COMPUTATIONAL ANALYSIS OF PATIENT-SPECIFIC AORTIC VALVES

E. Stupak [1]***, A. Kačeniauskas***[1]***, V. Starikovičius***[1]***, A. Maknickas***[1]***,***
R. Pacevič[1]***, M. Staškūnienė***[1]***, G. Davidavičius*** [2] ***and A. Aidietis***[2]

[1] *Vilnius Gediminas Technical University, Vilnius, Lithuania*
[2] *Vilnius University Hospital "Santariškių Klinikos", Vilnius, Lithuania*

1. Introduction

Aortic stenosis is one of the most common valvular disorders encountered in clinical practice and its prevalence is expected to increase in industrialized countries [1]. Sufficiently accurate computational models combined with medical imaging can serve as an inexpensive tool for medical research providing fundamental information for the improvement in patient care [2].

2. Methods

The electrocardiographically gated 4D images were acquired from a human subject by using the Philips iE33 ultrasonographic system (Philips Healthcare, Andover, MA, USA) to obtain the patient-specific geometry of the aortic valve. The geometric parameters of the aortic valve were extracted from acquired images by using the Medical Imaging Interaction Toolkit (MITK). A 3D geometric model was constructed from hypocycloid and epicycloid parametric curves according to the extracted parameters. The patient-specific geometry of the aortic valve was defined by NURBS surfaces generated by using Blender. In order to import the geometrical model into computational software some Boolean operations on NURBS surfaces were performed in Salome. Finally, the patient-specific geometry was imported into the ANSYS DesignModeler to generate the finite element meshes and finite volume meshes for solid and flow, respectively.

Pulsatile blood flow through the aortic valve was described by incompressible Navier-Stokes equations. The local Reynolds number of the flow past aortic valve reach high values during the deceleration phase following peak systole, therefore, turbulence models were applied to improve convergence of the numerical solution and to avoid the explicit simulation of the smallest scales. The traditional k-ε model and the shear-stress transport k-ω model enriched by low Reynolds number modifications and intermittency transition equation were investigated. Structural model for the determining of stress-strain state in aortic valve was constructed. The displacements formulation was used in equations of dynamics. The differential equations of the flow and the solid were numerically solved by using the ANSYS Fluent and the ANSYS Structural, respectively.

3. Results

The computational analysis was performed on the OpenStack cloud infrastructure. The blood was modelled as an incompressible Newtonian fluid with density set to ρ=1060kg/m^3. The dynamic viscosity coefficient was μ=0.004028kg/(m s). The systolic phase of the cardiac cycle was analyzed by applying a time-dependent velocity of the plug flow as the inflow boundary conditions. The velocity curve was defined by using Doppler measurements and analytic approximations. The no-slip boundary conditions were prescribed for velocity on the aorta walls and leaflet surfaces. At the outflow the prescribed pressure and zero velocity gradient normal to the boundary were applied. The nominal outflow boundary was located 0.1m downstream of the leaflets to recover toward the fully developed state. The shorter solution domain was also considered in the sake of comparison. Almost identical velocity values were observed in the case of both solution domains. In the case of laminar flow, mesh independent numerical solution was achieved on meshes consisting of about 1.5

million finite volumes. Figure 1 illustrates complex 3D flow patterns in the longitudinal section of aortic sinuses. The velocity field was visualized by using streamlines coloured according to pressure field, which helped identifying zones of vortices. A pair of vortices was developed in each sinus, while other vortices were formed behind the junction of two leaflets. It is worth noting that high Reynolds number turbulence model significantly smoothed the vortices. The results of the shear-stress transport *k-ω* model were significantly closer to the laminar flow.

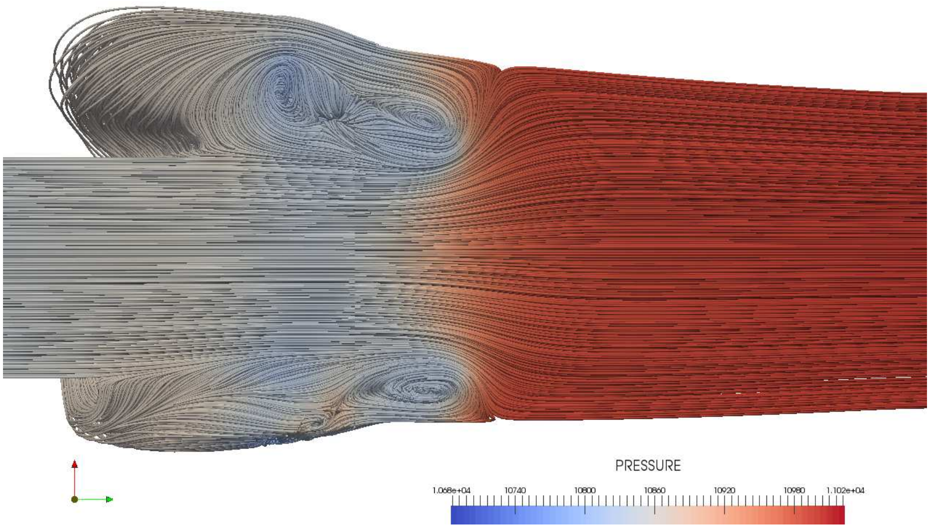

Figure 1. Complex flow pattern in the aortic sinuses.

Structural analysis was performed using three FE models consisting up to 0.4 million FEs with the prescribed pressure drop on leaflets as in the peak systole conditions. Anisotropic Fung-type hyper elastic material having specific weight of 1000kg/m^3 was considered. Solution was performed in iterative way using program controlled time substep size. The obtained Von Mises stresses are of similar values as in reference [3].

4. Conclusions

The computational analysis of the patient-specific aortic valve is presented. Backflow through the outlet and complex 3D blood flow patterns present challenges to numerical schemas and computational software. The shear-stress transport *k-ω* model improved convergence of the solution and revealed the vortex pattern close to the pattern observed in the laminar flow. The computed values of Von Mises stress were in agreement with the results found in the literature.

Acknowledgements

The presented research was supported by the Research Council of Lithuania in the framework of the project “Patient-specific simulation of the flow through the aortic valve” (MIP-052/2014).

References

[1] B. Iung and A. Vahanian (2011). Epidemiology of valvular heart disease in the adult, *Nat. Rev. Cardiol.*, **8,**162–172.

[2] C. Chnafa, S. Mendez and F. Nicoud (2012). Image-based patient-specific simulation: a computational modelling of the human left heart haemodynamics, *Comput. Methods Biomech. Biomed. Eng.*, **15**, 1–3.

[3] F. Sturla, E. Votta, M. Stevanella, C.A. Conti, A. Redaelli (2013). Impact of modeling fluid–structure interaction in the computation analysis of aortic root biomechanics, *Med. Eng. Phys.*, **35**, 1721–1730.