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Normal aortic valves stay open much longer in systole than porcine substitutes

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Abstract

Objective: To compare the opening mechanics of porcine valve substitutes with those of a normal human aortic valve.
Background: All commercially available porcine valves are pretreated with glutaraldehyde. This study was undertaken to evaluate the consequences of such treatment on valve mechanics.
Methods: The opening mechanics of the aortic valve, especially the time taken to open fully from a closed position, and the duration for which the valve is maximally open, were compared in a normal aortic valve, a stent-mounted porcine valve, and a stentless porcine valve, using a finite element model.
Results: Despite a 4-fold higher gradient, stent-mounted porcine valves were slower in attaining the fully open position, and the time for which the valve was fully open was almost 25% less than a normal valve. In stentless valves, the compliant root made the initial opening mechanics similar to those of a normal valve. Once this effect was over, the effect of porcine leaflet properties took over, and there was a corresponding delay in the valve opening.
Conclusions: Fixing the root with a stent and stiffening the leaflets with glutaraldehyde result in delayed valve opening and decrease the duration for which the valve is fully open, thus contributing to inferior hemodynamics.

Keywords

Aortic valve, biomechanics, bioprosthesis, computer simulation, finite element analysis

Introduction

The choice of a valve substitute has an important bearing on long-term survival after aortic valve replacement. Ideally, following valve replacement for aortic stenosis, there should be complete regression of left ventricular hypertrophy, normal transvalvular gradients at rest and peak exercise, normal coronary flow reserve, the largest possible effective valve orifice area, and complete freedom from valve-related complications and structural deterioration. Unfortunately, none of the currently available valve substitutes is able to achieve this goal, especially in smaller aortic roots; an eloquent testimony to the functional elegance of the normal aortic valve. While stentless porcine valves are closer to achieving these results than stented valves, they are far from close to the normal aortic valve.[1,2] The reasons attributed to the superiority of stentless valves include the placement of a larger valve for a given annular size, a design that mimics normal anatomy, lack of a rigid stent in a small aortic root, and the contribution of a compliant aortic root to the increase in effective valve orifice area.[3–6] However, an important question needs to be addressed. All commercial porcine valves are subjected to glutaraldehyde treatment before human implantation, causing inevitable alterations to their mechanical properties. The thickness distribution is also different. In addition, in stented valves, the root is noncompliant. What is the effect of these changes in mechanical properties on the hemodynamics of the valve? Because a very high frame rate is needed to capture the opening

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mechanism precisely, finite element modelling of a porcine aortic valve was performed, and compared to a normal human aortic valve. This has previously been shown to be a powerful tool to understand the mechanics of normal aortic valve function.[6,7]

Methods

The details of the modelling technique have already been published.[6,7] Three valves were simulated in this study. The first was a normal aortic valve (model A). The 2nd was a stentless porcine valve. This was achieved by substituting the mechanical properties and thickness of the normal valve leaflets with those of porcine leaflets, retaining the normal root properties.[8] Although such a model may seem unrealistic, the aim of this model was to highlight the effect of altered leaflet mechanical properties on valve opening. To a large extent, this model mimics a stentless porcine valve (model B). The 3rd was a stented porcine valve. One of the key issues in modelling a stented porcine valve is the boundary condition at the interface. For example, Luo and colleagues[8] used a hinged boundary condition. The boundary condition for the stented valve is crucial. Two extreme conditions are possible: one is to consider the joint to be hinged (model C); the other is to fix all degrees of freedom, rotation, and translation (model D). We feel the actual boundary condition is somewhere between these 2 conditions. However, both boundary conditions were simulated. These 2 boundary conditions are the 3rd and 4th models. The number of finite elements varied from one model to the other, approximately 3300 shell elements were used for a normal aortic valve, out of which 1100 occupied each leaflet area. These shell elements were based on reduced integration along with hourglass control. The dynamic analysis was performed with the explicit algorithm available in Abaqus/Explicit software. The model considers the open position as the stress-free position. The pressure cycle followed is from our earlier work.[6]

Orthotropic behavior has been assumed for the leaflets, summarized in Table 1. The orthotropic properties of the normal leaflets were taken from the work of Grande-Allen and colleagues.[9] The properties of the porcine leaflets were derived from data in the literature.[8] The slopes corresponding to 15% strain were used for this study. The root and the sinus were assumed to be isotropic, with a Young's modulus value of 2 MPa. The Poisson's ratio in all cases was assumed to be 0.45. The thickness of the leaflet is important and has been adapted from the literature.[10] The thickness variations in the 2 leaflets are shown in Figure 1.

Table 1. Properties of the valves used in the study.

Property	Normal aortic valve	Porcine valve
Ecirc (MPa)	6.885	8.453
Erad (MPa)	1.624	0.857
Elong (MPa)	8.89	8.89985
Gxy (MPa)	1.121	1.121
Gyz (MPa)	1.121	0.2955
Gxz (MPa)	0.56	0.295
nxy	0.106	0.45
nyz	0.106	0.101
nxz	0.45	0.45
Density ($kg{\cdot}mm^{-3}$)	$1{,}100 \times 10^{-12}$	$1{,}100 \times 10^{-12}$
Thickness (mm)	0.65–1.10	0.2–1.18

Ecirc: elastic modulus in the circumferential direction; Elong: elastic modulus in the longitudinal direction; Erad: elastic modulus in the radial direction; Gxy: shear modulus in the xy plane; Gxz: shear modulus in the xz plane; Gyz: shear modulus in the yz plane; ν: Poisson's ratio.

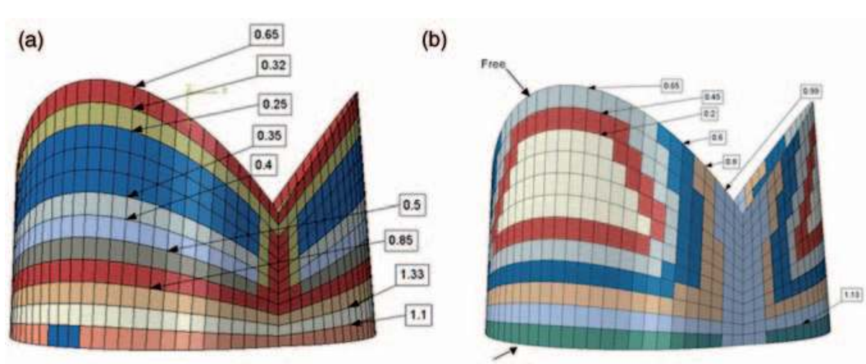

Figure 1. (a) Thickness distribution of a normal human aortic valve. (b) Thickness distribution of a normal porcine valve.

Results

The purpose of the study was to simulate and capture 2 main events: the opening characteristics of a normal aortic valve and a porcine valve, and the time taken to achieve their largest effective valve orifice area, starting from a closed position. Because the porcine leaflets, unlike those of the normal aortic valve, did not open freely at a transvalvular gradient of 2 mm Hg, the comparison that follows is at a gradient of 2 mm for the normal valve and 8 mm for the porcine valve.

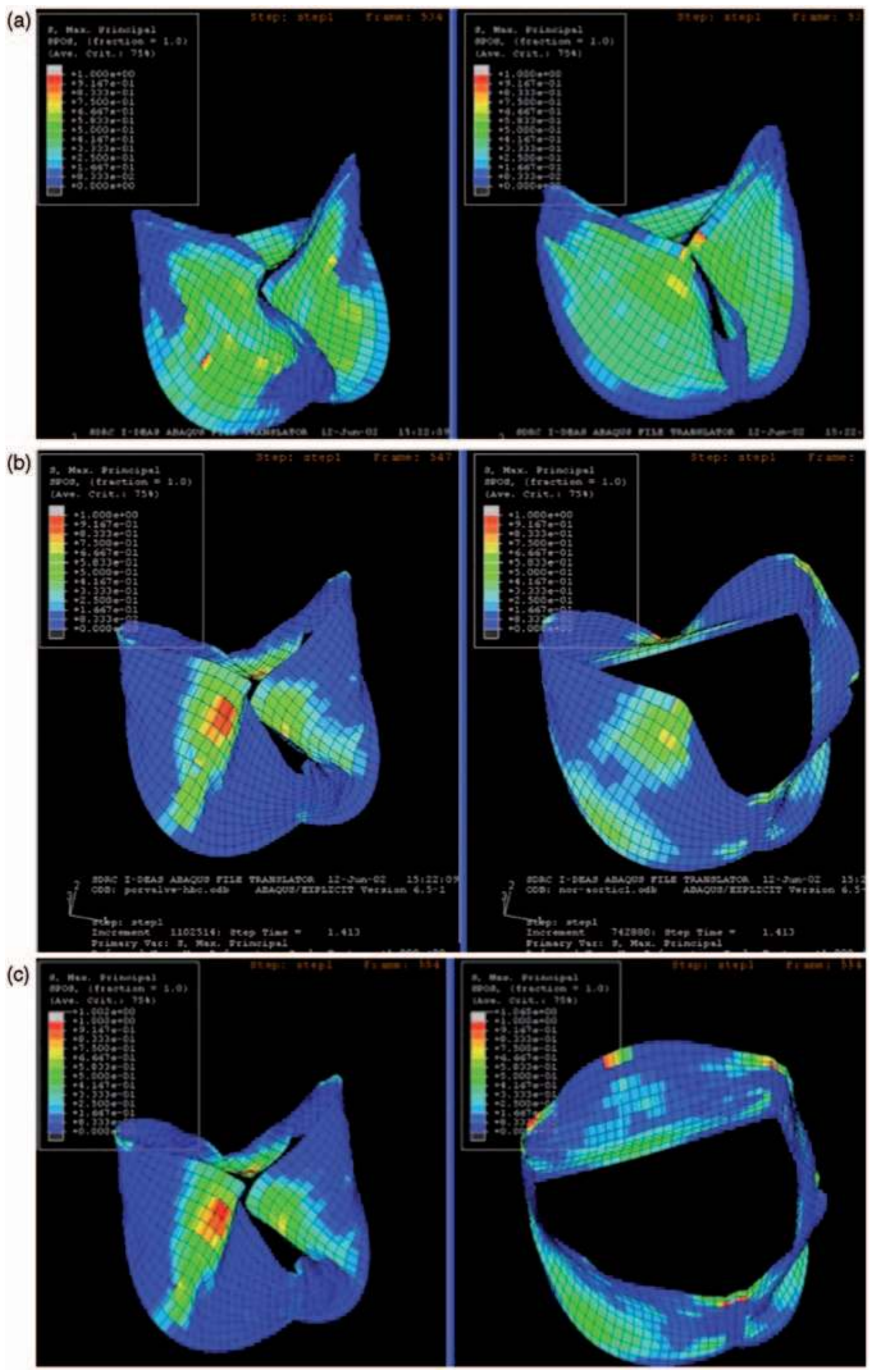

Figure 2. (a) The opening of a normal (right) and porcine aortic valve (left) captured at the same phase of the cardiac cycle. The time is 1.38 s from the beginning of systole. (b) Opening of both valves 1.413 s. The porcine valve is barely open, while the normal valve (right) is nearly open. (c) Opening position of both valves at 1.43 s. The normal valve (right) is fully open while the porcine valve is still struggling.

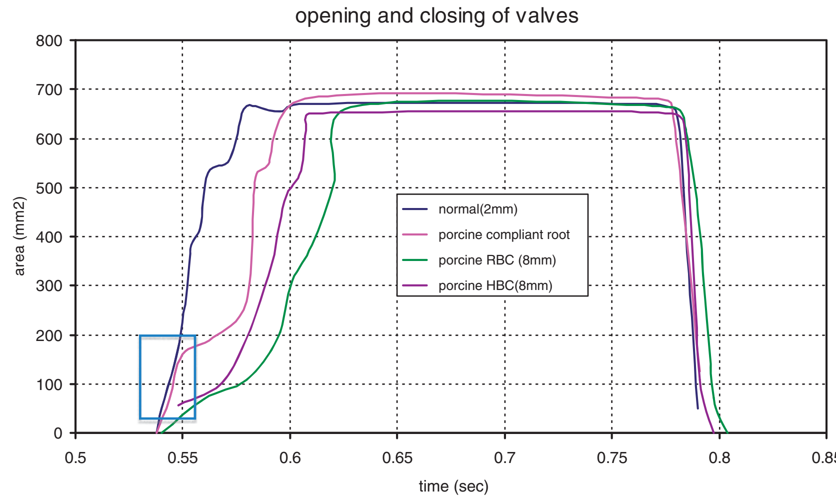

Figure 3. Model A (normal aortic valve) is clearly the earliest to attain a fully open position. Model B (porcine compliant root) mimicking a stentless valve, follows the graph of the normal valve as long as the influence of a compliant root is felt (box), then the porcine leaflet properties take over and the curve is parallel to the other porcine models. Models C and D (porcine valves) are clearly delayed in attaining a fully open position and the impact of a noncompliant aortic root on the valve opening can clearly be seen.

The striking finding of the study was the ease with which a normal aortic valve (model A) springs open at a minimal gradient of 2 mm Hg. In contrast, a porcine valve, especially the stented valve, even at a higher gradient of 8 mm Hg, is slow to open and seems to struggle sluggishly to attain the fully open position (Figure 2). The stentless valve had a behavior between these 2 extremes.

The time taken to attain the largest effective valve orifice area is really an inference of the first finding. The method of calculating the largest effective valve opening area (EVOA) for a given valve, for the purpose of this study, has been previously described.[7] The findings are shown in Figure 3. The normal aortic valve (model A), aided by a compliant aortic root and pliable leaflets, was the first to reach its fully open position. In contrast, the stented porcine valve opened much more lethargically. The boundary condition seemed to play a role: the valve with a rigid boundary condition (model D) seemed to open later than the valve with a hinged boundary condition (model C). The most interesting case was the porcine valve with a compliant root (model B). The valve opening was aided initially by expansion of the root, and the valve area started to increase immediately. This manifested as the same slope as a normal aortic valve in the EVOA diagram, marked by a circle for clarity. Once the effect of the compliant root on leaflet opening was exhausted, the porcine leaflet properties took over, and the slope shifted to be parallel to the stented valve (models C and D). Note the similarity between the magenta and green curves after the initial phase. The extent of the delay in attaining the fully open position was very significant. The stented porcine valves attained their largest EVOA 25% later in systole than a normal aortic valve, and therefore, the duration for which these valves were fully open was 25% less than normal (Figure 4).

Discussion

The quest for an ideal valve substitute continues. A detailed understanding of the functioning of the normal aortic valve, honed to perfection over millions of years of evolution, is likely to offer new insights into how to design reliable valve substitutes. Porcine valves, even the newer generation ones, continue to have inferior hemodynamics and clinical outcomes compared to normal valves. This study was essentially aimed at highlighting the contribution of the mechanical properties of the leaflets and aortic root to overall aortic valve function, and thus offer a clue as to where further improvement in valve design may be possible. Finite element modelling, as a tool for studying aortic valve function, has been well described, and has the unique advantage of very high frame rates of capture and performing "what if?" studies, making it possible to view events not otherwise easily visualized.

The inferior hemodynamics of porcine valves compared to normal aortic valves has been well documented. The normal aortic valve is an exquisitely designed structure where geometry and mechanical properties are fine-tuned to achieve the best possible design for the valve. Appreciation of the contribution of a compliant aortic root to normal aortic valve function has led to stentless valve designs. However, the contribution of altered leaflet mechanical properties due to glutaraldehyde fixation, the universally used

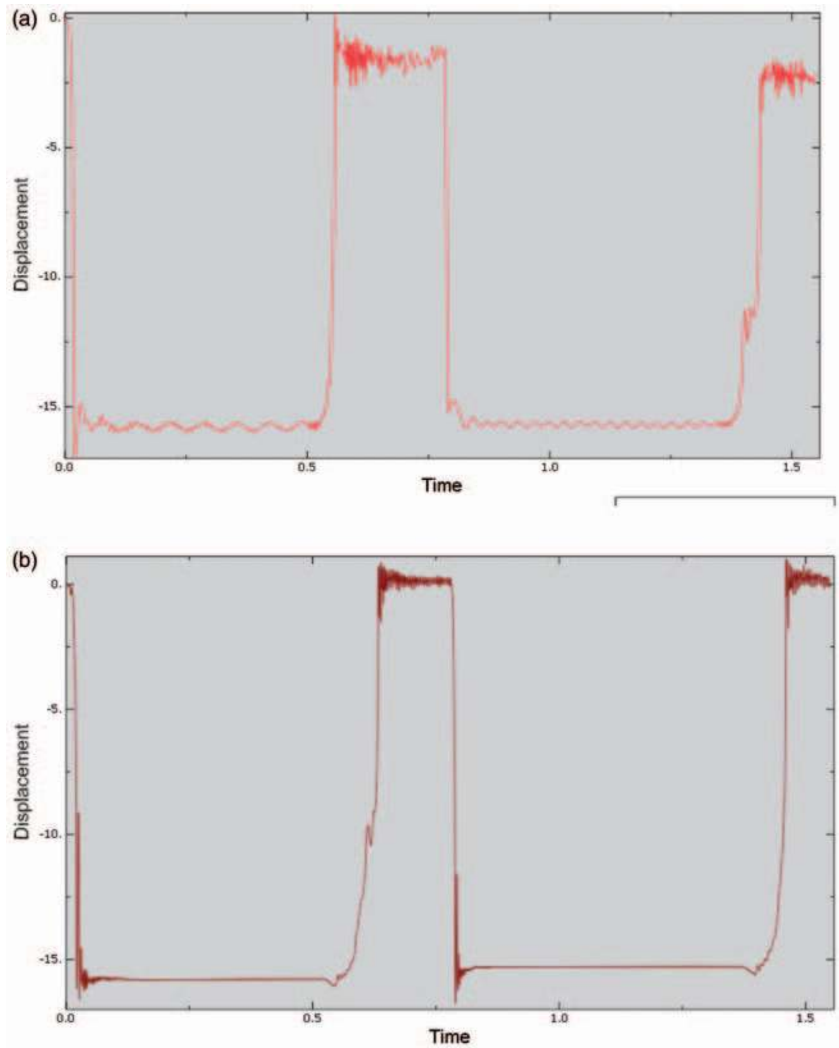

Figure 4. (a) Duration for which normal aortic valve is fully open. (b) Duration for which stent-mounted porcine valve is fully open. Clearly, the porcine valve is open for 25% less time than normal.

method by which porcine valves are sterilized and fixed, to leaflet opening, has not been previously studied. This study highlights the importance of the mechanical properties of the leaflets and the root to normal aortic valve function. Both seem to be important. Altering one alone, as in a stentless valve, only partially fixes the problem. Even at 4-times higher pressures needed to open the porcine valves, compared to normal, the valve opening was delayed by 25%, leading to a corresponding reduction in the time available during systole when the valve was fully open. This could be an important additional reason for the increased aortic flow velocities observed clinically in these valves, and the inferior hemodynamics. The impact of valve size has not been addressed in this study. That would need fluid flow to be incorporated into the model, and is currently being studied. It is also evident that the findings of this study have to be corroborated by other evidence in vivo and in vitro, using more sophisticated imaging modalities.

We consider that besides the geometry of design, the mechanical properties of the leaflets and the root also contribute enormously to the functional elegance of the normal aortic valve. This factor needs to be taken into consideration in designing biological valve substitutes.

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Conflicts of interest statement

None declared.

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