patients undergo valve replacement surgeries each year, and this number is estimated to reach 850,000 patients in the year 2050 (Hasan et al., 2014, Avelar et al., 2017, Coulter et al., 2019).

Biological prosthetic valves, as they are similar to natural valves, do not interfere with hemodynamics and reduce the possibility of rejection by the patient. However, they are characterized by having a low lifetime, mainly due to leaflet calcification and structural failures (Tayama, 2022, Rahmani et al., 2019).

The low lifetime of biological prosthetic valves, comprising 10-15 years, is related to the occurrence of valve leaflet oscillations, a phenomenon known as flutter (Avelar et al., 2019). Although the flutter is related to calcification, hemolysis, thrombus formation, and leaflet fatigue of biological heart valve prostheses, this phenomenon is not frequently studied in the dynamics of theseprostheses (Zhu et al., 2023).

Studies indicate that the geometric parameters of prosthetic heart valves directly impact the vibrational characteristics of leaflets (Lee et al., 2020, Johnson et al., 2020, Johnson et al., 2022). Lee et al. (2020) analyzed the influence of valve diameter and leaflet thickness on flutter behavior using an *in-vitro* methodology and numerically. Bioprotheses manufactured from bovine pericardium with 0.2, 0.4 and 0.6 mm in thickness and 21, 25 and 27 mm in diameter were tested. The authors have found that the smaller the diameter and the greater the thickness, the more oscillations occurred during the period in which the valve was open, that is, the greater the flutter frequencies for these cases. In addition, it was evident that the flutter is more sensitive to variation in thickness.

In other two studies that also focused on geometric parameters of prosthetic heart valves, Johnson et al. (2020) and Johnson et al. (2022) performed fluid-structural simulations to investigate the effect of varying leaflet thickness on flutter. The authors noticed that thinner leaflets are more susceptible to flutter, and this may be caused by a flow-induced instability that excites the valve bending vibration in these cases. Finally, due to the occurrence of leaflet oscillations, variables such as forces applied to the leaflets and strains revealed an irregular and oscillatory behavior, thus being a critical factor in the increase of the inherent fatigue caused by the cyclic loading during the cardiac cycle.

Numerical simulations appear as a powerful tool for the investigation, in a non-invasive and low-cost way, of the parameters that govern the flutter phenomenon in heart valves, since evaluating such characteristics by experimental means is hardly feasible (Siguenza et al., 2018). Simulations based only on the Finite Element Method (FEM) are simpler to perform and are sufficient to qualitative analyzes, however, they are focused only on the dynamics of the leaflets, reducing the interaction with the blood to a constant pressure term along the cusp (Grosh et al., 2018, Cai et al., 2021). Hou et al. (2021) investigated the effect of the protrusion height of aortic valves on the opening and closing movement of these valves to select the most appropriate protrusion height range for the total closure of the bioprostheses using the FEM technique. Sadrabadi et al. (2021) evaluated, using the FEM technique, the effect on the dynamics of the aortic valves due to advancing age and calcification. Qiu et al. (2021) analyzed the stress field and the dynamics of the leaflets of two different models of bioprostheses, considering the effect of the eccentricity of one of the models, using the finite element method.

The present work aims to evaluate the effect of thickness variation on the vibrational parameters of the leaflets using the finite element method. In this way, our aim is to contribute to the advancement of prosthetic biological heart valvedesigns, mainly by improving the understanding of the flutter phenomenon in these valves to contribute to the advancement of the durability of the biological prostheses.

2. METHOD AND MATERIALS

2.1 Geometry

In order to perform the simulations proposed in this work, it was necessary to develop a representative geometric model of the systemstudied. A geometric model available in the literature present in Abbasi and Azadani (2020) was considered, where the detailed step-by-step for obtaining this geometry is described. SolidWorks software (SolidWorks, Inc., Concord, MA, USA) was used to develop the CAD (Computer Aided Design) model of the aortic valve prosthesis. Figure 1 indicates some views of the valve geometry and a scheme that represents the positioning of the valve.

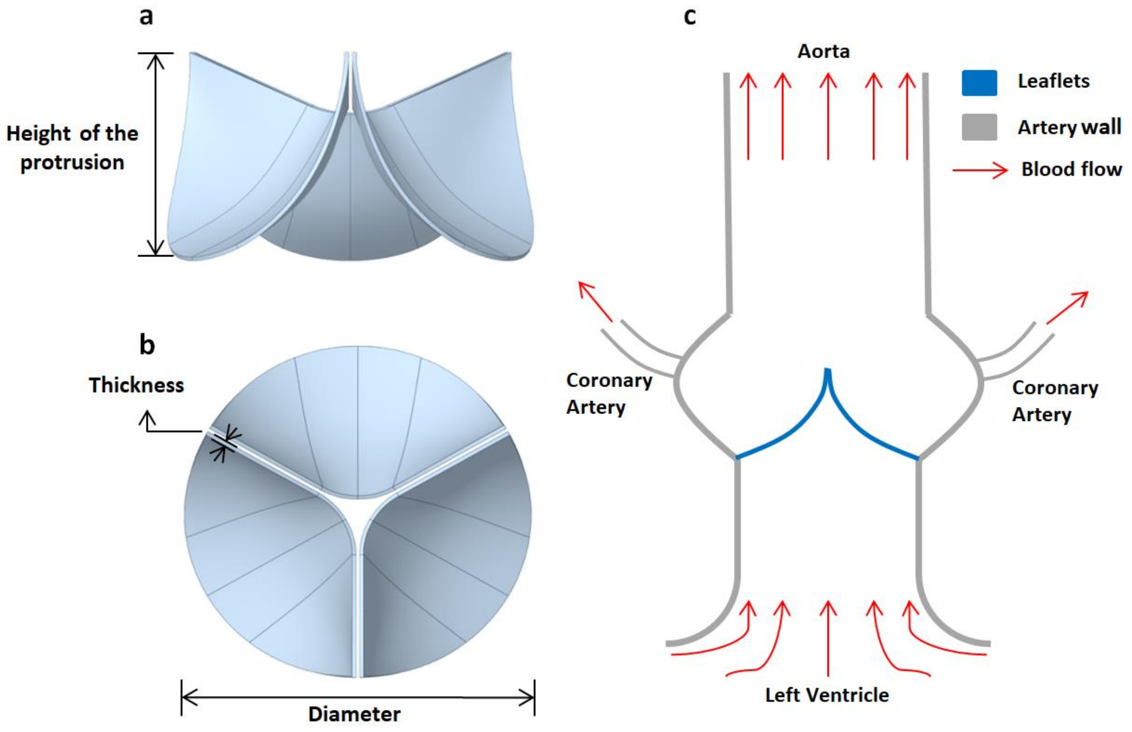

Figure 1.Some views of the valve geometry used and a scheme of the valve. (a) Frontal view. (b) Superior view. (c) Scheme representing the valve positioning.

The diameter of the valve and the height of the protrusion were considered fixed for all simulations with values of 25 and 15 mm, respectively (Chen and Luo, 2020, Lee et al., 2020, Kadel, 2020, Tango et al., 2018). To evaluate the relationship between thickness and flutter parameters, the following values for thickness were considered: 0.20, 0.25, 0.30, 0.35 and 0.40 mm (Chen and Luo, 2020).

2.2 Equation of motion

The equation of motion for deformable bodies in the Lagrangian framework can be written as (Sadrabadi et al., 2021):

$$\rho\frac{\partial^2 \boldsymbol{u}}{\partial t^2} - \boldsymbol{\nabla}\cdot(\boldsymbol{FS}) = \rho\boldsymbol{b}, \tag{1}$$

where ρ is the density of the deformable body material, **u** is the displacement vector, **F** is the deformation gradient tensor, **b** is the body force vector, **S** is the second Piola-Kirchhoff stress tensor determined from the strain energy function W as $\mathbf{S} = 2\frac{\partial \mathrm{W}}{\partial \mathbf{C}}$, where **C** is the Cauchy-Green deformation tensor.

2.3 Valve material

The isotropic hyperelastic model was adopted as constitutive behavior of the valve material, in which the strain energy function was modeled according to the first order Ogden model, Eq. (2). The model was adjusted according to the experimental data of the stress-strain curve obtained from an axial stress test of porcine aortic valve prostheses.

$$W = \frac{\mu}{\alpha}(\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 1) + \frac{1}{2}K(J-1)^2, \tag{2}$$

where, W is the strain-energy function per unit undeformed volume, λ_1, λ_2 and λ_3 are the principal stretches in the three directions of Cauchy-Green deformation tensor, K is the bulk modulus, $J = \lambda_1\lambda_2\lambda_3$ is the ratio of the deformed elastic volume over the reference (undeformed) volume, and μ and α are the empirical model constants. In the present work, it was considered that the valve material was incompressible, and μ = 59,782.4 and α = 12.97 were adopted (Joda et al., 2016). The incompressible was set considering the incompressibility parameter, $\mathrm{d} = 2/\mathrm{K}$, equal to zero. For the valve material density, the value of 1100 kg/m^3 was used (Borowski et al., 2022, Ma et al., 2022, Morany et al., 2023, Lavon et al., 2021).

2.4 Numerical solution

Simulations to obtain the vibrational parameters of the biological prosthesis leaflets were performed using the ANSYS-Mechanical® 2022R2 software (ANSYS-Mechanical Inc., Lebanon, NH, USA).

Transient simulations comprising only the systolic phase of 0.23 s were performed, because the flutter occursonly during the moment that the valve is open. A time step of 1x10^{-4} s was adopted, totaling 2300 time steps to complete the calculations, to ensure that the simulations are able to record flutter during the systolic period (Ìasbeck, 2019).

Frictionless contact was defined between the leaflets on the faces orientated to the left ventricle, since the coefficient of friction between the leaflets is negligible for the current analysis (Pasta et al., 2020, Emendi et al., 2020, Borowski et al., 2018, Wald et al., 2018). The Augmented Lagrange formulation was selected because it is recommended for frictionless contacts and when there are large displacements, and a value of 0.1 was used for the normal stiffness, since bending is dominant in the considered problem (Ansys, 2022). Furthermore, the symmetric contact behavior was selected because, in this case, there is no possibility of determining which face is the target and which is the source of the contact (Ansys, 2022).

The Newmark and Newton-Raphson methods were used to perform the temporal discretization of the governing equation of the analyzed physical problem and to linearize and solve the discretized equation, respectively.

2.5 Boundary conditions

Wherein the deformations on the faces of the leaflets that are fixed to the support ring of the valve are small and there are no values available in the literature, displacements and rotations of theses faces were disregarded, that is, they were used fixed support to model this behavior (Saleeb et al., 2013).

As a physiological boundary condition, it was applied the transvalvular pressure curve, obtained in the work of Ìasbeck (2019), to the faces of the leaflets orientated to the left ventricle. Figure 2a illustrates the transvalvular pressure curve, and Figure 2b represents a scheme that shows a resume of the boundary conditions.

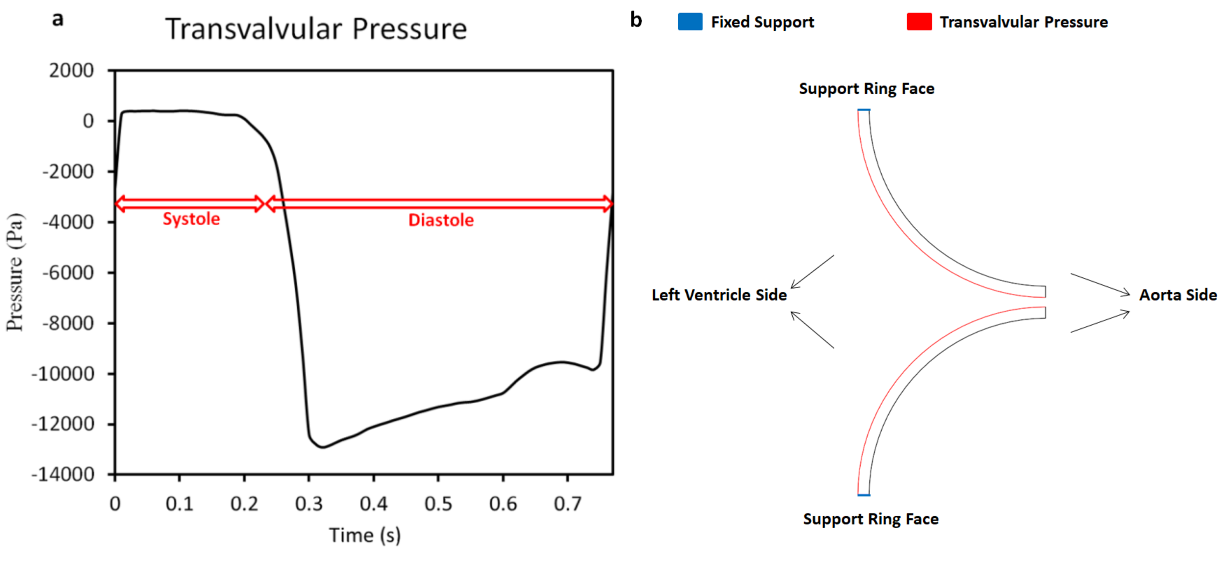

Figure 2. (a) Transvalvular pressure. (b) Scheme of the boundary conditions.

2.6 Mesh test convergence

For the mesh test, static structural simulations were performed, in which all the previously described boundary conditions and set-up were adopted. However, a constant pressure load of 400 Pa was applied on the leaflets (Íasbeck, 2019), a value corresponding to the systolic peak of transvalvular pressure. The grids were refined by doubling the number of elements. The displacements of the leaflet tips (Figure 3) and the maximum principal stress developed in the valve were monitored, and the meshes are made up of quadrilateral shell elements to reduce the computational cost. Table 1 describes the parameters generated for the three meshes tested and the relative errors between the meshes.

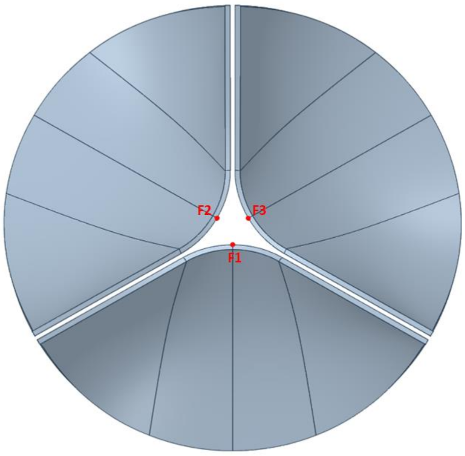

Figure 3. Locations where displacements were monitored for the mesh test convergence.

Table 1. The parameters of the meshes tested and the relative errors between the meshes.

Meshes			
	Mesh 1	**Mesh 2**	**Mesh 3**
# Nodes	11628	22641	44826
# Elements	7322	14492	29036
Maximum Skewness	0.57661	0.59859	0.62379
Minimum Ortogonal Quality	0.73005	0.71262	0.69097

Relative Error		
	Mesh 2 – Mesh 1	**Mesh 3 – Mesh 2**
Displacement F1 (%)	0.47	0.40
Displacement F2 (%)	0.63	0.23
Displacement F3 (%)	0.58	0.11
Maximum Principal Stress (%)	1.49	1.32

The present work considered a relative error between the meshes below 5% for all monitored quantitiesas sufficient to eliminate the mesh dependency in the simulations. Thus, mesh 1 was selected to perform the simulations, since this mesh has the lowest computational cost and met the criteria established for the relative error.

2.7 Flutter analysis

To perform the analysis of the flutter variables, such as frequency and amplitude of oscillations, a Fast Fourier Transform (FFT) of the radial displacements of the tips of the leaflets was performed using the MatlabMathworks® software.

3. RESULTS AND DISCUSSION

Figure 4 indicates the radial displacement of the cusps tips of the aortic valve prosthesis during the systolic period for the five thicknesses considered.

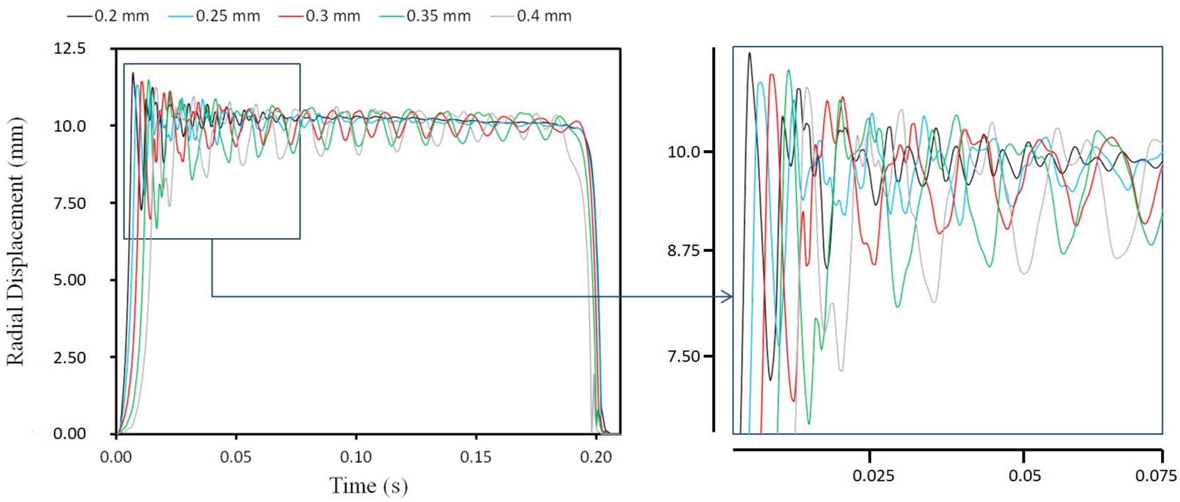

Figure 4. Radial displacement during systole.

First, regarding the dynamics of the opening and closing movement of the valves, it can be observed that with the increase in the thickness of the leaflets, there is an increase in the time for complete opening of the valve, 0.0065 s, 0.008 s, 0.01 s, 0.013 s and 0.016 s, respectively, as also, it is evident the decrease of the time of the complete closing, 0.015 s, 0.0135 s, 0.0075 s, 0.0055 s and 0.003 s, respectively, evidencing, in this way, that the thinner valves remained open for greater intervals of time. In addition, it is noticed that the thicker valves had a smaller mean opening area compared to the thinner valves, since the mean positions during systole were smaller, 10.2 mm, 10.1 mm, 10.0 mm, 9.94 mm and 9.92 mm, respectively. All these indicated facts can be explained by the increase in the flexural stiffness of the valve cusps due to the increase in thickness, making the valve opening process more difficult. This tendency was also observed in the works of Chen and Luo (2020).

With respect to the flutter behavior of the valve leaflet tip, it is possible to observe that oscillations occurred for the five simulated valves. From Figure 4, it can be seen that the greater the thickness of the cusps, the greater the amplitude of the tip vibration of these elements, 0.14 mm, 0.19 mm, 0.43 mm, 0.51 mm and 0.52 mm, respectively. This occurs because the thicker valves, due to their greater inertia and lower average values of opening at which vibration occurs, have greater space and capacity to move until reaching the maximum permissible opening level of the valve. As a consequence of this fact, the frequencies of oscillations decrease with increasing thickness, 628.4 Hz, 523.4 Hz, 433.6 Hz, 368.3 Hz and 322.6 Hz, as thinner valves are able to complete the cycle faster. These results are in contradiction with the results of Lee et al. (2020) and Johnson et al. (2020). In these two works, the authors performed experimental studies and computational calculations based on fluid-structure interactions (FSI), and concluded that the frequency of oscillations increases while the thickness of the leaflets also increases. It was expected that thicker leaflets would vibrate at a higher frequency and lower amplitude, since frequency has a direct relationship while amplitude has an inverse relationship with stiffness. This may indicate that, or a purely structural analysis is not correct for the study of flutter, and therefore it is only recommended to perform FSI calculations and experimental observations, or flutter in prosthetic heart valves is still a poorly understood phenomenon with low predictability of their behavior.

Finally, it was observed that for the valves with smaller thicknesses, the oscillations damped more quickly, and for these cases, due to the lower inertia, the exponential decay factor is greater, thus causing greater vibration damping.

Figures 5a and 5b illustrate the distribution of maximum principal stress and maximum principal strain that developed in the geometry over the systolic period, until time 0.2 s.

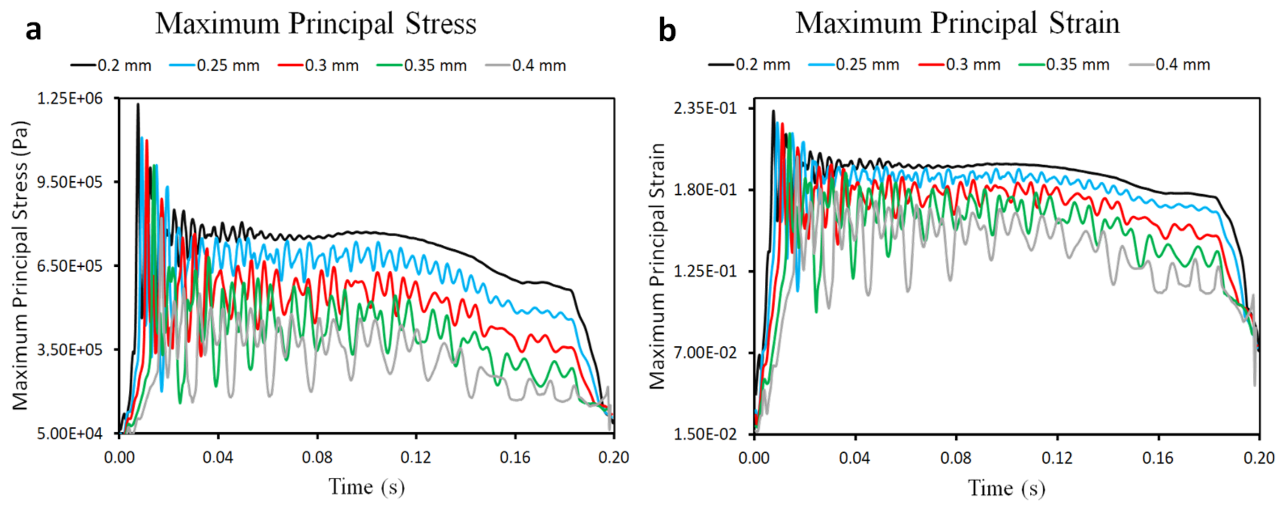

Figure 5. (a) Maximum Principal Stress during systole. (b) Maximum Principal Strain during systole.

Due to the occurrence of leaflet oscillations during the period that the valve is open, the maximum principal stress and strain developed in the geometry also exhibit an oscillatory and irregular behavior. In addition, it can be seen that the two quantities had a qualitatively similar distribution over time. For the same pressure loading for the five cases, the thinnest cusps developed higher mean strain values, 0.188, 0.180, 0.168, 0.155 and 0.141 respectively, and as a consequence, these valves also showed higher mean stress values, 0.719 MPa, 0.629 MPa, 0.516 MPa, 0.411 MPa and 0.325 MPa, respectively, having values of the same order of magnitude that the works of Ìasbeck (2019), Luraghi et al. (2019), Luraghi et al. (2018), Cao and Sucosky (2017).

The oscillatory behaviors of stress and strain were similar to those of radial displacements illustrated in Figure 4. The thinner leaflets had higher frequencies for the two variables, as well as the magnitudes damped more quickly, while the thicker leaflets had greater amplitudes and the oscillations occurred for a longer time. Thus, some points can be highlighted regarding the fatigue caused by flutter. The fatigue of a component dependsmainly on two factors: number of cycles and the value of the alternating stress (half the amplitude of the stress). In this case, for thinner leaflets, more cycles occurred, however, the alternating stress showed lower values, while for thicker leaflets, fewer cycles were observed with lower values of alternating stress. Therefore, there was a balance between the two factors with the thickness variation, so it is necessary to perform additional calculations to determine in which of the thicknesses the resistance to fatigue caused by flutter is more attractive.

The present work has the limitation of having only performed structural simulations, that is, it restricts the interaction of the valves with the blood to a constant pressure term along the leaflets. However, flutter is a more complex phenomenon and has a fluid structurenature, and it is necessary to perform FSI simulations and *in-vitro* bench tests to compare with the results presented here and verify their validity, since quantities that have an influence on flutter, such as velocity and pressure fluctuations (Becsek et al., 2020, Lu et al., 2020, Liu et al., 2020), are not considered here. Another factor that was removed from the analysis was the anisotropic behavior of the biprostheses material, and this condition may influence both the vibrational parameters of the leaflets and, mainly, the stress and strain distributions along the valves.

4. CONCLUSION

The present work proposes the study of flutter in aortic valve bioprostheses using structural simulations using the finite element technique, with the aim of increasing the understanding of this phenomenon to improve the design of these prostheses. The results indicated that the flutter phenomenon in heart valves is closely related to the variation in the geometric parameters of these valves, mainly in relation to the leaflet thickness variation. Valves with thinner cusps had higher vibration frequencies and lower amplitudes compared to valves with thicker leaflets. These results are in contradiction with the results available in the literature, therefore, further investigations on the subject are needed to increase the understanding and predictability of this phenomenon responsible for accelerating the process of deterioration of biological heart valve prostheses. Techniques based on fluid structure simulations are recommended, since the model considers important variables for the phenomenon, for example, the turbulent effects of local hemodynamics. Flutter also influences other variables, such as stress and strain, which exhibited oscillatory and irregular behavior during the systolic period, thus resulting in an additional term capable of accelerating the fatigue process of the prostheses. It was observed that thicker valves showed greater amplitudes of the alternating stress term and smaller complete cycles during systole. However, more meticulous investigation is necessary to indicate which thickness is more attractive in terms of fatigue.

5. ACKNOWLEDGEMENTS

The authors would like to thank FAPEMIG (APQ-02824-21) for their support in this project.

6. REFERENCES

Abbasi, M., and Azadani, A., 2020. A geometry optimization framework for transcatheter heart valve leaflet design, Journal of the Mechanical Behavior of Biomedical Materials, Vol. 102, pp. 103491.

Ansys, Structural Analysis Guide. Canonsburg: ANSYS, Inc., 2022.

Avelar, A.H.F., Canestri, J. A., Bim, C., Silva, M.G.M., Huebner, R., and Pinotti, M., 2017. Quantification and Analysis of Leaflet Flutter on Biological Prosthetic Cardiac Valves, Artificial Organs, Vol. 41, p. 835-844.

Avelar, A.H.F., Stófel, M.A.G.E., Vieira, G.D., Canestri, J.A., and Huebner, R., 2019. Analysis of Leaflet flutter in biological prosthetic heart valves using PIV measurements,ActaScientiarum. Technology, Vol. 42.

Becsek, B., Pietrasanta, L., and Obrist, D., 2020. Turbulent Systolic Flow Downstream of a Bioprosthetic Aortic Valve: Velocity Spectra, Wall Shear Stresses, and Turbulent Dissipation Rates, Frontiers in Physiology, Vol. 11.

Cai, L., Zhang, R., Li, Y., et al., 2021. The Comparison of Different Constitutive Laws and Fiber Architectures for the Aortic Valve on Fluid-Structure Interaction Simulation, Frontiers in Physiology, Vol.12.

Cao, K., and Sucosky, K., 2017. Simulation of Native, Diseased, and Prosthetic heart Valve Function: Computational comparison of regional stress and deformation characteristic in tricuspid and bicuspid aortic valve leaflets, International Journal for Numerical Methods in Biomedical Engineering, Vol. 33.

Chen, Y., and Luo, H., 2020. Pressure distribution over the leaftlets and effect of bending stiffness on fluid-structure interaction of the aortic valve, Journal of Fluid Mechanics, Vol. 883.

Coulter, F.B., Schaffner, M., Faber, J.A., Rafsanjani, A., Smith, R., Appa, H., Zilla, P., Bezuidenhout, D., and Studart, A.R., 2019. Bioinspired Heart Valve Prosthesis Made by Silicone Additive Manufacturing, Matter, Vol. 1, p. 266-279.

Bhagra, C., Souza, R., Silversides, C., 2016. Valvular heart disease and pregnancy part II: management of prosthetic valves, Heart, Vol. 103, p. 244-252.

Borowski, F., Ott, R., Oldenburg, J., Kaule, S., Oner, A., Schmitz, K.P., and Stiehm, M., 2022. Validation of a Fluid Structure Interaction Model for TAVR Using Particle Image Velocimetry, Current Directions in Biomedical Engineering, Vol. 8, p. 512-515.

Borazjani, I., 2013. Fluid-structure, immersed boundary-finite element method simulations of bio-prosthetic heart valves, Computer Methods in Applied Mechanics and Engineering, Vol. 257, p. 103-116.

Emendi, M., Sturla, F., Ghosh, R.P., Bianchi, M., Piatti, F., Pluchinotta, F.R., Giese, D., Lombardi, M., Redaelli, A., and Bluestein, D., 2021. Patient-Specific Bicuspid Aortic Valve Biomechanics: A Magnetic Resonance Imaging Integrated Fluid-Structure Interaction Approach, Annals of Biomedical Engineering, Vol. 49, p. 627-641.

Ghosh, R.P., Marom, G., Rotman, O.M., Slepian, M.J., Prabhakar, S., Horner, M., and Bluestein, D., 2018. Comparative Fluid-Structure Interaction Analysis of Polymeric Transcatheter and Surgical Aortic Valves Hemodynamics and Structural Mechanics, Journal of Biomechanical Engineering, Vol. 140, pp. 121002.

Hou, Q., Liu, G., Liu, N., Zhang, H., Qu, Z., et al. 2021. Effect of Valve Height on the Opening and Closing Performance of the Aortic Valve Under Aortic Root Dilatation. Frontiers in Physiology, v. 12, pp. 697502.

Hasan, A., Ragaert, K., Swieszkowski, W., Selimovic, S., Paul, A., Unal, G.C., Mofrad, M.R.K., Khademhosseini, A., 2014. Biomechanical properties of native and tissue engineered heart valve constructs, Journal of Biomechanics, New York, Vol. 47, p. 1949-1963.

Iásbeck, G., 2019. Modelagem Computacional de um Bioprótese de Válvula Cardíaca – Análise Dinâmica e Fenômeno de Flutter. Dissertação apresentada no Programa de Pós-Graduação em Engenharia Mecânica da Universidade Federal de Minas Gerais.

Johnson, E.L., Wu, M.C.H., Xu, F., and Hsu, M.C., 2020. Thinner biological tissues induce leaflet flutter in aortic heart valve replacements. Proceeding of the National Academy of Sciences of the United States of America, Vol. 117, p. 19007-19016.

Johnson, E.L., Rajanna, M.R., Yang, C.H., and Hsu, M.C., 2022. Effects of membrane and flexural stiffnesses on aortic valve dynamics: Identifying the mechanics of leaflet flutter in thinner biological tissues, Forces in Mechanics, Vol. 6, pp. 100145.

Kadel, S., 2020. Computational Assessment of Aortic Valve Function and Mechanics under Hypertension. A Thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in Renewable and Clean Energy Engineering.

Lavon, K., Morany, A., Halevi, R., Hamdan, A., Raanani, E., Bluestein, D., and Ali, R.H., 2021. Progressive Calcification in Bicuspid Valves: A Coupled Hemodynamics and Multiscale Structural Computations, Annals of Biomedical Engineering, Vol. 49, p. 3310-3322.

Lee, J.H., Scotten, L.N., Hunt, R., Caranasos, T.G., Vavalle, J.P., and Griffith, B.E., 2020. Bioprosthetic aortic valve diameter and thickness are directly related to leaflet fluttering: results from a combined experimental and computational modeling study, JTCVS open, Vol. 6, p. 60-81.

Liu, S., Cai, C., and Han, Y,. 2020. Time-domain simulations of turbulence effects on the aerodynamic flutter of long-span bridges, Advances in Bridge Engineering, Vol. 1, n. 1, p. 1-23.

Lu, Z.Y., Wong, C.W., Zhou, Y., 2020. Turbulence intensity effect on the axial-flow-induced vibration of an elastic cylinder, Journal of Fluids and Structures, Vol. 99.

Luraghi, G., Migliavacca, F., and Matas, J.F., 2018. Study on the Accuracy of Structural and FSI Heart Valves Simulations, Cardiovascular Engineering and Technology, Vol. 9, n. 4, p. 723-738.

Luraghi, G., Migliavacca, F., González, A.G., et al., 2019. On the Modeling of Patient-Specific Transcatheter Aortic Valve Replacement: A Fluid-Structure Interaction Approach, Cardiovascular Engineering and Technology, Vol. 10, n. 3, p. 437-455.

Ma, X., Gao, B., Tao, L., Ding, J., Li, S., Qiao, A., and Chang, Y., 2022. Hemodynamic study of the effect of the geometric height of leaflets on the performance of the aortic valve under aortic valve reconstruction, Journal of Thoracic Disease, Vol. 14, p. 1515-1525.

Morany, A., Lavon, K., Bardon, R.G., Kovarovic, B., Hamdan, A., Bluestein, D., and Ali, R.H., 2023. Fluid-structure interaction modeling of compliant aortic valves using the lattice Boltzmann CFD and FEM methods, Biomechanics and Modeling in Mechanobiology, Vol. 22, p. 837-850.

Oveissi, F., Naficy, S., Lee, A., Winlaw, D.S., and Dehghani, F., 2020. Materials and manufacturing perspectives in engineering heart valves: a review. Materials Today Bio, Vol. 5, pp. 100038.

Pasta, S., Cannata, S., Gentile, G., et al., 2020. Simulation study of transcatheter heart valve implantation in patients with stenotic bicuspid aortic valve. Medical & Biological Engineering & Computing, Vol. 58, p. 815-829.

Qiu, D., Barakt, M., Hopkins, B., Ravaghi, S., and Azadani, A.N., 2021.Transcatheter aortic valve replacement in bicuspid valves: The synergistic effects of eccentric and incomplete stent deployment. Journal of the Mechanical Behavior of Biomedical Materials, v. 121, pp. 104621.

Rahmani, B., McGregor, C., Byrne, G., and Burriesci, G., 2019. A Durable Porcine Pericardial Surgical Bioprosthetic Heart Valve: a Proff of Concept. Journal of Cardiovascular Translational Research, v. 12, p. 331-337.

Sadrabadi, M.S., Eskandari, M., Feigenbaum, H.P., and Arzani, A., 2021. Local and global growth and remodeling in calcific aortic valve disease and aging. Journal of Biomechanics, v. 128, pp. 110773.

Saleeb, A. F., Kumar, A.,and Thomas, V. S., 2013. The important roles of tissue anisotropy and tissue-to-tissue contact on the dynamical behavior of a symmetric tri-leaflet valve during multiple cardiac pressure cycles. Medical engineering & physics, Vol. 35, p. 23-35.

Singuenza, J., Pott, D., Mendez, S., Sonntag, S.J., Kaufmann, T.A.S., and Steinseifer, U., 2018. Fluid-Structure interaction of a pulsatile flow with an aortic valve model: A combined experimental and numerical study, International Journal for Numerical Methods in Biomedical Engineering, Vol. 34.

Tango, A.M., Salmonsmith, J., Ducci, A., and Burriesci, G., 2018. Validation and Extension of a Fluid-Structure Interaction Model of the Healthy Aortic Valve, Cardiovascular Engineering and Technology, Vol. 9, p. 739-751.

Tayama, E., Saku, T., Anegawa, T., Oryoji, A., and Negoto, S., 2022. Prosthetic cardiac valves: history and review of cardiac prostheses clinically available in Japan, Surgery Today, Vol. 52, p. 521-531.

Wald, S., Liberzon, A., and Avrahami, I., 2018. A numerical study of the hemodynamic effect of the aortic valve on coronary flow,Biomech Model Mechanobiol, Vol. 17, p. 319-338.

Xu, F., Johnson, E.L., Wang, C., Jafari, A., Yang, C.H., Sacks, M.S., Krishnamurthy, A., and Hsu, M.C., 2020. Computational investigation of left ventricular hemodynamics following bioprosthetic aortic and mitral valve replacement, Mechanics Research Communications, Vol. 102, pp. 103604.

Zhu, Y., Wilkerson, R.J., Pandya, P.K., et al., 2023. Biomechanical Engineering Analysis of Pulmonary Valve Leaflet Hemodynamics and kinematics in the Ross Procedure, Journal of Biomechanical Engineering, Vol. 145.

7. RESPONSIBILITY NOTICE

The authors are the only responsible for the printed material included in this paper.