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Toward modeling the effects of regional material properties on the wall stress distribution of abdominal aortic aneurysms

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Abstract

The overall geometry and different biomechanical parameters of an abdominal aortic aneurysm (AAA), contribute to its severity and risk of rupture, therefore they could be used to track its progression. Previous and ongoing research efforts have resorted to using uniform material properties to model the behavior of AAA. However, it has been recently illustrated that different regions of the AAA wall exhibit different behavior due to the effect of the biological activities in the metalloproteinase matrix that makes up the wall at the aneurysm site. In this work, we introduce a non-invasive patient-specific regional material property model to help us better understand and investigate the AAA wall stress distribution, peak wall stress (PWS) severity, and potential rupture risk. Our results indicate that the PWS and the overall wall stress distribution predicted using the proposed regional material property model, are higher than those predicted using the traditional homogeneous, hyper-elastic model ($p < 1.43E-07$). Our results also show that to investigate AAA, the overall geometry, presence of intra-luminal thrombus (ILT), and loading condition in a patient specific manner may be critical for capturing the biomechanical complexity of AAAs.

Keywords

Abdominal aortic aneurysm; regional material properties; wall stress distribution; finite element analysis; Physiological modeling; Vascular imaging; Biomechanical modeling; Image processing and segmentation

1. INTRODUCTION

Abdominal aortic aneurysms (AAA) are expansions of the infra-renal aorta by at least 50% of its normal diameter [1–3]. AAAs are one of the leading causes of death in the United States [1, 4] and the mortality of AAA rupture is increasing, while also expanding more toward younger populations [5]. If rupture happens, 50% of patients will die before arriving

to the emergency room; moreover, there is a 40–50% risk of mortality for patients undergoing AAA repair surgery [5].

The gradually increasing weakening of the aortic wall due to metabolic activity at the aneurysm site has been introduced as the main reason for AAA development [5, 6]. As such, researchers have pointed that rupture is likely to occur if the overall effective stress developed inside the aortic wall exceeds the mechanical strength of the vessel tissue. For decades, the $D_{max} > 5.5$ cm criterion has been used as the main diagnostic factor to study AAA severity; however, it has been shown, including our previous study [6], that the peak wall stress along with the whole geometry of the AAA provides a more reliable indicator of the rupture. Moreover, it was shown that the rupture site is related to the location of the AAA where the PWS occurs [5–7]; hence, this AAA rupture criterion can be implemented and estimated using biomechanical parameters such as peak wall stress (PWS), especially for aneurysms with $D_{max} < 5.5$ cm. As such, it is critical to faithfully study the AAA behavior and accurately assess its severity and potential risk of rupture according to the developed wall stress distribution. This process entails the reconstruction of a 3D AAA model, the estimation of the developed wall stress distribution, and the identification of the location(s) exhibiting peak wall stress in response to the loading conditions (i.e., internal pressure within the AAA), geometric, and biomechanical aneurysm parameters. Therefore, biomechanical parameters, including material properties of aortic wall at AAA bulge, wall thickness, and intraluminal thrombus, are needed to be considered in the study of AAA.

Two main material (properties) models have been used to model the behavior of the aortic wall: a linear elastic model and a non-linear elastic model. While the linear, elastic approximation is not obsolete and is still employed in some recent studies, it has been acknowledged that the biomechanical behavior of the aortic wall, similar to that many other soft tissues [8, 9], is better captured by a non-linear elastic model. This behavior is rendered by the collagen fibers that make up the different layers in the aortic wall [4, 8–10]. However, it has been recently suggested that regional material properties are more effective when analyzing AAAs and their behavior rather than assuming a uniform material property for the entire AAA wall.

Wilson *et al.* [11] conducted an experiment that illustrates from a mechano-biology perspective, how effects of intra-luminal thrombus (ILT) and regional damage in extracellular matrix due to degradation can result in regional material anisotropy, stiffness inhomogeneity, and regional wall thickness. Therefore, these effects need to be taken into account to study the AAA behavior and rupture risk [11]. Their study showed that the stiffest aortic material tends to have the slowest growth rate and lowest stretch in collagen fibers, while the least stiff aortas had highest stress and more expansion rate. However, it still remains controversial whether these findings conclude a lower rupture risk. This study used an idealized AAA geometry on a small sample size of 6 patients and did not consider effects of ILT [11].

Tierney *et al.* [7] used CT images acquired at both peak systole and diastole at the location of D_{max} and neck of AAA to calculate the strain. They showed that although the location of PWS was similar for a homogenous AAA and for an AAA model with regional material

properties, a difference of 5% in magnitude of strain was observed between the two models, resulting in a different PWS magnitude predicted by the two models. Their study was conducted on a small sample of three male patients using CT images featuring the AAA neck and D_{max} , they employed the same 3D geometric model for all three cases, did not consider ILT effects, and considered the regional material properties across four regions of the AAA [7].

Reeps *et al.* [12] used wall samples for 50 patients who underwent open surgery. They measured mechanical wall properties. They showed that wall thickness is related to metabolic activity of wall and also to ILT. They concluded that in order to more accurately predict the rupture risk of AAA, all of these parameters have to be considered. They mentioned some of their limitations as the fact that samples were collected from patients who had an open surgery or repair, which limited them with a small number of samples. Moreover, using PET/CT imaging for more accurate wall thickness prediction is not a standard clinical method for AAA and cannot be applied on a large sample size of patients [12].

Despite all efforts to study the effect of the material properties of aortic wall and AAA, a handful of limitations still remain to be addressed, which indicate the need for more investigations and developments. In this paper, we extend our previous work toward a more comprehensive biomechanical model that incorporates regional wall material properties and use it to study several patient-specific cases along with regional wall characterizations. We employ a uniform model for material properties (a two-parameter, hyper-elastic, isotropic, incompressible material formulation) and our proposed model with regional material properties, and compare the wall stress distributions to study the biomechanical behavior of the AAA wall.

2. MATERIALS AND METHODS

2.1. Imaging data

Contrast enhanced Computer Tomography Angiography (CTA) images of un-ruptured AAAs from nine patients 76 ± 10 years old were retrospectively obtained from the Division of Vascular Surgery at the University of Rochester following approval by the Human Subject Review Board. All CT images were acquired on a Philips Brilliance 64 CT scanner using a 512×512 scan matrix with an average pixel size of 0.7605 mm and average slice thickness of 3.74 mm (Figure 1a shows a sample of CTA images).

2.2. 3D AAA model generation and wall stress distribution reconstruction

Our computational workflow consists of two components which have been described in detail in [6]. The former component uses commercial software to construct a 3D mesh model of the AAA from clinical quality CT or MRI images (Figure 1b, c). In addition, this step includes the evaluation of the geometrical indices that describe the shape and size of the AAA [6]. The latter component consists of the developed finite element model (FEM) using commercial software based on the computational domain extracted from the previous step, along with prescribed biomechanical parameters and loading conditions to analyze the

biomechanical behavior of the AAA and compute the wall stress distribution under the prescribed conditions [6] (Figure 1d).

2.3. Incorporation of wall thickness and the intraluminal thrombus (ILT) effects

Due to the relatively low resolution of the clinical-quality medical images available for this study, extraction of the true wall thickness of the aneurysm was challenging. To mitigate this limitation, a 2-mm uniform wall thickness was prescribed throughout the entire length of the aorta, similar to previous studies [13–15]. To mesh the wall thickness, as part of our FEM[6], we use quadratic hexahedral element (SOLID186 element type in ANSYS), in contrast with previous studies which used either the shell elements (element SHELL93 in ANSYS) or the tri-linear hexahedral element (SOLID185 in ANSYS) [4, 9, 10], with two elements across the thickness (between the outer wall and inner wall of the aorta). This element type is a 20-node element having three degrees of freedom at each node. Usage of the quadratic hexahedral elements are better at handling deformation and bending in different area of the AAA due to its three degrees of freedom per node, ultimately producing more realistic simulations of the AAA behavior. 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.

We incorporate the effect of the intraluminal thrombus by “filling in” the volumetric mesh (a development from the shell meshing in our previous work [6]) 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, as it has been shown that ILT affects the overall wall stress distribution as well as the PWS and, in turn, the AAA rupture risk [4, 10, 16, 17]. To mesh the ILT region, we also used the quadratic hexahedral elements.

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 [3, 5]. In terms of the boundary conditions, to the best of our knowledge, there have been no specific studies available in the field that aim to realistically model the effect of the surrounding organs on the AAA in terms of pressure, force or deformation constraints. As such, as also reported in previous studies, we assume that both ends of the aneurysm were fully constrained from moving in any direction [5, 14, 18, 19].

2.5. AAA tissue modeling: strain measurement

In order to determine the patient-specific relative regional material properties, all nine patients underwent ultrasound evaluation of their AAA prior to their CT scans. B-mode ultrasound images of the abdominal aorta were obtained by a registered vascular technologist using a Sonix-Touch US system (Analogic), with an Ultrasonix C7–3/50 convex transducer at a center frequency of 5 MHz. The technologist visually determined the point of maximal diameter, in the circumferential orientation, and had the patient preform a 10 second breath hold. A 10-second sequence of raw RF ultrasound imaging data of the

pulsatile aorta was obtained. Non-invasive blood pressure measurements were obtained using a Welch Allyn Spot Vital Signs system, prior to and after completion of ultrasound image acquisition [3].

The 2D displacement field was calculated for the entire imaging field across the cardiac cycle. The mean principal strain was determined across the entire cardiac cycle, within four anatomical regions of the segmented vessel, as reference to the vessel centroid (Anterior +45° to -45°, posterior +135° to -135°, medial +45° to +135°, and lateral -45° to -135°) (Figure 2c). The image frame which corresponded to the mean maximum principle strain, during one cardiac cycles, was assumed to occur at systole, and was used to determine the maximum principal strain within each of the anatomic regions. More details on the strain measurements could be find in [3].

2.6. Stiffness surrogate and regional material property estimation

As described earlier, each AAA was divided into four anatomic regions and the maximum principle strain was measured for each region for each patient [3]. The elastic modulus (E) is the slope of the stress-strain curve, which provides information about the stiffness of the material as an indicator of material properties. In this case, a surrogate stiffness metric in lieu of the elastic modulus was computed and it was subsequently used to characterize each of the four material regions. The pulse pressure, as determined by the difference between the average systolic and diastolic pressures values, is used as the internal pressure acting normally to the inner surface of the aneurysm which is responsible for the internal stress as well as the inherent deformation of aortic wall. Using pulse pressure and regional maximum principal strain, the relative difference of elastic modulus between anatomic regions is calculated using [20, 21]:

$$Relative\ Modulus = \frac{Pulse\ Pressure(MPa)}{Experimental\ Strain} \quad (1)$$

However, in order to simulate an elastic modulus surrogate for our FE modeling, some normalization is needed. The relative modulus is normalized by a mean aortic linear elastic material property previously published as 500 KPa. Having this regional elastic modulus as a surrogate for four regions of AAA, along with Poisson's ratio (0.49), the regional material properties for each patient are imported into our finite element model to calculate the corresponding PWS.

We use two different material property models to calculate the wall stress distribution for each patient. The first model was a two-parameter hyper-elastic (Hyp model), isotropic, incompressible material behavior, adopting values from Raghavan and Vorp [22], which has been used as a standard model to study the behavior of AAAs for decades [5, 6]. The second model is the proposed regional material property model (four regional elastic moduli and Poisson's ratio of 0.49). The patient-specific pulse pressure was used as the loading condition for both models and was prescribed on the patient-specific 3D AAA model reconstructed from CTA images (Reg model).

2.7. Statistical analysis

Spearman's correlation is used for statistical analysis and the correlations with $p < 0.05$ are considered statistically significant. To compare the difference between two material models, Student t-test is used as well, and the comparison with $p < 0.05$ indicates that the two models differ significantly.

3. RESULTS

It has been shown that AAA severity, biomechanical behavior, and risk of rupture are dependent on both its overall geometry and biomechanical parameters, such as material properties. In our previous study, we showed how the variability of different geometrical parameters affect the wall stress distribution in AAA and how PWS may be a better criterion for assessing AAA severity. We now focus on the effects of mechanical behavior on the wall stress distribution, as well as PWS. We propose a regional material property model (Reg Model) and investigate its effect on PWS and risk of the rupture in compare with the standard hyper-elastic (Hyp model) model.

3.1. Parameter study: effect of Elastic modulus on PWS

To solely study how different material properties will affect the PWS, we used the reconstructed 3D geometric model of one of our patients as the reference AAA geometry, along with its patient-specific blood pressure. We then prescribed the regional material property configurations estimated for all nine patient datasets to this reference geometry. This approach enabled us to minimize the effect of the geometry and loading condition, and assess the changes in wall stress solely caused by the nine-different regional material property configurations. Table 1 provides four elastic moduli in 4 regions of the AAA for each patient, along with the mean elastic modulus and standard deviation, PWS for each patient with same AAA geometry.

In the parameter study, the mean and standard deviation of the regional elastic modulus shows a correlation of 0.67 ($p = 0.04$) and 0.71 ($p = 0.03$) with PWS, respectively. These results indicate that both mean elastic modulus and the standard deviation of the elastic moduli have a statistically significant positive correlation with the PWS. It also can be concluded that the diversity in between the 4 regions, which has been presented here by standard deviation, has a slightly stronger correlation with the PWS in comparison to the mean elastic modulus.

Figure 3a shows the change in PWS (for the same geometry) with change in mean elastic modulus (i.e., the mean of the four elastic moduli across the four material regions for each patient). Elastic modulus is an indicator of the material stiffness and a lower elastic modulus may lead to an increase in the rupture risk [21, 23]. However, the mean elastic modulus does not differ for different patients significantly, and is sufficiently close to 500 KPa, which is the standard elastic modulus in linear model which has traditionally been used to study AAA [6, 9].

Therefore, to have a better understanding of the effect of regional material properties on PWS, we compare the standard deviation of the regionally estimated stiffness for each

patient for the same reference AAA geometry (Figure 3b). The STDV presents a measurement of the stiffness discontinuity or variability in the AAA wall. As Figure 3b illustrates, higher stiffness variability correlates with higher wall stress.

3.2. Parameter study: effect of ILT on PWS

In our previous work [6], we performed a parameter study on the effect of the wall thickness on PWS, while all the other factors were unchanged. The results were in good agreement with previous studies [10, 17, 18], which had illustrated that the wall thickness has an inverse relation with PWS, meaning an increase in wall thickness causes a decrease in PWS. This behavior can also be explained by the principle of hoop stress (i.e., the circumferential stress in a vessel wall) under the first order assumption of a thin-walled vessel, given by the relationship: $\sigma_{hoop} = p r / t$, where p is the pressure in the vessel, r is the vessel mean radius, and t is its thickness. This assumption implies that the vessel wall is sufficiently thin, such that the inner and outer vessel radii are similar and can be described as a mean radius r .

Moreover, it has also been suggested that the ILT has an inverse relation with PWS [10, 14, 16]. From an engineering standpoint, based on the hoop stress principle law, ILT plays a role of increasing the wall thickness, which decreases wall stress. On the other hand, from a biological standpoint, ILT provides a barrier for oxygen transfer, recreating a hypoxia environment between thrombus and AAA wall [4], especially at the site of thicker ILT. This phenomenon increases inflammation of wall and wall weakening due to degradation of smooth muscle cells (SMCs) and extracellular matrix, which, in turn, increases the risk of rupture [24].

To study the effect of ILT in patient specific models, we used our model to compare the estimated wall stress and hence PWS predicted by both material models in this study – hyper-elastic (Hyp) and regional (Reg) material models both in absence of, and also considering the ILT. Table 2 shows a summary of the results.

In both models, the PWS is significantly higher in the absence of ILT (56.6 ± 17.6 % for the Hyp model and 58.2 ± 24.7 % for the Reg model). A Student t-test across all patients also showed a significant difference between the PWS in the absence and presence of ILT for both model ($p < 10E-6$ for both material models). It is also noted that the location of the PWS is different in the presence of ILT for all patients. Based on our results, in the case of no ILT, the PWS occurs closer to the location of D_{max} , while in the presence of the ILT, for both models, the PWS occurs closer the neck of the aneurysm.

3.3 Comparison between Hyper-elastic and Regional material model

We use the two different material properties model (hyper-elastic and proposed linear regional) for all patients to study how the regional material property is affecting the wall stress vs. the hyper-elastic model, which have been used as the standard material model to study AAAs. Table 3 shows a summary of patient specific characteristics including: maximum transverse diameter (D_{max}), loading pulse pressure, PWS for Hyp and Reg models, and Mean $\pm$ Std. Error., of the wall stress for each patient specific geometry as well as γ (is the ratio of the volume of AAA to the volume of ILT. It is used as an indicator for amount of ILT).

Figure 4 illustrates the wall stress distribution (characterized by mean $\pm$ std error) along with PWS for both linear regional material and the hyper-elastic models (blue and orange dots). As seen, both models follow the same trends for different patients, indicating the effect of AAA geometry of each specific patient on its PWS. Also, it can be seen that for all the patients the PWS for regional material model is higher; once again confirming the effect of more discontinuity on the PWS. Moreover, for all patients, the wall stress predicted by the regional material model is higher than the stress predicted by the hyper-elastic material model.

Our results show an average increase of $24.75 \pm 5\%$ in PWS from Hyp model to Reg model across the nine patients analyzed in this study. The location of PWS is same for 8 patients for both models. A t-test was performed for each patient to compare the wall stress distribution predicted by the two material properties model ($p < 1.43E-07$) and indicated that the wall stress distributions predicted by the two models are significantly different.

4. DISCUSSION

Traditionally, the maximum transverse diameter of AAA was used to monitor disease and assess severity and need for repair and/or surgery [5, 22]. However, in recent years, investigations have illustrated that a combination of the overall geometry of AAA along with the biomechanical parameters such as loading condition and material properties have to be considered for a comprehensive study of AAA. It was also shown that the PWS as a metric for wall stress distribution has a critical role in determining rupture risk of AAA.

A two-parameter hyper-elastic (Hyp model), isotropic, incompressible material behavior, adopting values from Raghavan and Vorp [22] has been used as a standard model to study the behavior of AAAs [5, 6]. Assumption for this model is a uniform material property throughout the entire aneurysm. However, growth of AAA is due to the metabolic activity of the metalloproteinase matrix at the aneurysm site which causes weakening of the wall [19, 25, 26]. This metabolic activity is influenced by different extent of inflammation and subsequent degeneration of the elastin and collagen within the connective tissue of the aortic wall, which, in turn, results in a remarkable structural deformation of aortic wall [16, 24]. On the other hand, all of AAAs suffer from different amounts of ILT which has been shown to have a remarkable impact on the wall degradation and its weakening [16, 27]. As a result of these biological activities, different parts of AAA will have different material properties such as different degree of stiffness and distensibility.

Therefore, to incorporate realistic material properties to study AAA, the regional material properties are needed. Several studies have previously been conducted to investigate the regional material properties, however, they were associated with several limitations as mentioned earlier, including an idealized AAA geometry on a small sample size patients, not considering effects of ILT using CT images featuring the AAA neck and D_{max} . Collected samples from patients who had an open surgery or repair, using PET/CT imaging for more accurate wall thickness prediction which is not a standard clinical method for AAA and cannot be applied on a large sample size of patients [7, 11, 12]. In this work, we used a subset of nine cases from a larger population of AAA patient datasets retrospectively

obtained from the Division of Vascular Surgery at the University of Rochester. The contrast enhanced CT images and blood pressure at the time of CT acquisition for each AAA are provided, hence enabling us to construct a patient-specific AAA model, while also considering the effect of ILT extracted from each individual CT image set. Through a non-invasive experiment, the principal strains were calculated for four regions of AAA (anterior, posterior, lateral and medial) [8]. In the present work, we used these reported strains and the patient-specific blood pressure, to calculate the elastic modulus of each region as a representation of the regional material property (Reg).

As mentioned, two material models are used to study the AAA's behavior: the standard hyper-elastic model and the proposed regional material properties model. Figure 5 shows the full statistics (mean, median, variance, inter-quartile range, and maximum) of the wall stress distribution for all nine patients of this study for both material models. This plot indicates that the PWS can be used as a more reliable tool to study the rupture risk since less than 5% of the nodes constitute the region deemed as exhibiting the peak wall stress.

This plot also indicates that for all nine patients the overall wall stress distribution as well as the PWS is higher in Reg model than the Hype model, implying for a need for more accurate material property model to study the AAA more precisely.

Our results, also, illustrate that while a Regional material property model presents higher PWS for all patients, the location of the PWS did not change between the two models. For 7 patients, the location remains the same. For one patient the location changed entirely from anterior of aneurysm in Hyp model to the medial wall of aneurysm for Reg model. For the other patient, the PWS happens on the anterior wall, but close to the neck of AAA for Reg model and near the end of the AAA for Hyp model. However, for all patients except one, the PWS happens next to the neck of AAA on the thinner side of the aortic wall with less amount of the ILT. And only for one patient PWS happens at the D_{max} location with less amount of ILT as well.

It was concluded that for a set of four elastic moduli, the more diverse the four moduli, the higher the risk of the rupture. The mean elastic modulus has a correlation of 0.67 ($p = 0.04$) and the standard deviation of the 4 moduli for 4 regions shows a correlation of 0.71 ($p = 0.03$) with PWS while all the other parameters were kept fixed which both indicate that the PWS has a direct relation with the material properties of AAA.

Figure 6 shows the stress distribution as well as the location for PWS for Patient # 4 in our dataset. Figure 6a, b show the wall stress distribution in the absence of ILT, for Hyp model and Reg model, respectively. In this case the PWS happens on the anterior region of AAA for both model towards the distal end of AAA, and it is higher in Reg model.

Figure 6c–f illustrate the wall stress distributions for the same patient in the presence of the ILT. Figure 6c shows the wall stress distribution on the outer wall of AAA for the Hyp model while Figure 6d demonstrates the corresponding wall stress distribution on the lumen wall of AAA for Hyp model. Similarly, Figure 6e, f show the wall stress distribution for Reg model on the outer and lumen wall of AAA, respectively. It can be seen that in the presence of ILT the PWS happens on the lumen wall for both model. Also, it locates on the posterior

region of AAA towards the neck of the aneurysm, and the PWS is higher in the Reg model as well.

The location of PWS changes in the absence of ILT compare to the presence of ILT for both material models. It also illustrates that PWS is higher in Reg model in compare with Hyp model in both cases. Moreover, the PWS is higher in the absence of the ILT. These results indicate that in order to have a comprehensive model to monitor AAAs behavior more appropriately, a model which incorporates both regional material properties and ILT is required.

Additionally, in both models, the PWS is significantly higher in the absence of ILT (56.6 ± 17.6 % for the Hyp model and 58.2 ± 24.7 % for the Reg model) than in the presence of ILT. Significant differences are observed between the overall wall stress distribution in the absence and presence of ILT for both model ($p < 10E-6$ for both material model). It is also noted that the location of the PWS is different in the presence of ILT for all patients. Based on our results, in the case of no ILT, PWS is happening more towards the location of D_{max} , while in the presence of the ILT, for both models, the PWS is located closer toward the neck of the aneurysm.

Figure 7 shows a summary of the regional stiffness across the four material regions for each of the nine patient datasets, along with the peak wall stress value and the region in which it occurs. Similarly, Table 3 shows a summary of patient specific data along with PWS for both models, and suggests that several different parameters are affecting PWS, as well as the rupture risk. For example, a comparison between Patient # 1 featuring a smaller D_{max} than Patients # 4 and # 5 featuring a larger D_{max} , shows that Patient # 1 has a higher PWS which could be a result of higher loading condition (blood pressure), a smaller γ (is the ratio of the volume of AAA to the volume of ILT. It is used as an indicator for amount of ILT). This result indicates that to have more accurate tool to investigate the AAAs behavior as well as estimating their rupture risk and need for repair, the patient specific data including specific geometry, loading condition, and biomechanical properties are necessary.

4.1 Finite Element Analysis Rupture Index (FEARI), comparison between Hyp and Reg model:

To better emphasize how the material property affects the AAA's behavior and its rupture risk, here we use a previously proven rupture index [2]. The Finite Element Analysis Rupture Index (FEARI) is an indicator of the rupture risk estimated based on the failure definition of the material, when the stress overcomes the strength of the material. A value close to 0 indicates a low risk of rupture, while a value close to 1 suggests a high risk of the rupture.

$$FEARI = \frac{FEA \text{ wall stress (PWS)}}{Experimental \text{ wall strength}} \quad (2)$$

The wall strength values from [2] are used. In their experiment, they performed a uniaxial test on samples from 149 AAAs to measure the average wall strength in four regions

(anterior: 0.7744 MPa, posterior: 0.8658 MPa, lateral: 0.9221 MPa, Medial: 0.9187MPa [2, 7]. Using the reported wall strength and the PWS predicted by our FE model, we calculated the FEARI for each patient. Table 4 shows that the FEARI index is higher for all patients using the regional material property model.

5. CONCLUSION

There are few limitations associated with this study. First the surrogate stiffness and elastic modulus estimate based on the principal strains estimated from ultrasound elastography images using the amplitude of the pressure wave as the stress is a simple, first order approximation, which will be subsequently refined using the theory of thin or even thick-walled pressure vessels. Moreover, an even more accurate approach would be to use an inverse problem formulation to reconstruct the vessel material properties, which is beyond the scope of our work at this stage. Moreover, the characterization of the entire AAA according to four distinct regions featuring a sharp transition may be somewhat unrealistic. As such a smoothing algorithm will be implemented to ensure less drastic transitions between adjacent material regions, which is most likely the case in actual vessel walls, despite the development of disease.

To accurately model and study AAA behavior, it is necessary to consider the effect of regional material properties and overall geometry of AAA. In the present study, a model incorporating the patient-specific regional material properties along with patient-specific AAA geometry and loading condition was presented. Our results indicate that the use of regional material properties lead to higher predicted peak wall stresses than those predicted by the homogeneous hyper-elastic material models, therefore also suggesting a higher risk of rupture, assuming wall strength remains unchanged. Moreover, we also showed that besides the overall geometry, consideration of ILT is also essential and it impacts both the predicted PWS, as well as the FEARI index, but the overall location of the peak wall stress and hence the potential rupture location does not change.

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Fig 1:

Example of (a) an axial CT slice of human AAA, and the 3D model generation of AAA (b) Segmented contours of the aortic lumen (blue) and outer (red) walls from CT images, (c) Smoothed 3D mesh of the outer wall, and (d) Distribution of the peak wall stress (PWS) for right posterior of (c) while black arrow shows the location of PWS at the inflection point.

Fig. 2:

(a) Cross section of AAA with black arrow 1 showing outer wall, black arrow 2 showing inner wall which is separated from the outer wall with the white line, 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. (c) B-mode ultrasound images which shows four arbitrarily defined anatomy regions of the segmented vessel (adopted from[3]).

Fig. 3:

(a) Correlation of PWS with mean stiffness across four material regions for all nine patient datasets mapped onto the same reference patient geometry, (b) PWS vs. mean stiffness across the four material regions for each patient.

Fig. 4: Wall stress distribution predicted by the hyper-elastic material model (orange) and regional material model (blue). The plot shows wall stress Mean and Standard Error (bar plots) and also the peak wall stress (corresponding dots above each bar plot).

Fig. 5:

Box and whiskers plots illustrating the wall stress distribution predicted by the Hyp (red) and Reg (blue) model. The plot shows the inter-quartile range, mean (X), median, and also the higher end of the wall stress, in the vicinity of the peak wall stress (the highest value shown on each individual plot).

Fig 6.

Anterior view of the AAA showing the wall stress distribution predicted by the Hyp (a) and Reg (b) model in absence of ILT. Wall stress distribution predicted by the Hyp outer wall (c), Hyp lumen wall (d) and Reg outer wall (e), Reg lumen wall (f) model on the posterior in presence of ILT. Black arrows show the location of PWS in each case. Note that wall stress distributions are scaled based on the Reg model for both with and without ILT)

Fig. 7:

Elastic moduli for 4 regions for each patient along with their PWS based on the patients' specific data. Black × shows the PWS for each patient and black symbol (purposely overlapping with the stiffness value of one of the four regions) indicates the region in which the PWS occurs for each patient.

Table 1:

Summary of results: effects of Elastic modulus on PWS

Patient	Elastic Modulus (KPa)				Mean Elastic Modulus across Four Wall Regions $\pm$ Std. Dev. of Elastic Modulus of Four Wall Regions (KPa)	PWS (KPa)
	Posterior	Anterior	Medial	Lateral		
1	475	487	566	470	494 $\pm$ 45	60.3
2	562	735	363	340	450 $\pm$ 186	59.6
3	490	573	372	565	495 $\pm$ 94	56.8
4	471	385	452	691	479 $\pm$ 133	60.8
5	394	656	533	415	500 $\pm$ 121	63.6
6	610	482	548	360	500 $\pm$ 106	59.9
7	925	303	370	402	414 $\pm$ 286	72.4
8	450	492	681	375	475 $\pm$ 130	69.5
9	515	568	448	468	495 $\pm$ 53	57.8

Table 2:

Model-predicted wall stress distribution using Hyper-elastic (Hyp) and regional (Reg) material models with and without considering the presence of ILT.

Patient	Hyp Model-predicted PWS without ILT (KPa)	Hyp Model-predicted PWS with ILT (KPa)	Reg Model-predicted PWS without ILT (KPa)	Reg Model-predicted PWS with ILT (KPa)
1	171	58.3	207	70.6
2	238	68.6	343	75.0
3	129	86.0	213	107.0
4	170	49.5	255	60.8
5	172	38.8	264	48.3
6	195	86.6	233	119.1
7	123	82.3	227	115.0
8	186	53.5	255	72.2
9	166	104.4	196	200.0

Table 3:

Summary of Specific Patient Data

Patient	D_{max} (cm)	Pulse Blood Pressure (mmHg)	γ^*	Hyper-elastic PWS (KPa)	Linear Regional PWS (KPa)	Wall Stress (KPa) Mean $\pm$ Std. Error Hyp Model	Wall Stress (KPa) Mean $\pm$ Std. Error Reg Model
1	4.96	73	0.50	58.3	70.6	29.1 $\pm$ 2.7	35.3 $\pm$ 2.4
2	7.02	64	0.63	68.6	75	34.5 $\pm$ 1.5	37.5 $\pm$ 1.6
3	7.67	42	0.33	86	107	43.7 $\pm$ 1.5	53.6 $\pm$ 1.8
4	5.82	60	0.63	49.5	60.8	24.8 $\pm$ 2.2	30.4 $\pm$ 2.7
5	5.71	56	0.52	38.8	48.3	19.7 $\pm$ 0	24.7 $\pm$ 0
6	5.41	59	0.42	86.6	119.1	43.4 $\pm$ 0.4	59.6 $\pm$ 0.3
7	5.06	63	0.61	82.3	115	41.4 $\pm$ 2.5	57.3 $\pm$ 3.5
8	6.34	63	0.57	53.5	72.2	26.8 $\pm$ 2.1	36.1 $\pm$ 0.9
9	6.61	76	0.53	104.4	200	52.2 $\pm$ 2.7	100.1 $\pm$ 6.8

Table 4:

Calculation of FEARI

Patient	Region of AAA which PWS happens	PWS Hyp model (KPa)	PWS Reg model (KPa)	FEARI Hyp model	FEARI Reg model
1	Medial	58.3	70.6	0.063	0.077
2	Lateral	68.6	75	0.074	0.081
3	Posterior	86	107	0.100	0.124
4	Posterior	49.5	60.8	0.057	0.070
5	Medial	38.8	48.3	0.042	0.053
6	Lateral	86.6	119.1	0.094	0.129
7	Anterior	82.3	115	0.106	0.149
8	Posterior	53.5	72.2	0.062	0.083
9	Posterior	104.4	200	0.121	0.231