# Simulating the Motion of Heart Valves Under Fluid Flows Induced by Cardiac Contraction

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## ABSTRACT

The motions of natural and replacement valve leaflets are complex functions of a large number of interactions. The principle concern of the simulation work is the investigation of how natural valves operate whilst attached to a deformable aorta close to its connection with the left ventricle. This involves the interaction of a fluid flow with soft, highly deformable structures. LS-DYNA was used to analyse the system using a series of models with fluid-solid interaction. A ventricle model helped create a detailed prediction of the temporal and spatial variation of flow into the aorta as the ventricle contracts. The principle input to this model was experimental data on displacements. This allowed a flow pattern to emerge naturally in the ventricle. This flow pattern was then available for input into the aortic valve model. The creation of this was a significantly non-trivial task. There are several aspects whose computational demands can be mutually destructive without care in the modelling. These include the effects of contact, spatially moving flow gradients, and the large deformations of the aortic wall and the sinuses. The development of the aortic valve model used linear elastic properties for the different solid materials. Subsequently the simulation progressed to using fully non-linear properties. The paper highlights some of the difficulties encountered and the solutions found, as well presenting some of the results.


## INTRODUCTION

In a natural aortic valve the blood ejected from the left ventricle during systole induces pressure gradients over the valve leaflet surface causing it to displace and change its shape. In turn, these displacements cause changes in the flow field, eventually shutting it off as the valve closes. This also happens in a replacement bioprosthetic valve. In the past, very strong coupling within the Fluid-Structure Interaction (FSI) effects has made it difficult to make detailed studies of bioprosthetic heart valves in anything like their expected operating conditions. Experimental methods (Schoephoerster and Chandran, 1991) allow the fluid and valve to be studied, but are often expensive and time consuming. Neither do they provide detailed structural data. Alternatively, computational studies can provide data for design evaluation. But traditional Computational Fluid Dynamics (CFD) analyses are restricted to flow fields around fixed geometries (Stevenson and Yoganathan, 1985) and Structural Finite Element Analysis (FEA) systems require fluid load contributions to be specified a priori (Patterson et al, 1996; Yoxall, 1997; Chew et al, 1999).

Two, separate but linked, cardiac investigations are described here. The first model studies the variation of fluid velocity at the aortic aperture outlet from a left ventricle. Imposing suitable displacements on the endocardium wall (see below for details) created the motions. The output velocity data was then used as the input “load” history for a second model, that of the aortic root, valve and sinus region.

The study of the details that emerge from analyses like this can help in making decisions as to potentially useful modifications to valve design (Black et al, 1989). For this to be acceptable, appropriate levels of confidence first need to be generated in the predictive accuracy of the numerical output. Comparison with experimental observations (Lawford, 1998) is used here and provides the most important of these confidence-building tests. The analyses demonstrate the enormous potential of the FSI facilities that are becoming available in LS-DYNA.

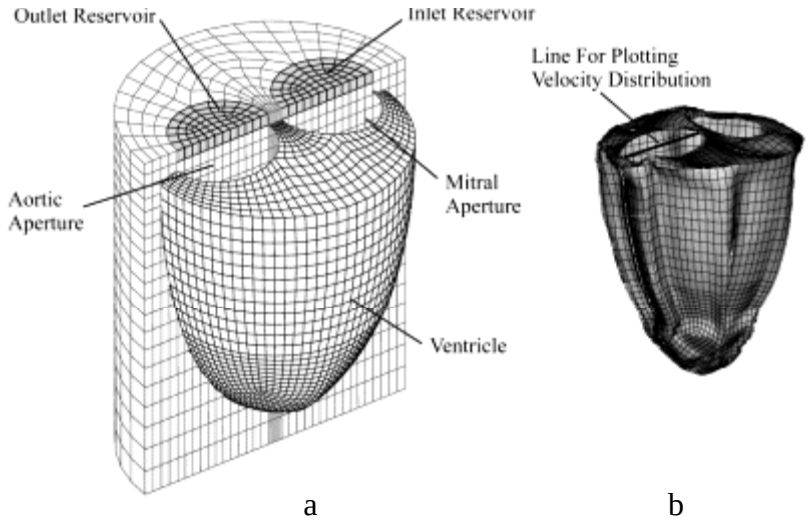


Figure 1. The model of the left ventricle: (a) a section of the fluid control volume revealing the undeformed mesh of the ventricle; (b) the shape of the ventricle after deformation.

## APPROACH

All geometry was created using ANSYS version 5.4 as the pre-processor. The neutral file output from ANSYS (relating to the finite element model) was converted into the required format for LS-DYNA using several filter programs written by the authors and their colleagues. Computational limitations dictated that the ventricle and valve models had to be solved separately.

*Ventricle Model*

The primary aim of the ventricle model was to provide temporal velocity data at the aortic outlet due to ventricle wall motions. It was decided to model the ventricle as the endocardium and to ignore material response. The ventricle model is shown in Figure 1a where the front portion of the fluid control volume has been removed so that the ventricle structure is more easily viewed. The dimensions used for the geometry were taken from some experimental work published by Schoephoerster and Chandran (1991) that concerned flow within the ventricle. Two apertures were placed in the top (base) of the ventricle to provide an inlet (mitral aperture) and outlet (aortic aperture) for the fluid. The ventricle was discretised using 3168 4-noded Belytschko-Tsay shell elements that were specified as 'shells-in-solids' so that interaction with the fluid was established. Material properties for the ventricle were assumed to be linear elastic with a modulus of 3 MPa and Poisson ratio of 0.45 with a constant shell thickness of 1mm throughout the structure. Details of ventricle wall motion were obtained from results of a Magnetic Resonance Image (MRI) study conducted by Maier (1992). This work gave details of rotations and radial motion during end diastole, mid systole and end systole at four different segments (posterior, anterior, septum and inferior) in three planes (base, mid and apex) of the left ventricle. The data was smoothed and mapped onto the relevant regions of the structure so that the ventricle wall was moved along a vector that represented the experimental results. Appropriate constraints were placed at the apertures to prevent axial motion of the structure and to keep it aligned

with the control volume mesh. The squeeze and twist effects of the contracted ventricle can be seen in Figure 1b.

The fluid control volume was constructed with three different meshed sections to match the ventricle structure placed inside it. In addition to the main flow domain, there were two separate inlet and outlet reservoirs that matched the mitral and aortic apertures of the ventricle (viewed as the shaded portions in Figure 1a). The control volume was meshed using a total of 8640 8-noded solid elements with the reservoirs specified as Eulerian ambient elements that permit fluid to be supplied or removed from the control volume. The fluid within the main control volume was specified as Newtonian with a density of 1000 $kg/m^3$ and a viscosity of $0.001Ns/m^2$.

Although the primary objective of the analysis was to record outlet flow due to ventricle wall motion, it was first necessary to fill the structure and then prevent fluid flowing back through the mitral orifice during the systole phase. Applying a pressure to inlet reservoir allowed filling to be achieved and specifying the velocity as zero across the mitral inlet prevented back flow. In all cases, loading was specified over a time span, which made it possible to activate it at the appropriate instant during the analysis. In addition to the wall motion analysis, the use of transient loading also enabled the same model to simulate pulsatile flow conditions within the ventricle. For this study, the ventricle wall motions were specified as zero and the pressure load was varied over time.

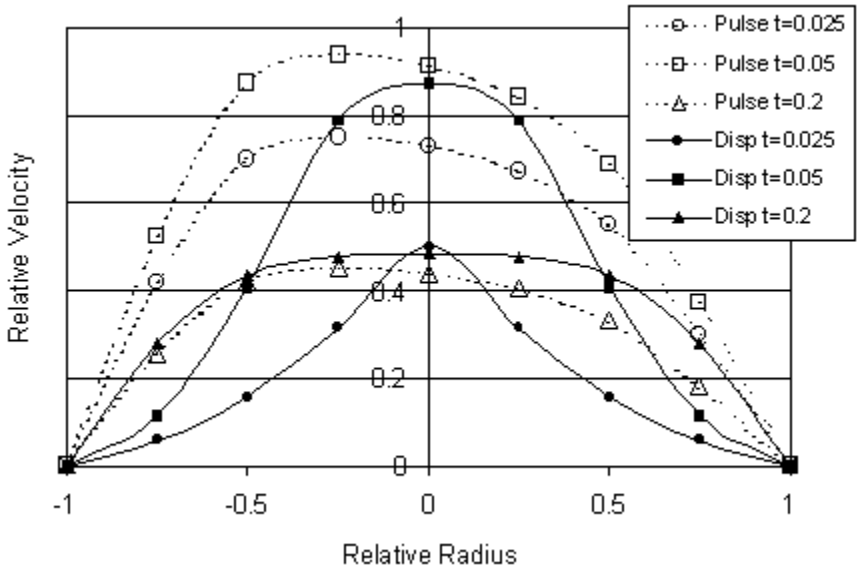


Figure 2. The normalised velocities at the aortic apertures according to the two ventricle models.

The aortic aperture velocity results (see Figure 1b for position) are shown in Figure 2. The velocity (relative to the maximum) is plotted against radius for both wall displacement and pulsatile models at three different times during systole (t=0.025 during early flow, t=0.05 during developed flow and t=0.2 during late systole). In both cases, the results were in agreement with physiological values (i.e. 0-1 m/s). It was further noted that the outlet velocity profile for the pulsatile analysis was not symmetrical however the corresponding results for the wall motion model showed much better symmetry.

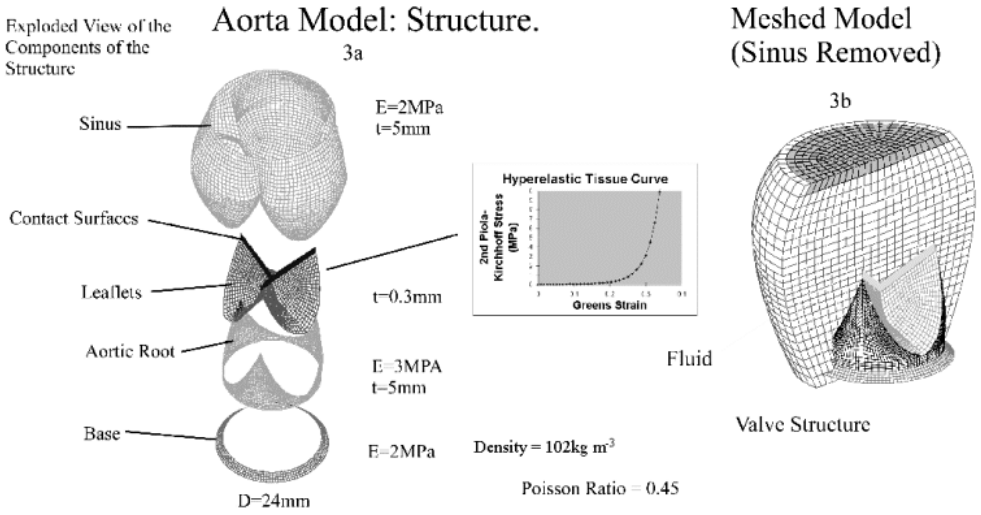


Figure 3. Exploded view of the aortic valve structure showing (a) the dimensions and material properties used, and (b) the leaflets of the valve embedded in the containing fluid control volume.

*The Valve Model*

The valve was taken as having a base diameter of 24 mm, with the remainder of the leaflets being generated in accordance with parameters taken from work done by Thubrikar (1990). Similarly dimensions used for the sinus regions were taken from work published by Swanson and Clark (1974). The model of the structure was sub-divided into several areas that enabled the aorta, base, leaflets, coaption regions (see Figure 3) to be selectively meshed with different material and sectional properties. The valve structure was meshed with 7608 4-noded Belytschko-Tsay shell elements and a further 480 elements were used in three sets of master-slave contact regions along the leaflet free edges. Non-linear hyperelastic tissue properties were assigned to the leaflet elements and linear elastic material characteristics were adopted for all other regions. The two ends of the structure were axially constrained to prevent rigid body motion.

The overall size of the control volume was such that it allowed for the radial expansion of the valve structure. It was created in four different sections using 10368 8-noded solid elements. The valve model is shown in Figure 4 where the front portion of the fluid control volume has again been removed together with the sinus region so that the valve structure is more easily viewed. The sections were the main flow domain with inlet, outlet and outer reservoirs that enabled the main region to remain filled at all times during the analysis. The fluid within the main control volume was specified as Newtonian with a density of 1000 $kg/m^3$ and a viscosity of $0.001Ns/m^2$. Loading of the control volume was achieved through taking the temporal velocity results from the outlet of the ventricle model and applying them over the inlet reservoir (which corresponds to the base of the valve) of the valve model. This was realised by writing some additional Fortran code.

The model was solved for 0.1 s (approximately one third of systole). Under normal circumstances an aortic valve is expected to open in a physiological time of about 0.04s (see Figure 5), at this point in the analysis the leaflets were very close to the fully open position. At approximately 0.05s the model valve had opened fully.

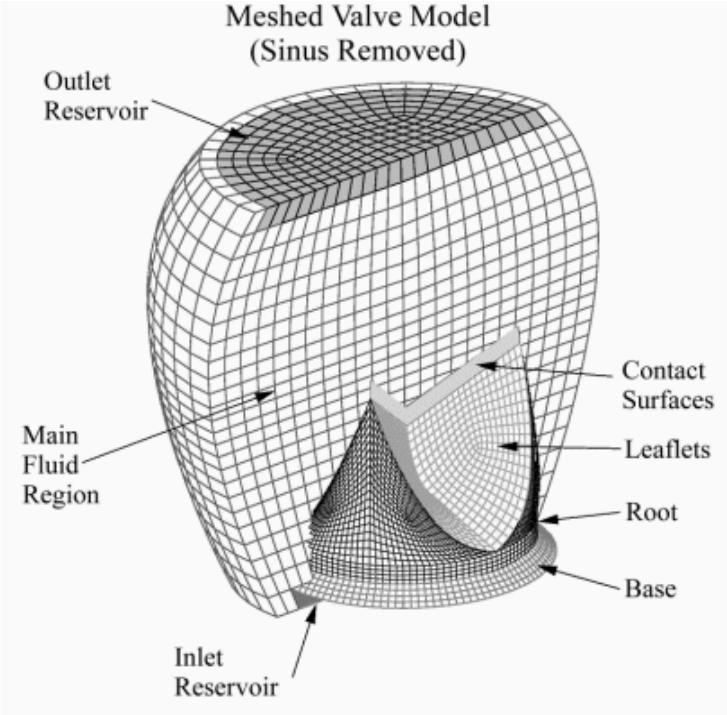


Figure 4. The aortic valve model, showing the inlet and outlet reservoirs. A third reservoir covers the lateral surface, allowing for lateral flow as the sinuses (not shown) and aortic wall expand and contract.

## DISCUSSION

Both models showed good agreement with physiological behaviour. The velocity results from both of the ventricle models are taken at times that relate to different phases of the cardiac cycle. At t=0.025s flow is accelerating and the valve has just started opening and at t=0.2s flow deceleration has started and the valve should be closing. The other condition (t=0.05s) relates to the fully opened valve where flow rates are usually at peak values. Both pulsatile and wall displacement models predicted velocities that would be considered in agreement with physiological velocities however the shape of the velocity profile differed considerably. The velocity profile for the wall displacement model remained symmetrical throughout the analysis (viewed as solid lines in Figure 2) however there was asymmetry present in the pulsatile model velocity profile. Throughout the pulsatile analysis, the peak velocity at any given time tended to be offset from the aperture centre towards the ventricle wall. The valve model aortic velocity predictions and leaflet opening observations were comparable with experimental work. These findings suggest that fluid-structure interaction models have the potential to be a beneficial part of the design process of bioprosthetic heart valves.

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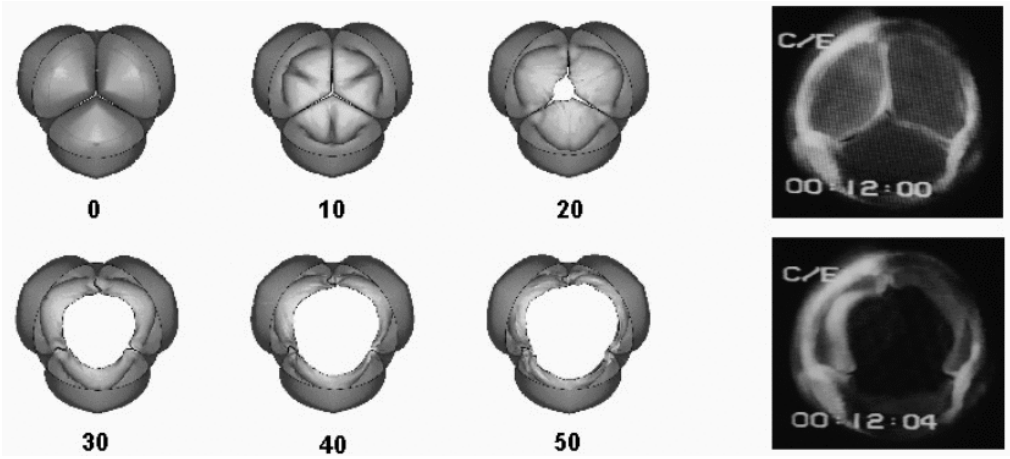

Figure 5. Opening sequence of the model (on the left) in comparison with two images (on the right) from a pulse duplicator study of the opening phase of a bioprosthetic valve.

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