Impact of Wall Thickness and Saccular Geometry on the Computational Wall Stress of Descending Thoracic Aortic Aneurysms

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Background—Wall stress calculated using finite element analysis has been used to predict rupture risk of aortic aneurysms. Prior models often assume uniform aortic wall thickness and fusiform geometry. We examined the effects of including local wall thickness, intraluminal thrombus, calcifications, and saccular geometry on peak wall stress (PWS) in finite element analysis of descending thoracic aortic aneurysms.

Methods and Results—Computed tomographic angiography of descending thoracic aortic aneurysms (n = 10 total, 5 fusiform and 5 saccular) underwent 3-dimensional reconstruction with custom algorithms. For each aneurysm, an initial model was constructed with uniform wall thickness. Experimental models explored the addition of variable wall thickness, calcifications, and intraluminal thrombus. Each model was loaded with 120 mm Hg pressure, and von Mises PWS was computed. The mean PWS of uniform wall thickness models was 410±111 kPa. The imposition of variable wall thickness increased PWS (481±126 kPa, P < 0.001). Although the addition of calcifications was not statistically significant (506±126 kPa, P = 0.07), the addition of intraluminal thrombus to variable wall thickness (359±86 kPa, P ≤ 0.001) reduced PWS. A final model incorporating all features also reduced PWS (368±88 kPa, P < 0.001). Saccular geometry did not increase diameter-normalized stress in the final model (77±7 versus 67±12 kPa/cm, P = 0.22).

Conclusions—Incorporation of local wall thickness can significantly increase PWS in finite element analysis models of thoracic aortic aneurysms. Incorporating variable wall thickness, intraluminal thrombus, and calcifications significantly impacts computed PWS of thoracic aneurysms; sophisticated models may, therefore, be more accurate in assessing rupture risk. Saccular aneurysms did not demonstrate a significantly higher normalized PWS than fusiform aneurysms. (Circulation. 2013;128(suppl 1):S157-S162.)

Key Words: aneurysm ■ finite element analysis ■ mechanical stress ■ modeling ■ rupture ■ risk stratification

Aortic diseases, including aortic aneurysms, are the twelfth leading cause of death in the United States. Although abdominal and ascending aortic aneurysms are more common, the incidence of descending thoracic aortic aneurysms (TAAs) is increasing, reaching 10.4 per 100,000 person-years. The overall mortality rate of ruptured TAAs has been reported as high as 97%, with ≈60% of patients dying in the prehospital environment.

Asymptomatic TAAs are typically operated when the maximum aortic diameter reaches 6.5 cm, which is associated with a 14.1% annual risk of rupture, dissection, or death. However, smaller aneurysms also rupture not infrequently: TAAs with maximum diameters between 5 and 6 cm have been shown to exhibit annual risks of ≈6.5%. Therefore, improved rupture risk stratification will allow more rational surgical intervention and will likely lead to improved survival in patients with TAAs.

The superiority of computational wall stress compared with a maximum diameter criterion in the assessment of rupture risk has been demonstrated in several studies. Wall stress has been shown in abdominal aortic aneurysms to be associated with rupture and aneurysm growth. Furthermore, peak wall stress (PWS) has been found to be a more sensitive and specific predictor of rupture than diameter alone. Despite the sparse application of numeric simulation techniques to the descending thoracic aorta, PWS in TAAs undergoing radiological and clinical surveillance has been found to correlate closely with observed aneurysmal growth rates.

The accuracy of numeric simulation in aortic aneurysms is predicated on precise reconstructions of aneurysm geometry, as well as realistic material properties of the arterial wall and other aneurysm components, such as intraluminal thrombus (ILT) and wall calcifications. Current preoperative imaging techniques are unable to discern the material properties...
of individual aneurysms, rendering high-fidelity geometric reconstructions even more essential.

A review of the methods used in the calculation of wall stress in aortic aneurysms yields a heterogeneous mixture of techniques that vary significantly in both geometric reconstruction and numeric simulation. Despite the fact that significant regional variations in aortic thickness have been demonstrated in pathological specimens,\(^{9,10}\) the vast majority of biomechanical analyses of aortic aneurysms assume uniform wall thickness. Reeps et al\(^{11}\) demonstrated significant variations in calculated wall stress with models of differing sophistication, but even their methods were based on uniform wall thickness aortic geometries. Although it may seem intuitive that varying wall thickness will significantly impact wall stress maxima and distribution, efforts at examining the effects of incorporating image-based, locally resolved, wall thickness in aortic aneurysm stress analysis have been limited.

Meanwhile, saccular aneurysms have historically been perceived by vascular surgeons as possessing a greater rupture risk than their fusiform counterparts, but most studies have exclusively looked at fusiform aneurysms. The true natural history of saccular aneurysms remains unknown, but recent retrospective studies have cast doubt on the malignant nature of these aneurysms.\(^{12}\) Our laboratory recently showed greater diameter-normalized wall stress in saccular compared with fusiform aneurysms in simple finite element models of the arterial wall only, with linear material properties and uniform wall thickness.\(^{13}\) However, it is unclear that these effects would be preserved in models incorporating locally resolved wall thickness, as well as intramural calcifications and ILT.

**Methods**

**Patients**

A total of 10 patients undergoing computed tomographic angiography for aneurysms of the descending thoracic aorta were retrospectively identified. Patients were identified by querying studies from the Radiology Information System (Centricity RIS-IC; GE Healthcare, Waukesha, WI) using PRESTO (Montage Healthcare Solutions, Philadelphia, PA). Approval was obtained from the Institutional Review Board of the University of Pennsylvania, and the need for informed consent was waived.

Five consecutive fusiform aneurysms and 5 consecutive aneurysms of saccular morphology were selected. Aneurysms were considered saccular if aortic dilation was confined to a discrete section of the arterial wall, as opposed to the entire circumference of the vessel. Aneurysms secondary to connective tissue disorders, aneurysms of infectious cause, and aneurysms with concomitant aortic dissections were excluded because the elastic properties of the aortic wall under these conditions are unknown.

**Image Segmentation and Aneurysm Reconstruction**

Three-dimensional TAA geometries were reconstructed from individual stacks of 2-dimensional computed tomographic angiography images using a series of custom MATLAB (The MathWorks, Natick, MA) algorithms. Automatic windowing was applied to improve the contrast between arterial structures and surrounding tissues. Subsequently, an anisotropic diffusion filter was applied to the images to reduce noise and enhance borders.\(^{14}\)

Lumen boundaries were obtained automatically by using an algorithm based on active contours without edges, originally described by Chan and Vese.\(^{15}\) Although this algorithm was largely accurate and user-independent, it had a tendency to bleed into adjacent high-intensity structures, such as the spinal column. Therefore, these areas were identified on precontrast image series and subsequently subtracted from the luminal segmentation.

The outer adventitial border was traced using a semiautomated method incorporating isolines constructed from smoothed pixel intensities. The algorithm automatically selected a threshold using curvature constraints and comparison with previous slices. In portions of the aortic wall where ILT was present, thrombus was differentiated from arterial wall using a combination of texture features, including mean pixel intensity, SD, and entropy. Areas corresponding to thrombus were defined as being outside the luminal boundary but inside the boundary marking the inner arterial wall. Because both inner and outer boundaries of the arterial wall were separately defined, local wall thickness could be resolved. The image segmentations were subsequently reviewed for each slice, and manual correction was used as necessary.

Gaussian curvature-based surface smoothing was applied to both the arterial wall and the ILT surfaces. The resulting surfaces were meshed into hexahedral and tetrahedral elements for the arterial wall and thrombus, respectively. Meshes consisted of 2.5x10\(^6\) to 1.25x10\(^6\) elements depending on thrombus volume. The reconstructed mesh representing the final mesh discretization of the aneurysm was then exported to Abaqus/CAE (v6.11, Simulia, Providence, RI) for finite element analysis (FEA).

**Material Properties**

Both ILT and the arterial wall were assumed to be hyperelastic, homogeneous, and isotropic. They were modeled using the nonlinear hyperelastic formulation derived from Raghavan and Vorp, who examined the uniaxial properties of excised abdominal aortic aneurysms (AAA).\(^{16,17}\) The strain energy functions used to model the arterial wall and ILT were as follows:

\[
W = C_W \left( I - 3 \right) + C_L \left( I - 3 \right)^2
\]

\[
W = D_W \left( I - 3 \right) + D_L \left( I - 3 \right)^2
\]

for the arterial wall

for ILT

In these formulations, \(W\) represents strain energy, and the constants \(C_W, C_L, D_W, D_L\) represent material parameters for the wall and ILT, respectively. Population mean values of \(C_W=0.174\) MPa, \(C_L=1.881\) MPa, \(D_W=0.026\) MPa, and \(D_L=0.026\) MPa were used. Calcified elements were modeled using linear elastic material properties, with Poisson’s ratio of 0.45 and Young’s modulus of 45 MPa.\(^{16}\)

**Numeric Simulation**

Wall stress was calculated using the finite element method, where complex geometry is parsed into small elements on which numeric simulations are more readily performed. A uniform pressure of 120 mm Hg was applied to the luminal surface in Abaques, and a nonlinear large deformation model was used. The arterial wall and ILT were composed of first-order hexahedral (C3D8) and tetrahedral (C3D4) elements, respectively. Contact with adjacent structures, including the spine and other adjacent organs, was not considered. Similarly, shear stress generated by blood flow was not examined because the effects have been previously determined to be minor. The aorta was translationally fixed at the proximal and distal ends, as well as by the head vessels. A no-slip condition was placed on the interface between the arterial wall and the ILT. The von Mises stress, an axis-independent scalar, is frequently used as an indication of material failure and was reported for that purpose.

**Model Construction**

For each individual aneurysm, an initial model was constructed using a uniform wall thickness representation of the arterial wall. The value chosen for wall thickness was the average wall thickness over the segmented portions of each individual aorta. Five additional experimental models added the effects of variable wall thickness (VWT), ILT, and calcifications. Model definitions are presented in Table 1.

**Statistical Analysis**

Statistical analysis was performed using MATLAB. Continuous variables were compared using paired \(t\) tests or Wilcoxon signed-rank
**Table 1. Model Definitions in Descending Thoracic Aortic Aneurysm Simulations**

<table>
<thead>
<tr>
<th>Model</th>
<th>Wall Thickness</th>
<th>Calcifications</th>
<th>Intraluminal Thrombus</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Uniform</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>2</td>
<td>Variable</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>3</td>
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</tr>
<tr>
<td>5</td>
<td>Uniform</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>6</td>
<td>Variable</td>
<td>Yes</td>
<td>Yes</td>
</tr>
</tbody>
</table>

Tests, where appropriate. All P values were 2-sided, and a value of P<0.05 was considered to be statistically significant. All results are expressed as mean±SD, unless otherwise specified.

**Results**

Ten aneurysms in 10 patients were analyzed under different combinations of wall thickness conditions and materials. Individual aneurysm characteristics are shown in Table 2. The average patient age was 69.3±12.1 years. The average maximum aortic diameter was 51±10 mm. The average wall thickness in aneurysmal portions of the aorta was 1.81±0.15 mm, which was comparable with prior tissue studies.5 A single representative saccular TAA reconstruction with minimal calcifications and thrombus, with the stress contour distribution overlaid, is shown in Figure 1.

**Wall Thickness**

The incorporation of VWT into aortic reconstructions resulted in overall higher PWSs in all simulated TAAs when comparing models 1 and 2 (410±111 versus 481±126 kPa, P<0.001). PWSs were increased by 19.7% (range, 10.5%–39.8%). Areas of high wall stress often colocalized with portions of the descending thoracic aorta with low wall thickness. This was most evident when directly comparing models 1 and 2, but this colocalization persisted even when considering the most sophisticated model (Figure 2). When comparing models 5 and 6 that included intramural calcifications and ILT, the effect of VWT remained significant (339.4±84 versus 363±88 kPa, P<0.001).

**Calcifications**

The inclusion of calcifications in model 3 did not significantly influence PWS compared with model 2 (481±126 versus 506±126 kPa, P=0.07). However, the presence of intramural calcifications was found to significantly alter the spatial stress distribution (Figure 3). Calcifications attract stresses, resulting in decreased wall stress in the arterial wall, but possibly leading to stress concentrations at the edges of the calcified areas.

**Intraluminal Thrombus**

The inclusion of ILT into the aortic geometry of model 2 significantly decreased PWS experienced by the arterial wall in model 4 (481±126 versus 368±88 kPa, P<0.001). A similar effect was found when comparing models 3 and 6 (P<0.001), reducing wall stress by an average of 38.3% (range, 18.5%–56.7%). The presence of thrombus decreases the luminal surface area and, therefore, decreases the area over which the force of blood pressure acts.

**Saccular Geometry**

In a majority of cases (n=9), the location of PWS did not coincide with the plane of maximum diameter. Rather, it was found that areas of high stress coincided with areas of abrupt change in aneurysm morphology, such as the aneurysm neck. No difference between saccular and fusiform aneurysms was found in the computed wall stresses in model 6 (363±87 versus 364±89 kPa, P=0.98). Similarly, diameter-normalized wall stresses were statistically indistinguishable (77±7 versus 67±12 kPa/cm, P=0.22). No significant differences between saccular and fusiform aneurysms were noted with respect to maximum aneurysm diameter (55±9.8 versus 46.7±9.6 mm, P=0.24), the degree of calcification (9.5±10.1% versus 9.6±8.8%, P=0.99), or the ILT fraction (23.4±16.1% versus 25.0±15.3%, P=0.86).

**Discussion**

The goal of structural analysis for aortic aneurysms is to provide better risk stratification than diameter measurements alone. Although there is growing evidence in the literature that wall stress may be a superior predictor of clinical outcome, there are no standardized methods for performing this analysis. The current study represents an attempt to determine the role of model complexity in the stress analysis of descending TAAs, with particular attention to the effects of aortic wall thickness and saccular geometry.

By comparing models assuming uniform wall thickness to VWT models, it was found that considering local wall thickness significantly increased estimates of PWS. This was not only evident when comparing the most simplistic models (1 and 2) that only considered the aortic wall but also when taking into account ILT and wall calcifications (models 5 and 6). Similarities were also found in the distribution of wall thickness and the stress contour plots of individual aneurysms, suggesting that wall thickness had a significant effect on stress.
distribution alongside aneurysm geometry. As a result, uniform wall thickness models may not accurately estimate wall stress and could potentially lead to incorrect assessments of rupture potential. This suggests that improvements in wall thickness determination, either through image analysis techniques or increased computed tomographic angiography scan resolution, will improve the accuracy of FEA models.

Our finding that inclusion of wall thickness significantly impacts computational hydrostatic pressure predictions echoes calls for inclusion of variable and regional wall thickness in modeling aortic aneurysms.\textsuperscript{20,21} Although other attempts have been made to study the influence of wall thickness on FEA-predicted aortic wall stress, prior models have not incorporated actual image-based regional VWT.\textsuperscript{7}

The inclusion of calcifications was found to dramatically alter the distribution of wall stress but did not significantly impact its magnitude. Given our low sample size (n=10) and a \( P \) value approaching statistical significance, this may represent a type II error. The inclusion of ILT was shown to have a buffering effect, systematically reducing PWS. These effects are consistent with other findings in the literature.\textsuperscript{11,18}

In a majority of the aneurysms examined, the location of PWS was found in areas with sudden morphological change. This correlates with computational and experimental studies performed using in vitro AAA models showing that points of maximum PWS and points of rupture coincide with regions of inflection and not with maximum aortic diameter.\textsuperscript{22} High wall stress in aneurysm necks has also been linked to increased aneurysm growth. Although an exact mechanism is unclear, it has been postulated that increased PWS leads to upregulation of matrix metalloproteinases, resulting in increased matrix degradation and progressive aneurysmal dilatation.\textsuperscript{6}

Wall stress was calculated using finite element techniques, which do not include the effect of blood flow. Traditionally, the impact of wall shear stress has been ignored because studies in the thoracic aorta have shown it to be several orders of magnitude lower than static stresses calculated using pressure–deformation analyses.\textsuperscript{19} Furthermore, because fluid structure interaction techniques involve a cosimulation of fluid flow fields and static stress, one would expect the effects of wall thickness, calcifications, and thrombus to persist. Although incorporating VWT in FEA models of the descending thoracic aorta significantly impacts calculated PWS, its effect on the predictive capability of the overall technique is unknown.

The modeling strategy described herein ignores the breaking or failure strength of the aortic wall; some investigators have attempted to compute a rupture potential index that compares FEA-predicted wall stress with a complicated formula for predicted wall strength.\textsuperscript{23} Furthermore, we have chosen to examine the influence of wall thickness, calcifications, and mural thrombus on peak von Mises wall stress. Although other components or invariants of aortic wall stress may be more

Figure 1. A. Raw computed tomographic angiogram cross-sectional image with segmentation into luminal (blue), inner arterial (red), and outer adventitial (green) surfaces. B. Stress contour plot of entire descending thoracic aorta.

Figure 2. A. Stress contour plot for an individual thoracic aortic aneurysm (TAA) under assumptions of model 1 (uniform wall thickness, no calcifications or thrombus). B. Model 2 (variable wall thickness [VWT], no calcifications or thrombus). C. Model 6 (VWT, with calcifications and thrombus). D. Wall thickness map of corresponding TAA. Note the colocalization of wall thickness distributions and spatial distribution of wall stress.
relevant, the importance of incorporating local wall thickness is expected to be consistent.

Saccular aneurysms were not found to have significantly elevated PWS compared with fusiform aneurysms, even when PWS was normalized to aneurysm diameter. In our series, these 2 groups were statistically indistinguishable in terms of diameter, thrombus content, and degree of wall calcification. Although this may represent the effect of a small sample size, it can be interpreted that individual aneurysm characteristics outweigh the classification of aneurysm as fusiform or saccular. The clinical implication is that aneurysms should be taken individually and not repaired or observed based on simple classification schemes.

There are important limitations to this study. One significant limitation is that the wall thicknesses calculated based on computed tomographic angiography data have not been validated against measurements of actual tissue. Unlike other arterial structures, such as the carotid artery where arterial thickness has been validated with duplex ultrasound, no universally accepted noninvasive method of measuring aortic wall thickness exists. Although we believe our wall thickness measurements to be accurate, validation of these image analysis techniques clearly represents a direction of further research, currently underway in our laboratory.

Similarly, despite manual review of segmentation of each image slice, we recognize that the distribution of wall calcification and ILT may also be subject to small degrees of error. Unfortunately, there is no standardized method regarding aortic aneurysm image segmentation and 3-dimensional reconstruction. The establishment of such a standard and its validation remain an area of future research.

An additional limitation is the necessary use of modeling assumptions. Both the aortic wall and ILT were assumed to have isotropic material properties, although several studies have demonstrated anisotropy in both materials. Unfortunately, the degree of anisotropy has been shown to vary significantly between individuals, and currently there exists no way of accurately estimating an individual patient’s aortic material properties in vivo.

Overall, the inclusion of locally VWT has been found to significantly impact finite element estimates of PWS and that aneurysms with saccular configurations do not have intrinsically higher wall stress. This information may prove to be useful in the construction of future finite element models and ultimately may improve the predictive capabilities of biomechanical analysis.

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Disclosures
None.

References


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