Aortic root numeric model: Annulus diameter prediction of effective height and coaptation in post–aortic valve repair

Gil Marom, MSc,[a] Rami Haj-Ali, PhD,[a] Moshe Rosenfeld, DSc,[a] Hans Joachim Schäfers, MD,[b] and Ehud Raanani, MD[c]

Objective: The aim of the present study was to determine the influence of the aortic annulus (AA) diameter in order to examine the performance metrics, such as maximum principal stress, strain energy density, coaptation area, and effective height in the aortic valve.

Methods: Six cases of aortic roots with an AA diameter of 20 and 30 mm were numerically modeled. The coaptation height and area were calculated from 3-dimensional fluid structure interaction models of the aortic valve and root. The structural model included flexible cusps in a compliant aortic root with material properties similar to the physiologic values. The fluid dynamics model included blood hemodynamics under physiologic diastolic pressures of the left ventricle and ascending aorta. Furthermore, zero flow was assumed for effective height calculations, similar to clinical measurements. In these no-flow models, the cusps were loaded with a transvalvular pressure decrease. All other parameters were identical to the fluid structure interaction models.

Results: The aortic valve models with an AA diameter range of 20 to 26 mm were fully closed, and those with an AA diameter range of 28 to 30 mm were only partially closed. Increasing the AA diameter from 20 to 30 mm decreased the averaged coaptation height and normalized cusp coaptation area from 3.3 to 0.3 mm and from 27% to 2.8%, respectively. Increasing the AA diameter from 20 to 30 mm decreased the effective height from 10.9 to 8.0 mm.

Conclusions: A decreased AA diameter increased the coaptation height and area, thereby improving the effective height during procedures, which could lead to increased coaptation and better valve performance. (J Thorac Cardiovasc Surg 2013;145:406-11)

Supplemental material is available online.

The local geometry of the aortic valve plays an important role in its performance. In the past, different factors influencing the aortic valve configuration have been the focus of many studies. The coaptation area and effective height (h_E) of the cusps have been identified as important parameters in determining the acute and long-term function of repaired aortic valves.[1-3] Aortic aneurysms have been found to reduce cusp coaptation, initiate aortic insufficiency, and increase cusp stress,[4,5] a mechanism that seems to be caused by dilatation of the sinotubular junction and/or aortic annulus (AA).

In valve-sparing aortic replacement procedures, the aortic root is replaced with a graft while retaining its native aortic valve. Two widely used, similar procedures have been developed by David and Feindel[6] and Sarsam and Yacoub.[7] Avoiding prosthetic valves and retaining the native cusps improves the hemodynamics, reduces the risk of thromboembolism and endocarditis, and eliminates the need for anticoagulation, thereby improving the overall quality of life.[8] The disadvantages of these techniques, however, lie in the risk of reduced valve durability owing to the nonphysiologic opening and closing characteristics, leading to incomplete restitution of normal valve configuration.[8]

Because the heart is arrested and empty during surgery, it is very difficult to predict the postoperative final results in terms of valve geometry and competence. For this reason, other tools such as numeric models are important to provide the surgeon with improved geometric information to perform valve repair or preservation procedures.

Previous numeric models have been used to study the effect of root geometries on the mechanical performance of aortic valves, with and without aortic sinuses,[9-11] for valve-sparing procedures,[4,12,13] and with aneurysmatic

From the School of Mechanical Engineering,[a] Tel Aviv University Faculty of Engineering, Tel Aviv, Israel; Department of Thoracic and Cardiovascular Surgery,[b] University Hospitals of Saarland, Homburg, Germany; and Department of Cardiothoracic Surgery,[c] Chaim Sheba Medical Center, Tel Hashomer, Israel.

This work was partially supported by a grant from the Nicholas and Elizabeth Slezak Super Center for Cardiac Research and Biomedical Engineering at Tel Aviv University. R. Haj-Ali received support from the European-Union Marie-Curie IRG grant.

Disclosures: Authors have nothing to disclose with regard to commercial support.

Received for publication Sept 26, 2011; revisions received Jan 6, 2012; accepted for publication Jan 24, 2012; available ahead of print Feb 27, 2012.

Address for reprints: Ehud Raanani, MD, Department of Cardiothoracic Surgery, Chaim Sheba Medical Center, Tel Hashomer 52621 Israel (E-mail: ehud.raanani@sheba.health.gov.il).

0022-5223/$36.00

doi:10.1016/j.jtcvs.2012.01.080

Abbreviations and Acronyms

2D = 2-dimensional
3D = 3-dimensional
AA = aortic annulus
d_{AA} = annulus diameter
h_C = coaptation height
h_E = effective height
FSI = fluid structure interaction
NCCA = normalized cusp coaptation area

roots.[4,13] In dry models,[4,9,12,13] it has been suggested that the sinuses are important for minimizing stress in the cusps. However, the influence of the flow field was ignored in these models, and, therefore, physiologic boundary conditions could not be used. Ranga and colleagues[10] and Katayama and colleagues[11] used fluid structure interaction (FSI) models to study different types of aortic roots. These models did not include contact interaction between the cusps, a factor that plays a critical role in aortic valve diastolic behavior, especially for valve-sparing procedures.[14]

It is still necessary to perform parametric simulations and to supply the surgeon with sufficient data to match the graft to the patient's pathology. Most importantly, there is currently no available information regarding which valve repair configuration yields the minimum amount of stress in the cusps. Furthermore, it is unclear to what degree the sinotubular junction and AA should be reduced. More recent studies in the reconstruction of regurgitant bicuspid aortic valves have shown that an enlarged AA is an important predictor of mid- to long-term failure of valve repair.[15] Although changes in AA diameter in valve-sparing procedures or in subcommisural annuloplasty directly affect the dynamics of the valve,[16] no studies on the influence of these changes have been reported. The objective of the present study was to determine the influence of changes in AA diameter on the geometry and dynamics of the aortic root and valve cusps using numeric methods.

METHODS

Physical Model

The 3-dimensional (3D) geometry of a normal aortic valve and root was reconstructed using geometric relationships similar to those suggested by Thubrikar.[17] Figure 1 illustrates a proposed aortic valve base geometry with dimensions scaled with respect to the annulus diameter (d_{AA}). The scaling values are taken as a base configuration for a healthy valve. Five additional geometries, with d_{AA} from 20 to 30 mm, were calculated from the base geometry with applied outer pressure that expands or shrinks the initial AA. These calculated geometries were used as the initial configuration of the aortic valve and root, assuming zero stress in the tissues. The cusps and root had a thickness of 0.3 and 0.6 mm, respectively. We assumed that the cusps and root had separate but tied geometries and that the cusps moved with the root at their joint boundaries. Two straight rigid tubes with a length of 1 and 4 cm were added upstream and downstream, respectively, to move the flow boundary conditions away from the regions of interest.

Although leaflet tissue is anisotropic and hyperelastic, isotropic and linear elastic material properties were assumed in the present model. This assumption is suitable for simulating coaptation,[12] the major goal of the present study. A Poisson's ratio of 0.45, Young's modulus of 1 and 2 MPa, and a density of 1100 and 2000 kg/m^3 were used for the cusps and root, respectively.[4,12] To account for the tissue damping effect, a stiffness-proportional damping coefficient ($\beta = 0.013$ s) was selected for the cusps according to previous sensitivity.[18] The aortic root was significantly stiffer; therefore, a value of $\beta = 0.15$ s was used to yield realistic results for the root. The flow was assumed to be laminar and the blood to be slightly compressible and Newtonian at a constant temperature of 37°C.[19] Blood compressibility[20] of $3.75 \cdot 10^{-10}$ m^2/N was used to improve the convergence of the FSI model.[21] Full Navier-Stokes equations were used to model the flow, and force and momentum equilibrium equations were solved using a displacement-based finite element for the structure.[19]

The present study focused on the closing diastolic phase of the valve, starting with an almost closed valve. The h_Es were calculated from no-flow ("dry") static models, similar to the clinical measurements, and the other parameters were calculated from fully compliant FSI models. The dry models were loaded with linearly increasing pressures, reaching a maximum load of 80 mm Hg after 20 ms. For the FSI models, a constant physiologic diastolic blood pressure of 0 and 80 mm Hg was specified, for a duration of 10 ms, as the boundary conditions at the left ventricle and the aorta ends, respectively. The initial flow conditions were obtained from a steady state solution of the initial configuration.

Parametric Geometry and Discretization

The structural solver used an implicit nonlinear dynamic analysis with an implicit direct displacement-based finite element method. A master–slave contact algorithm was used between the cusps themselves and with the sinuses, assuming a nonfriction contact. Initially, a shell mesh was created for the aortic root and the 3 cusps using a mesh generator, TrueGrid (XYZ Scientific Applications, Livermore, Calif). The surface shell mesh was extruded into 3 layers of 3D elements. Conversion of the shell to solid elements is required by the fluid solver to allow for different pressure values at both sides of the cusps. The mesh of the structural part has more than 80,000 elements, which was found to be adequate after a mesh refinement study.[18]

A finite volume method was used for solving the unsteady 3D Navier-Stokes and mass conservation equations using the Eulerian approach with a Cartesian mesh. The flow domain was discretized with a mesh of approximately 700,000 elements after a mesh refinement study.[18] The mesh was dynamically adapted in the vicinity of the walls, including near the moving cusps and compliant root. The implicit flow solver used a spatially second order upwind scheme and a second order temporal discretization.

The FSI model was solved by a partitioned solver with nonconformal meshes, which allows large structural motion through the fluid domain and simplifies the introduction of contact.[18] The sequential coupling process exchanges the structure geometry (kinematics) and the pressure load between the 2 solvers. A common adaptive time stepping increment was automatically calculated by negotiation between the flow and structure solvers. The maximum time step in each solver was limited to 0.05 ms, and at least 2 structural solver time steps were required for each flow time step.

The structural problem was solved by Abaqus 6.9-EF1 (Dassault Systems, Simulia, Providence, RI) finite element software, and the FlowVision HPC, version 3.07.02 (Capvidia, Leuven, Belgium) was the fluid dynamics solver. The FlowVision Multi-Physics manager (Capvidia) managed the coupling between the 2 codes.

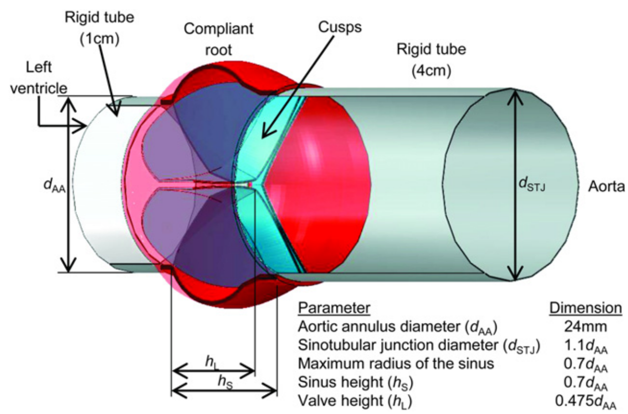

FIGURE 1. A schematic view of the aortic valve healthy model showing the compliant region and the added rigid tubes.

RESULTS

Dependency of h_E on d_{AA}

The h_E was calculated when the applied pressure load reached a value of 3 mm Hg. This value was the minimum load required to fully close the valve. The h_E was defined as the valve height (h_L in Figure 1) of the closed valve. Figure 2 plots the h_E as a function of d_{AA}. As expected, the h_E decreases with the increase in the d_{AA}. An h_E range of less than 12 mm but greater than 7 mm is believed to be normal for adults.[2] The cases with d_{AA} of 28 mm and 30 mm have an h_E of less than 9 mm (Figure 2, below the *dashed line*) and did not fully close under the prescribed simulation conditions. This is in agreement with the findings of Bierbach and colleagues,[2] who used transthoracic echocardiographic measurements to demonstrate that 96% of all patients with moderate or more severe aortic insufficiency had an h_E of less than 9 mm.

Dependency of Coaptation on d_{AA}

Coaptation was calculated at the point when the valve was fully closed for 1 ms. Those cases that were not fully

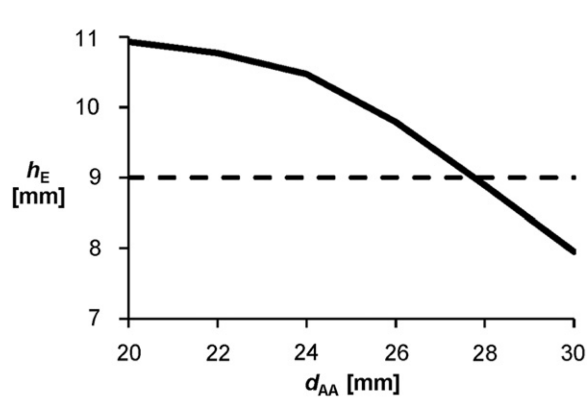

FIGURE 2. The effective height *(h_E)* as a function of annulus diameter *(d_{AA})* under a pressure load of 3 mm Hg.

closed (d_{AA} of 28 and 30 mm) were compared after 1 ms of the constant opening area. The projected 2-dimensional (2D) deformed configurations on C-C plane are presented in Figure 3. The C-C plane location was selected to best illustrate valve coaptation in a similar fashion to echocardiographic measurements. The partially opened valves are prolapsed into the left ventricle and have a ‘‘belly region’’ in the cusps. The coaptation height (h_C) on the C-C plane increases with the decrease in d_{AA}. This can also be seen in Figure 4, which compares the h_C on the C-C plane and the average h_C as a function of d_{AA}. Coaptation is also marked in Figure 4 on a sample cusp from the case with a d_{AA} of 24 mm. The average h_C is defined as the coaptation area divided by the free-edge length. The average h_C also decreased with the increase in d_{AA}. The normalized cusp coaptation area (NCCA) is defined as the coaptation area divided by the total cusp surface area. An additional graph of the NCCA is published online and presents similar findings to the average h_C.

Maximum Principal Stress and Energy Density

The maximum principal stress in the various d_{AA} is compared along 2 lines: the cusp free-edge and the cusp symmetry line. The maximum principal stress as a metric for cusp performance is suitable for both isotropic and anisotropic materials. The various graphs along these 2 lines, as well as the maximum principal stress contours for the $d_{AA} = 24$ mm case, can be found online. The highest stress values are likely to be found on the free-edge line near the commissures. Maximum principal stress on the symmetry line is an important metric because it can be used to compare cusp stress values under partially open and fully closed valves. Stress in the partially open valves ($d_{AA} = 28$ mm and 30 mm) is greater than the fully closed valves on both of the lines, although the model with $d_{AA} = 22$ mm has the lowest stress. The largest stress was found in the 28-mm model, in which the valve is partially open but there is a relatively large coaptation area (Figure E1). For all cases, the largest stress is found on the free edge near the commissures. On the symmetry line, the stress is higher in the noncoapting region with much greater values when the ‘‘belly’’ exists (28- and 30-mm cases). The symmetry line also shows an increase in stress in the coaptation area, excluding the 30-mm case in which there is no coaptation near the symmetry line.

Figure 5 compares the maximum principal stress on the cusps (σ_{max}) and the strain energy density for the various cases. The addition of energy density as a second metric is introduced to provide a total (per cusp) deformation measure. The maximum principal stresses are calculated at 1 ms after closure, similar to our previous results, but the energy density is computed 5 ms after the beginning of the solution. Energy density increases with the increase of d_{AA}, implying that valves with a smaller d_{AA} are more efficient. The largest

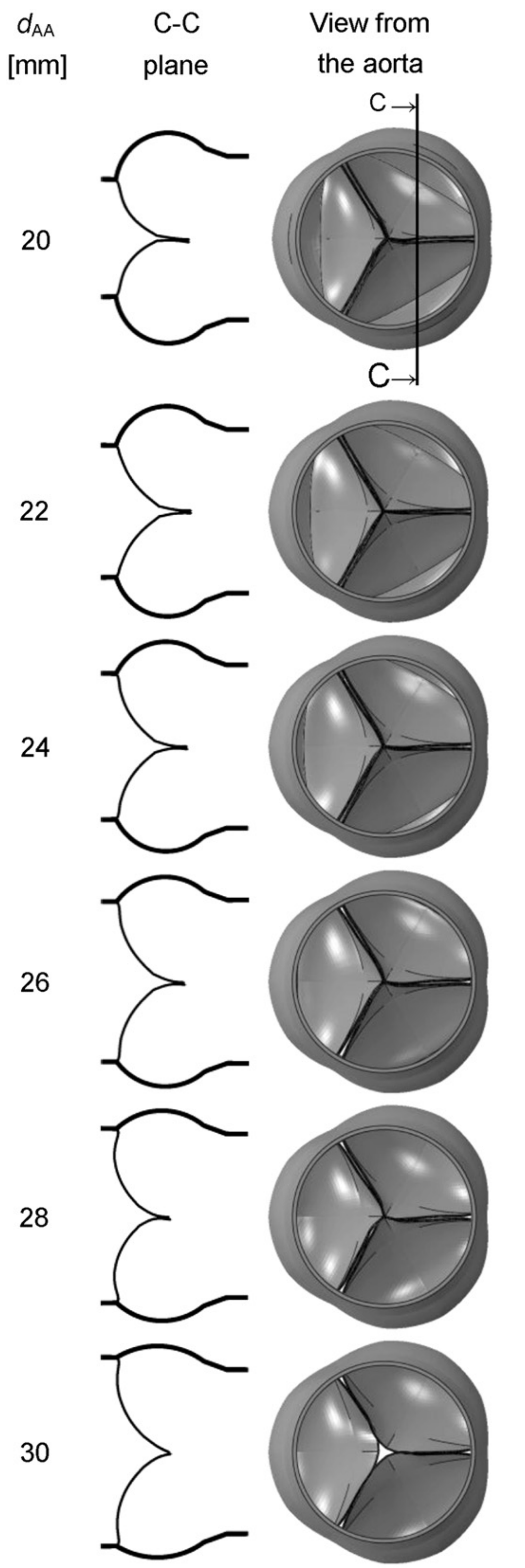

FIGURE 3. Projected deformed configurations of the closed valves on the C-C plane with $r = 5$ mm and views from the aorta. The case of $d_{AA} = 28$ mm is open with discontinuous contact, although the aorta view dose not disclose this fact.

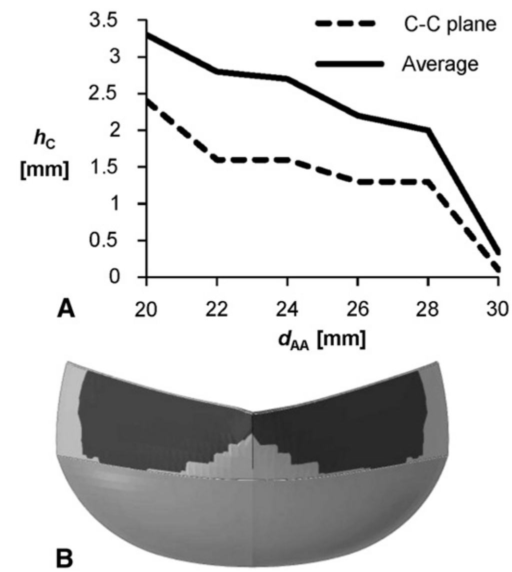

FIGURE 4. A, Coaptation heights *(h_C)* as a function of annulus diameter *(d_{AA})* and (B) coaptation area marked in black for the case of fully closed (1-ms duration) $d_{AA} = 24$ mm valve.

stress in the 28-mm case might be the result of the prolapse of the valve or the contact pressure that acts on a smaller coaptation area (Figure E2). The larger twists in the cases of 20 and 22 mm might cause larger maximum stress relative to the 24-mm case.

DISCUSSION

To prevent regurgitation, it is considered desirable to have the valves completely closed with a relatively larger coaptation area. It was found that increasing the d_{AA} results in a reduced coaptation height from the 2D views (C-C plane), findings similar to the echocardiographic measurements. This appears plausible and is consistent with assumptions in valve-preserving surgery[22] and recent observations in aortic valve repair.[15] More meaningful parameters, which can be calculated from current 3D models, are defined according to the entire coaptation area.

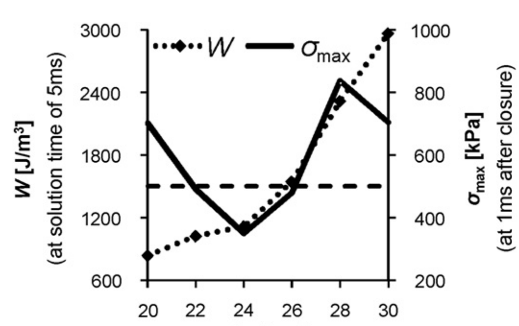

FIGURE 5. Comparison of energy density (W) and the maximum principal stress *(σ_{max})* for the various annulus diameters *(d_{AA})*. The *dashed line* indicates the $\sigma_{max} = 500$ kPa.

ACD

Although these parameters have similar findings to the 2D measurements, they can capture discontinuous contact that might be hidden in the 2D view, such as was found in the 28-mm case. It appears that many in the medical field estimate the NCCA to be in the range of 30% to 50%,[14] although our study places the NCCA values between 3% and 27%. Our current estimated NCCA values correlated well with other computational models.[23]

The quality of the closure can be seen from the maximum principal stress, because excessive stress values can damage the valve and reduce its durability.[17] The 28-mm case has a maximum principal stress of up to 2 times that of the 26-mm case (the largest d_{AA} case with a fully closed valve). This suggests that partially open valves might have lower durability. The case of 22 mm has the lowest stress on the 2 lines defined for comparison (Figure E1). A comparison of the maximum stress in the entire cusp (Figure 5) suggests that 22, 24, and 26 mm are preferable, with stress less than 500 kPa (Figure 5, *dashed line*), which represents a level of stress in which tissue stiffening can develop.[24] Although a smaller d_{AA} produced a larger average h_C (Figure 4) with smaller strain energy (Figure 5), a reduction of d_{AA} from 24 to 22 mm increased the average h_C and decreased the strain energy by only 3.5% and 10%, respectively. Therefore, the healthy case (24 mm) had the best combination of stress, strain energy density, and coaptation that, to the best of our knowledge, is a new and important finding, clearly demonstrating the clinical implication of an optimal annulus diameter, a conclusion that concurs with the suggested valve-sparing goals.[22] Avoidance of annular dilation seems to be as important as overcorrection.

The h_Es were calculated from dry models and were found to differ from those of the FSI models, which were similar to the findings on coaptation and mechanical stress.[18] The main reasons for these differences are obviously caused by the pressure load on the cusps. The calculated blood pressure load in the FSI model is not as uniform as in the traditional dry models. Furthermore, dry models usually assume a transvalvular pressure load on 1 side of the cusp, but this assumption becomes invalid in the coapting regions, because contact pressure acts on the left ventricle side and not on ventricular pressure.

The trend found for the h_E as a function of d_{AA} (Figure 2) correlated well with that of the average coaptation height (Figure 4). This indicates that improving the h_E during valve repair and replacement could also lead to increased coaptation and better performance, especially for nonprolapsed valve geometries. The nonfully closed 28- and 30-mm cases of d_{AA}, both in the dry and FSI models, are similar to the in vivo measurements of valves with an h_E less than 9 mm.[2] These cases also demonstrated the largest stress. Therefore, a very low h_E might indicate that the cusps are prolapsed and the valve might suffer from lower durability. The observations in this model indicate that focused reduction of annular diameter alone might increase the h_E and thus normalize cusp configuration. This concept is different from current approaches, in which the h_E is only increased by cusp intervention.[16,25] Furthermore, this model demonstrates that the largest stress is found on the free edge near the commissures in all cases, implying that a cusp repair procedure near the commissures might fail early and should therefore be avoided.

The present model represents only a first step toward clinical guidelines that could be used in routine surgical practice. However, although the correlation between the h_E and coaptation provides a practical tool that can be implemented for surgery, additional studies should be performed to investigate whether highly focused intervention at the annular diameter could achieve the same results in a clinical setting.

CONCLUSIONS

From our results, it is evident that decreasing the AA diameter increases the coaptation height and area. Furthermore, our results also indicate that measuring the h_E during surgery will closely correlate with postoperative coaptation and improved valve performance.

References

1. Aicher D, Langer F, Lausberg H, Bierbach B, Schäfers HJ. Aortic root remodeling: ten-year experience with 274 patients. *J Thorac Cardiovasc Surg*. 2007;134: 909-15.
2. Bierbach BO, Aichera D, Issaa OA, Bomberga H, Gräber S, Glombitza P, et al. Aortic root and cusp configuration determine aortic valve function. *Eur J Cardiothorac Surg*. 2010;38:400-6.
3. Jeanmart H, de Kerchove L, Glineur D, Goffinet JM, Rougui I, Van Dyck M, et al. Aortic valve repair: the functional approach to leaflet prolapse and valve-sparing surgery. *Ann Thorac Surg*. 2007;83:S746-51.
4. Soncini M, Votta E, Zinicchino S, Burrone V, Fumero R, Mangini A, et al. Finite element simulations of the physiological aortic root and valve sparing corrections. *J Mech Med Biol*. 2006;6:91-9.
5. David TE. Aortic valve repair and aortic valve-sparing operations. In: Cohn LH, ed. Cardiac surgery in the adult. 3rd ed. New York: McGraw-Hill Medical; 2008. p. 935-47.
6. David TE, Feindel CM. An aortic valve-sparing operation for patients with aortic incompetence and aneurysm of the ascending aorta. *J Thorac Cardiovasc Surg*. 1992;103:617-22.
7. Sarsam MAI, Yacoub M. Remodeling of the aortic valve annulus. *J Thorac Cardiovasc Surg*. 1993;105:435-538.
8. De Paulis R, De Matteis GM, Nardi P, Scaffa R, Buratta MM, Chiariello L. Opening and closing characteristics of the aortic valve after valve-sparing procedures using a new aortic root conduit. *Ann Thorac Surg*. 2001;72:487-94.
9. Beck A, Thubrikar MJ, Robicsek F. Stress analysis of the aortic valve with and without the sinuses of Valsalva. *J Heart Valve Dis*. 2001;10:1-11.
10. Ranga A, Bouchot O, Mongrain R, Ugolini P, Cartier R. Computational simulations of the aortic valve validated by imaging data: evaluation of valve-sparing techniques. *Interact Cardiovasc Thorac Surg*. 2006;5:373-8.
11. Katayama S, Umetani N, Sugiura S, Hisada T. The sinus of Valsalva relieves abnormal stress on aortic valve leaflets by facilitating smooth closure. *J Thorac Cardiovasc Surg*. 2008;136:1528-35.
12. Grande-Allen KJ, Cochran RP, Reinhall PG, Kunzelman KS. Re-creation of sinuses is important for sparing the aortic valve: a finite element study. *J Thorac Cardiovasc Surg*. 2000;119:753-63.
13. Soncini M, Votta E, Zinicchino S, Burrone V, Mangini A, Lemma M, et al. Aortic root performance after valve sparing procedure: a comparative finite element analysis. *Med Eng Phys*. 2009;31:234-43.

14. Pethig K, Milz A, Hagl C, Harringer W, Haverich A. Aortic valve reimplantation in ascending aortic aneurysm: risk factors for early valve failure. *Ann Thorac Surg*. 2002;73:29-33.
15. Aicher D, Kunihara T, Abou Issa O, Brittner B, Gräber S, Schäfers HJ. Valve configuration determines long-term results after repair of the bicuspid aortic valve. *Circulation*. 2011;123:178-85.
16. El Khoury G, Vanoverschelde JL, Glineur D, Poncelet A, Verhelst R, Astarci P, et al. Repair of aortic valve prolapse: experience with 44 patients. *Eur J Cardiothorac Surg*. 2004;26:628-33.
17. Thubrikar M. The aortic valve. Boca Raton, FL: CRC Press; 1990.
18. Marom G, Haj-Ali R, Raanani E, Schäfers HJ, Rosenfeld M. A fluid-structure interaction model of coaptation in fully compliant aortic valves. *Med Biol Eng Comput*. 2012;50:173-82.
19. De Hart J, Baaijens FPT, Peters GWM, Schreurs PJG. A computational fluid-structure interaction analysis of a fiber-reinforced stentless aortic valve. *J Biomech*. 2003;36:699-712.
20. Wang SH, Lee LP, Lee JS. A linear relation between the compressibility and density of blood. *J Acoust Soc Am*. 2001;109:390-6.
21. Degroote J, Swillens A, Bruggeman P, Haelterman R, Segers P, Vierendeels J. Simulation of fluid–structure interaction with the interface artificial compressibility method. *Int J Numer Meth Biomed Engng*. 2010;26:276-89.
22. David TE, Maganti M, Armstrong S. Aortic root aneurysm: principles of repair and long-term follow-up. *J Thorac Cardiovasc Surg*. 2010;140:S14-9.
23. Grande KJ, Cochran RP, Reinhall PG, Kunzelman KS. Stress variations in the human aortic root and valve: the role of anatomic asymmetry. *Ann Biomed Eng*. 1998;26:534-45.
24. Broom ND. The stress/strain and fatigue behaviour of glutaraldehyde preserved heart-valve tissue. *J Biomech*. 1977;10:707-24.
25. Langer F, Graeter T, Nikoloudakis N, Aicher D, Wendler O, Schafers HJ. Valve-preserving aortic replacement: does the additional repair of leaflet prolapse adversely affect the results? *J Thorac Cardiovasc Surg*. 2001;122:270-7.

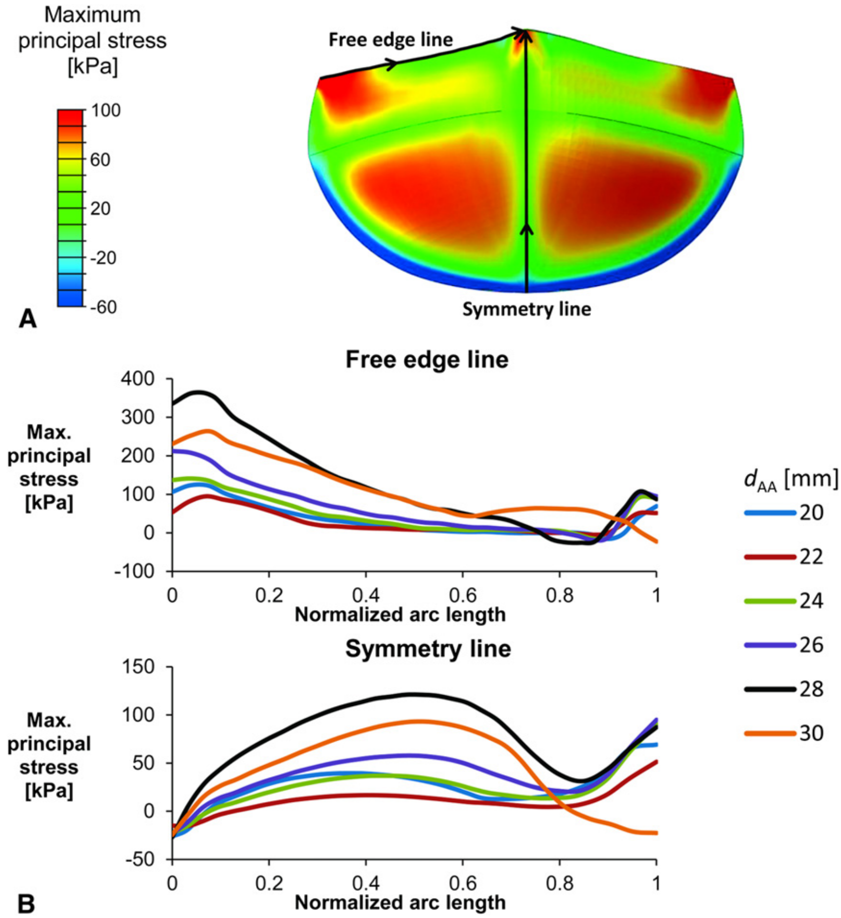

FIGURE E1. A, Contour of maximum principal stress on the cusp for annulus diameter *(d_{AA})* = 24 mm, also illustrating 2 defined lines. B, Maximum principal stress values at normalized locations for 2 predefined lines on the cusp.

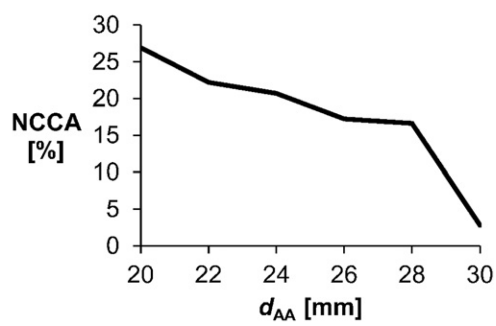

FIGURE E2. Normalized cusp coaptation area *(NCCA)* as a function of the annulus diameter *(d_{AA})*.