AORTIC VALVE ENGINEERING BY ADVANCED COMPUTATIONAL TECHNIQUES

J. de Hart (1,2), F.P.T Baaijens (2), B.A. de Mol (1,2) and F.N. van de Vosse (2)

(1) Academic Medical Center
Department of CardioThoracic Surgery
P.O. Box 22660, 1100 DD Amsterdam
The Netherlands

(2) Eindhoven University of Technology
Department of BioMedical Engineering
P.O. Box 513, 5600 MB Eindhoven
The Netherlands

INTRODUCTION

The development of prosthetic heart valves is primarily based on the time consuming trial-by-error concept involving many animal tests. The integration of computational analyses in aortic valve research is therefore very appealing as it is time and cost efficient. Moreover, these analyses enable to investigate performance characteristics, such as (shear) stresses, vortical flow and wash-out, which are difficult to evaluate experimentally. The mechanics, hemodynamics and kinematics of an aortic valve model have been studied using computational techniques, which incorporate the blood-valve interaction. The results can be used to define design criteria for improved (long-term) functioning.

METHODS

The valve model is based on the Finite Element Method (FEM) [1]. Trileaflet symmetry is adopted and the compliant aortic root is assumed to be isotropic. The leaflets are reinforced with collagen giving the structure physiological material characteristics (Figure 1). Time-dependent flow and pressure variables are applied to load the system during the systolic phase taking the interaction of the leaflets with the blood fully into account.

In modeling fluid-structure interaction, the fluid domain is most conveniently described with respect to an Eulerian reference frame while a Lagrangian formulation is more appropriate for the structure. These formulations, however, are incompatible. A solution to this is an Arbitrary Lagrange-Euler (ALE) formulation for the fluid domain, which involves a continuous adaptation of the mesh without modifying the topology. In this method the fluid domain follows the expansion of the aortic wall, which is caused by rising internal fluid pressures.

With respect to the large leaflet motion within the computational fluid domain it is generally difficult, if not impossible, to adapt the fluid mesh in such a way that a proper mesh quality is maintained without changing the topology. Alternatively, remeshing can be performed, either continuously, using a Lagrangian formulation, or in conjunction with an ALE formulation, where remeshing is performed if the mesh quality has degenerated too much. However, besides introducing artificial diffusivity, remeshing may be difficult to perform with sufficient robustness and accuracy for three-dimensional problems.

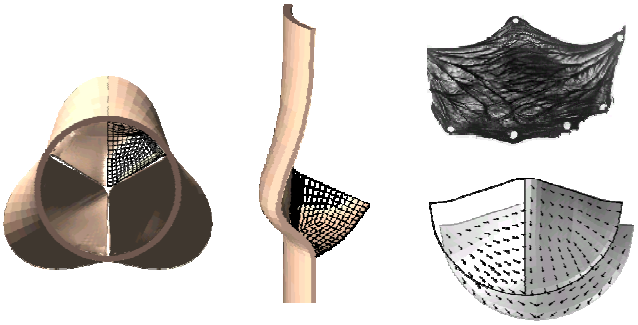
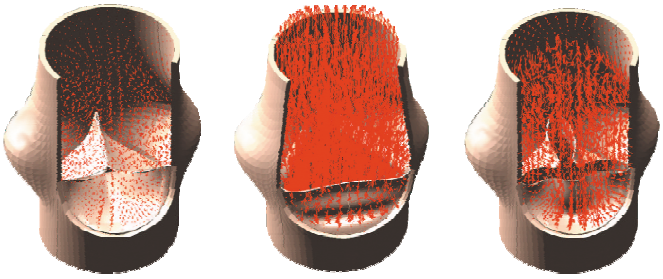


Figure 1: Aortic (a) and side (b) view of the model, and natural and implemented collagen reinforcement (c).

To circumvent these difficulties a "fictitious domain" (FD) method is used where the fluid and structure are described in an Eulerian and Lagrangian setting, respectively. The method is based on the imposition of velocity constraints associated with moving internal boundaries by means of Lagrange multipliers [2,3].

RESULTS

Aortic root dilation and the interaction with surrounding blood flow dominate opening and closing of the valve. In some cases aortic root compliance accounts for 80% of the valve opening provided that intercommissural stretch of the leaflets is prevented. The interaction of the leaflets with the blood flow is essentially present in mid- and late-systolic phase (see Figure 2).



(a) (b) (c)

Figure 2: Fluid dynamics at three successive time points in the cardiac cycle (a to c).

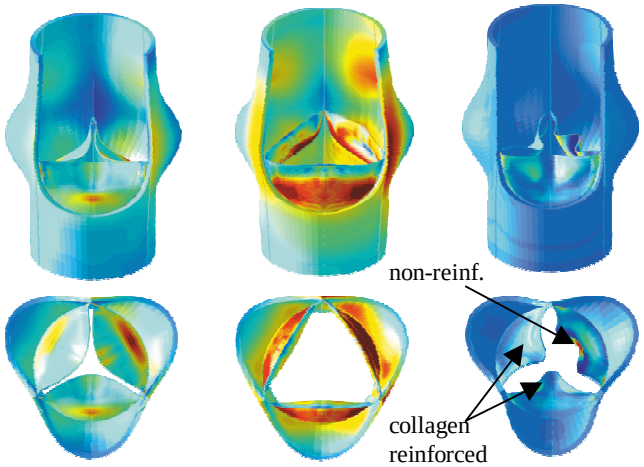
Collagen fiber reinforcement of the leaflets reduces mechanical stresses up to 60% and provides a much more homogeneous stress distribution over the leaflets (see Figure 3). Moreover, the motion of the leaflets is substantially stabilized by these fibers (reduced fluttering in the main stream), while the hemodynamics are preserved. Stress reversal caused by opposite deformation patterns of the closing valve is significantly reduced in late-systolic phase by the collagen (see Figure 3(c)). The low bending stiffness of the fibers and small strains during initial opening allows a transvalvular pressure gradient approximately similar to the non-reinforced case.

DISCUSSION

A combined FD/ALE method is used to analyze aortic valve functioning with fully coupled fluid-structure interaction. The method provides a setting, where conventional formulations for the fluid and structure can be maintained. The stability and accuracy of the method is, however, strongly dependent on the chosen discretizations of the fluid, structure and Lagrange multipliers [2]. An accurate prediction of flow near the leaflets remains complicated, since the fictitious sources that represent the internal boundaries require an interpolation of the fluid velocity to the leaflet interface.

During systole the valve leaflets are moving in an essentially kinematical process initially governed by aortic root expansion and subsequently by the fluid flow. The collagen fibers and aortic root compliance have a major impact on the long-term performance of the valve. The opening configurations are mainly determined by commissural outward motion and are similar for reinforced and non-reinforced leaflets. Compared with valves contained in a rigid root the leaflets are much less subjected to bending resulting in lower stresses. The collagen fibers stabilize the leaflet motion during mid-systole and prevent high bending deformations of the leaflet free edges during valve closing.

The model can serve as a design tool to improve prosthetic devices or to develop new valve concepts. Calcification, aortic (root) aneurysm or remodeling aspects can readily be studied using such computational models. Simulations can be made available to clinicians to diagnose the interaction between pathology and hemodynamics, and their future development with and without intervention.



(a) (b) (c)

Figure 3: Stress state at three successive time points in the cardiac cycle (a to c).

References:

1. Hughes, T.J.R., 1987. "The Finite Element Method". Prentice-Hall, Inc., New Jersey.
2. De Hart, J. , G.W.M. Peters, P.J.G. Schreurs and F.P.T. Baaijens, 2003. "A three-dimensional computational analysis of fluid-structure in the aortic valve". J. Biomech., Vol. 36, pp. 103-112.
3. Baaijens, F.P.T., 2001. "A fictitious domain/mortar element method for fluid-structure interaction". Int. J. Num. Meth. In Fluids, Vol. 35, pp. 743-761.