MECHANICAL STRESSES IN ABDOMINAL AORTIC ANEURYSM. MATERIAL ANISOTROPY A PARAMETRIC STUDY

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Summary. Biomechanical studies suggest that one determinant of abdominal aortic aneurysm (AAA) rupture is related to the stress in the wall. To date, stress analysis conducted on AAA is mainly driven by isotropic tissue models. However, recent biaxial tensile tests performed on AAA tissue samples demonstrate the anisotropic nature of this tissue. The purpose of this work is to study the influence of geometry and material anisotropy on the magnitude and distribution of the peak wall stress in AAAs. Three-dimensional computer models of symmetric and asymmetric AAAs were generated. A five parameter exponential type structural strain-energy function was used to model the anisotropic behavior of the AAA tissue. The anisotropy is determined by the orientation of the collagen fibers (one parameter of the model). The results suggest that shorter aneurysms are more critical when asymmetries are present. They show a strong influence of the material anisotropy on the magnitude and distribution of the peak stress.

1 INTRODUCTION

An abdominal aortic aneurysm (AAA) is an abnormal widening of the aorta which is related to weakness of the vessel wall (associated to degradation of connective tissue). AAAs are potentially life-threatening medical conditions often requiring surgical intervention. These interventions, however, continue to pose serious risk on patients with a mortality rate of about 5% on patients with stable AAA [1]. Therefore, (biomechanical) indicators of AAA rupture, and definitions of better criteria for surgical intervention are of pressing need.

At the moment, the major criterion that has been used for decision making whether an aneurysm should be operated or not is the aortic size (diameter)[2]. Studies conducted on abdominal aortic aneurysms[3, 4, 5] , suggest peak wall stress as a more reliable parameter
for the assessment of the risk of AAA rupture. Therefore, a method to find more reliably an estimate of the AAA wall stresses could result in a useful tool for clinicians in assessing the risk of AAA rupture. An accurate and reliable stress analysis of AAA requires not only a precise three-dimensional description of the lesion but also an appropriate constitutive law for the AAA material. In this regard, most of the previously reported AAA studies have used isotropic models [3, 4, 5]. Such models, however, are limited for AAA stress analysis since \textit{ex vivo} biaxial experiments on human AAA tissue, recently documented in [6], demonstrated that the aneurysmal degeneration of the aorta leads to an increase in mechanical anisotropy, with the circumferential direction being the preferential stiffening direction. Therefore, it is advantageous to use anisotropic constitutive models for AAA stress analysis, as, e.g., in [6].

The particular aim of this work is to determine how geometry and material anisotropy influence the magnitude and distribution of the peak wall stress in the AAA. Three-dimensional AAAs were generated by using a parametrization [7] in which maximum diameter and aneurysm length can be individually controlled. A modified form of the three-dimensional strain-energy function (SEF), as proposed in [8], was used to model the anisotropic behavior of the AAA tissue with material parameters obtained by fitting the model to biaxial data recently reported in [6]. The results from this constitutive approach were then compared with results obtained by taking the AAA tissue as isotropic, and using the constitutive model proposed in [9].

2 METHODS

2.1 Geometric model

We use here a parameterized geometric aneurysm model to better identify the effect of each individual geometric variable (diameter, length, and asymmetry) on the overall mechanical response of the aneurysm wall. We generated (idealized) geometric models by means of the commercial software IDEAS such that the cross section at any axial position is circular. The shape of the aneurysm is defined by a ‘parabolic-exponential shape’ [7]

\[ R(Z) = R_a + \left( R_{an} - R_a - c_3 \frac{Z^2}{R_a} \right) \exp \left( -c_1 \frac{|Z|}{R_a} \right)^{c_2}, \]

where \( R_a \) is the radius of the healthy artery, \( R_{an} \) is the maximum radius of the aneurysm, \( c_1 \) is a constant to be taken as 5.0, and \( c_2, c_3 \) are dimensionless geometrical parameters depending on the geometry of the aneurysm according to

\[ c_2 = \frac{4.605}{(0.5L_{an}/R_a)^c_1}, \quad c_3 = \frac{R_{an} - R_a}{R_a} (0.8L_{an}/R_a)^2, \]

where \( L_{an} \) defines the length of the aneurysm.

In order to study the effect of the AAA geometry on the distribution of the wall stresses we introduce three (dimensionless) geometrical parameters, i.e. \( F_R = \frac{R_{an}}{R_a}, \quad F_L = \frac{L_{an}}{R_a}, \quad F_E = \frac{e^{(F_R-1)/F_L}}{F_R-1} \). The parameter \( F_R \geq 1 \) defines the ratio between the maximum AAA radius and the healthy arterial radius, \( F_L \) defines the ratio between the length of the aneurysm and the maximum AAA radius, while \( F_E \in [0, 1] \) is a measure of the aneurysmal eccentricity. The range of the values \( F_R \) and \( F_L \) were given in Table 1. For all models we assumed the wall thickness to be uniform, with 1.5 mm [3], and the arterial radius was considered to be \( R_a = 10.1 \) mm. The
overall length of the lesion, say $L$, was set to be $1.3L_{an}$, and, therefore, ranged from 64.6 mm for case 1, to 90.4 mm for case 9. This length was also found to be the minimum required for the boundary conditions to have no effect on the stress distribution in the lesion.

<table>
<thead>
<tr>
<th>Case</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
<th>9</th>
</tr>
</thead>
<tbody>
<tr>
<td>$F_R$</td>
<td>2.0</td>
<td>2.0</td>
<td>2.0</td>
<td>2.375</td>
<td>2.375</td>
<td>2.75</td>
<td>2.75</td>
<td>2.75</td>
<td>2.75</td>
</tr>
<tr>
<td>$F_L$</td>
<td>1.5</td>
<td>2.0</td>
<td>2.5</td>
<td>1.5</td>
<td>2.0</td>
<td>2.5</td>
<td>1.5</td>
<td>2.0</td>
<td>2.5</td>
</tr>
</tbody>
</table>

Table 1: Range of the geometric parameters $F_R = R_{an}/R_a$ and $F_L = L_{an}/R_{an}$ defining the aneurysmal shape.

### 2.2 Material model

Physiologic and biomechanical studies show that the AAA wall is a heterogeneous material undergoing large strains prior to failure, and to deformed nearly volume-preserving\(^2\),\(^2\). In this work the aneurysmal tissue is assumed to behave isochoric and anisotropic undergoing large strains. In order to address the effect of the anisotropy on the material response, isotropic and anisotropic constitutive relations for the AAA are considered. For the case of isotropy the material response of the aneurysm is characterized by the SEF \([9]\)

$$
\Psi = U(J) + \frac{C_1}{C_2} \left[ e^{\frac{C_2}{2} (I_1 - 3)} - 1 \right] ,
$$

where $C_1$ is a constant with the dimension of stress and $C_2$ is dimensionless, and $\bar{I}_1$ is the first invariant of $\mathbf{C}$. For the case of anisotropy, the isochoric part of the SEF is additively decomposed into an isotropic contribution, corresponding to the matrix material, and an anisotropic contribution, related to the (two families of) collagen fibers, i.e \([10]\).

$$
\Psi = U(J) + C_1 (\bar{I}_1 - 3) + \frac{k_1}{2k_2} \left\{ e^{k_2 ([1-\rho] (\bar{I}_1 - 3)^2 + \rho (\bar{I}_4 - \bar{I}_6^0)^2)} - 1 \right\} + \frac{k_3}{2k_4} \left\{ e^{k_4 ([1-\rho] (\bar{I}_1 - 3)^2 + \rho (\bar{I}_4 - \bar{I}_6^0)^2)} - 1 \right\} ,
$$

where $C_1$ is a stress-like material parameter, and $k_1, \ldots, k_4$ are material parameters corresponding to the fibers, while $\rho \in [0, 1]$ is a (dimensionless) measure of anisotropy. $\bar{I}_4 > 1$ and $\bar{I}_6 > 1$ are dimensionless parameters regarded as the initial crimping of the fibers. $\bar{I}_4 = \mathbf{a}_0 \cdot \mathbf{C} \cdot \mathbf{a}_0$, $\bar{I}_6 = \mathbf{b}_0 \cdot \mathbf{C} \cdot \mathbf{b}_0$ are the fourth and sixth invariants, and $\mathbf{a}_0$, $\mathbf{b}_0$ are the direction of the fibers. Table 2 shows the values of the material parameters for both constitutive models.

<table>
<thead>
<tr>
<th>MODEL</th>
<th>$C_1$ (kPa)</th>
<th>$C_2$ (kPa)</th>
<th>$\rho$</th>
<th>$k_1 = k_3$</th>
<th>$k_2 = k_4$</th>
<th>$\phi$</th>
<th>$\bar{I}_4 = \bar{I}_6^0$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Isotropic</td>
<td>1.04</td>
<td>280.40</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Anisotropic</td>
<td>0.12</td>
<td>-</td>
<td>0.14</td>
<td>244.90</td>
<td>1576.20</td>
<td>5.0</td>
<td>1.038</td>
</tr>
</tbody>
</table>

Table 2: Values of the material parameters for the isotropic model (3) and the anisotropic model (4).
Figure 1: Maximum principal stress fields in symmetric ($F_E = 0.0$) and asymmetric aneurysms ($F_E = 1.0$) with different relative lengths $F_L$, and for $F_R = 2.75$. Results were obtained with the anisotropic model (4).

3 RESULTS AND CONCLUSIONS

For the geometric parameters it was found that the maximum principal stress increases (almost) linearly with the diameter of the aneurysm (dependence on $F_R$), as has also been reported in [3]. This variation is observed independent of the degree of asymmetry of the aneurysm (value $F_E$), or the relative length $F_L$. However, the peak wall stress is more sensitive to changes in the diameter as the aneurysm asymmetry increases. The effect of aneurysm asymmetry is quite important; for AAAs with the same diameter and relative length, the peak wall stress can increase significantly as the aneurysm becomes more asymmetric. For example, in an aneurysm with $F_R = 2.75$ and $F_L = 1.5$, the peak wall stress increases from 330 to 457 kPa, more than 35% (the numbers refer to the isotropic calculation). On the other hand, a fusiform aneurysm ($F_E = 0.0$), with $F_R = 2.75$ and $F_L = 2.5$ has about the same peak wall stress as an asymmetric one ($F_E = 0.0$), with parameters $F_R = 2.0$ and $F_L = 1.5$. Regarding the relative aneurysm length, the effect of $F_L$ on the stresses seems to correlate with $F_E$. For symmetric ($F_E = 0.0$) and slightly asymmetric ($F_E = 0.5$) aneurysms, the peak wall stress increases with $F_L$ (see Figs 1(a) and (b)), while for $F_E = 1.0$ the maximum peak stresses are obtained for shorter aneurysms, i.e. for $F_L = 1.5$. These results suggest that shorter aneurysms are more critical when asymmetries are present.

REFERENCES


632-639.


