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(PEAK) WALL STRESS AS AN INDICATOR OF ABDOMINAL AORTIC ANEURYSM SEVERITY

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Abstract

Abdominal aortic aneurysms, which consist of dilatations of the infra-renal aorta by at least 1.5 times of its normal diameter, are becoming a leading cause of death worldwide. Rupture often occurs unexpectedly, before a repair procedure is conducted. The AAA maximum diameter has been used as a clinical criterion to monitor AAA severity. However, assessment of AAA rupture risk requires knowledge of wall stress and wall strength at the potential rupture location. We conducted a study on 37 patient specific CT datasets to investigate the benefits of using peak wall stress instead of D_{max} for AAA rupture severity. Correlation between PWS and 24 geometric indices and biomechanical factors was studied where eleven of them showed a statistically significant correlation with PWS. A Finite Element Analysis Rupture Index was used to conclude that the use of D_{max} as a single predictor of AAA behavior and severity may be insufficient based on our patient population with a D_{max} smaller than the 5.5 cm, clinically recommended repair threshold.

Index Terms

Abdominal Aortic Aneurysm (AAA); Computed Tomography (CT); geometrical indices; image processing

1. INTRODUCTION

Abdominal aortic aneurysms (AAA) are expansions of the infrarenal aorta by at least 1.5 times of its normal diameter. The mortality of AAA rupture is increasing, while also

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expanding more toward younger populations [1–4]. If rupture happens, 50% of patients will die before arriving to the emergency room; moreover, there is up to 80% 30-days risk of mortality for patients undergoing AAA repair surgery [5–7].

For decades, the maximal diameter (D_{max}) has been considered as the main factor responsible for the growing weakness of the aortic wall. According to Laplace's law for thin-walled tubes, which states a linear relationship between vessel diameter and wall stress, it is expected that wall stress will increase with an increase in the vessel diameter [8, 9]. As such, D_{max} has also served as the most relevant biomarker for clinical follow-up and a faithful predictor for assessing the need of surgery or repair, with D_{max} values higher than 5.5 cm suggesting a significant risk of rupture [10–12].

However, AAAs have more complex shapes than just simple cylinders, rendering Laplace's law as only a first-order approximation. Additionally, recent studies have suggested that D_{max} alone may be a necessary, but not a sufficient predictor of aneurysm severity, as 1% of the patients featuring maximal aneurysm diameters smaller than 5.5 cm have still experienced rupture each year [11, 13, 14].

However, from a biomechanical point of view, rupture occurs if the overall effective stress developed inside the aortic wall exceeds the mechanical strength of the vessel tissue [1, 7, 15]. Therefore, the models which utilize peak wall stress (PWS) as an indicator for severity of rupture are more promising. Additionally, it has been shown that the rupture site is related to the location of the AAA where the PWS occurs rather than the location of D_{max} [6]. Hence, peak wall stress can be implemented as a more effective AAA rupture criterion that incorporates geometrical parameters and biomechanical parameters such as blood pressure and intraluminal thrombus, especially for aneurysms with $D_{max} < 5.5$ cm that are deemed not at risk according to the clinical D_{max} criterion.

According to the AAA literature, model-based estimates of the peak wall stress are viewed as relevant, quantitative indicators of aneurysm severity, with a larger PWS indicating a higher severity and risk of rupture. As such, studying the effect of geometric and biomechanical parameters on the peak wall stress helps us better understand the location where the aneurysm is most likely to rupture and, in turn, better understand the underlying concept of aneurysm rupture and its causes.

In current work, we extended our previous model demonstrated on mice and synthetic data [16] and implemented it on patient specific 3D AAAs. Moreover, with the goal of emphasizing the benefit of using PWS beyond that of D_{max} as an indicator of aneurysm severity, a Finite Element Analysis Rupture Index was used on patient specific AAA models along with the patient specific blood pressures.

2. MATERIALS AND METHODS

2.1. Imaging data

Contrast enhanced Computer Tomography Angiography (CTA) images of AAAs from 37 patients (6 women and 31 men) 69 ± 9 years old were retrospectively obtained from the

Division of Vascular Surgery at the University of Rochester following approval by the Human Subject Review Board. Eleven of these patients have gone through repair and we consider them as ruptured AAA while 25 of them are un-ruptured. All CT images were acquired on a Philips --Brilliance 64 CT scanner using a 512×512 scan matrix with an average pixel size of 0.7 mm and average slice thickness of 3.53 mm (Figure 1a shows a sample of CTA images).

2.2. 3D AAA model generation

The first component of our framework used to construct a 3D model of the AAA is explained in detail below.

2.2.1 Segmentation—DICOM images (Figure 1a) from AAA CT were imported into MATLAB. We used different filtering methods to enhance the resolution and quality of the CT images: a Gaussian low-pass filter was implemented to reduce the level of noise in the original image to improve the effectiveness of the edge detection; a sharpening filter was used to increase the sharpness of the image features; and finally, a histogram equalization operation was performed to enhance image contrast.

The active contour method was used to perform the segmentation of outer and lumen wall [17]. The default mask for active contour segmentation is typically a rectangular mask, which, for low resolution and noisy images, such as clinical quality images, results in lots of unwanted points on the segmented image. We modified the process of mask selection by enabling the user to define a free-shape contour which is more compatible and adaptive to the wall of aorta for the first slice in AAA scan set. Then, our algorithm starts to segment the lumen wall using the active contour algorithm and displays the contour and the segmented wall.

After the segmentation process is completed for one slice, algorithm uses each segmented (lumen and outer) wall as the mask for the segmentation of the next slice and continues the segmentation process automatically until the last slice of each scan set resulting in a stack of contours for lumen (in blue in Figure 1b) and outer walls of AAA (in red in Figure 1b). At the end of the segmentation procedure, the user can evaluate if there is any discontinuity between different layers and, if needed, can perform the segmentation only for the discontinued layers.

2.2.2. Interpolation and point cloud generation—While no minimum number of sequential slices was required to build the 3D vessel geometry, up-sampling the dataset and introducing intermediate segmented slices in-between the native slices avoid large interpolation errors, yielding a smoother point cloud, and, consequently, a smoother 3D vessel model (Figure 2b). To this extent, our algorithm enables the user to select the number of intermediate slices to be “inserted” between the native sequential segmented CT slices, followed by interpolation for lumen (in blue in Figure 1c) and outer walls of AAA (in red in Figure 1c).

2.2.3. 3D mesh generation—The algorithm then generated the 3D mesh of the aorta wall by connecting the neighboring nodes via four-node surface elements. We used volume

preserving 3D smoothing tools to create a smooth full 3D mesh of the aorta wall, as shown in Figure 1d, which subsequently served as input into the ANSYS finite element model (FEM) [18], along with nodal coordinates, element connectivity information, material properties, boundary conditions, and blood pressure (loading conditions).

2.2.4. Calculation of geometrical indices—Once the 3D model was constructed, the outer wall surface of the aorta model was used to calculate the 24 geometrical indices that characterize the AAA according to its 1D size, 2D shape, 3D size, 3D shape and second-order curvature-based characteristics using the definitions and formulations adapted from Martufi *et al.* [10], Shum *et al.* [9], and Tang 2013 [11, 12].

2.3. Wall stress distribution reconstruction

While the stress distribution developed within simple geometries may be estimated using analytical solutions, complex geometries require numerical solutions typically obtained using the finite element method (FEM) [12]. The latter component of our computational model consisted of the FEM solver that uses the ANSYS commercial software to analyze the biomechanical behavior of the AAA and quantify the stress and deformation developed in the vessel wall.

The 3D AAA mesh shown in Figure 1d was imported into ANSYS. The number of elements used for each AAA was determined based on a mesh sensitivity analysis. While a higher number of smaller elements may improve accuracy, our mesh sensitivity analysis showed that roughly doubling the number of elements (i.e., from ~ 9,000 to ~ 20,000 elements) only resulted in a 1.2% change in PWS. To ensure sufficient accuracy in our analysis, our human AAA meshes ranged from 30,000 to 130,000 elements, depending on the geometry and characteristics of each individual AAA. Unlike previous studies which used linear hexahedral element (solid 185 element type in ANSYS) [4, 12], we used the quadratic hexahedral element (solid 186 element type in ANSYS), two elements across the wall thickness of 2-mm [19]. The advantage of using this element type was its ability to handle deformation and bending more accurately due to its three degrees of freedom per node and quadratic displacement behavior, ultimately resulting in a more accurate simulation for hyper-elastic material properties.

We then used a two-parameter ($d = 17.4 \text{ N/cm}^2$ and $\mu = 188.1 \text{ N/cm}$) hyper-elastic, isotropic, incompressible material model similar to the one proposed by Raghavan and Vorp [8] to model the behavior of the AAA wall.

2.3.1. Incorporation of intraluminal thrombus (ILT)—It has been shown that the presence of intraluminal thrombus (ILT) impacts both aortic wall degeneration and wall stress distribution in the AAA. ILT increases inflammation of wall and wall weakening due to degradation of smooth muscle cells (SMCs) and extracellular matrix [8, 15]. It has been proved that a larger ILT is associated with lower wall stress while causing an elevation on wall stress on the thinner portions of AAA wall.

Therefore, to have a comprehensive understanding of AAA's behavior, the inclusion of ILT is necessary. We incorporate the effect of the intraluminal thrombus by “filling in” the

volumetric mesh (a development from the shell meshing in our previous work [16]) of the region between the inner wall and the lumen. This provides a better estimation of PWS for AAAs which exhibit different amount of ILT and correspondingly different thicknesses throughout the aneurysm. To mesh the ILT region, we also used the quadratic hexahedral elements. The quadratic hexahedral element type is a 20-node element having three degrees of freedom at each node. Figure 2a shows a cross sectional image of the AAA including inner wall, outer wall and ILT, while Figure 2b illustrates volume meshing for the wall and ILT.

2.4. Loading and boundary conditions

The loading condition is represented by the pulse pressure (PP) which is the difference between the peak systole pressure and the diastole pressure. We used pulse pressure to stimulate the zero- pressure AAA geometry at the beginning of the measurement. In terms of the boundary conditions, as also reported in previous studies, we assume that both ends of the aneurysm were fully constrained from moving in any direction

2.5. Finite Element Analysis Rupture Index (FEARI):

To better emphasize the benefit of using PWS to study the AAA's behavior and its rupture risk, here we calculate a previously proven rupture index. The Finite Element Analysis Rupture Index (FEARI) is an indicator of the rupture risk estimated based on the failure definition [1] - when the stress overcomes the strength of the AAA's wall. A FEARI value close to 0 indicates a low risk of rupture, while a value close to 1 suggests a high risk of the rupture.

$$FEARI = \frac{FEA \text{ wall stress (PWS)}}{Experimental \text{ wall strength}}$$

To compute the FEARI index, we used the wall strength values from the study in [1], in which the authors performed a uniaxial test on 149 AAA samples and measured the average wall strength in four regions of AAA: anterior, posterior, lateral, and medial. Using the reported wall strength and the PWS predicted by our FE model, we computed the FEARI index for each patient.

3. RESULTS

In current work, we evaluated our previous study [16] to model patients specific AAAs including ILT. Figure 3 illustrates the stress distribution and location of PWS for one patient.

To have a better understanding of the AAA behavior and the factors which affect the development of AAA, its severity, and, accordingly, its likelihood of rupture, we performed the full analysis on the correlation between 24 different geometrical indices and loading blood pressure with PWS. The range of PWS computed from our model is consistent with PWS values reported in previous studies [4, 12, 13]. Table I shows the summary of the correlation analysis between PWS and all geometric indices. It can be seen that PWS shows statistically significant correlation ($p < 0.05$) with 10 geometrical indices and the blood

pressure. These results are also consistent with findings reported in previous work [9, 19, 20].

Table II shows a summary of the peak wall stress, D_{max} , and FEARI index for all patients. It can be seen that patients with smaller D_{max} are showing larger FEARI meaning higher rupture risk; even some of them have been ruptured compared to patients with larger D_{max} .

4. DISCUSSION AND CONCLUSION

In current work, we implemented a two-parameter model of AAA wall behavior on patient specific AAAs, with the goal to assess the changes in peak wall stress and maximum diameter D_{max} as a predictor of AAA severity and risk of rupture. We evaluated our model using patient specific geometric AAA models extracted from CTA data, along with patient specific blood pressures to analyze the wall stress distribution as a measure of the rupture severity of AAAs alternative to D_{max} . These data suggested that PWS shows correlation with different geometrical parameters as well as loading pressure, therefore, it could be utilized as a more promising criterion for rupture severity validation rather than D_{max} alone.

Moreover, we also computed the rupture index (FEARI) using the model-predicted peak wall stress values and compared the rupture index predicted by the peak wall stress to the clinical decision with regards to the need for AA repair.

Future efforts will focus on increasing the complexity of the model, such as accounting for regional material properties, and studying their effects on the predicted peak wall stress and its correlation with the maximum diameter D_{max} and rupture index identified based on the clinical decision with regards to the need for repair surgery.

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Fig. 1:
Example of AAA modeling: (a) an axial CT slice of human AAA, (b) Segmented contours of the aortic lumen (blue) and outer (red) walls, (c) Interpolated point cloud of the aortic lumen (blue) and outer (red) walls (d) Smoothed 3D mesh of the outer wall, and (e) Distribution of the peak wall stress.

Fig. 2:

(a) Cross section of AAA with black arrow 1 showing outer wall, black arrow 2 showing inner wall, black arrow 3 showing the lumen wall (ILT lies between the inner wall and the lumen wall), and red arrow indicates the location of PWS, (b) the volume meshing of the wall and ILT.

Fig 3:
Posterior view of the AAA showing the wall stress distribution predicted by our model.

TABLE I:

Spearman's correlation coefficient between geometrical indices and PWS

Geometrical Indices	Correlation Coeff. (r)	P value, Statistically Significant * (if $p < 0.05$)
1D size indices		
D _{max}	0.79	<0.001*
Dneck 1	0.38	0.02*
Dneck 2	0.40	0.01*
Hsac	0.27	0.09
Hneck	-0.24	0.15
Hb	0.001	0.10
Lsac	0.34	0.03*
Lneck	-0.19	0.26
Dc	0.16	0.35
D	0.19	0.27
2D size indices		
DDr	0.42	0.009*
DHr	-0.22	0.18
Hr	0.38	0.02*
BL	-0.32	0.05*
T	0.14	0.40
β	-0.002	0.99
3D size indices		
S	0.63	<0.001*
V	0.55	<0.001*
VILT	0.38	0.01*
γ	-0.22	0.20
Three-dimensional (3D) shape indices		
IPR3D	0.03	0.86
2nd order curvature-based indices		
GAA	0.03	0.88
MAA	0.04	0.83
GLN	-0.03	0.85
MLN	0.20	0.24
Patient Specific Blood Pressure		
BP	0.54	<0.001*

*** The definitions and formulations adapted from Martufi *et al.* [4], Shum *et al.* [18], and Tang 2013 [11].

TABLE II:Summary of FEARI, PWS, and D_{max} for all patients

Patient	PWS (kPa)	D_{max} (cm)	FEARI	Ruptured
1	128.0	7.53	0.17	
2	62.5	4.26	0.07	
3	69.5	6.05	0.08	
4	54.9	4.96	0.06	
5	111.5	6.95	0.12	1
6	51.2	4.77	0.06	
7	68.5	5.28	0.09	
8	56.4	4.14	0.06	1
9	51.8	5.82	0.06	
10	45.0	5.36	0.05	
11	55.2	5.71	0.06	
12	64.3	4.84	0.07	1
13	71.0	4.93	0.08	1
14	61.2	5.52	0.07	
15	31.7	4.54	0.04	
16	81	6.34	0.10	
17	50.3	4.83	0.06	
18	65.5	5.24	0.08	
19	77.6	5.13	0.09	1
20	126	7.33	0.15	1
21	24.0	4.02	0.03	
22	97.1	6.61	0.11	
23	79.8	6.49	0.10	1
24	125.0	6.04	0.14	
25	65.5	5.50	0.08	1
26	75.8	6.56	0.10	
27	27.2	3.76	0.03	
28	65.5	5.80	0.08	1
29	55.7	5.38	0.06	
30	77.2	6.57	0.09	1
31	66	7.47	0.08	
32	72.9	6.09	0.08	
33	74.7	5.33	0.09	
34	33.6	3.79	0.04	
35	60.7	6.30	0.07	1
36	70.5	6.08	0.08	

Patient	PWS (kPa)	D_{max} (cm)	FEARI	Ruptured
37	50.1	4.68	0.06	

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