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Geometric Surrogates of Abdominal Aortic Aneurysm Wall Mechanics

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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

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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.

Keywords

aneurysms; geometry quantification; stress; biomechanics; medical image analysis

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 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 and colleagues [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 and co-workers [8] also reported that PWS was significantly higher in ruptured AAA (mean PWS = 102 N/cm²) than in non-ruptured AAA (mean PWS = 62 N/cm²) by analyzing 27 AAA models (12 ruptured and 15 non-ruptured). Moreover, Khosla and colleagues [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),

$$W = c_1(I_1 - 3) + c_2(I_1 - 3)^2 \quad (1)$$

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/cm² and $c_2 = 188.1$ N/cm², was implemented numerically with nearly incompressible material properties (Poisson's ratio $\nu = 0.499$). 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 hours using a Windows workstation with up to 24 cores and 96 GB of RAM. Post-processing of the FEA simulations was performed with EnSight (CEI Inc., Apex, NC) by calculating the spatial distributions of the first principal stress. The maximum of the first principal stress distribution was taken to be the PWS. The PWS is considered a good predictor of AAA that are at high risk of rupture, although the location of the PWS does not always predict the rupture location [12]. Instead, the area most prone to rupture has been reported as the posterior wall [4, 12].

2.3 Geometry quantification

The 30 sets of segmented images were used for quantitative geometry characterization by calculating 47 indices representative of the size, shape and wall thickness of each aneurysm (see Supplementary Material for the mathematical description of each index). These indices include 1D size, 2D shape, 3D size and shape features, as well as indices to quantify the surface curvature and wall thickness distributions of the AAA. The 1D size and 2D shape indices, as well as regional distributions of wall thickness, were calculated directly from the segmented images [19]. Wall thickness was assessed at each cross-section at 72 equidistant points located along the inner wall boundary identified by the user during image segmentation. The 3D size and shape indices were calculated from the volume meshes generated from the segmented images [20]. The surface curvatures were calculated using a biquintic finite element (BQFE) method that requires generating BQFE meshes of the AAA outer wall surface and evaluating the principal curvature distributions based on the spatial location of the points within each element of the BQFE mesh [21].

2.4 Data analysis

The mean PWS for the 30 AAA were compared within the three population groups to determine if differences in this biomechanical metric were statistically significant. This was performed using an analysis of variance in SAS (SAS Institute Inc., Cary, NC) for which the null hypothesis was defined as all groups having the same mean PWS. A rejection of the null hypothesis was considered when the probability (p-value) was less than the prescribed significance level of $\alpha = 0.05$.

A statistical treatment of the geometric data was performed in SAS based on feature reduction, stepwise discriminant analysis, and nonlinear non-parametric correlations with PWS. The objective of these analyses was to identify which geometric indices are able to discriminate among the population groups and if they are correlated with PWS. The 47 geometric indices were used in a stepwise control method with a significance level of 0.40 for the entry of a variable into the discriminating set and of 0.25 for preserving a variable in the discriminating set. This method was used only for feature selection of the indices as it does not assess how well the selected indices actually discriminate the three groups. To evaluate if any of the indices obtained from the stepwise control method differentiate the AAA population groups, a subsequent examination was done with a discriminant analysis. The latter was based on the assumption that the distribution of the indices within each group

was multivariate normal, which allowed the discriminant function to be determined by a parametric method (a measure of generalized squared distance). In addition, a co-dependency matrix was built and used to assess the outcome of this analysis to determine if any discriminants are co-dependent. The geometric indices resulting from the analysis (i.e., the discriminants) were subsequently used for nonlinear non-parametric correlations with PWS and the significance of these evaluated against the prescribed significance level of $\alpha = 0.05$. This was performed in SAS for each population group using the geometric indices as independent variables and PWS as the dependent variable. The objective of these correlations was to obtain a polynomial equation that predicts PWS as a function of the discriminants and quantify the relative effect of each discriminant within a specific population group.

3. RESULTS

3.1 Wall stress

Figure 1 illustrates an exemplary AAA wall spatial distribution of the first principal stress at peak systole. Table 1 describes the PWS as mean with standard deviation for each of the population groups. PWS is higher for Group III compared to Groups I and II, with the differences in PWS being statistically significant between the three groups ($p < 0.05$).

3.2 Feature reduction, discriminant analysis and non-parametric correlations of geometry and wall stress

Table 1 includes the values of all 47 geometric indices as mean with standard deviation for each of the population groups. These geometric measures were subject to the feature reduction protocol, which yielded the following 11 indices as statistically significant ($p < 0.05$) amongst the 30 AAA (see Table 2): maximum diameter (D_{max}), tortuosity (T), maximum diameter to proximal neck diameter ratio (DDr), aneurysm outer wall surface area (S), non-fusiform index (NFI), median Gaussian curvature (K_{median}), maximum compactness (C_{max}), asymmetry of the abdominal aorta (β), area averaged Mean curvature (MAA), mode of the Mean curvature (M_{mode}), and centerline length of neck (L_{neck}). The discriminant analysis was used to assess if any of these indices differentiate the AAA population into the prescribed groups, resulting in 7 indices being discriminants with no error (error = 0%), as shown in Table 3. These 7 indices (D_{max} , T , DDr , S , K_{median} , C_{max} , and M_{mode}) can discriminate the three AAA groups with 100% certainty. Since D_{max} is a control variable (due to maximum aneurysm diameter being the clinical criterion used to differentiate one patient group from another), a second discriminant analysis was performed with the remaining 6 features. The result of this analysis indicated that the error for assessing how well these 6 geometric indices discriminate the three groups was again 0%. Therefore, the 6 remaining features (T , DDr , S , K_{median} , C_{max} , and M_{mode}) can also discriminate the AAA groups with 100% certainty.

Table 3 also describes the goodness of fit (R^2 coefficient) of the nonlinear non-parametric correlations established between PWS and each of the 7 discriminant geometric indices, as well as the p-values for the statistical significance of these correlations. The maximum compactness of an AAA (C_{max}) had the lowest percentage of variance explained by the

nonlinear correlation while also $p > 0.05$. However, it was decided to keep this feature as part of the discriminant geometric indices as removing it and repeating the discriminant analysis with the remaining 6 features (D_{max} , T , DDr , S , K_{median} and M_{mode}) resulted in a discrimination error of 3%.

3.3 Nonlinear parametric regression analysis

A polynomial function of the form described by Eq. (2) was used for the nonlinear parametric correlations for each population group:

$$PWS = \{D_{max}^{a1}\} * \{(T \cdot 10)^{a2}\} * \{(DDr \cdot 10)^{a3}\} * \left\{\left(\frac{S}{5}\right)^{a4}\right\} * \{(-K_{median} \cdot 100)^{a5}\} * \{(C_{max} \cdot 10)^{a6}\} * \{(-M_{mode} \cdot 100)^{a7}\}, \quad (2)$$

where the power coefficients $a1$, $a2$, ..., $a7$ establish the parametric correlation between the 7 discriminants and PWS. The order of magnitude of some of the independent variables in Eq. (2) was modified with the rationale that when a correlation consists of a group of bases and exponents, the influence of each variable can be assessed by the exponents alone if the bases are greater than 1 and have the same order of magnitude. For this purpose, multiples or submultiples of the original geometric indices were used. Table 4 shows the power coefficients and the goodness of fit for the correlation of each AAA group. Comparing the relative magnitudes of these exponents, it is evident that in addition to the maximum diameter, T and M_{mode} are important discriminants for Group I. Similarly, T and DDr are important discriminants for Group II, while T and S are important discriminants for Group III. The power coefficients provide an indication that the nonlinear parametric correlations follow closely the mean PWS calculated for the AAA groups. To emphasize this finding, predicted vs. observed value charts for PWS are illustrated in Figure 2, where the predicted PWS was obtained from Eq. (2) while the observed PWS was the direct output of the FEA simulations. Figure 2 shows that many of the predictions fall within $\pm 21\%$ of the observed stresses, underscoring the high PWS predictability of the 7 geometric discriminants based on the aforementioned polynomial functions.

4. DISCUSSION

Vascular surgeons typically recommend elective AAA repair through continued surveillance of abdominal ultrasounds or CT scans. When an aneurysm reaches a 5.5 cm maximum diameter (5.0 – 6.0 cm depending on the clinical center), elective repair is recommended. This one-size-fits-all approach does not take into account other individual AAA features that may differentiate aneurysms at high risk for rupture [22]. Quantification of AAA geometry highly correlated with wall mechanics could prove to be more relevant for assessing rupture risk than the maximum diameter criterion alone. Previous studies suggest that maximum diameter alone does not provide sufficient information to assess the individual aneurysm risk of rupture since the aorta varies greatly in size from one subject to another. Nussbaumer proposed that a dilatation ratio defined with the diameter of the normal aorta and the maximum AAA diameter is an important factor to evaluate for rupture risk assessment [23].

Giannoglou and colleagues [24] report that internal aortic tortuosity is a key factor in the prediction of rupture risk, while the mean centerline curvature was found to be a significant predictor of PWS by Georgakarakos et al [25]. In addition, infrarenal aortic volume appears to predict growth rate and correlates stronger with an estimated biomechanical rupture risk, according to Lindquist and associates [26].

The present work highlights the importance of taking into account AAA geometric features other than maximum diameter when differentiating the clinical status of three major population groups. These groups were identified with a retrospective review of medical records of AAA patients who were either placed under surveillance, had an elective repair, or an emergent repair based on the most recent set of available CT images. The outcome of the FEA simulations revealed that the differences in mean PWS were statistically significant among the groups. The most important outcome of the present study is the correlation of geometry with wall mechanics based on the 7 geometric indices that yield no discriminant error among the AAA groups. The regression analyses with PWS revealed higher correlation coefficients and higher predictive powers for all groups when using the 7 geometric discriminants compared to maximum diameter alone. While higher correlation coefficients do not necessarily indicate that the dependent and independent variables vary concomitantly, the predicted vs. observed values of PWS provide an assessment of such variances.

The correlations of geometry and PWS for the three distinct AAA groups underscore the importance of evaluating non-invasively these 7 important characteristics of AAA geometry: maximum diameter (D_{max}), tortuosity (T), maximum diameter to proximal neck diameter ratio (DDr), aneurysm outer wall surface area (S), median Gaussian curvature (K_{median}), maximum compactness (C_{max}), and the mode of the Mean curvature (M_{mode}). While DDr is co-dependent with D_{max} , it was retained as discriminator since removing it would have yielded a 4% error in the ability of the model to discriminate between the AAA groups. Noteworthy is that D_{max} , DDr , C_{max} , and S are geometric indices related to the size of the aneurysm, while T , K_{median} , and M_{mode} are related to the curvature or shape of the aneurysm. The first three of these indices can be estimated with standard clinical image processing software, while the other four indices require applying the algorithms described by Martufi et al [19], Shum et al [15, 20] and Lee et al [21] after segmentation of the CT images. This outcome is in agreement with the work of Washington and others [13], who found that PWS is highly correlated with D_{max} , AAA sac volume, and DDr . Nevertheless, finding that D_{max} is one of the significant discriminants of AAA geometry is an expected outcome of the study as the three population groups are created for the clinical management of the disease on the basis of measuring the aneurysm's maximum diameter.

One important limitation of this work was the inability to include intraluminal thrombus (ILT) in the geometric and FEA models, due to the lack of a contrast agent when performing the abdominal CT exams. It was reported previously that inclusion of ILT in a FEA model leads to lower PWS compared to FEA models devoid of ILT [27]. Likewise, when ILT is modeled as an incompressible linear elastic material, Speelman and others [28] found that for a relative thrombus volume of 30%, the reduction in PWS is 22% with a thrombus shear modulus of 40 kPa, 8% with a shear modulus of 10 kPa, and only 4% with a shear modulus of 4 kPa. The role of ILT in AAA rupture potential is controversial, with ILT reducing wall

stress locally [29, 30], but also increasing rupture risk [31, 32]. In some of these previous studies [28, 29], it was also reported that thrombus transmits pressure to the vessel wall. In vivo measurements of pressure reveal that the aneurysm wall is subjected to almost all ($91 \pm 10\%$) of the intraluminal pressure [31]. Hence, excluding the ILT in the geometric and FEA models is not an optimal choice since the thrombus may affect the gradients of wall stress across the aortic wall. Conversely, ILT inclusion in the FEA model would result in regional variations of the material properties of the wall in contact with the ILT. Such variations are unknown on a patient-specific basis and would be rather challenging to quantify from ECG-gated clinical imaging, which is not a standard of care procedure for AAA patients. Therefore, the material properties are typically measured ex vivo by means of planar biaxial tensile testing, which was recently used to determine that the AAA wall has greater stiffness in the circumferential than in the axial direction [33]. Evidently, relying on such ex vivo methods leads to the use of population-averaged homogeneous material properties for the AAA FEA models, which is a common shortcoming of any biomechanical modeling approach. An alternative approach, which eliminates the need for patient-specific material properties while using standard of care CT images and optional non-standard of care MR images, is described by Joldes and colleagues [34, 35].

The number of subjects in each population group represents another limitation of this work as it is not possible to obtain statistical significance of the correlations of geometry with wall stress by using all 47 geometric indices. Since the number of independent variables is greater than the sample size of the population, statistical significance can only be achieved by reducing the number of variables, hence the justification for performing a feature reduction analysis. Therefore, the outcome of the correlations should be taken with caution as it is bounded by said analysis. The relative effect of each geometric discriminant highly correlated with PWS is quantitatively given by the power coefficients and goodness of fit described in Table 4. While D_{max} is an important index in the correlations, six additional discriminants (T , DDr , S , K_{median} , C_{max} , and M_{mode}) should also be considered surrogates for PWS, in lieu of complex finite element modeling.

Future work in geometric and biomechanical modeling of AAA should be directed toward testing the predictive value of the geometric surrogates of PWS in a large population size using prospectively acquired human subjects data. While there are important ethical considerations to be taken into account when implementing these surrogates as rupture risk predictors, the prospective nature of such data acquisition would allow for longitudinal assessment of the geometric indices during imaging follow-up of each patient, which typically occurs every 6 months as a standard of care. The quantification of these indices may lead to the creation of a database consisting of geometric surrogates that statistically discriminate between at-risk and “safe” AAA, from which any future AAA models can be compared against by means of classifier models such as those created with machine learning techniques. The clinical translational potential of the computational tools described in this work is high as meshing, geometric index quantification and classification models can be fully automated and executed on a standard Windows laptop computer in minutes. Nevertheless, the most important roadblock in achieving fully automated outcomes is the image segmentation step, which at the time of this writing is a user-dependent process and time consuming, particularly as it relates to the identification of the thrombus/wall boundary.

5. CONCLUSION

Computed tomography angiography images of thirty patient-specific AAA corresponding to three population groups (10 subjects under surveillance, 10 electively repaired and 10 ruptured who received an emergent repair) were retrospectively acquired to generate computational models for finite element analysis and geometry quantification. The first principal stress and 47 geometric indices characteristic of the individual vessel geometry were calculated for each AAA model. After reducing the geometric indices to a subset of 7 discriminants (D_{max} , T , DDr , S , K_{median} , C_{max} , and M_{mode}) that differentiate the geometry of the three groups with no error, a non-linear regression analysis was performed with these seven indices and PWS. We conclude from this analysis that the aforementioned 7 statistically significant geometric discriminants can be used as surrogates for AAA PWS predictions. This work represents initial evidence in support for a geometry-based approach that can be used in the clinic to assess the at-risk condition of individual AAA.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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HIGHLIGHTS

- A combined biomechanics and geometry quantification approach is proposed to find the surrogates of aortic aneurysm mechanics.
- There are 7 geometric variables that are highly correlated with peak wall stress and differentiate the aneurysm population groups with zero error.
- The proposed 7 discriminants can be used as surrogates of peak wall stress in lieu of complex finite element modeling for aneurysm rupture risk assessment.

Figure 1.
Exemplary first principal stress (PS1) [in N/cm²] AAA wall distribution.

Figure 2.

Predicted (by the correlation polynomials given by Eq. (2)) vs. observed (by the FEA simulations) PWS [N/cm²]: (a) Group I, (b) Group II, and (c) Group III. The predicted PWS fall within $\pm 21\%$ of the observed PWS for each population group.

Table 1

Peak wall stress (PWS) and geometric indices for the three AAA population groups (mean $\pm$ standard deviation).

Biomechanical or Geometric Variable	Group I (n = 10)	Group II (n = 10)	Group III (n = 10)
PWS (N/cm ²)	41.3 $\pm$ 10.0	68.7 $\pm$ 24.9	80.1 $\pm$ 25.2
D_{max} (cm)	4.1 $\pm$ 0.2	5.1 $\pm$ 0.2	7.3 $\pm$ 0.9
$D_{neck,p}$ (cm)	2.5 $\pm$ 0.4	2.6 $\pm$ 0.6	3.3 $\pm$ 0.8
$D_{neck,d}$ (cm)	2.7 $\pm$ 0.4	2.9 $\pm$ 0.6	2.7 $\pm$ 1.6
H (cm)	9.3 $\pm$ 1.3	9.8 $\pm$ 1.2	11.5 $\pm$ 2.0
L (cm)	12.0 $\pm$ 1.7	12.2 $\pm$ 1.5	16.7 $\pm$ 3.5
H_{neck} (cm)	2.3 $\pm$ 1.7	2.1 $\pm$ 1.6	1.4 $\pm$ 1.3
L_{neck} (cm)	2.9 $\pm$ 2.1	2.9 $\pm$ 2.0	2.0 $\pm$ 1.7
H_{sac} (cm)	7.0 $\pm$ 1.5	7.7 $\pm$ 1.8	10.1 $\pm$ 2.0
L_{sac} (cm)	9.0 $\pm$ 2.0	9.3 $\pm$ 2.6	14.7 $\pm$ 3.3
H_b (cm)	5.2 $\pm$ 1.9	5.9 $\pm$ 1.6	5.7 $\pm$ 1.8
d_c (cm)	0.2 $\pm$ 0.2	0.4 $\pm$ 0.3	1.1 $\pm$ 0.9
DHr	0.5 $\pm$ 0.1	0.5 $\pm$ 0.1	0.6 $\pm$ 0.1
DDr	1.7 $\pm$ 0.3	2.0 $\pm$ 0.5	2.3 $\pm$ 0.6
Hr	0.2 $\pm$ 0.2	0.2 $\pm$ 0.2	0.1 $\pm$ 0.1
BL	0.6 $\pm$ 0.2	0.6 $\pm$ 0.1	0.5 $\pm$ 0.1
β	0.9 $\pm$ 0.2	0.9 $\pm$ 0.1	0.8 $\pm$ 0.1
T	1.3 $\pm$ 0.1	1.2 $\pm$ 0.1	1.4 $\pm$ 0.1
C_{ave}	1.1 $\pm$ 0.0	1.1 $\pm$ 0.0	1.1 $\pm$ 0.2
C_{min}	1.0 $\pm$ 0.0	1.0 $\pm$ 0.0	1.0 $\pm$ 0.1
C_{max}	1.6 $\pm$ 0.5	1.4 $\pm$ 0.6	1.8 $\pm$ 0.6
V (cm ³)	98.1 $\pm$ 18.5	150.2 $\pm$ 32.3	331.0 $\pm$ 116.1
S (cm ²)	143.8 $\pm$ 20.6	185.7 $\pm$ 31.7	269.9 $\pm$ 60.3
IPR	6.8 $\pm$ 0.4	6.6 $\pm$ 0.6	5.7 $\pm$ 0.4
NFI	0.8 $\pm$ 0.1	0.8 $\pm$ 0.1	0.8 $\pm$ 0.1
GAA (mm ⁻¹)	-0.6 $\pm$ 0.1	-0.5 $\pm$ 0.2	-0.4 $\pm$ 0.2
MAA (mm ⁻²)	-0.3 $\pm$ 0.0	-0.3 $\pm$ 0.0	-0.2 $\pm$ 0.0
GLN	19.9 $\pm$ 5.5	23.3 $\pm$ 8.6	29.0 $\pm$ 9.2
MLN	0.8 $\pm$ 0.2	0.8 $\pm$ 0.3	0.8 $\pm$ 0.3
K_{min} (mm ⁻¹)	-46.5 $\pm$ 3.0	-45.1 $\pm$ 5.0	-44.3 $\pm$ 5.6
K_{max} (mm ⁻¹)	6.6 $\pm$ 5.2	8.3 $\pm$ 8.4	11.1 $\pm$ 16.1
K_{ave} (mm ⁻¹)	-0.6 $\pm$ 0.1	-0.5 $\pm$ 0.2	-0.4 $\pm$ 0.2
K_{median} (mm ⁻¹)	-0.1 $\pm$ 0.1	0.0 $\pm$ 0.0	-0.1 $\pm$ 0.1
K_{mode} (mm ⁻¹)	0.0 $\pm$ 0.0	0.0 $\pm$ 0.0	0.0 $\pm$ 0.0

Biomechanical or Geometric Variable	Group I (n = 10)	Group II (n = 10)	Group III (n = 10)
K_{var} (mm ⁻¹)	6.4 ± 2.5	5.5 ± 3.1	4.4 ± 2.4
M_{min} (mm ⁻²)	-31.1 ± 13.3	-25.5 ± 14.2	-21.0 ± 13.8
M_{max} (mm ⁻²)	32.2 ± 13.1	22.4 ± 14.6	22.2 ± 15.0
M_{ave} (mm ⁻²)	-0.3 ± 0.0	-0.3 ± 0.0	-0.2 ± 0.0
M_{median} (mm ⁻²)	-0.4 ± 0.1	-0.3 ± 0.0	-0.2 ± 0.0
M_{mode} (mm ⁻²)	-0.4 ± 0.0	-0.4 ± 0.0	-0.2 ± 0.1
M_{var} (mm ⁻²)	1.8 ± 1.1	1.4 ± 1.2	1.0 ± 0.9
$t_{w,ave}$ (mm)	1.5 ± 0.3	1.8 ± 0.4	2.0 ± 0.7
$t_{w,min}$ (mm)	0.5 ± 0.2	0.6 ± 0.1	0.4 ± 0.3
$t_{w,max}$ (mm)	3.9 ± 1.9	4.1 ± 2.0	4.8 ± 1.9
$t_{w,Dmax}$ (mm)	1.5 ± 0.3	1.8 ± 0.4	2.2 ± 0.8
$t_{w,mode}$ (mm)	1.4 ± 0.2	1.6 ± 0.4	1.8 ± 0.8
$t_{w,median}$ (mm)	1.5 ± 0.2	1.7 ± 0.3	1.9 ± 0.7
$t_{w,meanVar}$ (mm)	0.1 ± 0.1	0.2 ± 0.2	0.3 ± 0.3

Table 2

The eleven geometric indices obtained by feature selection following the stepwise control method: D_{max} (maximum transverse diameter), T (tortuosity), DDr (maximum diameter to proximal neck diameter ratio), S (surface area of the AAA), NFI (non-fusiform index), K_{median} (median of the Gaussian curvature), C_{max} (maximum lumen compactness), β (asymmetry factor), MAA (area averaged Mean curvature), M_{mode} (mode of the Mean curvature), and L_{neck} (centerline length of AAA neck). These are the statistically significant indices amongst the 30 AAA models.

Geometry index	p-value
D_{max}	$\ll 0.0001$
T	$\ll 0.0001$
DDr	$\ll 0.0001$
S	$\ll 0.0001$
NFI	$\ll 0.0001$
K_{median}	$\ll 0.0001$
C_{max}	$\ll 0.0001$
β	$\ll 0.0001$
MAA	$\ll 0.0001$
M_{mode}	$\ll 0.0001$
L_{neck}	$\ll 0.0001$

Table 3

The seven discriminant geometric indices and their goodness of fit based on a nonlinear non-parametric correlation with PWS: S (surface area of the AAA), D_{max} (maximum transverse diameter), T (tortuosity), M_{mode} (mode of the Mean curvature), K_{median} (median of the Gaussian curvature), DDr (maximum diameter to proximal neck diameter ratio), and C_{max} (maximum lumen compactness). These indices are capable of discriminating the three AAA groups with no error.

Geometry index	Goodness of fit (R^2)	p-value
S	0.549	$\ll 0.0001$
D_{max}	0.464	$\ll 0.0001$
T	0.450	$\ll 0.0001$
M_{mode}	0.284	0.0024
K_{median}	0.236	0.0065
DDr	0.152	0.0333
C_{max}	0.091	0.1061

Power coefficients and goodness of fit (R^2) for the non-linear parametric polynomial correlations represented by Eq. (2) for each population group.

Table 4

AAA Group	a1	a2	a3	a4	a5	a6	a7	R ²
I	1.264	0.281	-0.063	-0.048	-0.050	0.070	-0.368	0.579
II	1.038	1.647	-1.172	-0.039	-0.029	0.137	-0.238	0.642
III	-0.859	2.260	0.275	0.534	-0.051	-0.344	0.036	0.780