The use of LS-DYNA fluid-structure interaction to simulate fluid-driven deformation in the aortic valve

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ABSTRACT

The physiology and anatomy of the human body is so complicated that viable simulations of even only parts of it require severe simplification. Accordingly, for our work on the mechanics of natural and replacement aortic heart valves, we have simplified into two sub-models the interactive complexity by which blood is ejected from the left heart ventricle to open and close the aortic valve. One model generates the distribution of velocity across the aortic aperture, and this velocity field controls the fluid input into the second model, that of the aorta and valve.

The three-dimensional model of the left ventricle is driven by applied wall displacements and it generates data for the spatially and time-varying blood velocity profile across the aortic aperture. This data then forms the loading conditions in another three-dimensional model, that of the aortic valve and its surrounding structures. Both models involve fluid-structure interaction and simulate the cardiac cycle as a dynamic event. Confidence in the models was obtained by comparison with data obtained in a pulse duplicator. The results show a circulatory flow being generated in the ventricle, a flow that produces a substantially axial motion through the aortic aperture. The aortic valve behaves in an essentially symmetric way under the action of this flow, so that the pressure difference across the leaflets is approximately uniform.

These results support the use of spatially uniform but temporally variable pressure distribution across the leaflets in dry or structural models of aortic valves. Many valve design studies have relied upon such dry modelling, and so the evidence of the present work helps to underpin the value of previous dry work. Scientifically, the study is a major advance through its use of truly three-dimensional geometry, spatially non-uniform loading conditions for the valve leaflets and the successful modelling progressive contact of the leaflets in a fluid environment.

INTRODUCTION – natural and replacement aortic valves

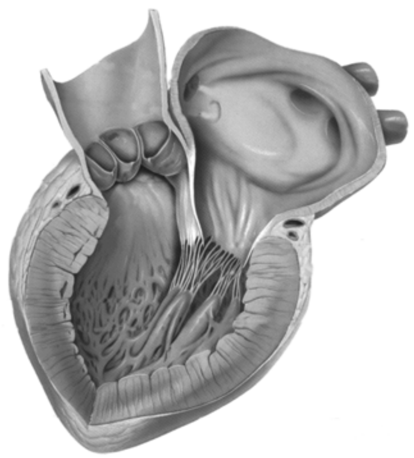

Figure 1: Section through the left ventricle of the heart showing the leaflets, sinuses and base of the aortic valve, the aorta and the mitral valve

The heart of a resting human adult pumps about 5 litres of blood per minute, a rate that is equivalent to at least 7,200 L per day, more than 100 times the weight of the body. Furthermore this rate can increase to 20 or 30 litres per minute during strenuous exercise. This flow pumps blood containing oxygen and nutrients to the organs and tissues of the body. It takes waste products to the kidneys and de-oxygenated blood to the right side of the heart itself from where it is pumped to the lungs. Four major valves control the flow inside the heart, preventing blood from flowing in the wrong direction within the heart chambers. Figure 1 shows in diagrammatic form the layout of the heart with the positions of two of the valves.

Some people are born with defective valves, and others suffer illnesses, like rheumatic fever, that damage their previously healthy valves. People are living longer In western societies, but increasing numbers lead sedentary life styles and consume highly processed food, factors that many experts now believe contribute to the coronary problems that can bedevil middle- to late-life. Amongst these general coronary problems, valve disease is a significant factor. All these people can be helped by operations that replace the defective valve with an artificial one, and, of these, the aortic valve is the one that needs the greatest number of replacements.
Replacement valves can be either mechanical or tissue (see Figures 2 and 3), a choice that depends both on the needs of the patient and the preferences of the surgical team performing the operation. Statistically, mechanical valves last longer than ones made from tissue. Furthermore, the risk of early post-operative death increases substantially with a second heart valve replacement. Both of these reasons currently favour mechanical over tissue valves. However, tissue valves are more biocompatible than mechanical ones, their haemodynamic performance is superior, and they have a much softer failure mode in comparison to mechanical valves which tend to fall catastrophically. Increasing the durability of tissue valves to the present level of mechanical valves would probably make them the main choice for most valve replacements with very significant improvement for the health of the patient.

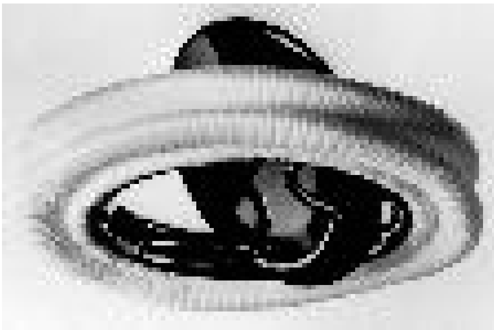

Figure 2: Example of a mechanical valve - a rotating disc design. This design, whilst providing a reasonably large area of fully-open valve through which the blood can flow, still occludes the natural flow and generates a highly un-natural flow field.

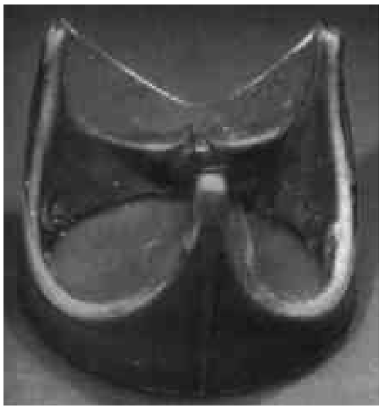

Figure 3: Example of a tissue valve - a prototype fully synthetic design that opens and closes in ways that mimic the natural, living counterpart. The flow is close to that in the living system.

To be competitive, these replacement tissue valves must endure great periods of service. The heart of a normal human beats about 75 times a minute [1], on average, so that the valve cycles about 40 million times a year. Since the average lifetime of a mechanical valve is about 15 years, a good commercial target for the average life for replacement tissue valves is 20 years. This means that the average valve must survive 800 million cycles, or more, of operation, with the leaflets undergoing large strain motions every cycle.
This increased durability is very design-sensitive, a fact that has driven a large number of computer-based design simulations over the last decade or so, (for example, [2] - [9]). All these have used "dry" modelling, a strategy that concentrates upon the

valve leaflets, and studies their strains induced by time-varying but spatially uniform pressures imposed on the leaflet surfaces. These models, though computationally expensive, could be run in useful times on the contemporary computers available, to give results that compared various design features with each other. An implied principle underlying this is that the relative ordering of the predictions of the "dry" models would be the same as those of more realistic models.
Recently, the development of numerical methods for modelling fluid-solid interaction (FSI) has allowed the blood flow to be modelled as well as the material of the valve, e.g. [10] - [12]. This now allows the study of, amongst other things, several features of the more realistic operation of fluid-driven tissue valves whose presence could not be assessed in the previous dry models. For example, the asymmetry of the geometry and aperture motion of the left ventricle suggests that the flow regimes are unlikely to be symmetric. The complexities of three-dimensional natural geometry are therefore likely to be important. So is the possibility that each of the valve leaflets is loaded differently by the complex flow into the aortic root from the left ventricle.
The simulation described here takes all of these factors into account. It uses an idealisation of the natural aortic valve explored in detail by Howard et al, [9] and which was itself based on the geometrical description set out by Thubrikar [1]. We modelled the ventricle separately as a fluid-filled vessel in order to obtain information about the flow delivered by the ventricle to the aortic valve. This information in the form of time-varying, spatial distributions of fluid velocity comprised the loading conditions for the aortic valve model, which also includes the aorta and sinuses as flexible structures that contain the flow.

Modelling strategy

The modelling of the structural components used Belytchko-Lin-Tsai shell elements with a Lagrangian formalism. These structural sections of the model were surrounded by a fluid control volume represented by eight-noded hexahedral brick elements with an Eulerian formation. The flow domain consisted of a control volume and two reservoirs (ambient elements capable of supplying and removing fluid either though specified velocity or pressure conditions), one attached to the inlet and the other to the outlet of the control volume.
The models and their meshes were generated using the pre-processor of the ANSYS finite element package and translated into LS-DYNA using a conversion program [13] written specially for the task. Constraints on computational resources forced the simulation to be divided into two parts, one being the aortic valve and its surrounding structure and the other, the left ventricle.

The Ventricle

Since the only function of the left ventricle model was to supply fluid velocity data as loading conditions for the aortic model, there was no need to represent its material properties accurately. Experimental wall-displacement data provided the ventricle shape-change conditions, and the computations then provided the resulting aortic fluid velocity profiles. We achieved a smoothly changing volumetric contraction of the ventricle by imposing a thermal shrinkage using a virtual decrease in temperature.
The geometry of the ventricle in its state of maximum expansion followed the design used by Schoephoerster and Chandran [14] in their experimental flow studies. Figure 1 shows the model, where 2768 shell elements representing the ventricle wall completely contained within a fluid control volume discretised using 8640 brick elements.

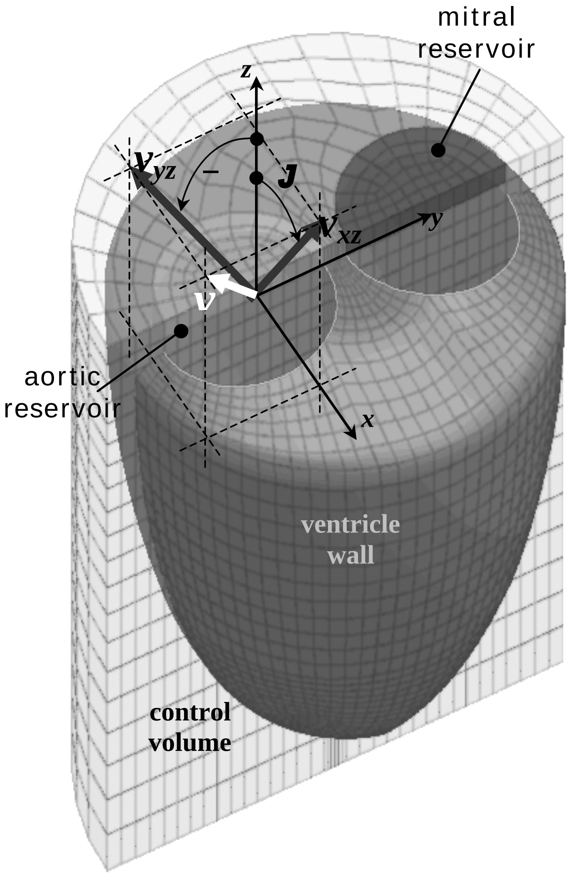

Figure 4 – Section through the ventricular flow domain showing shell elements of the ventricle model contained within it, and the alignment of the inlet and outlet reservoirs.

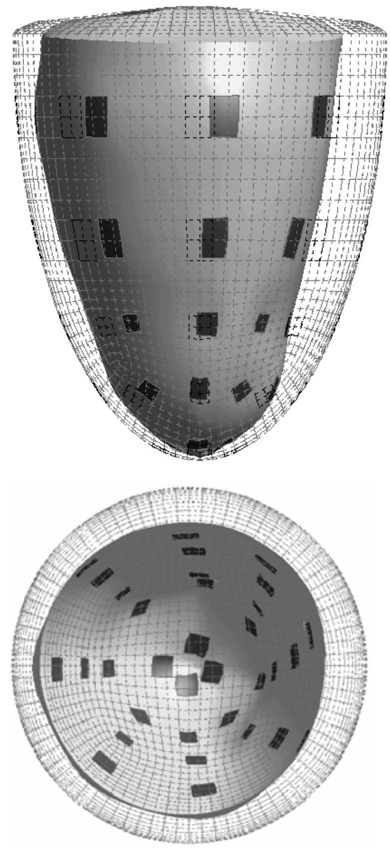

Figure 5 – Views of the ventricle model in its undeformed state (*wire-mesh*) and at its maximum contraction (*solid*).

There were inlet and outlet reservoirs at the positions of the mitral and aortic orifices, the apertures in the upper surface of the ventricle shown in Figure 4. The ventricle wall was linearly elastic with a Young's modulus of 15 MPa, zero Poisson's ratio and orthotropic thermal properties. Since ventricular contractions occur predominantly in the circumferential direction, the coefficient of thermal expansion was taken to be 10 x 10^3 in the circumferential and 5 x 10^3 in the axial direction. A shell thickness of 5mm prevented leakage in the mesh densities employed, and ensured good continuity in the wall curvature during contraction. The fluid was modelled as Newtonian with a density of 1000 kg.m^{-3} and a viscosity of 0.001 Ns.m^{-2}.

An imposed thermal shrinkage provided the ventricular contraction, by decreasing the temperature of the shell elements in the ventricle, so that the same time-varying reduction in volume occurred as that observed experimentally in physiological conditions (where the volume of the ventricle reduces to about one half of its maximum capacity). Forcing the movement of ventricle wall nodes to reproduce the displacements determined by Maier et al [15] produced the required motion of the ventricle, including wringing; see Figure 5.

The Aortic Valve

Earlier work by Yoxall [16] and Howard et al [9] pioneered the geometry of the aortic valve model used here. It all followed Thubrikar [1] and Tindale [17] in a strategy that allows the geometry of the valve, valve root and sinus to be described by a relatively small set of parameters; Figure 6.

Figure 7 shows the particular valve studied; its diameter was 24 mm and the ratios required to define the geometry shown in figure 6 are given in Table 1. The structure was described by 3564 4-noded Belytchko-Lin-Tsai shell elements. Regions in the structure where contact was likely to occur were identified from past experience e.g. Patterson et al [6] and master-slave relationships were defined between the potential contact or coaption surfaces. There were three pairs of frictionless contact surfaces such that rotational symmetry was maintained by defining half of the inflow or up-stream surface of each leaflet to act as master and half as slave, since each leaflet was expected to contact with both of its neighbours.

Table 1 – Aortic Valve Geometry Ratios (see figure 6 for definitions)

R_c/R_b	H/R_b	H_s/R_b	a	f	D_a/D_o	D_b/D_o	L_a/D_o	L_b/D_o
0.9	1.15	0.75	20	37	1.55	2.00	1.81	0.91

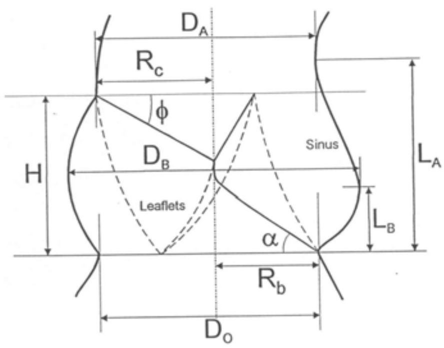

Figure 6 – Schematic diagram of the idealisation of the aortic valve and its surrounding structure showing parameters used to define the geometry.

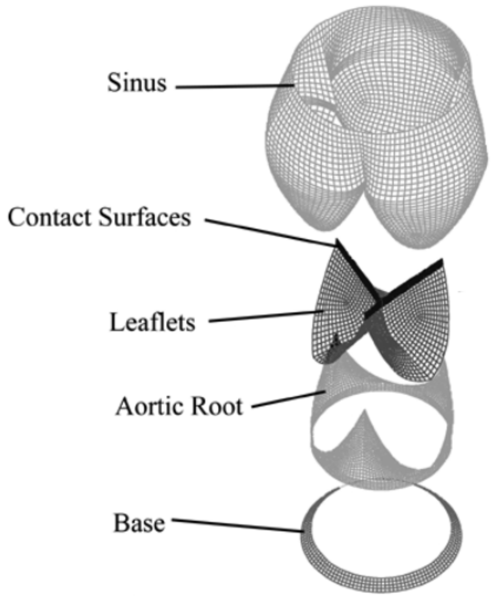

Figure 7 – Exploded view of the mesh used to model the aortic valve.

The base, aortic root and sinus were all 5mm thick linearly elastic material with Young's moduli of 2, 3, and 2 MPa respectively. The leaflets were 0.3 mm thick with non-linear elastic behaviour described by the model developed by Black et al [4].

The fluid control volume consisted of 36,864 elements completely surrounding the structure mesh. (See Figure 8). Small fluid elements both allowed the leaflets to coapt, and prevented 'leakage' of fluid through the shell. The flow domain included two reservoirs at each end of the structure model that supplied and removed fluid from the control volume. The base of the valve connected with the edges of the inlet reservoir with motion in the direction of flow constrained. This ensured that all of the fluid supplied entered the aorta and that the aorta remained connected to the reservoir. The fluid was Newtonian with a density of 1000 kg.m^{-3} and a viscosity of 0.001 Ns.m^{-2}.

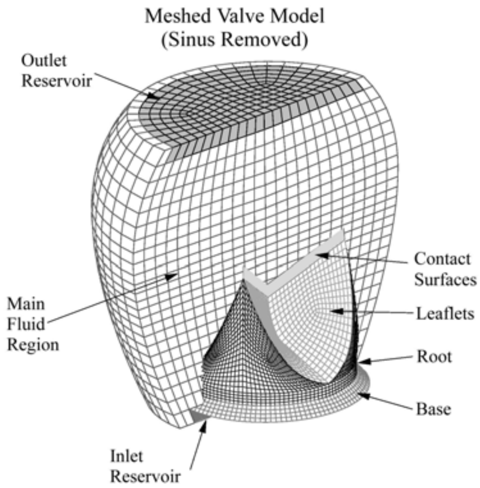

Figure 8. The aortic valve model, showing the inlet and outlet reservoirs. The Control volume allows for lateral flow as the sinuses (not shown) and aortic wall expand and contract.

The loading was applied as the time-varying velocity profiles for the aortic aperture obtained from the ventricle model described above. Since the spatial discretisation of the aortic aperture in the two models was different, the data were filtered by a sequence of smoothing and interpolation routines in order to generate velocity-time curves for the elements in the input reservoir of the valve model. A Fortran program was written to perform this operation automatically. In practice, the full cardiac cycle required an excessive amount of CPU time and so the pressure-time curves were modified by removing an interval of 0.1 seconds over which the velocity was approximately constant when the valve was fully open. This modification allowed the solution to be achieved in about 24 hours of CPU time.

Results

Figure 9 illustrates the sort of output available from the ventricle model. It shows the flow patterns in the beginning and at the end of the simulation designed to provide velocity data for input to the aortic valve model. Detailed examination of the flow patterns at the aortic orifice reveals the remarkable fact that they are very closely radially uniform across the aortic aperture. As a result, the flow into the aortic root, responsible for opening the valve, is symmetrical, to a very close approximation. This, of course, has important implications for how the valve operates, both for natural and replacement valves. Figure 10 shows displacements of the valve leaflets during both opening and closing, compared with pulse duplicator studies of a porcine valve at similar times during the cardiac cycle.

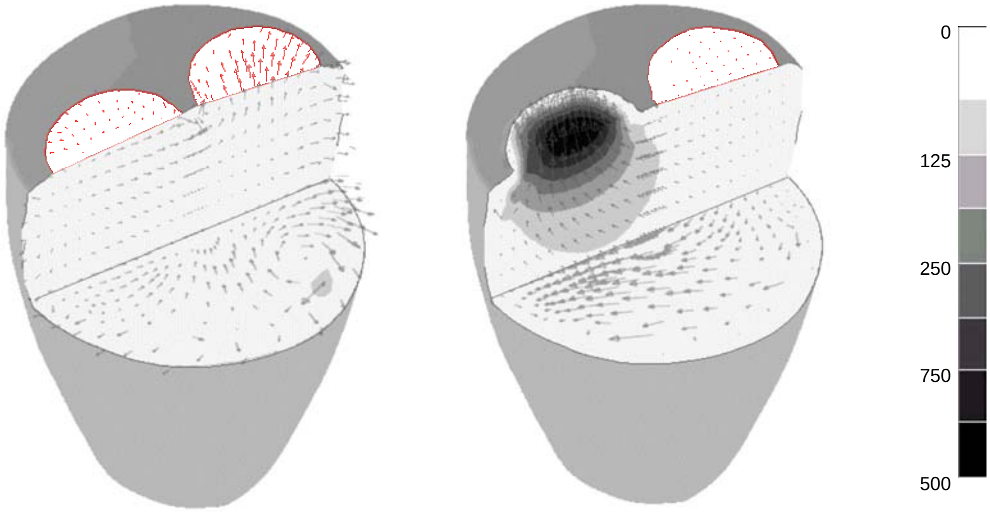

Figure 9 – Maps of velocity on part of a meridional and an equatorial plane in the ventricle at the beginning *(left)* and end *(right)* of the cardiac cycle.

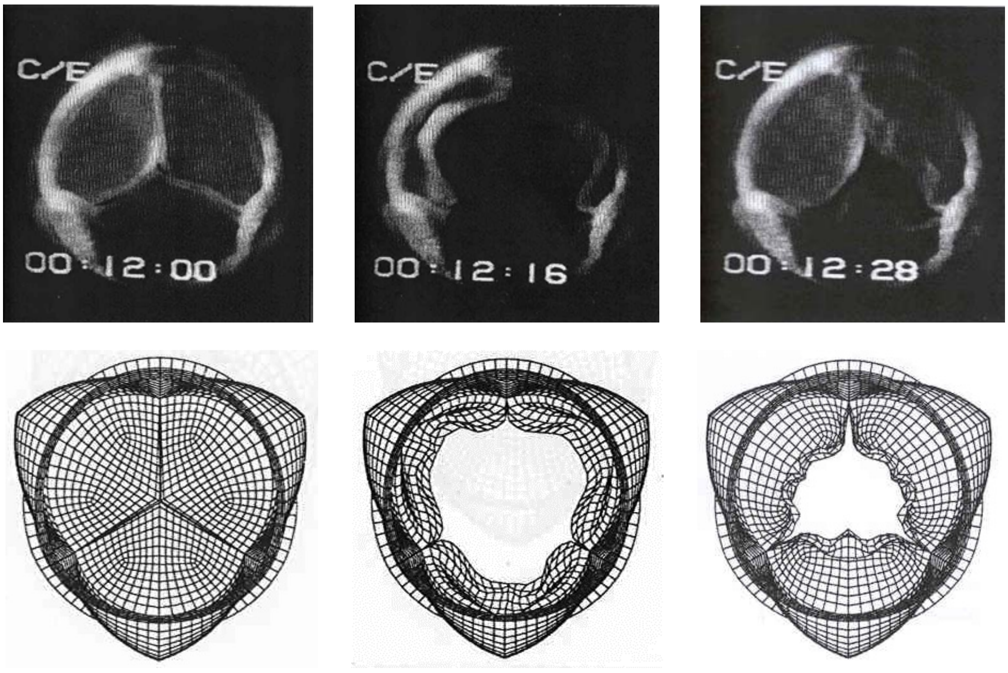

Figure 10 – Views, from the downstream or aorta, of the displacements of the valve leaflets at selected instants in the cardiac cycle. Data from the simulation (*below*) reflect that shown in pictures from a pulse duplicator study of a porcine valve (*above)*. The pictures and plots show the opening and subsequent closing of the valve in nominally 40ms increments.

Pressure data from this model is valuable for validating the dry models widely used for valve design studies. Doing this is not straightfoward, since sampling pressure differences across a valve leaflet at any given time depend upon the relative proximity of the sampling points in the mesh of the control volume. Despite this coarseness, the models examined here display, for the first time, that loading dry models with spatially constant pressure differences across the valves are reasonably accurate reflections of the more complex loading states revealed by the wet models studied here.

Summary and Conclusions

The work reported here demonstrates the potential of the Solid-Fluid Interaction routines available within LS-DYNA. It has involved highly deformable and flexible shell structures interacting with fluid and with themselves through the contact algorithms. Its successful outcome illustrates the power of the code to deal with these difficult biomechanical problems. The work demonstrates the truth of the following conclusions.

Ventricle wall motion generates a circulatory flow in the chamber, which in turn produces a substantially axial flow at the aortic aperture.

The pressure difference across the leaflets of the valve is essentially uniform, except for a short period just before it is fully open.

The conclusions above support the use of a spatially uniform but temporally varying pressure difference across the leaflets in dry or structural models of the aortic valve.

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