Finite element analysis of abdominal aortic aneurysms: geometrical and structural reconstruction with application of an anisotropic material model

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Computational biomechanics of abdominal aortic aneurysms (AAAs) made it possible to investigate several aspects of the disease and to provide information that would otherwise be difficult to obtain from experiments; the determination of wall stress distributions and rupture risk are two examples. A very few anisotropic strain–energy functions aim to capture vascular biomechanics and involve some coding to specify the collagen fibre orientations. In this study, we developed a solid mechanics framework for the use within Abaqus v. 6.10 (SIMULIA, Providence, RI, USA) with the aim to model the anisotropic response of AAAs in a robust and straightforward way. The proposed framework contains: (i) geometry reconstruction allowing flexible meshing; (ii) generation of 3D centrelines for each arterial branch; (iii) robust assignment of 3D collagen fibre orientation; (iv) AAA parameters for the Holzapfel–Gasser–Ogden model implemented in Abaqus. In the result section, we reproduce published stresses of an idealized geometry under physiological pressure with a difference of 4.41%, and apply the framework to patient-specific geometries. Finally, the simulation of an AAA deformed by two catheters during endovascular aortic repair is demonstrated.

Keywords: aneurysm; biomechanics; anisotropy; Holzapfel–Gasser–Ogden; Abaqus.

1. Introduction

Abdominal aortic aneurysms (AAAs) are localized arterial expansions occurring on the long-term and leading to a life-threatening rupture risk. This disease, identified as the 13th most common cause of death in the USA (Patel et al., 1995), is receiving a great deal of the clinical attention and research efforts. Current AAA repair procedures carry significant morbidity and mortality risks.

In particular, numerical analyses involving arterial biomechanics have increased in number and complexity, from isotropic constitutive models (Raghavan & Vorp, 2000) to anisotropic ones (Rodríguez et al., 2008; Basciano & Kleinstreuer, 2009; Xenos et al., 2010; Humphrey & Holzapfel, 2012).
However, this profusion of proposed models might bring some confusion when starting a new AAA analysis. In regard to anisotropic models, the orientation of collagen fibres often relies on ‘perfectly shaped’ elements, with local coordinate systems defined on element corners (Mortier et al., 2010). Another proposed strategy is to orientate the elements by projecting local coordinate systems based on the 3D centreline (Vande Geest et al., 2008), which is reproduced in this study; however, we also improved that approach so that it can also be applied to highly tortuous arteries. Another rather robust algorithm for incorporating the collagen fibre orientation, which was applied to the carotid bifurcation, was proposed by Kiousis et al. (2009).

Most of the numerical analyses involve solid elements, despite the fact that the anisotropic models developed so far account for collagen fibres working in tension. Indeed, AAA can be seen as 3D composites subject to blood pressure and axial stretch, where additional tension due to stent-graft deployment after endovascular repair may occur. In the present study, we use shell elements according to Prasad et al. (2013). Also calcifications (accumulations between the intima and media layers) could be included in composite shell elements by adding their contributions to the arterial wall. Shell elements allow to capture highly tortuous (patient-specific) shapes, especially in iliac arteries, which is more intricate with solid elements.

In this study, we developed a framework for a biomechanical AAA simulation based on the constitutive model proposed by Holzapfel et al. (2000), with the extension to incorporate fibre dispersion according to Gasser et al. (2006). In Abaqus v. 6.10 (SIMULIA), in which this model is also implemented, this model is called the Holzapfel–Gasser–Ogden (HGO) model. The framework was applied to patient-specific geometries and to a (simple) parametric virtual geometry, already presented in the literature, to show how our approach correlates with published wall stresses. The main parts of the framework are: (i) geometry reconstruction allowing flexible meshing; (ii) 3D centrelines for each arterial branch; (iii) robust assignment collagen fibre orientation; (iv) aneurysm parameters for the anisotropic model (Holzapfel et al., 2000; Gasser et al., 2006) implemented in Abaqus.

2. Methods

2.1 Geometry

First, a patient-specific AAA geometry of an arterial branch (labelled as PSG-1) was used to develop a robust ‘geometry/reconstruction’ algorithm. A tortuous geometry was chosen in order to illustrate the robustness of the developed algorithm and to build local coordinate systems accurately (on each element). Secondly, a (simple) parametric virtual geometry, as used in a precedent work (Rodríguez et al., 2008), was reproduced for the sake of comparison. Finally, a geometry from a different patient was used but with (much) longer iliac arteries, labelled as PSG-2. With this geometry PSG-2 the simulation of a catheter insertion, which typically occurs during endovascular aortic repair, was performed.

For the PSG-1 and PSG-2 geometries below the aortic bifurcations, a wall thickness of 1 mm was used for the iliac arteries. This number is based on a ratio of outer diameter to wall thickness of 7.7 mm for aged human iliac arteries, reported by Schulze-Bauer et al. (2003), and an average diameter of 8.0 mm measured at both PSG-1 and PSG-2. For the remaining walls of the two AAA geometries and for the parametric virtual geometry presented in this study, a uniform thickness of 1.5 mm was considered.

2.1.1 Patient-specific geometry

PSG-1 was selected from our database. The luminal domain was extracted from the anatomy, and subsequently considered as a realistic ‘tortuous’ AAA geometry.
For this purpose, the medical imaging software ORS (Object Research Systems, Montréal, QC, Canada) was used.

When performing the 3D segmentation of AAA directly from medical imaging platforms, the result is a discretized geometry with triangular facets, often exported in a STL (file) format. This obviously makes the subsequent meshing operation for the finite element model quite limited because the facets impose their boundaries as seed supports. To get around this drawback, the ‘Virtual Topology’ capability of Abaqus contributed to merge all the facets into a single (parametric) surface, as shown in Fig. 1, which in turn allowed free element type and size definitions. A Python script was written to accelerate and to make this process automatic, see Appendix A.

The 3D centrelines were defined for each arterial branch, i.e. aorta, internal and external iliac arteries, while the renal bifurcations were discarded for the sake of simplicity. The open source Vascular Modeling Toolkit (VMTK) suite of algorithms was used for this purpose. The result can be seen in Fig. 1. PSG-2 was prepared in a similar way as PSG-1.

### 2.1.2 Parametric virtual geometry

The function providing the mathematical definition of this geometry was given by Elger et al. (1996), and subsequently used by Rodríguez et al. (2008) and Toungara (2011, p. 64). Thus,

$$R(Z) = R_a + \left( R_{an} - R_a - c_3 \frac{Z^2}{R_a} \right) \exp \left( -c_2 \frac{Z}{R_a} \right), \quad (2.1)$$

where $c_1$ is a constant to be taken as 5.0, and $c_2$ and $c_3$ are dimensionless parameters depending on the aneurysm geometry, given as

$$c_2 = \frac{4.605}{(0.5 L_{an}/R_a)^{c_1}}, \quad c_3 = \frac{R_{an} - R_a}{R_a (0.8 L_{an}/R_a)^2}, \quad (2.2)$$

where $R_a$ is the radius of the healthy artery, $R_{an}$ is the maximum radius of the aneurysm and $L_{an}$ defines the length of the aneurysm (see Fig. 2). Finally, three additional parameters, i.e. $F_R \geq 1$, $F_L$ and $F_E \in [0,1]$, are introduced

$$F_R = \frac{R_{an}}{R_a}, \quad F_L = \frac{L_{an}}{R_{an}}, \quad F_E = \frac{e}{(F_R - 1) R_a}, \quad (2.3)$$

where $e$ denotes the aneurysmal eccentricity (see Fig. 2). The values used in this study are summarized in Table 1. These values lead to ‘case 7’ as defined by Rodríguez et al. (2008).
Fig. 2. Idealized geometric model of a AAA: \( R_a \) denotes the radius of the artery, \( R_{an} \) the maximum radius of the aneurysm, \( L_{an} \) the length of the aneurysm and \( e \) is the eccentricity between the aneurysm and the healthy artery.

Table 1  Parameters defining the virtual geometry

<table>
<thead>
<tr>
<th>( F_R ) (mm)</th>
<th>( F_L ) (mm)</th>
<th>( F_E ) (mm)</th>
<th>( R_a ) (mm)</th>
<th>( R_{an} ) (mm)</th>
<th>( L_{an} ) (mm)</th>
<th>( L ) (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.75</td>
<td>3.00</td>
<td>1.00</td>
<td>10.10</td>
<td>27.28</td>
<td>83.33</td>
<td>138.32</td>
</tr>
</tbody>
</table>

2.2 Collagen fibre orientation

As illustrated in Fig. 3, a local coordinate system is built for each finite element. Doing so, the process starts with the definition of the (average) element plane equation of the form \( ax + by + cz + d = 0 \). The coordinates of the points \( A \) (between nodes 1 and 2), \( B \) (node 3) and \( C \) (node 4) are used to define the coefficients \( a \), \( b \), \( c \) and \( d \) of the plane equation. Then the centre point (i.e. point \( a \)) is identified, and the precedent and next closest points from the centreline are projected on the element plane, which defines the local \( X \)-axis (tangent to the centreline). Finally, it is straightforward to define the local \( Y \)-axis via a simple vector cross product. When automated via a Python script, these algebraic operations are repeated for the whole set of elements, and it only takes a few minutes to complete.

At this point, another criterion was necessary to select the closest point along the centreline with regard to the centre point \( a \) for a given finite element. Indeed, a frequent situation, as illustrated in Fig. 4, might arise when the closest (centreline) points to a given finite element are located on a different vessel branch.

In Fig. 4, for example, an element (with centre point \( a \)) still located on the AAA, but close to the iliac bifurcation, might find itself closer to the centreline points from the iliac branch of the centreline. Therefore, in order to avoid such a situation and to enforce the selection of centreline points that would be located on the right vessel branch, a 3D cone of selection was implemented in our script. This cone was defined by a minimum angle \( \beta \) (within the range 0–180°) between the local \( Z \)-axis (along the normal vector) and the vectors starting from the element centre and ending on the centreline points. As a consequence, the script is ‘looking inside the geometry’ for centreline points, thus excluding non-relevant vessel branches. Figure 4 indicates the local coordinate systems, and how the two families of collagen fibres are oriented with an angle \( \alpha \) with respect to the local (circumferential) \( Y \)-axis. Appendix B shows how the cone of selection was implemented by defining the local coordinate system in which the collagen fibres are oriented.
Fig. 3. Definition of the element coordinate system: the $X$-axis is defined by the projection of the closest point of the centreline, the $Z$-axis is derived by the cross-product of the vectors aligned between the points $A$, $B$ and $A$, $C$, while the $Y$-axis is finally defined by the right-hand rule.

Fig. 4. Concept of 'cone of selection' aiming at a proper selection of the closest centreline points to define the $X$-axis of each element. This feature helps to orientate the collagen fibres properly, especially when accounting for tortuous iliac arteries. Both $X$-axis and $Z$-axis of each element are represented by blue and red lines (for related colour see the online version), respectively.

2.3 Parameters for the anisotropic constitutive model

To the best of the authors’ knowledge, there are no parameters yet available for the specific constitutive model of Holzapfel et al. (2000) using mechanical AAA data. The ‘missing’ parameters are now identified by means of a multiple non-linear regression on published biaxial tests performed on 26 tissue samples of AAAs (Vande Geest et al., 2006).
Each arterial sample is regarded as incompressible (Carew et al., 1968), requiring $\lambda_r, \lambda_\theta, \lambda_z = 1$, where $\lambda_r, \lambda_\theta$ and $\lambda_z$ are principal stretches (without shear) related to the radial, circumferential and axial directions, respectively. The constitutive model is described by a strain–energy function (SEF), say $\Psi$, that relates the energy per unit reference volume to strain and stress measures. Starting with the definition of the deformation gradient $F = dx/dX$ between a reference (undeformed) configuration, with position vector $X$, and a current (deformed) configuration, with related position $x$, the right Cauchy–Green tensor is then defined as $C = F^T F$. The directions of two families of collagen fibres in the reference configuration are denoted by the unit vectors $M_i, i = 1, 2$. From these premises, the following form of the SEF can be postulated as

$$\Psi = \Psi(C, M_1, M_2). \quad (2.4)$$

This function may be decomposed into a volumetric response $\Psi_{\text{vol}}$ and an isochoric response $\Psi_{\text{iso}}$, i.e.

$$\Psi = \Psi_{\text{vol}}(J) + \Psi_{\text{iso}}(\bar{I}_1, \bar{I}_4, \bar{I}_6), \quad (2.5)$$

where $J = \det F$ denotes the volume ratio. The invariants $\bar{I}_1, \bar{I}_4$ and $\bar{I}_6$ are defined as

$$\bar{I}_1 = \text{tr} \tilde{C}, \quad \bar{I}_4 = M_1 \cdot \tilde{C} M_1, \quad \bar{I}_6 = M_2 \cdot \tilde{C} M_2, \quad (2.6)$$

where the tensor $\tilde{C}$ is associated with volume-preserving deformations of the material according to

$$\tilde{C} = J^{-2/3} C. \quad (2.7)$$

We assume here a specific state, namely no shear, and that the mean fibre direction is embedded in the tangential surface of the tissue (the mean fibre has no components in the radial direction), so that the three invariants can then simply be expressed as

$$\bar{I}_1 = \tilde{\lambda}_\theta^2 + \tilde{\lambda}_z^2 + (\tilde{\lambda}_\theta \tilde{\lambda}_z)^{-2}, \quad \bar{I}_4 = \tilde{I}_6 = \tilde{\lambda}_\theta^2 \cos^2 \alpha + \tilde{\lambda}_z^2 \sin^2 \alpha, \quad (2.8)$$

where $\tilde{\lambda}_\theta = J^{-1/3} \lambda_\theta$ and $\tilde{\lambda}_z = J^{-1/3} \lambda_z$ are the modified stretches, while the parameter $\alpha$ denotes the angle between the collagen fibre reinforcement with regard to the local $Y$-axis of each element, which, subsequently, acts as a phenomenological parameter.

Frequently, in non-linear biomechanics the stresses and strains are expressed in terms of the second Piola–Kirchhoff stress tensor $S$, and the Green–Lagrange strain tensor $E = (C - I)/2$, respectively. Both tensors are related to the SEF via (Ogden, 1997; Holzapfel, 2000)

$$S = \frac{\partial \Psi}{\partial E} = 2 \frac{\partial \Psi}{\partial C}. \quad (2.9)$$

Moreover, the Cauchy stress tensor $\sigma$ can simply be obtained from the second Piola–Kirchhoff stress tensor by $\sigma = J^{-1} FSF^T$. For our approach (2.9) can be reduced to the in plane components, simply stated as

$$S_\theta = \frac{1}{\lambda_\theta} \frac{\partial \Psi}{\partial \lambda_\theta}, \quad S_z = \frac{1}{\lambda_z} \frac{\partial \Psi}{\partial \lambda_z}. \quad (2.10)$$

The in plane components of the Green–Lagrange strain tensor $E$ are

$$E_\theta = (\lambda_\theta^2 - 1)/2, \quad E_z = (\lambda_z^2 - 1)/2. \quad (2.11)$$
Subsequently, we use the specification of (2.5) according to

\[ \Psi_{\text{vol}} = \frac{1}{D} \left( \frac{J^2 - 1}{2} - \ln J \right), \]  

(2.12)

\[ \Psi_{\text{iso}} = C_{10}(\bar{I}_1 - 3) + \frac{k_1}{2k_2} \sum_{i=1}^{2} \left\{ \exp [k_2(\bar{E}_i)] - 1 \right\}, \]  

(2.13)

with

\[ \bar{E}_1 = \kappa(\bar{I}_1 - 3) + (1 - 3\kappa)(\bar{I}_4 - 1), \quad \bar{E}_2 = \kappa(\bar{I}_1 - 3) + (1 - 3\kappa)(\bar{I}_6 - 1). \]  

(2.14)

This formulation is the extended version of Holzapfel et al. (2000) according to Gasser et al. (2006), sufficiently general to capture the typical features of the arterial response. The used notation is partly taken from the documentation of Abaqus v. 6.10 in which this model is also implemented. The Macaulay brackets \( \langle \cdot \rangle \) in the two-term potential (2.13) are defined as \( \langle x \rangle = (|x| + x)/2 \). According to the Abaqus documentation, see Abaqus 6.13 Analysis User’s Guide (2013), only a positive value of \( \bar{E}_1 \) and/or \( \bar{E}_2 \) leads to a contribution of the exponential function in (2.13) to the strain energy \( \Psi_{\text{iso}} \). The parameter \( D \) in (2.12) is related to the bulk modulus \( K \) as \( K = 2/D \). As per Abaqus documentation, a value of \( D = 1 \times 10^{-6} \text{kPa} \) is recommended to ensure incompressibility; hence (2.12) can be seen as a kind of penalty function.

Hence, the following parameters remain to be identified by means of a multiple non-linear regression performed on the above mentioned biaxial experimental curves: \( C_{10} \) in kPa, \( k_1 \) in kPa, \( k_2 \) (dimensionless), the dispersion parameter \( \kappa \in [0, 1/3] \) (dimensionless) and \( \alpha (\circ) \) defining the orientation of collagen fibres. The parameter \( \kappa \) may take on values between 0 and 1/3, and describes the dispersion of the fibre directions, with 0 and 1/3 indicating perfect alignment and ‘complete’ dispersion, respectively (Gasser et al., 2006). A complete dispersion of fibres means that the material is isotropic.

Our approach to identify these parameters consists in the minimization of the error between the second Piola–Kirchhoff stresses \( S^\text{exp}_\theta, S^\text{exp}_z \), computed from experimental data (data taken from Vande Geest et al., 2006), and the stresses \( S^\text{mod}_\theta, S^\text{mod}_z \) predicted by the anisotropic model through (2.10). The ‘objective function’ to be minimized reads

\[ \chi^2 = \sum_{p=1}^{n} \left[ w_1 (S^\text{exp}_\theta - S^\text{mod}_\theta)^2 + w_2 (S^\text{exp}_z - S^\text{mod}_z)^2 \right], \]  

(2.15)

where \( n \) denotes the number of data points and \( w_1 \) and \( w_2 \) are weighting factors (set to 1.0 in this study).

The best fit operation was realized with a ‘Simulated Annealing’ (direct search) algorithm available in Mathematica 8.0 (Wolfram Research, Inc., Champaign, IL, USA). The obtained results are presented in Fig. 5, for different circumferential to axial stress ratios \( T_{\theta\theta}/T_{zz} \) (the value \( T_{\theta\theta}/T_{zz} \) denotes the ratio of engineering stresses in the circumferential and axial directions). A Simulated Annealing algorithm was preferred to the (conventionally used) Levenberg–Marquardt algorithm because the latter is based on the gradient and might lead to local minima, whereas the former guarantees a global minimum.

The resulting parameters are summarized in Table 2, where the low value for \( C_{10} \) (basically describing the isochoric contribution of the matrix material) may be related to the known fact of elastin loss during AAA evolution. Taking into account all the fitted curves, a global coefficient of determination \( R^2 = 0.86 \) was identified.
Fig. 5. Stress–strain data for a AAA (labelled as ‘Exp.’) provided by Vande Geest et al. (2006), see Fig. 1 therein, and the anisotropic model (labelled as ‘Mod.’) according to (2.12–2.14); for related colour see the online version.

Table 2 Parameters for the constitutive model (2.12–2.14). The parameter D in (2.12) was chosen to be $1 \times 10^{-6}$ kPa

<table>
<thead>
<tr>
<th>$C_{10}$ (kPa)</th>
<th>$k_1$ (kPa)</th>
<th>$k_2$</th>
<th>$\kappa$</th>
<th>$\alpha$ ($^\circ$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$1.1 \times 10^{-6}$</td>
<td>2853.6</td>
<td>9322.0</td>
<td>0.325</td>
<td>5.0</td>
</tr>
</tbody>
</table>

2.4 Finite elements

Linear fully integrated shell elements (S4) were used, although linear reduced ones (S4R) could also be used for efficiency purposes, and without sacrificing the accuracy. A total of 15,092, 30,460 and 53,758 finite elements were used for the mesh of the virtual geometry, PSG-1 and PSG-2, respectively. PSG-2 has the same finite element density as PSG-1 but uses more elements because of the longer iliac arteries modelled. Regarding the simulation of the catheter release both the left and right catheters were modelled with linear beam elements (B31).

2.5 Numerical analysis with Abaqus: used parameters

We have used Abaqus/Explicit for all our analyses. An explicit scheme was preferred because an increasing number of analyses dedicated to the simulation of stent-graft deployment has been presented recently (De Bock et al., 2012; Prasad et al., 2013), and an explicit solver is required for such analyses for a proper management of contacts. Therefore, conducting our study in an ‘explicit’ mode appeared to be relevant.

The quasi-static nature of all models was verified by making sure that the kinetic energy (ALLKE) is smaller than 5% of the total internal energy (ALLIE) during the analyses; no mass scaling was applied. A smooth load increase (sigmoid function, i.e. loads are prescribed slowly at the beginning and the end of the analysis) was used to avoid any (unrealistic) dynamic effects, while a time period of 1 s
was prescribed. A transverse shear stiffness of 10 kPa was attributed to the used shell elements, as recommended in the literature, see, e.g. Vande Geest et al. (2008).

In order to keep the maximum aspect ratio (thickness to the element edge or diagonal length) of 1 required by Abaqus (via the option ‘max ratio = 1’), the thickness was ‘artificially’ decreased by a factor of 2. Therefore, with a prescribed finite (shell) element size of 1 mm and a thickness of 1.5 mm decreased to 0.75 mm, the final aspect ratio was 0.75/1 = 0.75. Consistently, the thickness of the iliac arteries in PSG-1 and PSG-2 was decreased from 1 to 0.5 mm. Note that an increase of the finite element size would lead to a mesh which is not fine enough to capture the complex curvatures of PSG-1 and PSG-2, and it would not provide the right stress distributions. In addition, the parameters with a dimension of pressure (inverse of pressure) of the constitutive models were increased (decreased) by the same factor in order to maintain a similar stiffness. Since an explicit solver, based on dynamic equations involving mass matrices, was used, the (required) density was also increased by the same factor. Overall, this strategy led to a stable finite element analysis.

2.6 Loading and boundary conditions

The virtual geometry was not only modelled with the constitutive law described above, but also with the isotropic hyperelastic model of Raghavan & Vorp (2000), which allowed a first verification. These two finite element analyses were executed with a pressure of 120 mmHg, and both end sections were fixed.

PSG-1 and PSG-2 were also subjected to a blood pressure of 120 mmHg. For these cases, the proximal and distal end sections were constrained so that the circumferential and axial degrees of freedom were fixed, thus allowing a free radial expansion. This set of constraints did not influence the stresses in the region of interest, i.e. the AAA bulge.

Prior to pressurize PSG-2 and simulate the interaction of the artery with the catheters, a rough ‘zero pressure’ version of this geometry was assessed by a single ‘forward and backward’ analysis cycle, as indicated by Bols et al. (2013). Also, both internal iliac end sections were attached to a non-linear spring mimicking a portion of healthy artery, and based on material properties provided by Raghavan et al. (1996). The left (leg) and right (body) catheters were attributed to Young’s moduli of 5.07 × 10^7 kPa and 7.90 × 10^7 kPa, respectively. These moduli were determined experimentally from three-point bending tests, see Roy (2014). This corresponds to typical catheters supporting iliac and body stent-grafts, having a diameter of 5 mm and 7 mm, respectively. These devices were moved laterally to the pair of corresponding centrelines and through the vessel directly, which ensured a proper initial positioning and which avoided potential numerical instabilities due to a (more realistic) axial introduction. Once the devices were moved onto the centrelines, they were then released while fixing only their most distal nodes, and the contact (including self-contact) was activated to allow an interaction.

3. Results

3.1 Pressurized virtual geometry

The maximum principal (Cauchy) stress distributions for the isotropic and the anisotropic constitutive models are presented in Fig. 6. The maximal principal stresses are meaningful for both models and provide a sense of the highest stresses in a given direction, which is here clearly the circumferential direction. As can be seen in Fig. 6, there is a smaller portion of high stresses for the anisotropic model (see the arrows therein) that is stiffer in the circumferential direction. This is related to the collagen fibres oriented preferentially along that direction, as a natural mechanism of defense against higher...
hoop stresses. Obviously, high stresses also occur where the gradient of the radius is large, which acts as a stress concentration factor.

The peak wall stress is higher for the anisotropic model, with a value of 759 kPa, against 658 kPa for the isotropic model. This trend was also observed by Rodríguez et al. (2008).

3.2 Pressurized patient-specific geometry—PSG-1

The overall stress distribution (maximal principal Cauchy stress) is presented in Fig. 7, where a peak wall stress of 1087 kPa was identified at the bifurcation domain. This peak stress appears in a region marked by multiple strong curvatures. For comparison purposes, the maximal principal stresses were also plotted specifically for a region of interest characterized by a high gradient of the radius, which is depicted in Fig. 8. A local peak wall stress of 792 kPa was observed in this region.

3.3 Pressurized patient-specific geometry including two catheters—PSG-2

The deformed PSG-2 geometry using the anisotropic model (2.12–2.14) can be appreciated in Fig. 9. The present model allowed to capture a realistic large deformation pattern and a folding behaviour. This compares well with clinical records.

4. Discussion

The peak wall stress of 759 kPa for the anisotropic model, as shown in Fig. 6(b), can be compared with 794 kPa, as documented by the model ‘case 7’ of Rodríguez et al. (2008). This represents a difference of 4.41%. Moreover, the PSG-1 geometry implemented with the anisotropic model showed a (local) peak wall stress of 792 kPa (Fig. 8) in a region of similar shape (similar double curvature) compared with the ‘case 7’ of Rodríguez et al. (2008).

In this work, we aimed to provide an analysis to specifically substantiate the different blocks of our workflow, as opposed to a ‘fully’ realistic elaboration. However, as a matter of future work to improve our proposed workflow, we suggest to account for the aspects discussed hereafter.
Fig. 7. Maximum principal (Cauchy) stresses for the anisotropic patient-specific model, with peak stresses in a region marked by multiple strong curvatures.

Fig. 8. Maximum principal (Cauchy) stresses for the anisotropic patient-specific model in a region of high gradients of the radius.

Regard the identification of the unloaded geometry (or the so-called zero-pressure geometry) several investigations were performed to tackle this structurally inverse problem (Govindjee & Mihalic, 1998; de Putter et al., 2007; Speelman, 2009; Gee et al., 2010; Weisbecker et al., 2014). However, such
approaches imply access to the finite element source code, which is not always possible. Fortunately, it is possible to converge towards an acceptable approximation of the zero-pressure geometry by means of a fixed point method in which forward analyses and nodal coordinate updates are iteratively performed (Raghavan et al., 2006; Bols et al., 2013).

It is well known that residual stresses are accompanied with healthy arteries, although they are not trivial to quantify (Holzapfel et al., 2007) and to model (Holzapfel & Ogden, 2010). According to Greenwald et al. (1997) residual strains present in the unloaded configuration of the (thoracic) aorta are mainly located in the elastin (matrix material). Moreover, He & Roach (1994) found that the media decreases by 91% in volume fraction in AAA. Therefore, we might assume that residual stresses in AAAs are small (and hence negligible) in a first approximation, as suggested by Raghavan et al. (2006). Interestingly, this assumption was confirmed after investigating five open surgeries, during which no significant opening was recorded. The contribution of calcifications was not accounted for in the present study. Indeed, these should be included if they are massively present within the wall (de Putter, 2006); material properties are provided by Maier et al. (2010).
For PSG-1 and PSG-2, the same parameters in the constitutive anisotropic model were considered for the whole geometry, whereas a different set of parameters should be attributed to the iliac arteries. Since these arteries remain healthy in most of AAA patients, an anisotropic model for aged and healthy iliac arteries should be used, for more details see the study by Schulze-Bauer et al. (2003). To the best of the authors’ knowledge, there is not yet a specific anisotropic constitutive model available for iliac arteries in a AAA patient.

We need to mention two challenges the community should also address to model AAA biomechanics: the actual thickness distribution, and the difficult issue of the inhomogeneity in stiffness. In addition, a pressing need is also the development of computational models that can predict the evolving wall stress and strength of AAA based on clinically available patient-specific data; for more details on related open problems see Humphrey & Holzapfel (2012).

The shell elements used are (3D) four-node, double-curved and general-purpose elements that provide solutions to problems adequately described within the Kirchhoff–Love or the thick Mindlin–Reissner shell theory, with a switch between such formulations. Those elements include hourglass control (though not necessary for the fully integrated S4 elements) and are valid for non-linear material responses as well as finite strains. They are also suitable for large rotations. These elements have the ability to model the bending behaviour of composites. As per Abaqus documentation, thickness changes are possible as a function of the in plane deformation, and these shell elements do not suffer from transverse shear locking. As long as the wall thickness does not exceed one tenth of a characteristic arterial dimension, typically its diameter, the Mindlin–Reissner theory provides an acceptable first approximation of stresses (with aortic and iliac thicknesses of 0.75 and 0.5 mm, respectively, used in the computation). However, since AAAs are 3D continua, and because the ratios of thickness to diameter might vary significantly from the average values considered in this study, a continuum approach (via 3D solid elements) is certainly an option for the sake of accuracy.

Finally, more realistic boundary conditions would include the spine as a rigid domain and the influence of the surrounding organs, either via an equivalent pressure of 12 mmHg (Hinnen et al., 2005) or by means of more sophisticated multi-dimensional analogies with electrical circuits (Moireau et al., 2012).

5. Conclusion
We have developed a simple and reproducible workflow for the numerical simulation of AAA biomechanics. The stress distributions with different (patient-specific) geometries and loading conditions were demonstrated by using established constitutive models. We provided the parameters for the HGO anisotropic model implemented in Abaqus that is a recurrent FEA package in biomechanics, and frequently used for the simulation of vascular prostheses implantation.

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References


Appendix A. Flow chart describing AAA geometry reconstruction in Abaqus

Appendix B. Flow chart describing the orientation of collagen fibres for each element