Deformational Dynamics of the Aortic Root
Modes and Physiologic Determinants

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Background—Current surgical methods for treating aortic valve and aortic root pathology vary widely, and the basis for selecting one repair or replacement alternative over another continues to evolve. More precise knowledge of the interaction between normal aortic root dynamics and aortic valve mechanics may clarify the implications of various surgical procedures on long-term valve function and durability.

Methods and Results—To investigate the role of aortic root dynamics on valve function, we studied the deformation modes of the left, right, and noncoronary aortic root regions during isovolumic contraction, ejection, isovolumic relaxation, and diastole. Radiopaque markers were implanted at the top of the 3 commissures (sinotubular ridge) and at the annular base of the 3 sinuses in 6 adult sheep. After a 1-week recovery, ECG and left ventricular and aortic pressures were recorded in conscious, sedated animals, and the 3D marker coordinates were computed from biplane videofluorograms (60 Hz). Left ventricular preload, contractility, and afterload were independently manipulated to assess the effects of changing hemodynamics on aortic root 3D dynamic deformation. The ovine aortic root undergoes complex, asymmetric deformations during the various phases of the cardiac cycle, including aortoventricular and sinotubular junction strain and aortic root elongation, compression, shear, and torsional deformation. These deformations were not homogeneous among the left, right, and noncoronary regions. Furthermore, changes in left ventricular volume, pressure, and contractility affected the degree of deformation in a nonuniform manner in the 3 regions studied, and these effects varied during isovolumic contraction, ejection, isovolumic relaxation, and diastole.

Conclusions—These complex 3D aortic root deformations probably minimize aortic cusp stresses by creating optimal cusp loading conditions and minimizing transvalvular turbulence. Aortic valve repair techniques or methods of replacement using unstented autograft, allograft, or xenograft tissue valves that best preserve this normal pattern of aortic root dynamics should translate into a lower risk of long-term cusp deterioration. (Circulation. 1999;100[suppl II]:II-54–II-62.)

Key Words: aorta ■ valves ■ surgery ■ structure ■ physiology

Many alternatives currently exist for surgically treating diseased aortic valves and aortic root aneurysms. These alternatives include different methods of valve-sparing procedures, aortic valve replacement with mechanical valves, and stented pericardial or porcine bioprostheses, as well as allografts, pulmonary autografts (Ross procedure), or unstented xenograft valves used in either the scalloped subcoronary position (as a root inclusion method) or as full aortic root replacement (with or without root tailoring). Long-term assessment has focused on hemodynamic performance, valve-related complications,1 cardiac-related complications,2 and structural valve deterioration (SVD).3–8 Suboptimal hemodynamic function, manifested by high transvalvular pressure gradients, may be associated with a higher probability of SVD due to dystrophic calcification and collagen fatigue induced by turbulence and/or abnormal bending or closing stresses on the aortic cusps. Nonetheless, numerous clinical and experimental studies implicate perturbation of the normal dynamic geometric properties of the aortic root as a factor portending earlier SVD, independent of valve hemodynamic performance3,4,9–12; altered normal root dynamics are thought to create abnormally high stresses on the cusps, which can lead to premature leaflet calcification and SVD.13,14

A better understanding of the dynamic 3D conformational changes of the normal aortic root may help clarify the link between aortic cusp stress and resultant valve failure after various surgical valve repair or replacement procedures. Such information may prove relevant in the future design of new surgical procedures to spare the aortic valve during aortic root aneurysm graft replacement, better methods of aortic annular

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enlargement, new techniques to tailor the size of the dilated aortic root to match a normal pulmonary autograft, and different methods of allograft and unstented xenograft valve implantation; furthermore, this information would be pertinent to the design of new stents for xenograft bioprosthetic valves and possibly new techniques for preparing and implanting unstented xenograft, allograft, or autograft valves.

We characterized the precise conformational changes occurring in the left, right, and noncoronary (NC) aortic root structures during the cardiac cycle in an ovine model. Finally, we assessed the response of aortic root 3D mechanics to changes in left ventricular (LV) pump performance and loading conditions.

Methods

Six adult, castrated, male sheep (72±4 kg) underwent implantation of miniature radiopaque markers in the aortic root, mitral annulus, and left ventricle using cardiopulmonary bypass (cross-clamp time 109±16 minutes). After a recovery period of 7.5±1.5 days, each animal was taken to the experimental animal cardiac catheterization laboratory.

Surgical Preparation

The animals were intubated and ventilated mechanically (Servo Anesthesia Ventilator, Siemens-Elema AB), and general anesthesia was maintained with inhalational isoflurane (1% to 2.2%). Through a left, fifth-intercostal space thoracotomy, pneumatic occluders (In Vivo Metric Systems) were placed around the superior and inferior vena cavae. The heart was suspended in a pericardial cradle, and 5 miniature radiopaque tantalum markers (internal diameter, 0.8 mm; outer diameter, 1.35 mm; length, 1.5 to 3.0 mm) were inserted into the LV epicardium and septum. Four markers were placed at the LV equatorial level circumferentially on the anterior, lateral, posterior, and septal walls, and 1 marker was placed on the LV apex. Epicardial echocardiography with color Doppler was used to assess the competence and anatomy of the aortic valve. Animals were placed on cardiopulmonary bypass using the left carotid artery and right atrial venous cannula. After recovery, studies were conducted in the experimental animal cardiac surgical intensive care unit.

Experimental Design

After recovery, studies were conducted in the experimental animal cardiac catheterization laboratory. The animals were premedicated with ketamine, intubated, and ventilated mechanically (veterinary anesthesia ventilator 2000, Hallowell EMC) with 100% oxygen. Simultaneous biplane videofluoroscopic and hemodynamic data were acquired with the animal in the right lateral decubitus position. Animals were studied in normal sinus rhythm after autonomic blockade and with ventilation arrested at end-expiration during data acquisition runs to minimize the effects of respiratory variation. Data acquired in this state constituted the control condition. Data were then acquired after (1) inotropic augmentation (calcium chloride 1 g IV bolus); (2) a 30% increase in LV afterload (aortic systolic pressure) with phenylephrine; and (3) a 30% decrease in LV afterload with sodium nitroprusside. On completion of the data acquisition, an aortic root angiogram was used to confirm aortic valve competence.

All animals received humane care in compliance with the principles of laboratory animal care formulated by the National Society for Medical Research and the Guide for Care and Use of Laboratory Animals, prepared by the National Academy of Sciences and published by the National Institutes of Health. This study was approved by the Stanford Medical Center Laboratory Research Animal Review Committee and conducted according to Stanford University policy.

Figure 1. Anatomy of aortic annulus and marker placement. The aortic annulus is the coronet-like fibrous structure that supports aortic leaflets, with leaflet commissures attached to peaks of coronet. Three nadirs of coronet form annulus, or aortoven-tricular junction. Pathological studies confirm that, in both human and ovine aortic roots, ventricular myocardium extends up into aortic root, especially in left and right annular sectors.

Sinuses of Valsalva are circumscribed by markers 2, 3, and 4; markers 4, 5, and 6; and markers 6, 1, and 2.

Figure 2. Average dimensions (mean±1SD) of aortic root for the 6 animals in this experiment at end-diastole. All lengths and angles are drawn to scale (as shown). Left commissure (LC), right commissure (RC), and NC commissure (NCC) denote, respectively, commissures between NC and left sinuses of Val-salva, left and right sinuses, and right and NC sinuses. Left nadir (L Nadir), right nadir (R Nadir), and NC nadir depict nadirs of central (belly) portions of left (L), right (R), and NC aortic (Ao) cusps, respectively. Definitions of left, right, and NC STJ segments are also shown, as are left, right, and NC annular sectors. Note that annular circumference (64.4±6.5 mm) was larger than STJ circumference (58.9±6.7 mm) in these sheep using this analytical framework; because we used linear chord segment lengths to compute circumference, calculated values underestimate true circumference by ~17%. Respective annular and sinotubular diameters calculated from marker coordinates were 24.7±2.5 and 22.6±2.6 mm.
Data Acquisition
A Philips Optimum 2000 biplane Lateral ARC 2/poly DIAGNOST C2 system (Philips Medical Systems) was used to record videofluoroscopic images at 60 Hz, with the image intensifiers in the 9-inch mode. Two-dimensional images from each of the 2 x-ray views (45° right anterior oblique and 45° left anterior oblique) were digitized and merged to yield 3D coordinates for each radiopaque marker every 16.7 ms. Analog LV pressure (LVP) and ECG voltage were simultaneously digitized and recorded on the videotape along with marker images during data acquisition.

Data Analysis
End-diastole was defined as the videofluoroscopic frame containing the peak of the ECG R-wave. End isovolumic contraction (IVC), or beginning ejection, was defined as the videofluoroscopic frame showing a decrease in LV volume greater than 3 mL. End ejection was assumed to occur at end-systole, which was defined as the videofluoroscopic frame preceding maximum negative dP/dt (−dP/dt max). End isovolumic relaxation (IVR) was defined as the videofluoroscopic frame containing the minimum LVP after systole.

Left Ventricular Volume
We calculated an instantaneous estimate of LV volume every 16.7 ms from the epicardial LV and mitral annular markers using a multiple tetrahedral model reconstructed from the marker coordinates and corrected for LV convexity. Although epicardial LV volume calculated in this manner overestimates true LV chamber volume (because it incorporates LV muscle mass), the change in epicardial LV volume is an accurate measurement of the relative change in LV chamber volume.15

Definition of Anatomic Regions and Aortic Root Deformations
The left, right, and NC annular sectors were arbitrarily defined as the segments around the annulus that corresponded to the lengths between the pairs of markers placed between the naidis of the NC and left, the left and right, and the right and NC cusps, respectively (Figures 1 and 2). Similarly, the terms left, right, and NC aortic root regions denote, respectively, the areas of the aortic root encompassed by the left, right, and NC annular sectors and the STJ marker at the top of the commissure between the annular markers. At the level of the STJ, the left, right, and NC segments corresponded to the respective sinuses, eg, the lengths between the pairs of markers on the commissures that spanned the left, right, and NC sinuses of Valsalva (Figure 2). The oblique longitudinal distances between the markers placed at the top of the commissures and at the bottom of the left, right, and NC cusps defined the respective aortic root longitudinal lengths.

Aortic root strain was decomposed into 5 modes of deformation: annular (aortoventricular junction) circumferential strain, STJ circumferential strain, aortic root longitudinal strain, aortic root shear, and torsional deformation. These 5 modes of deformation were computed separately for all 3 aortic root regions during 4 time intervals (IVC, ejection, IVR, and diastole). Lagrangian strain,16 defined as the relative percent change in length with respect to initial segment length, was used for all strain computations except shear and torsion.

The annular circumferential strains were computed from linear 3D chords between the 3 markers placed at the naidis of the aortic cusps that defined the left, right, and NC annular sectors. The STJ circumferential strains were computed from the markers placed at the top of the commissures that defined the left, right, and NC STJ segments. The distances between the markers placed at the top of the commissures and at the bottom of the cusps on the annulus were used to calculate the left, right, and NC aortic root longitudinal strains. Aortic root shear deformations of the left, right, and NC root were calculated from the triplet of markers (1 commissure marker and 2 cusp naidis marker) defining the aortic root regions (Figure 2). For each triplet of markers, the 2D Green’s strain tensor was used to calculate shear deformation during the cardiac cycle.17

**TABLE 1. Hemodynamic Data**

<table>
<thead>
<tr>
<th></th>
<th>Control</th>
<th>Range Over All Study States</th>
</tr>
</thead>
<tbody>
<tr>
<td>Heart rate, min⁻¹</td>
<td>115±8</td>
<td>89–136</td>
</tr>
<tr>
<td>ESP, mm Hg</td>
<td>81±11</td>
<td>57–110</td>
</tr>
<tr>
<td>EDP, mm Hg</td>
<td>24±5</td>
<td>11–34</td>
</tr>
<tr>
<td>LVPmin, mm Hg</td>
<td>12±4</td>
<td>1–20</td>
</tr>
<tr>
<td>ΔTVG, mm Hg</td>
<td>60±9</td>
<td>36–98</td>
</tr>
<tr>
<td>EDV, mL</td>
<td>140±47</td>
<td>58–191</td>
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<tr>
<td>ESV, mL</td>
<td>114±39</td>
<td>41–155</td>
</tr>
<tr>
<td>SV, mL</td>
<td>25±9</td>
<td>12–52</td>
</tr>
<tr>
<td>EF</td>
<td>0.18±0.03</td>
<td>0.15–0.30</td>
</tr>
<tr>
<td>dP/dt_{min}, mm Hg/s</td>
<td>1309±186</td>
<td>943–3125</td>
</tr>
<tr>
<td>dP/dt_{max}, mm Hg/s</td>
<td>-1231±244</td>
<td>-845–1728</td>
</tr>
</tbody>
</table>

Data are expressed as mean±SD. ESP indicates end-systolic pressure; ESV, end-systolic volume; SV, stroke volume; min, minimum; and max, maximum.

Statistical Analysis
All data are reported as mean±1SD. For each animal, the reported data represent the average of 3 consecutive cardiac cycles. Comparison of root deformation between different periods in the cardiac cycle was made using repeated measures ANOVA. Differences in deformation between the left, right, and NC aortic root regions were assessed using a 3-factor repeated measures ANOVA model. Hemodynamic predictors of each mode of root deformation during each time period in the cardiac cycle were assessed using a multivariate general linear model (GLM). For each mode of deformation and for each time period during the cardiac cycle, the independent variables tested in the multivariate GLM included the physiological parameters listed in Table 1 and the LV, aortic, and transvalvular pressure changes measured during that period. Forward and backward stepwise regression on those variables was used to identify a subset of variables to construct each model. The threshold for regression coefficient significance for inclusion or exclusion of a particular variable was set as 0.10. The level of significance for all statistical comparisons was P<0.05.

Results
Hemodynamics and Aortic Root Dimensions
Three animals had trace aortic insufficiency on the aortic root angiogram at the end of the data acquisition catheterization study. Table 1 shows all the hemodynamic parameters for the 6 animals in the control state and the range of values of those parameters obtained during the 3 interventions (calcium chloride, phenylephrine, and sodium nitroprusside). Figure 2 illustrates the baseline geometry and dimensions at end-diastole of the ovine aortic root, drawn to scale for these 6 animals. It also depicts the actual angular relationships between the various marker pairs.

Aortic Root Deformation During IVC
During IVC, both the annulus and the STJ underwent rapid circumferential expansion (Figure 3, Table 2). This was paralleled by an increase in longitudinal aortic root length, but it was not accompanied by shear or torsional deformation of the aortic root. Thus, during IVC, the aortic root “cylinder” underwent simple circumferential and longitudinal dilatation. However, circumferential expansion of the annulus was not uniform during IVC. Expansion was greatest in the left sector and was least in the NC annular sector, which reflects possible differences in the tensile properties and/or myocyte...
content of each commissural region or external structural coupling effects. For example, the NC annular sector is continuous with the anterior mitral annulus (aortic-mitral coupling effects. For example, the NC annular sector is continuous with the anterior mitral annulus (aortic-mitral coupling effects. For example, the NC annular sector is continuous with the anterior mitral annulus (aortic-mitral coupling effects. For example, the NC annular sector is continuous with the anterior mitral annulus (aortic-mitral coupling effects. For example, the NC annular sector is continuous with the anterior mitral annulus (aortic-mitral coupling effects. For example, the NC annular sector is continuous with the anterior mitral annulus (aortic-mitral coupling effects. For example, the NC annular sector is continuous with the anterior mitral annulus (aortic-mitral coupling effects.

The terms enclosed in parentheses represent the GLM regression coefficients ± 1SD. This GLM may be used to calculate the effect of independent or combined changes in EDV or dP/dtmax on the left annular sector expansion during IVC. For example, over the combined range of EDV and dP/dtmax studied (Table 1), the left annular sector expansion during IVC predicted by the GLM increased from 7.3 ± 2.0% to 24.0 ± 6.6%. However, no significant relationship between right and NC annular sector expansion and the hemodynamic parameters examined was found.

With respect to STJ expansion during IVC, EDV was again significantly linked with expansion of the STJ in the left (P=0.011) and right (P=0.001) sinus regions but not with STJ expansion in the NC sinus, as shown below:

left STJ IVC expansion = (0.025 ± 0.009) × EDV + (3.1 ± 1.3)
right STJ IVC expansion = (0.028 ± 0.007) × EDV + (4.1 ± 1.1)

The GLM predicted an increase in left STJ expansion from 4.5 ± 1.8% to 7.8 ± 3.0% and an increase in right STJ expansion from 5.7 ± 1.5% to 9.4 ± 2.4% over the range of EDV studied.

### Aortic Root Deformation During Ejection

During ejection, the annulus underwent circumferential contraction, whereas the STJ continued to expand (Figure 3, Table 3). This annular contraction, however, was again not homogeneous: the left and right annular sectors contracted significantly more than did the annulus in the NC sector. In addition, the aortic root underwent nonuniform shearing during ejection that resulted in torsional deformation of the root. The left and NC sinuses underwent clockwise torsion, and the right aortic root underwent counterclockwise torsion (as viewed from above by the surgeon; Figure 3).

A multivariate GLM inspecting circumferential strain, shear, and torsional deformation was used to identify which hemodynamic variables correlated with aortic root deformation during ejection; ejection fraction (EF) was the only independent predictor of annular circumferential contraction during ejection, yielding the following general linear models (left, P<0.001; right, P=0.002; and NC, P=0.02):

- left annular sector circumferential contraction during ejection = (40.4 ± 8.0) × EF
- right annular sector circumferential contraction during ejection = (24.9 ± 7.0) × EF
- NC annular sector circumferential contraction during ejection = (17.4 ± 6.8) × EF

Thus, during ejection over the range of EF studied, the GLM predicted increases in left annular circumferential contraction from 6.1 ± 2.2% to 12.1 ± 2.4%, right annular contraction from 3.7 ± 1.1% to 7.5 ± 2.1%, and NC contraction from 2.6 ± 1.0% to 5.2 ± 2.0%.

### Table 2. Annular and STJ Circumferential Deformations and Aortic Root Longitudinal, Shear, and Torsional Deformations During IVC

<table>
<thead>
<tr>
<th></th>
<th>AV-Circ.*</th>
<th>ST-Circ.*</th>
<th>Long.*</th>
<th>Shear, degrees</th>
<th>Tor, degrees</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left</td>
<td>11.2 ± 2.5</td>
<td>8.3 ± 2.1</td>
<td>4.9 ± 2.7</td>
<td>0.4 ± 1.7</td>
<td>1.1 ± 3.5</td>
</tr>
<tr>
<td>Right</td>
<td>4.8 ± 2.3</td>
<td>7.8 ± 2.5</td>
<td>3.6 ± 1.5</td>
<td>0.7 ± 0.5</td>
<td>1.1 ± 0.8</td>
</tr>
<tr>
<td>Posterior</td>
<td>3.2 ± 1.1</td>
<td>7.0 ± 2.3</td>
<td>3.2 ± 1.3</td>
<td>−0.2 ± 1.4</td>
<td>−0.6 ± 2.9</td>
</tr>
</tbody>
</table>

Data are expressed as mean ± SD. AV indicates aortoventricular/annular; Circ, circumferential; ST, sinotubular; Long, longitudinal; and Tor, torsional.

*Changes during isovolumic contraction are significant at P<0.001 by repeated measures ANOVA.

†Differences between the 3 groups (left, right, and NC) in the measured change are significant at P<0.001 by a 3-factor repeated measures ANOVA.
Maximum LV dP/dt was the only variable that correlated significantly with aortic root shear deformation during ejection. The general linear models were as follows (left, $P=0.004$; right, $P=0.02$; and NC, $P=0.05$):

- left aortic root shear during ejection
  $= (0.00342\pm 0.00104) \times \frac{dP}{dt_{\text{max}}}$

- right aortic root shear during ejection
  $= -(0.00103\pm 0.00040) \times \frac{dP}{dt_{\text{max}}}$

- NC aortic root shear during ejection
  $= (0.00037\pm 0.00017) \times \frac{dP}{dt_{\text{max}}}$

The negative regression coefficient between shear in the right aortic root region implies that the right root region underwent increasing counterclockwise shear with a higher maximum LV dP/dt, whereas the left and NC regions underwent increasing clockwise shear deformation. During ejection over the range of dP/dt_{max} studied, the clockwise shear of the left aortic root predicted by the model increased from 3.2$\pm$0.6° to 10.7$\pm$3.3°, counterclockwise shear of the right aortic root increased from 1.0$\pm$0.3° to 3.2$\pm$1.3°, and shear in the NC root increased from 0.35$\pm$0.16° to 1.2$\pm$0.5°.

LV end-diastolic pressure (EDP) and LV dP/dt_{max} were independent predictors of torsional deformation of the left and right aortic root regions, but no variables were linked significantly with torsional deformation of the NC root region. The general linear models for the left and right root torsion were as follows (left, $P=0.001$; right, $P=0.004$):

- left aortic root torsion during ejection
  $= (0.42\pm 0.12) \times \frac{\text{EDP}}{\text{dP/dt}_{\text{max}}} + (0.0050\pm 0.0013) \times \frac{\text{dP/dt}}{\text{dP/dt}_{\text{max}}}$

- right aortic root torsion during ejection
  $= -(0.12\pm 0.03) \times \frac{\text{EDP}}{\text{dP/dt}_{\text{max}}} - (0.0007\pm 0.0004) \times \frac{\text{dP/dt}}{\text{dP/dt}_{\text{max}}}$

Over the combined range of EDP and LV dP/dt_{max} studied, the clockwise torsion of the left aortic root predicted by the model increased from 9.3$\pm$2.5° to 29.9$\pm$8.1°, and counterclockwise torsion of the right aortic root increased from 2.0$\pm$0.7° to 6.3$\pm$2.2°.

Longitudinal deformation during ejection did not change significantly over the range of hemodynamic conditions examined.

### Aortic Root Deformation During IVR

During IVR, the aortic root underwent further circumferential contraction at both the annulus and the STJ, and it underwent further shearing and torsional deformation (Figure 3, Table 4). During IVR, the root underwent longitudinal compression as well. The greatest annular circumferential contraction occurred at the left annular sector, and the least occurred at the NC annular sector. In contrast to the asymmetric annular circumferential contraction, the left, right, and NC sinuses at the STJ contracted symmetrically. Longitudinal compression of the aortic root was also symmetric among the left, right, and NC regions of the aortic root.

Multivariate GLM identified that minimum LVP at the end of IVR and change in transvalvular pressure gradient ($\Delta$TVG) during IVR were the only hemodynamic variables related significantly to left and right annular contraction. Annular contraction, however, was not uniformly dependent on minimum LVP or $\Delta$TVG; it was asymmetric. The left annular sector underwent more circumferential contraction with a decrease in minimum LVP and an increase in the $\Delta$TVG, whereas the right annular sector contracted less as minimum LVP decreased and aortic $\Delta$TVG increased. The general linear models were as follows (left, $P<0.001$; right, $P=0.05$):

- left annular sector IVR contraction
  $= -(12.6\pm 1.9) + (0.49\pm 0.08) \times \text{LVP}_{\text{min}} - (0.067\pm 0.025) \times \Delta\text{TVG}$

- right annular sector IVR contraction
  $= -(5.5\pm 2.3) - (0.20\pm 0.1) \times \text{LVP}_{\text{min}} + (0.044\pm 0.031) \times \Delta\text{TVG}$

Circumferential contraction of the NC annular sector was independent of changes in the hemodynamic variables exam-
The range of longitudinal aortic root compression calculated shown below (shear, region were significantly affected by minimum LVP, as NC STJ contraction increased approximately uniformly from Over the range of minimum LVP studied, the left, right, and longitudinal sectors during IVR as minimum LVP decreases and TVG increases. Over the combined range of minimum LVP annular sectors during IVR as minimum LVP decreases and TVG, to 18.7±4.3%, at low minimum LVP and high aortic gradients. Conversely, contraction of the right annular sector decreased from 7.9±3.1% to 1.4±0.6%.

Circumferential contraction of the STJ, however, increased in a symmetrical and uniform manner, with decreasing minimum LVP giving rise to the following GLM equation fits (left, P=0.005; right, P=0.010; and NC, P=0.07):

left STJ IVR contraction=(0.24±0.07)×LVP\textsubscript{min}
right STJ IVR contraction=(0.24±0.08)×LVP\textsubscript{min}
NC STJ IVR contraction=(0.25±0.08)×LVP\textsubscript{min}

Over the range of minimum LVP studied, the left, right, and NC STJ contraction increased approximately uniformly from 0.24±0.07% to 4.8±1.6%.

Longitudinal compression of the left and NC aortic root regions increased as minimum LVP fell, whereas longitudinal compression in the right root region did not correlate significantly with minimum LVP, again suggesting a redistribution of dynamic stresses during IVR, with the left annular and root region sustaining proportionally greater amounts of stress, as shown below (left, P=0.029; NC, P=0.001):

left longitudinal IVR compression=0.26±0.11×LVP\textsubscript{min}
NC longitudinal IVR compression=0.16±0.04×LVP\textsubscript{min}

Only shear and torsional deformation in the left aortic root region were significantly affected by minimum LVP, as shown below (shear, P=0.028; torsion, P=0.042):

left shear during IVR=−(0.18±0.08)×LVP\textsubscript{min}
left torsion during IVR=−(0.31±0.14)×LVP\textsubscript{min}

The range of longitudinal aortic root compression calculated from the model was 0.18±0.08° to 3.6±1.6° for shear and 0.31±0.14° to 6.2±2.8° for torsion.

**Aortic Root Deformation During Diastole**

During early diastole, the aortic root recoiled from its dynamically loaded configuration at the end of IVR: the annular sectors and STJ segments expanded, and the root elongated longitudinally. Annular expansion during early diastole was asymmetric, however, with the NC annular sector having the least expansion. In addition, the aortic root untwisted and exhibited shearing and torsional deformation in a direction opposite to that seen during ejection and IVR (Figure 3, Table 5). The multivariate GLM did not identify a significant correlation with any of the hemodynamic variables for any of the 5 modes of aortic root deformation during diastole, suggesting that diastolic deformation is predominantly a recoiling process that restores the aortic root to its static equilibrium configuration.

**Discussion**

In this sheep experiment, we observed that a simple model of aortic root deformation during the cardiac cycle can be decomposed into 5 modes of deformation; however, several were asymmetric, and many were heterogeneous. The magnitude of any mode of strain deformation depends on the applied stress in a predictable viscoelastic manner and also on the contractile state of the ventricle, which suggests that a dynamic ventricular modulating effect on root deformation must occur in normal ovine root function. Prior published work has considered predominantly only 2 modes of deformation in an axisymmetric model of the aortic root: aortoven- tricular (which we called annular) and STJ deformation; moreover, our observations of the behavior of these 2 well-studied modes of deformation differ in several aspects from those described in older reports.14,18–22

Commissural expansion at the STJ has been the most extensively studied deformation of the aortic root. Brewer et al.,18 working in Stanton Nolan’s laboratory in the early 1970s, were the first to characterize accurately the changes in sinotubular diameter during the cardiac cycle. They demonstrated in dogs that the STJ underwent a 16% diameter change during the cardiac cycle, and they postulated that the increase in STJ diameter before ejection might facilitate aortic cusp opening. Subsequent work by Thubrikar and colleagues14,19,20 showed in dogs that this radial displacement of the commissures correlated closely with aortic pressure:

**TABLE 5. Annular and STJ Circumferential Deformations and Aortic Root Longitudinal, Shear, and Torsional Deformations During Diastole**

<table>
<thead>
<tr>
<th></th>
<th>AV-Circ,* † %</th>
<th>ST-Circ,* † %</th>
<th>Long,* † %</th>
<th>Shear,* degrees</th>
<th>Tor,* degrees</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left</td>
<td>11.1±1.6</td>
<td>4.0±1.3</td>
<td>2.8±0.9</td>
<td>−3.2±2.1</td>
<td>−6.1±4.4</td>
</tr>
<tr>
<td>Right</td>
<td>10.3±2.8</td>
<td>5.8±2.7</td>
<td>3.3±1.0</td>
<td>2.3±0.6</td>
<td>3.8±0.8</td>
</tr>
<tr>
<td>Posterior</td>
<td>3.7±1.5</td>
<td>3.1±0.9</td>
<td>3.1±1.2</td>
<td>−2.2±1.5</td>
<td>−4.6±3.5</td>
</tr>
</tbody>
</table>

Data are expressed as mean±SD. Abbreviations as in Table 2.

*Changes during diastole are significant at P<0.001 by repeated measures ANOVA.
†Differences between the 3 groups (left, right, and NC) in the measured change are significant at P<0.001 by a 3-factor repeated measures ANOVA.
when aortic pressure increased during systole, the commissures moved radially outward; they subsequently moved inward when aortic pressure decreased during diastole. These investigators also found that at any instant in the cardiac cycle, STJ perimeter decreased as the systemic arterial pressure decreased. More recent work by Lockie et al.\(^{21}\) explored the relationship of dilation pressure on the dimensions of cadaveric homograft aortic roots. They demonstrated a 33% increase in sinotubular diameter when aortic root pressure rose from 0 to 80 mm Hg, and a 44% diameter increase when the pressure was increased to 120 mm Hg.

Hansen and colleagues\(^{22}\) explored both the STJ radial and longitudinal distensibility of the porcine aortic root in both a static model and in a suspended heart model. In the suspended heart model, they showed similar STJ radial strains of the left, right, and NC sinuses, with maximum strain occurring at peak aortic pressure and minimum strain occurring at end diastole. Longitudinal strains did not change in a consistent pattern of deformation, which these investigators partially attributed to limitations in their experimental preparations.

The conformational changes in the STJ of sheep that were seen with the high-precision 3D marker technology differed in various respects from these older observations. One finding, however, was consistent: STJ diameter was maximal near the time of peak LVP, which would facilitate aortic valve opening and possibly reduce cuspal opening or bending stresses. Minimum STJ diameter, however, did not occur at the end of systole with valve closure, but rather at the end of LVP decay when LVP was minimal (end-IVR and very early diastole) and the diastolic TVG was maximal. Furthermore, the sinotubular diameter at the end of LVP decay fell with increasing TVG. That relationship suggests that the annular commissures participate in load sharing to reduce peak stresses on the aortic cusps during the rapid increase in TVG immediately after valve closure. Immediately after the end of LVP decay, we found that the STJ underwent significant, rapid, early reexpansion, which was possibly due to the commissures recoiling from their deformed state during IVR caused by the rapid increase in transvalvular load after LVP decay. These observations suggest that the maintenance of annular and commissure flexibility may be a key component that allows the aortic root to dissipate cuspal stresses during valve closure. The magnitude of STJ expansion during IVC increased significantly with larger LV EDV, suggesting that the aortic root adapts to accommodate a greater preejection volume, which might assist with early rapid LV ejection. The increase in STJ diameter. Thus, during IVC contraction, the aortic root underwent a cylindrical dilation (dilation of the STJ and annulus together with root elongation) that accommodated a greater preejection LV volume. We conjecture that the shape change improves transvalvular hemodynamics and reduces abnormal stresses on the cusps. On the basis of those teleological interpretation, aortic valve substitutes and repair procedures that disturb the aortic root's physiological response to increased pump function and stroke volume may lead to accelerated SVD.

During ejection, annular contraction paralleled LV contractility and, subsequently, during IVR, annular diameter decreased further as a function of lower minimum LVP. These annular changes were asymmetric during those periods; during IVR with decreasing minimum LVP, contraction of the left annular sector increased, whereas contraction of the right annular sector actually decreased. Thus, an increase in transvalvular pressure load during IVR will cause a redistribution of load from the right annular sector to the left annular sector, which strongly implicates the structural role of the aortic root in determining its functional behavior. This complex dependence of annular dynamics on ventricular volumes, pressures, and contractility differs substantially from the results observed by previous investigators.\(^{14,18–23,25}\)

Aortic annulus enlarging procedures; total root replacement with homografts, autografts, or stentless xenografts; and rigid stents in porcine and allograft bioprostheses all may disrupt the complex normal annular dynamics and, therefore, predispose the valve to accelerated cuspal failure. However, subcoronary implantation of allograft, autograft, and stentless xenograft valves should not be associated with that theoretical drawback.

An important novel finding in this study was the characterization of a torsional component of aortic root deformation. The left aortic root region (the area under the NC-left commissure) underwent the greatest amount of circumferential and torsional deformation during the cardiac cycle, and the NC root region (under the right-NC commissure) deformed the least. Despite the large annular complex asymmetric deformations, the deformations at the STJ were symmetric, without shear or torsion. The normal sheep aortic root seems to dissipate the strains (dimensional changes) and shear strains created by LV contraction and torsion within its short length. Through that important mechanism, the dynamic aortic root may minimize opening (or even closing) shear stresses on the valve cusps near their insertion along the commissures that might occur if the commissures underwent torsion in response to the torsion occurring at the annulus.
Semirigid stenting of stentless xenograft valves (eg, the Dacron mesh covering of either the TSPV [St. Jude Medical, Inc] or Freestyle [Heart Valve Division, Medtronic, Inc] valves) may substantially alter aortic root compliance to torsional deformation. Those changes could, in theory, couple the torsional deformations of the annulus to the STJ, creating abnormal distribution of closing stresses on the aortic cusps; those stresses may portend accelerated cusp fatigue and eventual SVD.

Because of the morphological and functional asymmetry of the aortic root, regional differences in the response of the structures within the aortic root to changes in hemodynamics were observed. The most interesting pattern was the dependence of left annular sector deformation (subtending the NC-left commissure) on LV contractility, independent of other hemodynamic variables, and the absence of any interaction between contractility or hemodynamics and deformation of the NC annular sector. At the level of the STJ, however, dilation and expansion during the cardiac cycle were symmetric and uniform. Subsequently, during diastole, the aortic root underwent substantial deformation that was independent of changes in LV contractility or hemodynamics. During that period, the degree of deformation was best predicted by the extent of annular deformation preceding diastole, as the aortic root recoiled into its baseline configuration.

The intricate physiological changes within the aortic root and their contribution to aortic valve leaflet function seem to follow the postulate of Zimmermann,26 which states that the heart is a structure-function continuum. The dynamic aortic root seems to be designed to reduce cusp stresses and, thereby, fatigue and the likelihood of eventual SVD of substitute valves13,14,21,27 through its role in the complex opening and closing mechanics of the cusps. Christie and Barratt-Boyes12 used finite element analysis to study the effects of the aortic root on cusp stress distribution because previous morphological studies linked the orientation of the collagen fibers within the cusps to their stress-strain characteristics.28 The collagen orientation within the cusps allows significant radial extensibility at small loads to facilitate coaptation during valve closure. Loss of radial extensibility or alterations in cusp-loading conditions due to changes in aortic root 3D dynamics can produce poor coaptation that redistribute the stresses from the cusp bellies toward the commissures, predisposing to accelerated SVD.11,29 Christie and Barratt-Boyes12 also demonstrated that stent-post flexibility is an important component of the aortic root when the cusps have low radial extensibility and poor coaptation; under those conditions, flexible stent posts help preserve the normal distribution of high cusp stresses in the bellies and low stresses at the commissures.

The mathematical models used in the present study demonstrated that the mechanical properties of the aortic root must be balanced with the mechanical characteristics of the cusps for optimal load sharing; thus, in addition to preserving the dynamic nature of the aortic root, aortic valve replacement using tissue valves must also strive to balance the mechanical properties of the root with those of the cusps to prevent premature SVD. Given that hypothesis, surgical implantation of allograft, autograft, or unstented xenograft valves in a subcoronary, scalloped fashion that minimally disrupts aortic root dynamics will theoretically translate into a lower probability of SVD or other pathological changes.

Preservation of the mechanical advantages of the normal aortic root and cuspal complex requires that surgeons and manufacturers of substitute aortic valve replacements be cognizant of the unique structural asymmetry of aortic root dynamics and the implied functional asymmetry. The structural asymmetry may be due to the aortic root’s connections with adjoining cardiac structures. In addition, anatomical studies of the aortic root have repeatedly demonstrated geometric and histologic differences between the left, right, and NC regions.30–32 Histologic studies of the aortic root have confirmed that LV myocardium extends into the root and abuts the anatomical annulus, especially along the annulus of the muscular anteroseptal root region.25,31 Disturbing these structural differences may alter the natural asymmetry of aortic root physiology and thereby create abnormal loading and shearing stresses on the aortic valve cusps. The complexity of aortic root structure and the role of its structural elements in maintaining normal root function suggest that preservation of the normal aortic root structure during aortic valve preservation or replacement surgical procedures may be a critical determinant of long-term tissue durability.

Study Limitations

Comparisons among the various experimental results on aortic root dimension changes during the cardiac cycle is complicated by differences in species, differences in preparation (ranging from explanted cadaveric aortic roots to isolated heart models to closed-chest studies), and major methodological limitations in both temporal and spatial resolution and accuracy in older experiments. Identification of physiological determinants of aortic root deformation using a multivariate GLM between modes of deformation and variables reflecting hemodynamics and contractility is not sensitive to the bilinear stress-strain response of the aortic root nor to the known viscoelastic behavior of the root.24 Furthermore, regional variation in root deformation has been attributed to regional variation in morphology and histologic properties without consideration of the effect of regional variations in LVP and flow hemodynamics. Comparative anatomy of the human and ovine aortic valve differs somewhat in terms of the relative myocardial contribution to the aortoventricular circumference or annulus (~43% in humans and 58% in sheep) and the density of myocardial tissue extension into the aortic root.32 That difference suggests that the effect of the myocardial elements within the aortic root on aortic root and valve function may possibly be less pronounced in humans than in sheep.

Acknowledgments

We thank Mary K. Zasio, BA, Carol W. Mead, BA, and Erin K. Schultz, BS, for their technical assistance. This work was supported in part by Grants HL-29589 and HL-48837 from the National Heart, Lung, and Blood Institute. Drs Dagum and Green were supported by National Heart, Lung, and Blood Institute Individual Research Service Awards HL10000-01 and HL-09569, respectively. Dr Timek is a recipient of the Thoracic Surgery Foundation Research Fellow-
ship Award. Drs Dagum, Green, and Timek are Carl and Leah McConnell Cardiovascular Surgical Research Fellows.

References


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Circulation. 1999;100:II-54-II-62
doi: 10.1161/01.CIR.100.suppl_2.II-54

Circulation is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231
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Print ISSN: 0009-7322. Online ISSN: 1524-4539

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