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AORTIC VALVE GEOMETRY MODELING – REVIEW

ABSTRACT

The present work is a review of publications covering computer simulation of aortic valve operation and material properties of aortic valve components studies. Particular attention is paid to the anisotropy of material and geometric properties. The methods of geometric models developing by using specified research methods and/or diagnostic imaging devices are presented. The microstructure of the aortic valve is also described and its impact on material properties definition introduced. The various ways of describing the aortic valve leaflet anisotropic properties are mentioned. Often exploited simplifications and their impact on the simulation results is also presented.

Keywords: *aortic valve simulation; aortic valve geometry; heart valve tissue mechanical properties; aortic valve anisotropy*

INTRODUCTION

Aortic insufficiency being a consequence of root dilation or cusp deformation [1] is the main cause determinant undertaking a surgical repair procedure. Remodeling and reimplantation are currently in practice. Located between left ventricle of the heart and the aorta aortic valve prevents reversing blood flow to the heart chamber. The development of materials science towards increasingly sophisticated biomaterials allowed to replace the mechanical caged-ball and tilting-disc prosthesis with the prosthesis with a geometry similar to the native valve. Currently implantable prostheses are more functional and often do not require anticoagulation therapy. Computer methods to date allow to improve minimally invasive valve replacement approaches (Transcatheter Aortic Valve Implantation) [2, 3, 4] to improve native valve repair methods [5, 6, 7, 8, 9, 10, 11] to increase the functionality of prostheses already in use by changing their geometry [12] or to predict the optimal surgical solution [13]. The Functional aortic valve prosthetic operates dynamically which is conditioned by the nature of the blood flow at the left ventricle outlet. This involves the regions of high tensile and bending stresses formation that after exceeding the critical values can accelerate tissue structural fatigue damage. Stress concentrations within the leaflet may contribute to initiation of calcification by causing structural disintegration [14]. Computer simulation of aortic valve prosthesis operation allows to evaluate its functionality and to determine the extent of its efficiency under defined environment. An apt description of the initial geometry is an essential condition - aortic valve and a blood vessel geometric model, the type of analysis (static/dynamic), a description of the material behavior both valve leaflets and aortic and the behavior of the fluid - blood description. An accurate geometric description of aortic valve greatly influences on the nature of its operation mapping, and selection of simplifications on the numerical computation duration. The present work is a review of publications representing the natural aortic valve construction in terms of both macro- and microscopic. It includes issues which should be paid a special attention to taking on aortic valve simulation in terms of its geometry and way of material properties describing.

INITIAL GEOMETRY

Previous investigations of aortic valve geometry were carried out by casting silicone rubber molds under pressure [15, 16, 17] or utilizing hearts and a freezing technique under the pressure. Swanson and Clark [15] based on experimental studies using the first aforementioned method present the geometric dimensions of the human aortic valve variation as a function of pressure during cardiac cycle. On this basis, they introduced a parametric description of the valve and aortic root. As a reference dimension aortic annulus diameter was chosen because of approximately 10% of it variation with pressure from 20÷120 mmHg. The authors define the various dimensions of the aortic valve with respect to the reference dimension: length measured along the leaflet at the center of the sinus region, the width of the coaptation at the center and free edge length. The dimensions of the blood vessel where the valve is positioned are also presented in the parametric form respectively: sinus diameter, aortic root diameter, sinus of Valsalva length and height from ventricular tract base plane to top of annulus fibrosis. Swanson and Clark [15] results became the basis of other researchers geometric model description [18, 19, 20, 21]. Labrosse et al. [16, 17] have enhanced their database with dimensions [15] of closed normal human valves pressurized at 80mmHg measured by Swanson and Clark because they were obtained in very similar experimental conditions. The results obtained through simulations are consistent with the experimental result. Geometric orifice area measurement error was less than 5% in the experiment and in the simulation case. The authors point out that initial geometry does not include local variations with thicker tissue along the leaflet attachment line what slowed down closing of the simulated valve. This is probably the result of depending on the thickness of the tissue variable leaflet flexural properties.

Aortic valve leaflet is composed of three layers: the ventricularis, spongiosa and fibrosa that are predominantly composed of radially, loose and parallel to the cuspal free edge aligned collagen fibers respectively [22]. The role of collagen fibers is to reduce stresses in the leaflets during cardiac cycle and to provide smoother opening and closing valve [23]. This makes an important to take into account anisotropy of valve leaflets in terms of geometry and material properties in order to map the stress distribution. The leaflet thickness is varied within the physiological range depending on the region and type of leaflet (Fig.1). Regions of the noncoronary leaflet are assigned a greater thickness than corresponding regions in the right and left leaflet [24]. Thickness measurements were performed using micrometer [24], magnetic resonance imaging (MRI) [25] or using contacting profilometry [26]. Joda et al. [21] has simulated three models of aortic valve with different materials description each and concluded that valve model with the physiological leaflet thickness predicted 33% lower stress at the commissure and a 100% higher stress at the belly compared to the uniform thickness model. Similarly, the valve model with the non-uniform leaflet thickness predicted 27% lower strain at the commissure and 80% higher strain at the belly com- pared to the uniform thickness model. Grande et al. [27] based on simulation of aortic valve model generated by MRI concluded that leaflet and sinuses asymmetry affect the asymmetrical stress distribution. Coronary ostium in the right and left sinuses provide relief from the blood pressure loading additionally.

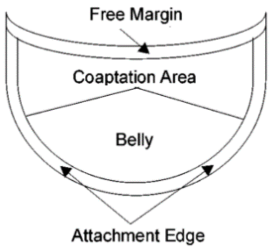

Fig. 1. Aortic valve leaflet regions [27]

The left leaflet is the smallest in terms of area, perimeter, free-margin length, and thickness [24, 28]. Vollenbergh and Becker [29] by measurement of the width and maximal height of each aortic valve's cusp of being in an unfixed state showed that inequality in cusp size occurs in 97,5% of cases. Kunzelman et al. [28] drew similar conclusions. Based on the experimental studies involving the laying a suture along the specified dimension and comparing the suture length to a ruler concluded that the noncoronary leaflet tends to be the largest of the three leaflets. Currently assumed to be equal the dimensions of the three valve leaflets in order to simplify, which amounts to a necessity modeling one-sixth of the valve, extending between one commissure and the middle of a sinus. This assumption simplified the modeling and analysis by solving one-sixth of the model, and reducing computation time. This approach ignores asymmetric deformation during non-linear dynamic analysis [30]. Grande et al. [27] presented the importance of anatomic asymmetry taking into account. Turns out that it affects not only the regional variation of stress and strain especially at the commissure and the belly regions [21] but the coaptation results in the three valve leaflet also. This is particularly important for assessment of the aortic valve model effectiveness. Hammer et al. [31] measured the average pattern of the collagen network in porcine aortic valve leaflet and based on the results of the simulation concludes that the concave pattern of curved fibers produces a closed valve with a 40% increase in central leaflet coaptation height and with decreased leaflet billow, resulting in a more physiological closed valve shape.

Currently initial geometry of aortic valve is generated often based on the data obtained in the diagnostic imaging methods. The great challenge in aortic valve geometry developing based on medical scans is to determine the leaflet thickness. This is due to the approximate density of the tissues and the surrounding blood. One of the methods is magnetic resonance imaging (MRI) measurements of excised human AV and root specimens [25, 27] or measurements often in two-dimensional sections at the same relative time of the cardiac cycle. Many possibilities for simulating the blood flow through the prosthetic valve entails the solution presented by Ionasec et al. [32] who proposed a novel system for patient-specific modeling and clinical assessment of the aortic and mitral valves using computed tomography (CT) and transesophageal echocardiography (TEE). Soncini et al. [33] based on 2D transthoracic echocardiographic (2D TTE) data developed 3D model of the sinuses of Valsalva. Morganti et al. [34] using designed and developed by themselves an algorithm for 3D modeling of the aortic root based on 2D echocardiography (2D Echo) measures presents generated circumferentially asymmetric geometrical models of the aortic root. Haj-Ali et al.

[35] developed 3D geometrical representation of the entire valve, including cusps and root compared to 3D-TEE measurements from non-pathological tricuspid aortic valve (TAVs). TEE and CT has recently become available in 4-D however aortic valve biomechanics simulation presents with challenges that have yet to be met and such volumetric time-resolved data due to its size and complexity is barely exploited in clinical practice [32].

Most of the studies that utilized AV parametric relationships to describe the valve geometry have focused on the cusps and the commissures, but did not include the geometry of the sinuses [16, 36]. Sinuses of Valsalva asymmetry ensures high valve efficiency through appropriate load bearing distribution during diastole [9]. The specific aortic root geometry provides different levels of expansion during ejection in the characteristic geometry planes. Lansac et al. [37] based on four-dimensional study of the aortic root dynamics makes a thesis that these chronologic, geometric changes are the mechanism to maximize ejection. Geometric models defined based on the experimental investigations unlike those based on diagnostic imaging methods often do not include the asymmetrical construction of valve leaflets and corresponding sinuses. To date, no studies on the effects of initial model position depending on the cardiac cycle phase but it is desirable that the initial model would be in the stress-free condition. As the unstressed valve state popular choice is a nearly closed [18, 25, 27, 38, 39, 40, 41] or fully-open configuration [16, 17, 20]. Fully-closed position selection (loaded tissues behavior) [27, 42, 43, 44] requires to assume the size and shape of the contact area in the coaptation regions of the leaflets. Experimental validation of aortic valve prosthesis initial geometry is important for independent validation of the results of the analysis with either experimental data or prior published results in the application of such analyses for complex biological flow phenomena.

MATERIAL PROPERTIES

The anisotropic and nonlinear material properties of aortic valve leaflet tissue, arising from the presence of oriented collagen fiber, affect the leaflet dynamics and stress distribution thereby the associate failure mechanisms. Highly nonlinear stress-strain relationships, large deformations, viscoelasticity, and strong axial coupling are particular challenges in constitutive modeling of aortic valve tissue. Despite this many previous works considered isotropic, linear [33, 45] or non-linear [43] elastic material property to define the constitutive behavior of valve leaflet tissue. Nicosia et al. [25] used linear elastic material properties but developed based on MRI geometric model includes leaflet thickness and normal variations in collagen fiber orientation. De Hart et al. [38] despite that the purpose of research is to analyze the impact on the structural stress state and the associated fluid dynamical flow, the aortic valve leaflets are assumed to behave linear elastic and isotropic. Marom et al. [46, 47] the choice of an isotropic and linear elastic leaflet tissue material properties justify the purpose of conducted simulation, ie. modeling the evolving nonlinear contact mechanism between the valve leaflet. Grande et al. [27] justify the assumption of linear material properties that the aortic valve function mainly in the higher elastic modulus region of its stress–strain curve during diastole. The author argues that the elastic region of the valve stress–strain curve has a greater role in the opening and early closing motion of the valve leaflets, while during the rapid final phase of valve closure, the collagen-dependent region of the stress–strain curve governs the deformational behavior. De Hart et al. [23] draws attention to the fact that fiber-reinforcement has an impact on transvalvular pressure gradient. In many previous works the anisotropic nature of the leaflet tissue is represented by means orthotropic [25, 27, 42, 48], transverse anisotropic [5, 6, 49] or hyperelasticity, what according to other researchers [27, 50] is not capable of capturing the real deformation-dependent anisotropy of aortic valve tissue. Burriesci et al. [51] concluded that even a relatively low presence of orthotropy will significantly alter the stress distributions of the valve. Commonly used simplification that collagen fibers are approximately normally distributed about a mean preferred fiber direction determine the choice of transverse isotropy properties of aortic valve tissue model [52]. Freed et al. [53] developed a transverse isotropic nonlinear constitutive model accounting for dispersion of collagen fibers observed in some tissues and demonstrated the feasibility of the model to represent aortic valve tissue. Koch et al. [20] like Patterson et al. [54] consider that leaflet tissue material models described as the nonlinear elastic are more responsive to the time-varying pressure wave which induced lower compressive but higher tensile stresses in the leaflets. By conducting static finite element analysis investigated the influence of the non–linear and anisotropic material properties of aortic valve leaflet. Koch analyzed four material models: isotropic linear elastic (ILE), transversely isotropic linear elastic (TILE), isotropic hyperelastic (IHYP), and transversely isotropic hyperelastic (TIHYP). Differences in the results of the four models were readily seen in diastole what confirms the validity of the obtained results and made thesis by predecessors. Peak values of stress were greatly reduced in the transversely isotropic models, and TIHYP model is characterized by the highest value of the amount of coaptation surface (45% considering the ratio of coaptation area to total leaflet area). Labrosse at al. [17] develop a novel element model of the natural aortic valve with transversely isotropic and hyperelastic material properties. It is also used in aortic root tissue modeling. Despite the fact that aortic root wall is known to be highly nonlinear and anisotropic [55, 56, 57] material models are frequently simplified as isotropic with no directional dependence [21, 58]. The basis for determining of aortic root material constants is commonly used tensile test.

SUMMARY

Depending on the specific aim of the study, certain simplifications may be reasonable as they will not greatly influence the results of that study. Geometrical simplifications and assumptions of symmetry are also reasonable if the researchers are not concerned with local variability and aim to characterize general patterns or to isolate the effects of one globally applied variable. Junction area of aortic valve with ascending aorta called commissure is exposed to the high stress therefore an accurate definition of material properties model is a major issue. Native heart valve tissue is most accurately described as a nonlinear, pseudoelastic, and anisotropic multilayered material. Selection of appropriate material model describing aortic valve leaflet behavior affects the assessment of valve efficiency e.g. by analyzing Effective Orifice Area changes and also to evaluate the functionality through coaptation surface area analysis. For a dynamic study of the full cardiac cycle, the nonlinear elastic and anisotropic properties of the valve should be considered to obtain realistic results.The geometrical and material properties should be defined to most accurately model the valve for the purpose of the study, while still maintaining computational costs at an acceptable level.

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