Influence of Medial Collagen Organization and Axial In Situ Stretch on Saccular Cerebral Aneurysm Growth

A model for saccular cerebral aneurysm growth, proposed by Kroon and Holzapfel (2007, “A Model for Saccular Cerebral Aneurysm Growth in a Human Middle Cerebral Artery,” J. Theor. Biol., 247, pp. 775–787; 2008, “Modeling of Saccular Aneurysm Growth in a Human Middle Cerebral Artery,” ASME J. Biomech. Eng., 130, p. 051012), is further investigated. A human middle cerebral artery is modeled as a two-layer cylinder where the layers correspond to the media and the adventitia. The immediate loss of media in the location of the aneurysm is taken to be responsible for the initiation of the aneurysm growth. The aneurysm is regarded as a development of the adventitia, which is composed of several distinct layers of collagen fibers perfectly aligned in specified directions. The collagen fibers are the only load-bearing constituent in the aneurysm wall; their production and degradation depend on the stretch of the wall and are responsible for the aneurysm growth. The anisotropy of the surrounding media was modeled using the strain-energy function proposed by Holzapfel et al. (2000, “A New Constitutive Framework for Arterial Wall Mechanics and a Comparative Study of Material Models,” J. Elast., 61, pp. 1–48), which is valid for an elastic material with two families of fibers. It was shown that the inclusion of fibers in the media reduced the maximum principal Cauchy stress and the maximum shear stress in the aneurysm wall. The thickness increase in the aneurysm wall due to material growth was also decreased. Varying the fiber angle in the media from a circumferential direction to a deviation of 10 deg from the circumferential direction did, however, only show a little effect. Altering the axial in situ stretch of the artery had a much larger effect in terms of the steady-state shape of the aneurysm and the resulting stresses in the aneurysm wall. The peak values of the maximum principal stress and the thickness increase both became significantly higher for larger axial stretches. [DOI: 10.1115/1.3200911]

Keywords: aneurysm, artery, cerebral, collagen, membrane, saccular growth

1 Introduction

Saccular cerebral aneurysms are detected in less than 5% of the human population and are usually diagnosed in elder people between the 5th and the 7th decade. High blood pressure, which is rather specific for man, appeared to have some influence on the development of cerebral aneurysms [1]. A subarachnoid hemorrhage due to the rupture of an intracranial aneurysm is a devastating event associated with high rates of morbidity and mortality. Approximately 12% of patients die before receiving medical attention, 40% of hospitalized patients die within 1 month after the event, and more than one-third of those who survive have major neurological deficits [2].

Cerebral aneurysms generally form and grow at arterial bifurcations in connection to the Circle of Willis, where the internal elastic membrane is partially destroyed and where the media is diminished [3]. Approximately 80% of all these aneurysms occur in one of the three main sites, i.e., the carotid/posterior communicating, respectively, anterior choroidal artery junction, the anterior communicating artery, and the middle cerebral artery main bifurcation [4].

Determination of the structure of the aneurysm wall is a necessary precursor to establish suitable constitutive relations for this type of tissue. For this purpose, experimental investigations of aneurysmal tissue, i.e., in terms of tensile testing and histological examinations, are necessary but theoretical modeling may also provide important insights. Structural changes in artery walls and, more specifically, structural evolution of developing aneurysms were addressed in previous theoretical studies [5–10]. Kroon and Holzapfel [8,9] proposed a theoretical model for the growth of a saccular cerebral aneurysm. In this model, the aneurysm wall was assumed to consist of a number of distinct collagen fiber layers. The continuous turnover of collagen in the layers was responsible for the growth of the aneurysm and the collagen production in the layers was governed by the embedded fibroblasts.

In the present paper, this aneurysm model is further developed. A realistic 3D setting is now employed in the form of a human middle cerebral artery. More specifically, the middle cerebral artery is modeled as a two-layer cylinder, where the layers correspond to the media and the adventitia. The constitutive behavior of the adventitia is governed by the aneurysm growth model and the media is modeled as a neo-Hookean material reinforced by two families of collagen fibers [11]. The structural integrity of an artery or aneurysm is to a large extent determined by the organization of the collagen fabric. The collagen of the adventitia or an aneurysm wall is mainly type I [3,12], whereas the collagen of the media is mainly of type III [13,14]. In a previous study, the influence of the organization of the type I collagen fabric of the ad-

---

1 Corresponding author.

Contributed by the Bioengineering Division of ASME for publication in the JOURNAL OF BIOMECHANICAL ENGINEERING. Manuscript received November 24, 2008; final manuscript received July 8, 2009; published online September 4, 2009. Review conducted by Michael Sacks.
ventitia was investigated [9]. Instead, the present study focuses on the influence of the orientation of the medial collagen on aneurysm growth. In addition, the consequences of different axial in situ stretches are studied. The influence is quantified in terms of stress distributions, wall thickness distributions, and aneurysm shape at steady-state.

A short review of the aneurysm growth model is given in Sec. 2. The current problem is then formulated, followed by a presentation of the numerical model and the numerical results. Finally, a discussion and some concluding remarks are provided.

2 Growth Model of a Saccular Cerebral Aneurysm

The saccular cerebral aneurysmal wall is considered to be the development of the adventitia of the originally healthy parent artery. The aneurysm wall is modeled as a hyperelastic material and is characterized by a strain-energy function \( \Psi \). The only load-bearing constituent is the collagen that is produced by fibroblasts spread throughout the collagen network. The aneurysm wall is assumed to consist of \( n \) distinct and discrete layers of collagen fibers that can be considered as plies forming a laminate. The collagen fibers within layer \( i \) are perfectly aligned in direction of \( \phi_i \), and as the fibroblasts are aligned in the same direction as the collagen fibers, newly produced collagen will be deposited at this angle as well during the growth process. The collagen mass production rate per unit reference volume, say \( m_i \), depends on both the stretching of individual fibroblasts and the proliferation of fibroblasts, which are taken to be governed by the global stretch-termy. The aneurysm wall is modeled as a hyperelastic material and the stiffness of collagen fibers. The fibers are considered to have two-layered cylinder, i.e., media and adventitia (see Fig. 1). The thicknesses of the media and the adventitia are denoted by \( H_{med} \) and \( H_{adv} \), respectively. Aneurysm growth is initiated by the removal of the media in a circular region, characterized by the radius \( R_{an} \) as shown in Fig. 1. The angle between the exposed adventitia in this circular region, plane \( B_2 \), and the cut in the remaining media, plane \( B_3 \), is 135°. This plane sheet is then mapped onto a cylindrical surface with outer radius \( R_0 \). In that mapping, the geometry is also scaled in the \( X_3 \)-direction by a factor of \( 1/\lambda_{L} \), giving the cylindrical structure the length \( L \). Thus, the cylindrical structure, as shown at the top right in Fig. 1, constitutes the reference configuration of the posed problem.

The external loading imposed on the model aneurysm consists of an internal pressure \( p \) and an axial stretch \( \lambda_{L} \). Boundary conditions in terms of tractions \( t \) and prescribed displacements \( u \) are thus imposed at time \( t=0^+ \) according to (accounting for symmetry)

\[
B_1, B_2; t_3 = -p, \quad X_1 = -R_0; u_1 = 0, \quad X_2 = 0; u_2 = 0
\]

\[
X_3 = 0; u_3 = 0, \quad X_3 = L; u_3 = (\lambda_{L} - 1)L
\]

where \( B_1 \), \( B_2 \), and \( B_3 \) are the surfaces defined in Fig. 1, which the pressure acts and \( t_n \) is the normal component of the traction vector \( t \).

3.2 Stress Response. Aneurysm growth is initiated by a local loss of media. This damage process occurs in a loaded state in which the artery is exposed to a blood pressure and an axial in situ...
stretch. Subsequently, we model the following processes: (i) a healthy (undamaged) artery is exposed to an (internal) pressure \( p \) and an axial stretch \( \lambda_L \); (ii) a local loss of media occurs in a region defined by the radius \( R_{\text{int}} \); and (iii) growth of the aneurysm starts. In order to model this process, the reference geometry is initially defined as a plane sheet with length \( \lambda_L L \) and with a circular damage zone. This plane sheet is then scaled in the axial direction by the factor \( 1/\lambda_L \) (and mapped onto a cylindrical shape) to obtain the mentioned reference configuration.

The adventitia of a healthy artery (and the aneurysm wall) is not able to withstand any (or very little) bending, whereas the media of healthy arteries has a significant bending stiffness. Therefore, the adventitia is modeled as a membrane and the media as a three-dimensional continuum (tetrahedral solid elements). The adventitia is built up of \( n \) layers with distinct collagen fiber angles \( \phi_i \) for each layer \( i \), where \( i = 1, \ldots, n \). The fiber angles are defined by the 2D reference coordinate system \( \zeta_1 \zeta_2 \), as shown in Figs. 1 and 2, in which \( \zeta_1 \) follows the circumferential direction of the artery and \( \zeta_2 \) the axial direction; \( X_2 \) and \( \zeta_1 \zeta_2 \) are local coordinate systems defined in every point on the artery surface. As shown in Fig. 2, the fiber distribution is uniform and the fiber direction \( \phi_i \) is taken to coincide with the \( \zeta_1 \)-axis, see Fig. 1.

For the strain-energy function (5) that governs the constitutive response of the adventitia, the in-plane second Piola–Kirchhoff stress components \( S_{ab} \) are given as

\[
S_{ab} = 2 \sum_{n=1}^{N} \frac{\partial \Psi}{\partial C_{ab}} = 2 \sum_{n=1}^{N} \int \gamma \tilde{g}(t,t_{\text{dp}}) \frac{\partial \Psi}{\partial C_{ab}} \, dt_{\text{dp}}, \quad a, b = 1, 2
\]

(8)

where \( C_{ab} \) are the components of the 2D right Cauchy–Green tensor, and indices \( a \) and \( b \) pertain to the local 2D in-plane reference coordinate system \( \zeta_1 \zeta_2 \) of the plane of the adventitial membrane.

When modeling the media, the components are smooth muscle cells, elastin, and collagen (type III) [16,17]. Elastin and smooth muscle cells are expected to have a fairly linear response. As both the smooth muscle cells and the collagen tend to be aligned approximately in the circumferential direction, the total response of the media is, in general, anisotropic (see Refs. [13,18–21]). In order to model the anisotropic mechanical behavior of the media, the strain-energy function as proposed by Holzapfel et al. [11] was adopted. Thus,

\[
\Psi_{\text{med}} = \frac{\mu_M}{2} (I_1 - 3) + \frac{k_{1,\text{med}}}{2k_{2,\text{med}}} \sum_{i=1}^{8} \{ \exp[k_{2,\text{med}}(l_i - 1)^2] - 1 \}
\]

(9)

where the parameter \( \mu_M \) denotes the shear modulus of the media describing the isotropic noncollagenous matrix material (mainly elastin and passive response of smooth muscle). The anisotropic part is related to the response of the collagen and described by \( k_{1,\text{med}} \) and \( k_{2,\text{med}} \), where \( k_{1,\text{med}} > 0 \) is a stresslike parameter and \( k_{2,\text{med}} > 0 \) is dimensionless. They do not depend on the geometry or fiber angle; those effects are instead introduced through the invariants \( I_2 \) and \( I_6 \) and are defined as

\[
I_1 = C \cdot 1, \quad I_2 = C \cdot A_1, \quad I_6 = C \cdot A_2
\]

(10)

The structure tensors \( A_1 \) and \( A_2 \) are

\[
A_1 = a_{10} \otimes a_{20}, \quad A_2 = a_{12} \otimes a_{22}
\]

(11)

where the column matrices \( [a_{10}] = \begin{bmatrix} \cos \beta & \sin \beta \end{bmatrix}^T \) and \( [a_{i0}] = \begin{bmatrix} \cos \theta & -\sin \theta \end{bmatrix}^T \) collect the components of the unit vectors \( a_{10} \) and \( a_{i0} \), respectively, with \( 2\beta \) being the angle between the collagen fibers, as shown in Fig. 3.

The principal values of the second Piola–Kirchhoff stress that correspond to the media [15] are calculated as

\[
S_a = \frac{1}{\lambda_a} \frac{\partial \Psi_{\text{med}}}{\partial \lambda_a}
\]

(12)

where the three principal directions are indexed \( a = 1, 2, \) and \( 3 \).

A stress measure that is physically relevant for the aneurysm wall is the corotated Cauchy-like stress measure \( \sigma_{ab} \) [15]. The in-plane membrane stress is defined as

\[
\sigma_{ab} = \frac{1}{J} \Upsilon_{ab} S^*_{\theta \gamma} \Upsilon_{\theta \gamma} = \frac{2}{\lambda_2 \lambda_1} \Upsilon_{ab} \sum_{i=1}^{N} \frac{\partial \Psi}{\partial C_{ab}} \Upsilon_{\gamma \gamma}
\]

(13)

where \( \alpha \) and \( \beta \) again pertain to the local 2D in-plane reference coordinate system. The deformation tensor \( \Upsilon^* \) with components \( C^*_{\gamma \gamma} \) are given as \( C^*(t) = C_{\text{loc}}(t, t_{\text{dp}} = t_{\text{dp}}) \), where \( C^*(t) = U^* - 2(t) \) and \( J^*(t) = \det U^*(t) \). This is the deformation experienced by the “oldest” and most stretched fibers in the aneurysm wall. The components \( S^*_{\theta \gamma} \) are seen as modified second Piola–Kirchhoff stress components that result from a differentiation of the strain-energy function with respect to \( C^*_{\gamma \gamma} \). In addition, the thickness change in the membrane (due to material growth) is introduced as \( \lambda_2 \), which

---

**Fig. 2** Uniform fiber distribution in the aneurysm wall shown for eight layers. The coordinate system \( \zeta_1 \zeta_2 \) corresponds to the tangential and axial directions, as shown in Fig. 1.

**Fig. 3** Orientations \( a_{10} \) and \( a_{20} \) of two families of fibers in the media symmetrically disposed with respect to the cylinder axis. The parameter \( \beta \) is the angle between the collagen fiber and the circumferential direction \( \zeta_1 \).
is defined as the ratio between the current and initial aneurysm wall thicknesses. This ratio is estimated as

$$\lambda_3 = \frac{1}{n \lambda_1 \lambda_2} \sum_{i=1}^{n} \frac{m_i}{m_0}$$  \hspace{1cm} (14)$$

where \( \lambda_1 \) and \( \lambda_2 \) are the total principal in-plane stretches of the membrane and \( m_i \) and \( m_0 \) denote the current and reference collagen mass content, respectively. It is important to emphasize that it is the production of new tissue that is described by the entity \( \lambda_3 \) and not an actual stretching. Thus, material parameters that need to be supplied for the adventitia are \( \mu_M, \mu_M^\alpha, \alpha, \lambda_\text{pre}, n, \) and \( \phi_1, \ldots, \phi_n, \mu_M, k_1, \text{med,} \) and \( k_2, \text{med} \) for the media.

In the half-closed interval \( t \in (\infty, 0) \), the modeled reference configuration is unloaded, which for the adventitia corresponds to a uniform deformation \( C = I \), where \( I \) is the 2D identity tensor. A uniform deformation in turn corresponds to a constant collagen production rate of \( \dot{m}_i(t) = \beta_i \), a constant fiber deformation of \( C_{\text{fib}}(t) = \lambda_0^2 \) and a constant strain energy per unit reference volume, i.e., \( \Psi(t) = 0 = \mu_M(\lambda_\text{pre} - 1)^2 \). At time \( t = 0^+ \), prescribed boundary conditions are imposed and the aneurysm starts to evolve.

### 3.3 Model Specification and Numerics.

The considered model geometry is according to a healthy human middle cerebral artery [22,23], which are \( R_0 = 1.2 \) mm, \( H_\text{med} = 0.25 \) mm, and \( H_\text{adv} = 0.30 \) \( \mu \)m. The size of the damage region, where the media is absent, is characterized by the radius \( R_\text{med} \) set to be \( \pi R_0/2 \). On the basis of investigations conducted by Monson et al. [22], material stiffness parameters are set to \( \mu_M = 0.3 \) MPa and \( \beta_0 = 14 \) MPa (in ref. [22], there is a substantial variation in results and the parameters chosen are from test specimens with a relatively low stiffness). The parameters \( k_1, \text{med} \) and \( k_2, \text{med} \) are based on the material parameters for a carotid artery from a rabbit [24], which are \( k_1, \text{med} = 2.3632 \) kPa, \( k_2, \text{med} = 0.8393 \) kPa, and \( \mu_M = 3.0 \) kPa, where the superscript "cb" stands for "rabbit." The non-dimensional parameter is taken to be the same for a human middle cerebral artery as for the rabbit \( k_1, \text{med} = k_1, \text{med}^{\text{a}'} \), whereas \( k_1, \text{med}^{\text{a}'} \) for a human cerebral artery is estimated by assuming that the relation

$$\frac{k_1, \text{med}}{\mu_M} = \frac{k_1, \text{med}^{\text{a}'}(t')}{\mu_M(t')}$$  \hspace{1cm} (15)$$

holds, leading to the estimation \( k_1, \text{med}^{\text{a}'} = 0.24 \) MPa. The length of the (quarter) model is \( L = 8R_0 \), which is considered to be sufficient in order for the ends of the artery not to have any influence on the stress distribution in the aneurysm wall. The stability properties of the aneurysm model were investigated by Kroon and Holzapfel [8]. It was found that the stabilization of the evolving aneurysmal wall was drastically increased when the parameter \( \alpha \) was in the range 1.5 < \( \alpha \) < 2. Therefore, in the present study \( \alpha \) is set at 1.7, as most aneurysms do grow in a stabilizing manner. It was also found that the prestretching of fibers in the aneurysm wall \( \lambda_\text{pre} \) should be set to a relatively low value when compared to the in situ stretch of arteries. Hence, the prestretch is, therefore, set to \( \lambda_\text{pre} = 1.02 \). The influence of the number of layers \( n \) in the media was also investigated by Kroon and Holzapfel [8,9] and it was found that as long as \( n \geq 4 \), the number of layers does not influence the model behavior to a large degree. Hence, the number of layers \( n \) is, therefore, set to \( n = 8 \).

The internal pressure \( p \) is set to 7 kPa, which is in accordance to the internal pressure for human carotid arteries [25]. The axial in situ stretch of human arteries depends on the location and is in the range 1.0–1.5 [26,27]. An in situ mean stretch for cerebral arteries of 1.31 was found by Monson et al. [28]. Three axial stretches are investigated, namely, \( \lambda_1 = 1.0, \lambda_2 = 1.2 \) and \( \lambda_3 = 1.4 \).

The open source finite element analysis program (FEAP) [29] is used to analyze the problem and the growth model of the aneurysm was implemented as a user membrane element. Three different finite element meshes are used in the study pertaining to the different axial stretches. The resulting mesh sizes are 13,492, 13,608, and 13,886 finite elements. Collapsed, four-node bilinear membrane elements are used to model the adventitia, which includes the developing aneurysm wall, and four-node trilinear tetrahedral solid elements for the media. The surfaces of \( B_1, B_2, \) and \( B_3 \), surface pressure elements are used to impose the pressure \( p \), which acts on the deformed configuration. In the region of the aneurysmal expansion, the mesh is refined and the elements there have a characteristic size of \( \pi R_0/80 \), which is sufficient to obtain converging results. In the solution scheme, a time independent procedure calculating the steady-state solution directly is used.

### 4 Numerical Results.

In the present numerical study we investigate the influence of the medial collagen organization, i.e., the fiber angle \( \beta \), and the axial in situ stretch \( \lambda_1 \) of the artery on the growth of the saccular cerebral aneurysm.

#### 4.1 Influence of Medial Collagen Organization.

In Fig. 4, the distributions of the maximum principal Cauchy stress \( \sigma_1 \) are displayed. Solutions are shown for the cases with \( \beta = 0 \) deg, 5 deg, and 10 deg (Figs. 4(a)–4(c)), and, as a reference, the solution with no medial collagen fibers is also included (Fig. 4(d)). The axial stretch is \( \lambda_1 = 1.2 \). For all cases the stress distribution varies smoothly over the aneurysm surface with a peak value at the fundus. When fibers are included in the media, the peak value is lower when compared to a model without fibers. For an axial stretch of \( \lambda_1 = 1.2 \) and \( \beta = 0 \) deg, the maximum principal stress reaches a peak value of 0.622 MPa (Fig. 4(a)). When the fiber angle in the media is increased to 5 deg and 10 deg, the maximum principal stress becomes 0.624 MPa for both cases (Figs. 4(b) and 4(c)). This is a very small change when compared to the model without medial fibers, where the maximum principal stress is 0.670 MPa (Fig. 4(d)). Thus, including collagen fibers in the media decreases the peak stress by 7.2% compared to the solution without medial fibers. The size of the aneurysm does not differ much among the cases with fibers (Figs. 4(a)–4(c)) but the aneurysm without medial fibers is noticeably larger. Including collagen fibers in the media decreases the peak stress by 7.2%. The peak values are at the fundus.

**Fig. 4** Distributions of maximum principal Cauchy stress \( \sigma_1 \) (axial in situ stretch \( \lambda_1 = 1.2 \)). The fiber angle of the medial collagen varies according to (a) \( \beta = 0 \) deg, (b) 5 deg, and (c) 10 deg; in (d) no collagen fiber is included in the media and the related aneurysm size is noticeably larger. Including collagen fibers in the media decreases the peak stress by 7.2%. The peak values are at the fundus.
media (Fig. 5(d)), which has a thickness increase of 4.56 at the fundus.

The maximum in-plane Cauchy shear stress \( \tau \) for the four investigated cases is plotted in Fig. 6. The maximum values are 0.093 MPa, 0.094 MPa, and 0.095 MPa for the cases with fibers in the media, as shown in Figs. 6(a)–6(c), respectively. These values are all lower compared to the case with no fibers in the media, which experienced a maximum shear stress of 0.102 MPa (Fig. 6(d)). The peak values do not appear at the fundus but are located close to the neck at the long side of the aneurysm, as can be seen in Fig. 6. The largest shear stress is about 15% of the largest maximum principal stresses. The minimum values are located between the fundus and the neck in the plane \( X_2 = 0 \). It may be noted that the maximum shear stress quantifies the difference between the two in-plane principal stresses. Thus, we may conclude that the maximum difference between the principal stresses is about 0.2 MPa and occurs close to the neck of the aneurysm. We emphasize that the stress distributions in Fig. 6 are symmetric with respect to the \( X_1 \)-plane and \( X_3 \)-plane, even though this is not obvious from Fig. 6.

4.2 Influence of Axial In Situ Stretch. In this part of the study a constant fiber angle \( \beta = 0 \) deg is used, and solutions for three different axial stretches \( \lambda_L = 1.0, 1.2 \) and 1.4 are compared. For the different axial stretches, the maximum principal Cauchy stress \( \sigma_1' \) (again occurring at the fundus) is found to be 0.580 MPa, 0.622 MPa, and 0.626 MPa, as shown in the Figs. 7(a)–7(c), respectively. The difference in the resulting steady-state geometry of these three cases is clearly shown in Fig. 7. No axial stretch \( \lambda_L = 1.0 \) results in a more spherical shape (Fig. 7(a)), whereas an axial stretch of \( \lambda_L = 1.4 \) results in a more elliptic shape (Fig. 7(c)).

The thickness increases \( \lambda_3 \) for the cases with the three axial stretches reach values of 4.07, 4.34, and 4.55 at the fundus in Figs. 8(a)–8(c), respectively. However, the maximum thickness increase is not always at the fundus. For \( \lambda_L = 1.4 \), i.e., the maximum value of \( \lambda_3 \) is not located in the aneurysm but rather in the intact artery close to the neck of the aneurysm. The value of the maximum thickness increase in that point is 5.64.

The resulting maximum in-plane Cauchy shear stress \( \tau \) reaches the maximum values 0.115 MPa, 0.0933 MPa, and 0.116 MPa, respectively, for Figs. 9(a)–9(c). As can be seen, the maximum shear stress is lowest for \( \lambda_L = 1.2 \), whereas the peak values are almost identical for \( \lambda_L = 1.0 \) and \( \lambda_L = 1.4 \). The location of the maximum value also changes when altering the stretch. The location of the maximum values for \( \lambda_L = 1.0 \) and 1.2 is at the long side of the aneurysm, whereas it is on the short side for \( \lambda_L = 1.4 \) (see Fig. 9).
whereas the peak values are almost identical for
with a very sharp neck. For the largest stretch investigated
value of the maximum principal Cauchy stress
with no axial stretch
seen by altering the axial stretch imposed on the model. The case
the long side of the aneurysm and on the short side for 1.4.

1.4. The location of the maximum values for
occur initiating aneurysm growth. In previous works, the proposed
loss of the media and the internal elastic lamina is assumed to
interation are not considered explicitly. Instead, an initial and instant
and mechanochemical processes leading to aneurysm growth ini-
present aneurysm model, processes that relate to fluid dynamics
may cause degradation of the media and the internal elastic lamina
strength of cerebral aneurysmal tissue was ex-
lagen fiber angles are 0.58–0.63 MPa and are of the same order.

In summary, saccular cerebral aneurysm growth has been mod-
rolled. In particular, the influences of the medial collagen organiza-
(137 µm (λL=1.0) to 130 µm (λL=4.34) for λL=1.2. As the thickness increase is governed by the total stretch of the material (with respect to the reference configuration) and as the inclusion of medial fibers reduced the aneurysm stretching by stiffening the borders between aneurysm and artery, this is an expected outcome. Changing the fiber angle in the media resulted in a minimal increase in thickness and, thereby, a minimal increase in thickness of the adventitia, whereas increasing the axial stretch of the artery resulted in relatively large stretches in the aneurysm, leading to increase in thickness. For the lowest axial stretch con-
cluded (λL=1.0), the thickness was 122 µm and for the largest was (λL=1.4) 137 µm. The thickness increases are in the range of experimentally determined values [31], where the thickness of larger cerebral aneurysms is between 116 µm and 212 µm.

5 Discussion

In the process of saccular cerebral aneurysm growth several stages can be identified. During the initial stage, the wall shear stress, which is induced by the blood flow, act on the intima and may cause degradation of the media and the internal elastic lamina of the artery. As a result, an increased load has to be carried by the adventitia, which is triggered to dilate. If this process is continued, this dilatation may develop into a saccular aneurysm. In the present aneurysm model, processes that relate to fluid dynamics and mechanochemical processes leading to aneurysm growth-initiation are not considered explicitly. Instead, an initial and instant loss of the media and the internal elastic lamina is assumed to occur initiating aneurysm growth. In previous works, the proposed aneurysm growth model was assessed for axisymmetric growth [8] and also for a more realistic 3D setting in the form of a human middle cerebral artery [9]. In the present paper the saccular aneurysm growth model is extended to include collagen fibers in the media of the parent artery surrounding the aneurysm. A parameter study is performed to investigate the influence of collagen fiber organization in the media and axial in situ stretch of the artery on the aneurysm growth. The driving mechanism for the aneurysm growth is the continuous turnover of collagen fibers in the aneurysm wall. The model response is quantified in terms of the principal Cauchy stresses, the thickness increase in the aneurysm wall, and the maximum in-plane Cauchy shear stresses. The model parameters are chosen on the basis of experiments and previous numerical results.

By introducing fibers in the media, the size of the developed aneurysm decreases noticeably. However, in terms of aneurysm size, there was no significant difference among the models with different medial fiber angles. It was also noted that the maximum stress, appearing at the fundus of the aneurysm, decreased by introducing fibers in the media. Fibers add stiffness to the vessel and, thereby, reduce the compliance at the boundary between the aneurysm and the artery. With decreased boundary compliance, the resulting aneurysm size becomes smaller and the aneurysm wall stresses decrease. By increasing the fiber angle $\beta$ the peak value of the maximum principal Cauchy stress $\sigma_{11}$ increases somewhat. When $\beta$ increases, the compliance in the circumferential direction of the artery increases and this results in a slightly wider aneurysm neck. This may turn in explain why the aneurysm wall stress increases with increasing $\beta$.

The most drastic change in the aneurysm geometry is, however, seen by altering the axial stretch imposed on the model. The case with no axial stretch ($\lambda_{L}=1.0$) produced a berry-shaped aneurysm with a very sharp neck. For the largest stretch investigated ($\lambda_{L}=1.4$), the neck was much less pronounced and the aneurysm clearly became less berry-shaped. Due to the loss of media in the aneurysm region, there is localization in the remaining media below the aneurysm. The reference geometry and the modeling method correspond to an instant loss of media in a circular damage region. The size of this damage region is defined by the radius $R_{an}$, which is independent of the axial stretch. However, it should be noted that the level of applied axial stretch will still affect how strong the localization becomes in the remaining media below the aneurysm. The character of this localization will strongly affect the shape of the aneurysm, where a higher axial stretch tends to enhance the localization and make the aneurysm more ellipsoidal.

The axial in situ stretch of arteries is known to vary and the values used here are in accordance with clinical observations [28]: $\lambda_{L}=1.0$ was observed for cerebral arteries. The largest axial stretch had a stronger influence on the maximum principal stress in the aneurysm than the fiber angle; higher stretches resulted in higher stresses. The surface area of the aneurysm on which the internal pressure acts is larger for larger axial stretches, which explains the higher stresses. The stress, however, is also influ-
ence by the thickness increase in the aneurysmal wall, which is larger for higher axial stretches and thereby reduces the stress to some extent. The strength of cerebral aneurysmal tissue was experimentally estimated at 0.5–2.0 MPa [30,31]. The peak stresses in our model for the various axial stretches and the medial collagen fiber angles are 0.58–0.63 MPa and are of the same order.

Adding fibers in the media reduced the maximum thickness at the fundus from 137 µm ($\lambda_{L}=4.56$) to 130 µm ($\lambda_{L}=4.34$) for $\lambda_{L}=1.2$. As the thickness increase is governed by the total stretch of the material (with respect to the reference configuration) and as the inclusion of medial fibers reduced the aneurysm stretching by stiffening the borders between aneurysm and artery, this is an expected outcome. Changing the fiber angle in the media resulted in a minimal increase in stretch and, thereby, a minimal increase in thickness of the adventitia, whereas increasing the axial stretch of the artery resulted in relatively large stretches in the aneurysm, leading to increase in thickness. For the lowest axial stretch con-
cluded ($\lambda_{L}=1.0$), the thickness was 122 µm and for the largest was ($\lambda_{L}=1.4$) 137 µm. The thickness increases are in the range of experimentally determined values [31], where the thickness of larger cerebral aneurysms is between 116 µm and 212 µm.

In summary, saccular cerebral aneurysm growth has been mod-
elle. In particular, the influences of the medial collagen organiza-
(fiber angles) and the axial in situ stretch on the aneurysm growth have been investigated. The previously proposed aneu-
ysm model was extended to include fibers in the media of the parent artery surrounding the aneurysm and a parameter study was performed by changing the collagen fiber angle in the media and the axial in situ stretch of the artery. When collagen fibers were included in the media, the peak stress in the aneurysm was re-
duced by 7.2% (compared to a case without fibers). Increasing axial stretch led to increasing steady-state aneurysm wall stresses. The numerical results predicted by the model are in good agree-
ment with experimental data documented in literature. The present study indicates that the improved estimations of the mechanical properties of the medial collagen and, in particular, of the axial in situ stretches of arteries are necessary for a refined prediction of aneurysm growth.

Acknowledgment

Financial support for this research was partly provided by @neurIST, an Integrated EU Project (Call Identifier No. FP6-2004-IST-4). This support is gratefully acknowledged.

References


