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Modeling leaflet correction techniques in aortic valve repair: A finite element study

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ABSTRACT

In aortic valve sparing surgery, cusp prolapse is a common cause of residual aortic insufficiency. To correct cusp pathology, native leaflets of the valve frequently require adjustment which can be performed using a variety of described correction techniques, such as central or commissural plication, or resuspension of the leaflet free margin. The practical question then arises of determining which surgical technique provides the best valve performance with the most physiologic coaptation. To answer this question, we created a new finite element model with the ability to simulate physiologic function in normal valves, and aortic insufficiency due to leaflet prolapse in asymmetric, diseased or sub-optimally repaired valves. The existing leaflet correction techniques were simulated in a controlled situation, and the performance of the repaired valve was quantified in terms of maximum leaflets stress, valve orifice area, valve opening and closing characteristics as well as total coaptation area in diastole. On the one hand, the existing leaflet correction techniques were shown not to adversely affect the dynamic properties of the repaired valves. On the other hand, leaflet resuspension appeared as the best technique compared to central or commissural leaflet plication. It was the only method able to achieve symmetric competence and fix an individual leaflet prolapse while simultaneously restoring normal values for mechanical stress, valve orifice area and coaptation area.

1. Introduction

The aortic valve is made of three leaflets (cusps) that open when the left ventricle of the heart contracts (systole) to eject blood into the aorta. The aortic valve function is to close and prevent backflow when the left ventricle relaxes (diastole). Diseased valves can develop a leak (aortic insufficiency–AI) and may have to be replaced, but some can be repaired. The number of indications for aortic valve repair keeps rising (Matalanis et al., 2010), whether it is by reimplantation (David and Feindel, 1992) or remodeling (Yacoub et al., 1998). Both procedures, however, may be associated with cusp prolapse (Boodhwani et al., 2009). Cusp prolapse arises when contact (or coaptation) between neighboring leaflets occurs below the physiologic plane of coaptation of a normal aortic valve (Boodhwani et al., 2008). Conversely, cusp restriction arises when contact occurs above the physiologic plane of coaptation. Although abnormal cusp coaptation can exist without significant AI, normal coaptation is associated with optimal long-term function of the valve (le Polain de Waroux et al., 2009) and should be restored as much as possible. This is the focus of the present study. The coaptation of individual leaflets can be adjusted using different leaflet correction techniques, such as central or commissural plication, or resuspension of the leaflet free margin (Boodhwani et al., 2008). The practical question then arises of determining which surgical technique provides the most physiologic coaptation along with the best valve performance. In the context of aortic valve repair, the tools to evaluate coaptation are visual inspection during the operation, and transesophageal echocardiography (TEE) intra- and perioperatively, with little room for trial-and-error. In contrast, the present study makes use of finite element (FE) analysis permitting the study of what-if scenarios. Recent FE aortic valve models provide increasingly accurate descriptions of the deformations, stresses and dynamics experienced by the valve (Conti et al., 2010; Koch et al., 2010; Labrosse et al., 2010). To shed light into the coaptation characteristics of repaired valves, simulation tools must be able to address asymmetric valves of various sizes, implement corrections of specific leaflet dimensions and represent the valve transformations from unpressurized to physiologically loaded. As detailed in the following, such a tool was created for the present study, with the ability to simulate physiologic function in normal valves as well as AI due to leaflet prolapse in

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diseased or sub-optimally repaired valves. Different existing leaflet correction techniques were simulated in a controlled situation, and the performance of the repaired valve was quantified in terms of maximum leaflets stress, valve orifice area, valve opening and closing characteristics. To further gauge the quality of coaptation depending on the leaflet correction technique employed, the total surface area of coaptation at a given time and the equivalent contact pressure acting on this area were considered.

2. Methods

2.1. FE model of a typical human aortic valve

Building on geometric modeling of symmetric aortic valves (Labrosse et al., 2006), a novel method was devised (see Supplementary Appendix 1) to produce a typical trileaflet aortic valve model without imposed symmetries starting from 8 parameters (namely, the heights and free margin lengths for the 3 leaflets, and the aortic and leaflet thicknesses) and the three-dimensional (3-D) coordinates of 15 landmark points (Fig. 1a). From these points, lines and curves were drawn using MatLab (The MathWorks, Natick, MA, USA) to sketch the aortic root. Then, the leaflet contours were created in partially closed position using a straight line for the leaflet height, and a sine curve of known length for the leaflet free margin (Fig. 1b). This configuration was chosen because it eliminates artificially high stresses along the attachment line as the leaflets open and close, an improvement over previously published models (Labrosse et al., 2010).

All the curves created formed the edges of linear and bicubic Coons surfaces (Salomon, 2006), each of which was discretized using structured meshing. The process was repeated to produce 4 (resp. 3) surfaces at a small distance from each other to allow for the creation of 3 (resp. 2) finite elements through the aortic (resp. leaflet) thickness, as a structured hexahedral FE mesh of the whole valve was generated by connecting the nodes from adjacent surfaces (Fig. 1c). The final mesh consisted of approximately 12,000 nodes and 8500 8-node solid elements.

In the future, the coordinates of patient-specific landmarks might be obtained from 3-D imaging such as 3-D TEE. However, for simplicity, the coordinates of the landmarks herein were determined to reproduce dimensions measured from rubber silicone molds cast in a symmetric unpressurized human aortic valve: series 7 in Swanson and Clark (1974) (Table 1, Supplementary Appendix 1).

All numerical analyses of the valve were carried out with commercial FE software LS-Dyna (LSTC, Livermore, CA, USA) on a Pentium 4 3.4 GHz processor PC with 2.48 GB of RAM. The mass density of the aortic and leaflet tissues was set at 1000 kg/m^3. Both tissues were modeled as hyperelastic, transversely isotropic materials using a Fung-like model with strain energy function

$$W=\frac{C_1}{2}[\exp(c_2E_\theta^2+c_3(E_z^2+E_r^2+E_{rz}^2+E_{zr}^2)+c_4(E_{\theta z}^2+E_{\theta r}^2+E_{r\theta}^2+E_{z\theta}^2))-1]+\frac{P}{2}(J-1),$$

where $c_1,\ldots,c_4$ are material constants (see Supplementary Appendix 2 and Table 1), $E_{\ldots}$ are deformations (Green strain components modified to only include the effects of volumetric work), and subscripts θ, z, r refer to the circumferential, longitudinal and radial directions, respectively. P is a Lagrange multiplier numerically enforcing the material near-incompressibility whereby J, the determinant of the deformation gradient tensor, is almost equal to 1. Although this material model was initially developed to represent myocardial tissue (Guccione et al., 1991), it has been shown to give accurate representations of human aortic tissue (Labrosse et al., 2009). It also lends itself to representing aortic leaflet tissue (Labrosse et al., 2010), with limitations highlighted in Supplementary Appendix 2.

The valve model was then studied over one cardiac cycle by application of known pressure pulses, as described and validated in Labrosse et al. (2010). Namely, since the analysis started from the unpressurized geometry, the pressure was ramped from 0 up to 80 mmHg before the cardiac cycle started in early systole. While the normal duration of diastole is approximately 0.60 s, it was shortened to 0.28 s in the simulation (Fig. 3) to save computational time. In addition, the simulation time was 1/10 of the real time, as analyses with real time or scaled time yielded results within 2% of each other, due to comparatively small inertial loads. The time step was automatically set and updated by LS-Dyna to achieve numerical stability of the solution.

By setting the value of the longitudinal stretch ratio at 1.00 (i.e. no pull), 1.20—expected normal physiologic conditions (Han and Fung, 1995)—and 1.30, it was possible to investigate the influence of this parameter on valve function and coaptation.

2.2. FE model of a valve with prolapse of one leaflet

Although three different leaflet dimensions is the norm rather than the exception (Silver and Roberts, 1985), in the interest of investigating leaflet correction techniques in a controlled environment, leaflet dimensions in the rest of the study were determined such that only one leaflet would prolapse. Specifically, in the unpressurized valve, the height of all three leaflets was 16 mm; the free margin of the right leaflet was set to 32 mm, while that of the left and non-coronary leaflets was 30 mm. In addition, the longitudinal stretch ratio was reduced from 1.20 until a central hole appeared in the closed valve, simulating AI.

2.3. Simulation of leaflet correction

Three existing leaflet correction techniques (Boodhwani et al., 2008) were implemented in the valve model with prolapse of the right leaflet and AI. The first technique was leaflet resuspension, either of the one-row type, where sutures run along the leaflet free margin (Fig. 2a), or of the two-row type, where a second row

Table 1
Typical human aortic valve model parameters. Dimensions in mm.

rb	**rc**	**h**	**rsm**	**LH**	**FE**
11.75	9.10	17.8	13.3	16	30
hs	**L. thickness**	**A. thickness**	**Hs**	**Height at rsm**	
18.33	0.50	1.52	7.1	10	
A. c_1 (kPa)	**A. c_2 (–)**	**A. c_3 (–)**	**L. c_1 (kPa)**	**L. c_2 (–)**	**L. c_3 (–)**
14.30	4.55	6.10	1.00	200	50

rb: Radius at the base of the valve, rc: radius at the commissures, h: valve height, rsm: maximum radius of aortic sinuses, LH: leaflet height, FE: leaflet free margin, hs: sinus height, L: leaflet; A: aorta, Hs: height of commissures (Appendix 1). c_i constants are material constants for the leaflet and aortic tissues (Appendix 2).

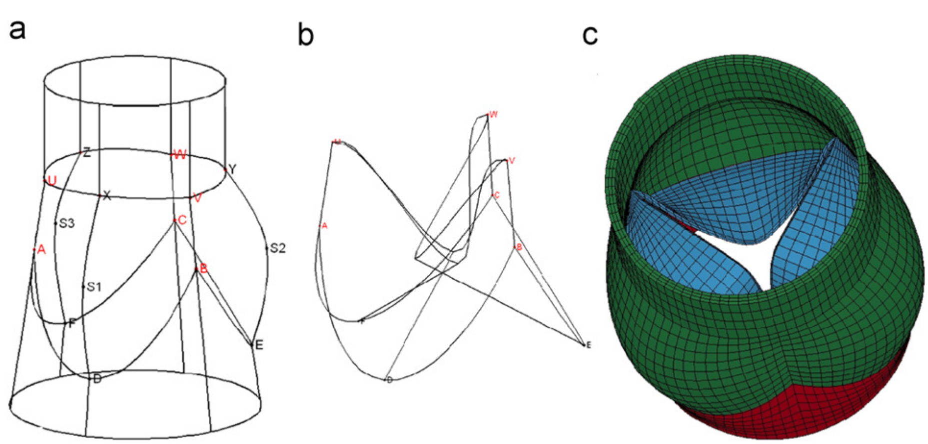

Fig. 1. (a) Line sketch of unpressurized aortic valve model, showing the left ventricular outflow tract, the aortic sinuses and the ascending aorta, including the 15 landmark points described in Appendix 1. (b) Line sketch of the aortic leaflets drawn in an assumed, unpressurized position. (c) Final finite element mesh of the aortic valve and root. Green: ascending aorta and aortic sinuses; blue: leaflets; red: base of the valve, including the left ventricular outflow tract.