Medical Engineering and Physics 000 (2018) 1–7

Contents lists available at ScienceDirect

Medical Engineering and Physics

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

Geometric surrogates of abdominal aortic aneurysm wall mechanics

Jesús Urrutia[a,b], Anuradha Roy[c], Samarth S. Raut[d], Raúl Antón[b], Satish C. Muluk[e], Ender A. Finol[f,*]

[a] The University of Texas at San Antonio, Department of Biomedical Engineering, San Antonio, TX, USA
[b] University of Navarra–Tecnun, Department of Mechanical Engineering, San Sebastian, Spain
[c] The University of Texas at San Antonio, Department of Management Science and Statistics, San Antonio, TX, USA
[d] Carnegie Mellon University, Department of Mechanical Engineering, Pittsburgh, PA, USA
[e] Allegheny Health Network, Department of Thoracic and Cardiovascular Surgery, Pittsburgh, PA, USA
[f] The University of Texas at San Antonio, Department of Mechanical Engineering, Room EB 3.04.08, One UTSA Circle, San Antonio, TX 78249, USA

ARTICLE INFO

Article history:
Received 1 February 2017
Revised 6 June 2018
Accepted 29 June 2018
Available online xxx

Keywords:
Aneurysms
Geometry quantification
Stress
Biomechanics
Medical image analysis

ABSTRACT

The maximum diameter criterion is the most important factor in the clinical management of abdominal aortic aneurysms (AAA). Consequently, interventional repair is recommended when an aneurysm reaches a critical diameter, typically 5.0 cm in the United States. Nevertheless, biomechanical measures of the aneurysmal abdominal aorta have long been implicated in AAA risk of rupture. The purpose of this study is to assess whether other geometric characteristics, in addition to maximum diameter, may be highly correlated with the AAA peak wall stress (PWS). Using in-house segmentation and meshing algorithms, 30 patient-specific AAA models were generated for finite element analysis using an isotropic constitutive material for the AAA wall. PWS, evaluated as the spatial maximum of the first principal stress, was calculated at a systolic pressure of 120 mmHg. The models were also used to calculate 47 geometric indices characteristic of the aneurysm geometry. Statistical analyses were conducted using a feature reduction algorithm in which the 47 indices were reduced to 11 based on their statistical significance in differentiating the models in the population ($p < 0.05$). A subsequent discriminant analysis was performed and 7 of these indices were identified as having no error in discriminating the AAA models with a significant nonlinear regression correlation with PWS. These indices were: D_{max} (maximum diameter), T (tortuosity), DDr (maximum diameter to neck diameter ratio), S (wall surface area), K_{median} (median of the Gaussian surface curvature), C_{max} (maximum lumen compactness), and M_{mode} (mode of the Mean surface curvature). Therefore, these characteristics of an individual AAA geometry are the highest correlated with the most clinically relevant biomechanical parameter for rupture risk assessment. We conclude that the indices can serve as surrogates of PWS in lieu of a finite element modeling approach for AAA biomechanical evaluation.

1. Introduction

An abdominal aortic aneurysm is a dilatation of the infrarenal aorta which, if left untreated, can rupture. When this occurs, there is a 50% mortality rate before reaching the operating room and an additional 25% intra-operative mortality rate [1]. AAA disease is ranked as the 13th cause of death in the U.S. [2] and is most common in patients 65 years of age and older [3]. This high rate of mortality is due to most AAAs remaining asymptomatic until they rupture. Risk factors associated with AAA rupture include smoking, obesity, high blood pressure, and high cholesterol, but timely diagnosis of asymptomatic AAA is still a difficult task. Many of them are diagnosed unintentionally when an abdominal ultrasound or computed tomography (CT) exam is performed for an unrelated condition. Nevertheless, prevention of rupture is the most important aspect of clinical management once the AAA is diagnosed. Patients with a known AAA undergo periodic abdominal ultrasound or CT scans as part of a voluntary surveillance program. Clinicians have determined that an AAA is at high risk of rupture when it grows to a diameter of 5.0–5.5 cm or shows an expansion rate greater than 1 cm/year [3]. The growth of an aneurysm over time requires close monitoring as it has been reported that AAA growth rate increases with increasing diameter [4].

Autopsy studies have shown that 10–24% of all ruptured AAA had a diameter smaller than 5.5 cm, while some larger than 5.5 cm never ruptured [5,6]. Hence, the maximum diameter criterion is not always sufficient to assess whether an aneurysm is at a high

* Corresponding author.
E-mail address: ender.finol@utsa.edu (E.A. Finol).

https://doi.org/10.1016/j.medengphy.2018.06.007
1350-4533/

Please cite this article as: J. Urrutia et al., Geometric surrogates of abdominal aortic aneurysm wall mechanics, Medical Engineering and Physics (2018), https://doi.org/10.1016/j.medengphy.2018.06.007

risk of rupture. Interventional and post-surgical morbidity costs can be high, so the prediction of rupture could be improved if one focuses on other characteristics of the aneurysm beyond its maximum diameter. From a biomechanical point of view, an aneurysm rupture occurs when the wall stress surpasses its strength. AAA wall stress can be calculated on an individual basis using patient-specific finite element analysis (FEA). PWS is elevated in ruptured aneurysms and it can be a better indicator than maximum diameter for AAA that are at a high risk of rupture [7–9].

Fillinger *et al.* [7] calculated wall stresses in vivo for 48 patient-specific AAA; 10 ruptured, 8 urgently repaired for symptoms and 30 large enough to merit elective repair within 12 weeks of the CT scan. They found that those aneurysms closer to the time of rupture (ruptured and symptomatic) had significantly higher PWS. In another work, Fillinger et al. [9] analyzed rupture risk over time in patients under surveillance. They studied 159 CT scans from 103 patients during 14 ± 2 months per CT scan and concluded that PWS is more statistically significant than diameter when predicting rupture. They also found that PWS and gender were the only significant independent predictors of rupture. Venkatasubramaniam *et al.* [8] also reported that PWS was significantly higher in ruptured AAA (mean PWS = 102 N/cm^2) than in non-ruptured AAA (mean PWS = 62 N/cm^2) by analyzing 27 AAA models (12 ruptured and 15 non-ruptured). Moreover, Khosla *et al.* [10] performed a meta-analysis of 348 individuals concluding that PWS is greater in symptomatic or ruptured AAA than in asymptomatic intact AAA, even after adjustment for mean systolic blood pressure, which is typically standardized at 120 mmHg. A probabilistic rupture risk index (PRRI) was proposed by Polzer and Gasser [11] to account for the uncertainties in AAA wall thickness and wall strength, and tested in a diameter-matched cohort of 7 ruptured and 7 intact AAAs. The results of the latter study support the use of a probabilistic approach for rupture risk prediction, combined with AAA biomechanical modeling, rather than solely using a deterministic approach. PWS shows a close association with maximum diameter, leading to the notion that PWS may not necessarily serve as a sole predictor of rupture risk, but that other factors such as size and wall strength should also be considered [12].

The association between PWS and rupture appears to be sufficiently clear, but what are the factors driving high wall stress? Washington *et al.* [13] followed an AAA patient over 28 months with the purpose of evaluating the potential correlation between PWS and AAA morphology and how it is related to rupture potential. They found that PWS was highly correlated with three morphological features: maximum diameter, sac volume and maximum diameter to neck diameter ratio. In the present work, PWS and 47 geometric indices were calculated for each of 30 AAA patient-specific models. The objective of the study was to identify which of the geometric indices are highly correlated with PWS. The correlations yield the individual geometric features that can be used as statistically significant geometric surrogates of PWS.

2. Methods

2.1. Patient population

Thirty patient datasets were acquired retrospectively from existing medical records at Allegheny General Hospital - AGH (Pittsburgh, PA) following approval of the appropriate protocol by the Institutional Review Boards at AGH and University of Texas at San Antonio. The datasets were identified from three AAA population groups in the AGH database as follows:

- Group I: Surveillance AAA (n = 10), from subjects under watchful waiting at the time of acquisition of the abdominal CT exam used to generate the AAA model and who did not have their aneurysm repaired until at least 12 months after this imaging follow-up;
- Group II: Electively Repaired AAA (n = 10), from subjects who received an elective surgical or endovascular repair less than 6 months after acquisition of the abdominal CT exam used to generate the AAA model; and
- Group III: Emergently Repaired AAA (n = 10), from subjects who received an emergent aneurysm repair no more than 1 month after acquisition of the abdominal CT exam used to generate the AAA model.

The 30 sets of images were segmented following a previously validated image processing algorithm [14] to identify the lumen, outer wall and inner wall boundaries at each cross-section of the abdominal aorta. On average, the pixel size in the CTA scans was 0.762 mm and the spacing (collimation) between images was 3.0 mm. The average maximum diameter for the population group was 5.6 ± 0.6 cm as calculated post-image segmentation. Lumen and outer wall boundary segmentations were obtained with an intensity gradient based algorithm, while identification of the inner wall boundary relied on a neural network algorithm trained with user generated inputs on the likelihood of where the lumen-wall or thrombus-wall interfaces are likely to be found. Additional details on these algorithms are found in the work of Shum *et al.* [14].

2.2. Finite element modeling

Using an in-house meshing application (AAAVASC, University of Texas at San Antonio [15,16]), the 30 AAA FEA models were generated with quadratic hexahedral elements following the method and mesh sensitivity analyses described by Raut *et al.* [16]. In brief, the FE meshes ranged in size from 90,000 to 200,000 elements with two elements clustered across the thickness of the vascular wall. The justification for using the aforementioned element type follows a prior comparison performed with linear and quadratic tetrahedral elements and linear hexahedral elements [17], where quadratic hexahedral elements yielded the most accurate wall stress and strain results using idealized shapes such as cylinders and spheres with uniform wall thickness. In addition, mesh sensitivity analyses with patient-specific geometries demonstrated that a minimum of 66,000 elements with two elements across the wall thickness were sufficient to achieve mesh-independent results with less than 5% difference in the maximum first principal stress [16]. The AAA wall was assumed hyperelastic, incompressible, and defined by the isotropic constitutive material model proposed previously by Raghavan and Vorp [18], who performed uniaxial tensile tests of tissue specimens obtained from 69 AAA patients. The strain energy function of such a model is proportional to the first invariant of the left Cauchy-Green deformation tensor, as indicated by Eq. (1),

$$W = c_1(I_1 - 3) + c_2(I_1 - 3)^2 \tag{1}$$

where

W is the strain energy density,
I_1 is the first invariant of the left Cauchy-Green tensor, and
c_1 and c_2 are material parameters evaluated experimentally.

This second order Mooney–Rivlin material model, with c_1 = 17.4 N/cm^2 and c_2 = 188.1 N/cm^2, was implemented numerically with nearly incompressible material properties (Poisson's ratio ν = 0.499). The FEA simulations were performed with the solver ADINA (Adina R&D Inc., Watertown, MA) with an intraluminal loading pressure of 120 mmHg applied in 24 time steps at 5 mmHg intervals. The proximal and distal ends of the abdominal aorta were considered to be fixed to replicate anatomical tethering of the aorta. Computational times varied from 4 to 12 h using a