A FLUID STRUCTURE INTERACTION MODEL OF PORCINE AORTIC VALVE WITH PHSYOLOGIC TISSUES PROPERTIES AND FLOW BOUNDARY CONDITIONS

Gil Marom[*], Hee-Sun Kim[**], Rami Haj-Ali[*], Ehud Raanani[+], Hans-Joachim Schäfers[++] and Moshe Rosenfeld[*a)]
[*]*School of Mechanical Engineering, Tel-Aviv University, Tel Aviv 69978, Israel*
[**]*Department of Architectural Engineering, Ewha Womans University, Seoul 120-750, South Korea*
[+]*Department of Cardiothoracic Surgery, Chaim Sheba Medical Center, Tel Hashomer 52621, Israel*
[++]*Department of Thoracic and Cardiovascular Surgery, University Hospitals of Saarland, Homburg, Saarland 66421, Germany*

Summary A full fluid-structure interaction (FSI) model of porcine aortic valve and root is presented. The tissue model recognizes the anisotropic hyperelastic behaviour of the highly compliant leaflets and includes a contact algorithm in the coapting regions. Physiological blood pressures are imposed at the upstream and downstream boundaries. The model includes all the four phases of the valve dynamics: opening, systole, closing and diastole. The calculated flow field and the stress distribution are given in various times during those phases.

INTRODUCTION

The kinematics and dynamics of the aortic valve are highly dependent on the combined mechanical properties of the valve and the aortic root, as well as the blood flow. Previous numerical models of physiologic aortic valve and root, that included coaptation under the full cardiac cycle, ignored the influence of the blood flow (performing “dry” simulations). Fluid-structure interaction (FSI) models of compliant aortic valve were forced to use non-physiologic blood pressures or unrealistic material properties in order to overcome various numerical difficulties [1]. The aim of the present study is to present a full FSI model of healthy porcine aortic valve and root with physiologic blood pressure and tissues properties.

METHODS

Three-dimensional (3D) geometry of a porcine aortic valve and root was reconstructed using dimensions and geometric relationships similar to those suggested by Thubrikar [2]. The cusps and the root were assumed to have separate but tied geometries; the cusps moved with the root at their joint boundaries. Two straight rigid tubes were added upstream and downstream, respectively, to move the flow boundary conditions away from the regions of interest.

The structural solver employed an implicit nonlinear dynamic analysis with an implicit direct displacement-based finite element method. The Collagen Fiber Network (CFN) material model [3] was employed for the leaflets tissues. The CFN model explicitly recognizes the collagen fibers and elastin matrix using different layers of elements. The hyperelastic behavior of the two materials and the radii of the fibers in various regions of the leaflet were calibrated to published experimental stress-strain curves from porcine aortic valve. The tissue of the aortic root was assumed to be isotropic and hyperelastic based on available experimental data of porcine aortic sinuses. All the tissues were assumed to be almost incompressible. A master-slave contact algorithm was employed between the cusps themselves as well as with the sinuses, assuming a non-friction contact. The mesh of the structural part has more than 50,000 elements, which was found to be adequate following a mesh refinement study [3].

The flow was assumed to be laminar and the blood to be Newtonian and isothermal. To improve the convergence of the FSI model the blood was assumed to be slightly compressible. A finite volume method was employed for solving the unsteady 3D Navier-Stokes and mass conservation equations using the Eulerian approach with a Cartesian mesh. Physiologic time dependent pressures were employed at the upstream and downstream boundaries, representing the pressures at the left ventricle and the ascending aorta, respectively. The flow domain was discretized with a mesh of approximately 700,000 elements following a mesh refinement study [3]. The mesh was dynamically adapted in the vicinity of the walls, including near the moving leaflets and the compliant root. The implicit flow solver used a spatially second order upwind scheme as well as a second order temporal discretization.

The FSI model was solved by a partitioned solver with non-conformal meshes, which facilitated large structural motion through the fluid domain and simplified the introduction of contact. The sequential two-way coupling process exchanged the structure displacement and velocities, and the calculated traction load (including the fluid pressure and shear) between the solvers. The structural problem was solved by Abaqus 6.10 (Dassault Systems, Simulia Corp., Providence, RI, USA) finite element software, while FlowVision HPC 3.08 (Capvidia NV, Leuven, Belgium) was the flow solver. FlowVision Multi-Physics manager (Capvidia NV, Leuven, Belgium) managed the coupling of the two codes.

[a)] Corresponding author. Email: rosenf@eng.tau.ac.il.

RESULTS

Figure 1 presents the flow field and stress distribution at three different instances (starting from the beginning of the opening). For each time, the blood pressure distribution and the velocity vectors on the symmetry, similar to the Echocardiography measurement plane, are shown (second column) together with the maximum principal stress contours on the deformed valve and root in a cross section view (third column). The location of the collagen fibres are also marked in the third column by the red lines. The fibres are presented above the deformed valve since the fibres in the model are between the layers of elastin and cannot be seen. The plots of t=50ms represent the opening phase when the pressure in the ventricle is larger than the pressure in the aorta and the large flow velocity from the ventricle opens the valve. At the end of the systolic phase (t=150ms, second row) the direction of the flow starts to reverse. In both of these phases the low stress in the leaflets might be related to the relatively low pressure difference on the leaflets. During diastole (t=250ms), the large pressure difference on the leaflets closes the valve and the coaptation height can be clearly seen in the 2D section. The large pressure difference leads to the creation of larger stress in the belly region relative the stress in the coapting region. The pressure difference is negligible in this region.

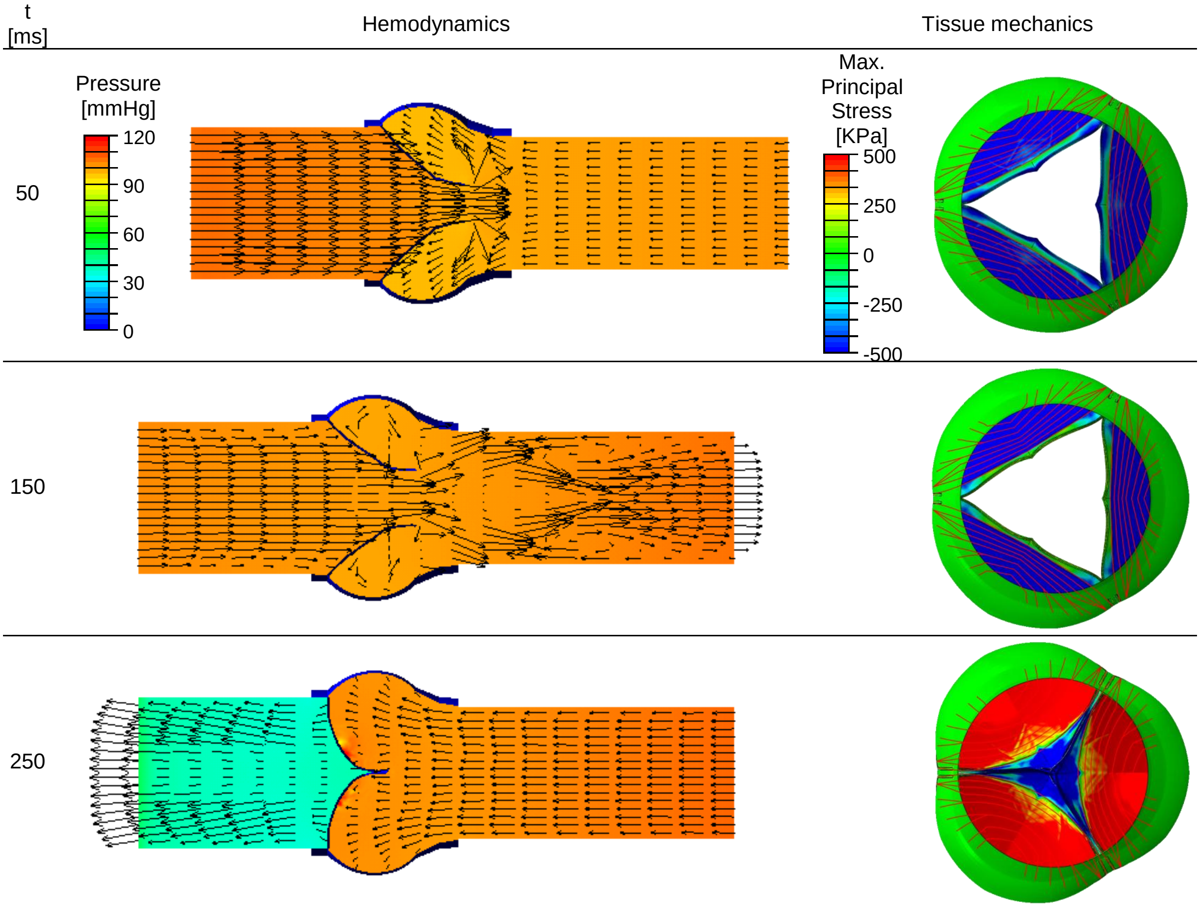

Figure 1 The flow field and stress distribution at three different instances

CONCLUSIONS

A 3D FSI model of natural aortic valve with compliant root, physiological blood pressure and realistic material properties has been developed. This model allows, for the first time, to model accurately the aortic valve kinematics and dynamics in realistic conditions.

References

[1] Chandran K.B.: Role of computational simulations in heart valve dynamics and design of valvular prostheses. *Cardiovasc. Eng. Technol.* 1:18-38, 2010.

[2] Thubrikar M.: The Aortic Valve. CRC Press, Boca Ralon FL 1990.

[3] Kim H.S.: Nonlinear Multi-Scale Anisotropic Material and Structural Models for Prosthetic and Native Aortic Heart Valves, *PhD thesis*, Georgia Institute of Technology, School of Civil and Environmental Engineering, Atlanta, GA, USA, 2009.

[4] Marom G., Haj-Ali R., Raanani E., Schäfers H.J., Rosenfeld M.: A fluid-structure interaction model of coaptation in fully compliant aortic valves. *Med. Biol. Eng. Comput.* DOI: 10.1007/s11517-011-0849-5, 2011.