Medical Engineering and Physics 53 (2018) 49–57

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Impact of isotropic constitutive descriptions on the predicted peak wall stress in abdominal aortic aneurysms

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ARTICLE INFO

Article history:
Received 28 November 2016
Revised 24 December 2017
Accepted 3 January 2018

Keywords:
Abdominal aortic aneurysm
Wall stress
Non-linear material model

ABSTRACT

Biomechanics-based assessment of Abdominal Aortic Aneurysm (AAA) rupture risk has gained considerable scientific and clinical momentum. However, computation of peak wall stress (PWS) using state-of-the-art finite element models is time demanding. This study investigates which features of the constitutive description of AAA wall are decisive for achieving acceptable stress predictions in it. Influence of five different isotropic constitutive descriptions of AAA wall is tested; models reflect realistic non-linear, artificially stiff non-linear, or artificially stiff pseudo-linear constitutive descriptions of AAA wall. Influence of the AAA wall model is tested on idealized ($n = 4$) and patient-specific ($n = 16$) AAA geometries. Wall stress computations consider a (hypothetical) load-free configuration and include residual stresses homogenizing the stresses across the wall. Wall stress differences amongst the different descriptions were statistically analyzed. When the qualitatively similar non-linear response of the AAA wall with low initial stiffness and subsequent strain stiffening was taken into consideration, wall stress (and PWS) predictions did not change significantly. Keeping this non-linear feature when using an artificially stiff wall can save up to 30% of the computational time, without significant change in PWS. In contrast, a stiff pseudo-linear elastic model may underestimate the PWS and is not reliable for AAA wall stress computations.

1. Introduction

Peak Wall Stress (PWS) [1,2] and related biomechanical parameters [3–5] receive increasing attention in assessing the rupture risk of Abdominal Aortic Aneurysms (AAA). Specifically, compared to the maximum diameter and its change in time, i.e. currently the most-used clinical risk assessment parameters, the biomechanics-based diagnostic parameters allow integrating many patient-specific risk factors [6].

PWS in AAA is nowadays calculated with patient-specific geometry, including morphology of the Intra-Luminal Thrombus (ILT – a pseudo-tissue occurring in almost all clinically relevant AAAs) and variable wall thickness [7]. Such simulations are typically based on Finite Element Method (FEM) and consider the AAA as loaded by Mean Arterial Pressure (MAP) [3], systolic pressure [4], or by 1.5 times increased MAP [8]. State-of-the-art FEM models also consider the (hypothetically) load-free geometry of AAA [4,9,10] and residual stresses (RS) in its wall [11]. Obviously, geometrical and mechanical properties of AAAs are diverse [12], and recently some of the FEM models for PWS predictions consider uncertainties of some input variables [8].

AAA histology and mechanical properties are complex [13,14]; in keeping with other model, a constitutive description of the AAA wall should include only the properties relevant to the overall simulation objective and disregard all the other information. An unnecessarily complex AAA model may of course be exploited for reference predictions; however, simpler (but still reliable) models are always preferred in clinical applications. The literature on AAA biomechanics presents wall models of very different degrees of complexity, and the key question "What complexity of the AAA wall model is necessary for a robust PWS prediction?" needs still to be answered. This answer would be of great value to AAA biomechanics researchers. Some AAA biomechanical models reported earlier [15], as well as some reported recently [16,17], proposed linear elastic wall models, but the most widely used constitutive description of the AAA wall is a Yeoh-type strain energy density function (SEDF) [18]. Tensile tests of AAA wall tissue highlighted its non-linear and anisotropic properties. Although the Yeoh model is isotropic, it captures data of uniaxial in-vitro testing reasonably well [19], especially in the context of their

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https://doi.org/10.1016/j.medengphy.2018.01.002
1350-4533/

intra- and inter-patient variability. Most important, for idealized AAA geometry, varying the two Yeoh parameters within their 90% confidence intervals (CI) yielded only 5% variations in PWS predictions [19]. Recently another study was published stating that the variation of these parameters by two orders resulted in 16% variation in PWS [20].

The in-vivo load of AAA wall induces stresses in circumferential and axial directions – conditions that can be established closely in planar biaxial tensile tests, but not in uniaxial tests. In addition, AAA wall constitutive descriptions derived from uniaxial and biaxial testing differ considerably. Under biaxial loading the AAA wall exhibits much lower initial stiffness, followed by a much steeper increase of stiffness with stretching (strain stiffening) [21]. AAA wall models reported elsewhere [22–24] are capable of capturing such stress–strain properties. However, this AAA walls steep strain stiffening results in FE models with bad conditioned system of equations, the numerical solution of which requires small time steps and extends computational time demands critically [8].

Some recent papers reported that accurate PWS results [16,17] can be obtained with a linear AAA wall model being overly stiff compared to the walls realistic non-linear description. The authors explain their finding with the fact that the stress state in statically determinate membrane-like structures is independent of the constitutive model. Unfortunately, the applicability of this assumption to AAA wall stress predictions has not been thoroughly discussed in literature, and conflicting data has been reported. In addition, many previous models [9,16,17] predicted considerable (and non-physiological) stress gradients across the AAA wall and violated thus the homogenous stress hypothesis [25].

The present study aims at exploring the degree of complexity of the AAA wall constitutive model that is required for a credible PWS prediction. This investigation is motivated primarily by the clinical need to speedup the patient-specific PWS calculations and stimulated by the previously reported findings [9,16]. The computational times needed to calculate PWS may differ by several orders of magnitude between simple linear [16] and more complex [8,11,14] AAA models. Moreover, if the wall model reflects the ascertained very low initial wall stiffness, the FE models may suffer from excessive mesh distortion that may lead to terminating the calculation. Therefore, a specific objective of the study is to test the applicability of artificially stiff wall descriptions, both non-linear and pseudo-linear; they would not only speedup the calculation but at the same time prevent high mesh distortion. This study is limited to isotropic constitutive models which are simple enough to have a potential to be transmitted into clinical applications.

2. Materials and methods

This study investigates wall stress in idealized and patient-specific AAA geometries. Idealized geometries are used to compare FEM-predictions and to offer a basic understanding of the impact of different wall models on the calculated wall stresses. Then the impact of different constitutive descriptions on wall stress predictions in patient-specific AAAs is explored using high-fidelity models.

2.1. Geometry representations

Idealized AAA geometries. The AAA wall can show a wide range of curvatures; we used recent AAA curvature data [26] to design meaningful idealized geometries. Four different axisymmetric tube-like shapes were analyzed (see Fig. 3). Geometry A represents a cylindrical tube with 100 mm length, 44 mm inner diameter and 2 mm wall thickness [9]. The other three geometries represent cylinders modified with an axisymmetric recess (B, C) or protrusion (D), with Gaussian curvatures (i.e. product of both principal curvatures) of $K = -5.1 \cdot 10^{-4}\,\mathrm{mm}^{-2}$, $K = -9.1 \cdot 10^{-3}\mathrm{mm}^{-2}$, and $K = 9.1 \cdot 10^{-3}\,\mathrm{mm}^{-2}$, respectively. Each tube was meshed in ICEM ver. 14.5 (ANSYS Inc., USA) with tri-linear hexahedral elements (SOLID 185 [27]) using 10 finite elements across the wall. A mesh convergence study confirmed that wall stress did not change by more than 3% when compared to a finer finite element mesh.

Table 1
Sets of parameters of material models characterizing the Abdominal Aortic Aneurysm (AAA) wall. Parameters refer to the fifth-order Yeoh model Eq. (1). Their stress–stretch curves under equi-biaxial stretching are illustrated in Fig. 2.

Material type	Material parameter (kPa)				
	c_{10}	c_{20}	c_{30}	c_{40}	c_{50}
Type 1[a]	5	0	0	2200	$13.741 \cdot 10^3$
Type 2[b]	5	0	0	2200	$13.741 \cdot 10^5$
Type 3[b]	5	0	0	2200	$13.741 \cdot 10^6$
Type 4[b]	5	0	0	2200	$13.741 \cdot 10^7$
Type 5[c]	1044	269	–	–	–

[a] Reflects non-linear in-vitro experimental data of the AAA wall reported elsewhere [21].
[b] Reflects artificially stiffened non-linear AAA wall properties.
[c] Reflects stiff pseudo-linear AAA wall properties.

Patient-specific AAA geometries. Patient-specific AAA geometries were reconstructed from Computed Tomography Angiography (CTA) images of 16 patients treated at the St. Annes University Hospital, Brno, Czech Republic. Here, the study was approved by the Institutional Ethics Committee and written informed consent was obtained from all participants. CTA images were segmented and STereoLithography (STL) files representing the wall and the ILT were acquired (A4clinics Research Edition vers. 3.0, VASCOPS GmbH, Graz, Austria). The STL files captured the aorta between renal arteries and aortic bifurcation and allowed us to generate a mapped finite element mesh (ICEM ver. 14.5, ANSYS Inc., USA) [8]. The FEM models used a constant AAA wall thickness of 2.0 mm with four tri-linear hexahedral (SOLID 185) elements across the thickness. For simplicity, the ILT was discretized with linear tetrahedral elements (SOLID 285 [27]) of characteristic size of three millimeters. This type of finite element formulation helps to prevent volume locking [27], an undesired numerical effect (numerical artifact) that dramatically decreases the accuracy of results [28]. The model considered a rigid contact between the wall and the ILT. Among all the 16 created FEM models, the number of finite elements used to discretize the wall and the ILT ranged from 5440 to 35,200 and from 1700 to 132,500, respectively. Fig. 5 presents the individual numbers of elements, and a typical mesh is shown in Fig. 1.

2.2. Constitutive modelling of AAA tissues

AAA wall constitutive description. Although the AAA wall shows a mild anisotropy [21], the present study captures its hyperelastic mechanical properties with the isotropic incompressible fifth-order Yeoh SEDF [18]

$$\Psi = \sum_{i=1}^{5} c_i (I_1 - 3)^i. \tag{1}$$

Here, $I_1 = \mathrm{tr}\,\boldsymbol{C}$ denotes the first invariant of the right Cauchy–Green deformation tensor $\boldsymbol{C}$, and AAA wall properties are specified by the stress-like material parameters c_i. For the present study five types of parameter sets shown in Table 1 were explored.

Type 1 represents the (isotropic) mean AAA response, where the least-square estimation of c_i, $i = 1, ..., 5$ was obtained from all pooled stress strain curves published elsewhere [21]. This model