Wall Stress Prediction of Abdominal Aortic Aneurysm: Influence of Geometry and Curve-Fitting Experimental Data

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Abstract: Biomechanics plays an important role for predicting risk of rupture in abdominal aortic aneurysm (AAA) as the peak wall stress is believed to be the most reliable and easy-to-use predictor. Many AAAs has asymmetric geometry which can be caused by the anterior bulging with posterior expansion limited by the vertebral column. Moreover, AAA is very often characterised mechanically using uniaxial testing which provides simplicity. The aim of this study was to investigate the influence of geometry and material parameters obtained from curve-fitting of experimental data of uniaxial test on the magnitude of peak wall stress in AAA. Three dimensional computer models of AAA in symmetric and asymmetric forms were generated in which the maximum diameter and length of aneurysm were determined based on the observation of AAA patients reported in the literature. For comparison, normal abdominal aortic vessel was also built. Here, the isotropic hyperelastic model proposed by Ogden was used and fitted to the experimental data obtained from the uniaxial testing of the circumferentially and longitudinally oriented specimens. The value of blood pressure was varied from 80 to 140 mmHg. Furthermore, the peak AAA wall stress was then computed using FEBio. The results showed that asymmetric form of AAA led to higher peak wall stress and possessed different stress distribution compared to that of AAA with symmetric form. The input of material parameters also affected the magnitude of peak wall stress. The results also confirmed that higher blood pressure could bring to higher rupture risk as the peak wall stress increasing. Care should be taken when simulating peak wall stress since some factors including geometry and magnitude of material parameters can affect the results significantly.
Keywords: AAA, Wall Stress, Nonlinear Finite Element, Hyperelastic Model, FEBIO

1. Introduction

Cardiovascular diseases (CVD) are the number one cause of death in worldwide, especially in the developed countries. One of CVD that frequently occur in the abdominal aorta is abdominal aortic aneurysm (AAA) which is a progressive enlargement of the abdominal aorta and is able to suddenly rupture leading to mortality. Smoking, advanced age, male gender, chronic obstructive pulmonary disease, hypertension, and a family history of aneurysmal disease are believed to be the primary risk factors that cause the formation of an aortic aneurysm (Blanchard et al., 2000). In Europe and US, the total number of cases of AAA reaches approximately 10% among those over the age of 65 (Sakalihasan et al., 2005) and has become the 13th most common cause of disease related death (Sanfellipo, 2003).

In current clinical practices, rupture risk of AAA is assessed by aneurysm diameter. Typically, when a threshold diameter has been attained which is about 5 to 5.5 cm (Humphrey, 2002; Upchurch and Schaub, 2006; Vorp, 2007), a surgical intervention can be made. However, rupture assessment based on the maximum diameter of aneurysm is not a reliable predictor for AAA rupture since there are some AAA patients who experienced rupture earlier even though the diameters have not achieved the threshold whereas some AAA cases with maximum diameter over than the threshold did not undergo rupture. Therefore, it is believe that factors

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other than size should be considered. There are some parameters have been proposed relating to AAA rupture predictor include the AAA expansion rate, wall stiffness, increase of intraluminal thrombus (ILT) thickness, wall tension, and peak AAA wall stress where the peak AAA wall stress has been becoming the most promising predictor (Vorp, 2007). A computational method using Finite Element Analysis (FEA) has been extensively employed to determine the distribution and magnitude of AAA wall stress. Furthermore, recent study conducted by Doyle et al. (2009) has presented a simple calculation to predict the rupture of AAA, known as finite element analysis rupture index (FEARI) which combines the FEA-computed peak wall stress with the wall strength of the specific region obtained by mechanical testing. The calculation of FEARI can be done by simply dividing the peak AAA wall stress with the wall strength at that region which are related to its maximum local stress and local strength.

In FEA, geometry plays a significant role to produce a detail stress analysis. In some studies, AAA is often modeled with a symmetrical fusiform. However, many AAA cases appear to be asymmetric, which can be caused by the anterior bulging with posterior expansion limited by the vertebral column. Although Vorp et al. (1998) has already investigated the effect of geometry in peak AAA wall stress, our study used different approaches to investigate the geometry effect by employing 3-D element with eight nodes and different constitutive model which has never been reported in any literature.

Even though AAA shows anisotropic behaviour, many researchers still employ uniaxial tension to mechanically characterise the mechanical behaviour of aorta. This can be understood as the uniaxial test provides a simplicity and the procedure can be done until the specimen fails which is still being a difficult task to do in biaxial testing. Furthermore, experimental data from the uniaxial data is very often fitted with isotropic hyperelastic model and thus, it can lead to confusion when the uniaxial testing data of AAA fitted to that model as the material behaves differently in circumferential and longitudinal directions.

The aim of this study is to determine how the geometry and material parameters obtained from curve-fitting of experimental data of uniaxial test influence the magnitude of peak wall stress in AAA. For comparison, a normal abdominal aortic vessel was also modelled. In addition, a higher blood pressure in AAA cases was also simulated related to higher rupture risk in a patient with hypertension.

2. Methods

Three-dimensional computer models of symmetric and asymmetric forms of AAA were generated using the commercial CAD software Catia V5 (Dassault Systèmes). The overall length and the diameter of AAA were 120 mm and 60 mm, respectively. Cross section at any axial position was idealised as in circular shape. Here, the intraluminal thrombus was not incorporated in the model. Since the wall thickness was not investigated in this study, the thickness was assumed to be uniform at 1.5 mm. Furthermore, normal Gaussian distribution (Bell curve) was used as the profiles of anterior and posterior of the AAA models. This shape of AAA was also used in the study by Vorp et al. (1998). Those AAA models are shown in fig. 1.

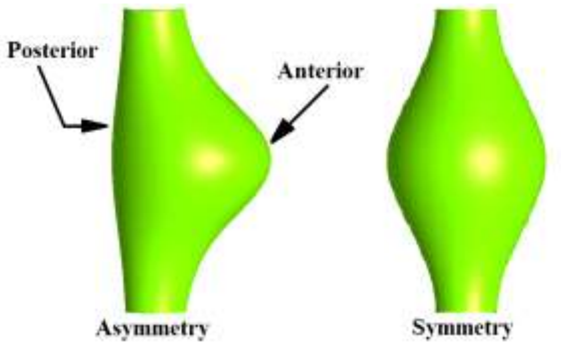

Figure 1. 3-D models of asymmetric (left) and symmetric (right) AAAs.

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Latter, the three models were imported to the pre-processor software Preview. In this study, all the FE simulations were performed using the finite element package FEBio, open source FE software (Maas et al., 2012). The models were then discretised using the brick element with eight nodes. The number of elements for symmetric AAA, asymmetric AAA and normal AA models was 13568, 13674 and 8160, respectively. Since the blood pressure within AAA acts on the inner wall of AAA, therefore, a peak systolic pressure load was applied uniformly to the inner surface of the models. Here, the peak systolic pressure load was varied from 80 to 140 mmHg (10 to 18 KPa). The shear stress induced by the blood flow was considered in this study as the influence is very small compared to the stress caused by the wall expansion [Inzoli et al. (1993), Thubrikar et al. (2001)]. All the models were fully constrained in the proximal and distal regions related to the attachment of abdominal aorta to the aorta at renal junction and iliac bifurcation. Also, residual stress was not considered, even though it may exist within the aortic wall *in vivo*.

Most studies show that the passive mechanical behaviour of arterial wall shows nonlinearity, anisotropy and incompressibility [for an overview see, e.g. Kalita and Schaefer (2008)]. However, some investigators such as Fillinger et al. (2003) and Doyle et al. (2009) still employed the isotropic constitutive model to simulate the AAA wall stress. In this study, we also used the isotropic hyperelastic model proposed by Ogden to model the behaviour of AAA and normal AA tissue which was aimed to investigate the reliability of the model in relation with the experimental data to simulate the AAA wall stress. The constitutive model is written as follows

$$W = \frac{2\mu}{\alpha}\left(\lambda_1^{\alpha} + \lambda_2^{\alpha} + \lambda_3^{\alpha} - 3\right) + \frac{K}{2}\left(\ln J\right)$$

$$J = \lambda_1 \lambda_2 \lambda_3$$

where μ is the shear modulus, α is a stiffening parameter (dimensionless) and K is bulk modulus. The first term of the equation is related to the isochoric elastic response and the second term is related to the volumetric elastic response. In the case of uniaxial tension, the deformation can be described as follows

$$\lambda_1 = \lambda ; \quad \lambda_2 = \frac{1}{\sqrt{\lambda}} ; \quad \lambda_3 = \frac{1}{\sqrt{\lambda}}$$

where λ is the uniaxial stretch. Thus, the general form of principal stresses can be calculated as follows

$$\sigma_i = \lambda_i \frac{\partial W}{\lambda_i} - p$$

where p is the hydrostatic pressure and can be calculated as the derivation of the volumetric part of the constitutive model. Latter, the model was fitted to the experimental data obtained from the study by Raghavan et al. (1996). The objective function of the residual between the experimental and the analytical values was then minimised to obtain the material parameters using custom Matlab code (Mathworks, Natick, MA). Here, *fsolve* was used and the residual function was defined as follows

$$residual = \sum_i \left[\sigma_{\exp}(i) - \sigma_{model}(i)\right]$$

where σ_{exp} and σ_{model} are the stresses obtained by the experiment and model, respectively. Table 1 shows the material parameters obtained from curve fitting for different experimental data from Raghavan et al (1996).

Table 1. The magnitudes of the material parameters for different experimental data.

	AAA Longitudinal	AAA Circumferential	Normal Longitudinal
μ (MPa)	0.554	0.314	0.207
α	9.918	10.266	9.133

To confirm that those material parameters were fit correctly and that the constitutive equations were properly input into the FEBio model, the uniaxial tension experiment was simulated in FEBio using those material constants. Fig. 2 shows a good agreement between the analytical modelling and the FE simulations for those magnitudes of material parameters.

Typically, hybrid element is employed to enforce incompressibilty in a material as the bulk modulus is infinite for a fully incompressible material which leads to infinite stiffness matrix in the standard FE formulation (Bower, 2008). Since hybrid element is not available in FEBio, a condition of nearly incompressible was applied to the FE modelling by inserting a large value of bulk modulus. To check the influence of the compressibilty factor, the value of K was varied in the simulations of uniaxial tension. Fig. 3 shows the influence of the magnitudes of K towards the stress value. It can be seen that the magnitudes of the stress for $K = 10^5$ MPa to 10^3 were very close, and thus the bulk modulus of 10^3 MPa was chosen to be input for the AAAs modelling.

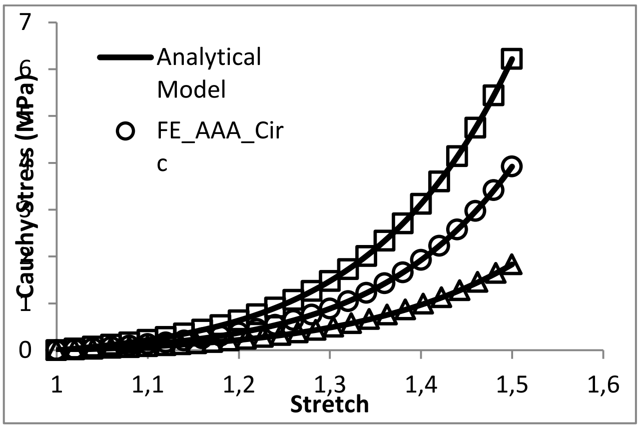

Figure 2. Comparison of analytical model and FE simulation of uniaxial testing for different experimental data.

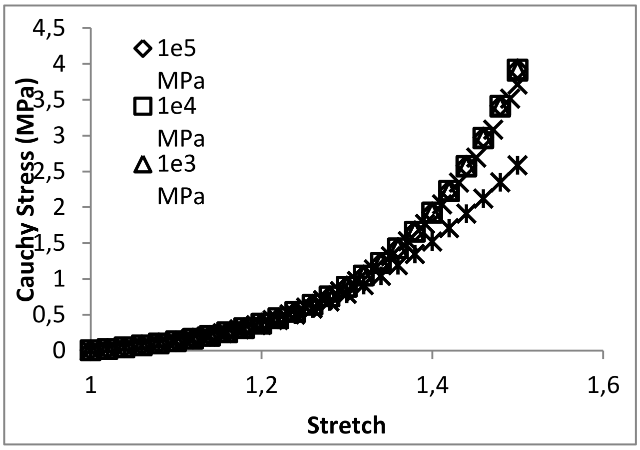

Figure 3. Influence of various magnitudes of bulk modulus in unaxial testing.

3. Results and Discussions

In the AAA simulations, the quantity of von Mises stress was utilised as the von Mises stress is more suitable failure analysis and is a function of the three principal stresses in the AAA model. The three-dimensional stress contour of asymmetric AAA model for various peak systolic pressures is shown in fig. 4 where the material parameters of circumferential sample were used. As the peak systolic increases, the peak mechanical wall stress also increases. In asymmetric model, the peak wall stress was developed at the midsection on the posterior surface and at the inflection points on the anterior surface. On the other hand, the symmetric model showed the peak wall stress was more uniformly distributed at the inflection

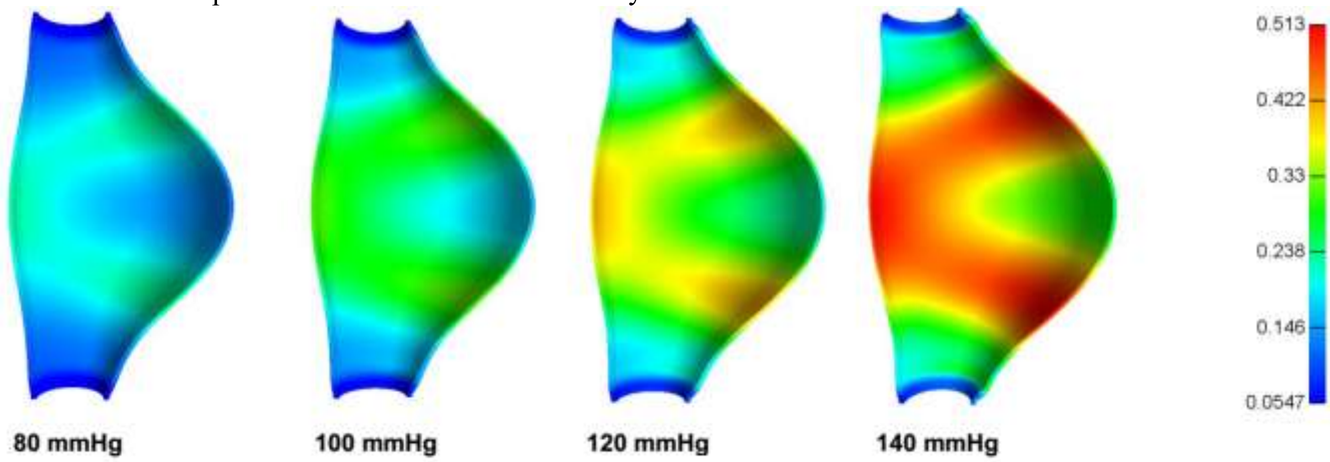

Figure 4. The simulated AAA wall stress of asymmetric model for various peak systolic pressures.

points on the posterior and anterior surfaces as shown in fig. 5 where the model was simulated for various systolic peak pressure and the material parameters of circumferential sample were employed. It was also observed that the asymmetric model possessed higher magnitude of peak wall stress compared to the symmetric one.

For comparison with a diseased state of abdominal aorta, a normal aorta model was also simulated. Fig. 6 shows the normal abdominal aorta subjected to various peak systolic pressures and we found that the AAA could lead the AA to have a high potential rupture as the peak wall stress in AAA increased significantly. Also, we found that the pressure-diameter relationship of normal AA at the peak systolic pressure of 80 mmHg to 140 mmHg was still linear whereas the symmetric and asymmetric models were discovered to be non-linear (fig. 7). It was also observed that the symmetric form was more distensible than the asymmetric one as the asymmetric showed a stiffening effect as the diameter increased.

Furthermore, table 1 shows the magnitude of the simulated peak wall stress for various material parameters magnitudes. The average stands for the averaged value of the material parameters from the circumferential and longitudinal data, which was also simulated to the models. This was aimed to investigate the influence of the input of material parameters obtained from curve fitting as the isotropic constitutive model does not considered different behaviour when a material is stretched in different directions. All the simulations on table 1 were conducted in peak systolic pressure of 120 mmHg. We found that the peak wall stress could vary depend on which the experimental data employed.

AAA rupture is a sophisticated process which is involved biological, biomechanical and biochemical processes. However, biomechanical process is still poorly understood, even though many studies have more concerned on it. In general, the AAA will undergo rupture when the peak wall stress exceeds the wall strength. Latter, investigation of *in vivo* wall stress distribution in AAA may lead to more accurate clinical estimations in order to predict risk of

AAA rupture and thus it can help clinician to determine the decision for elective repair.

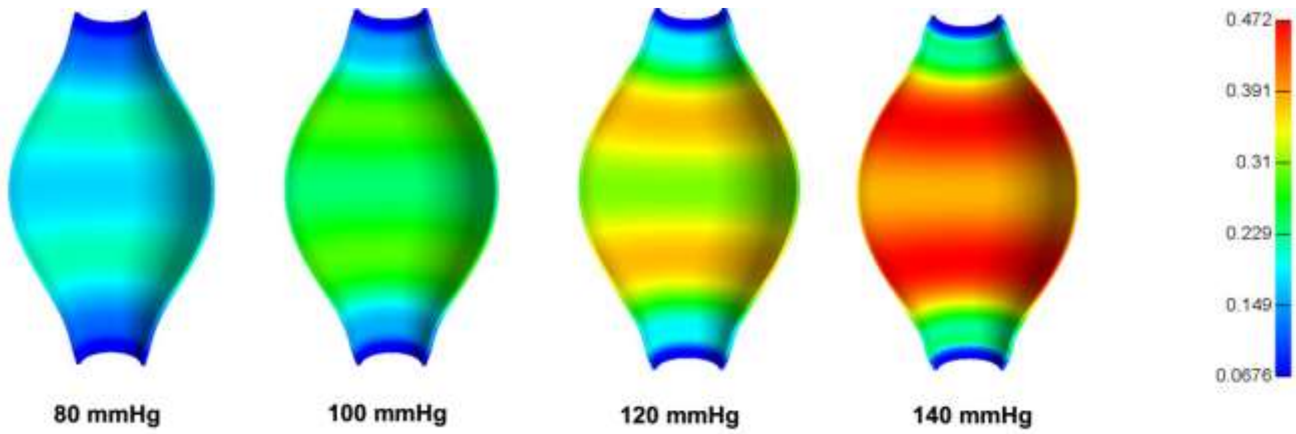

Figure 5. The simulated AAA wall stress of symmetric model for various peak systolic pressures.

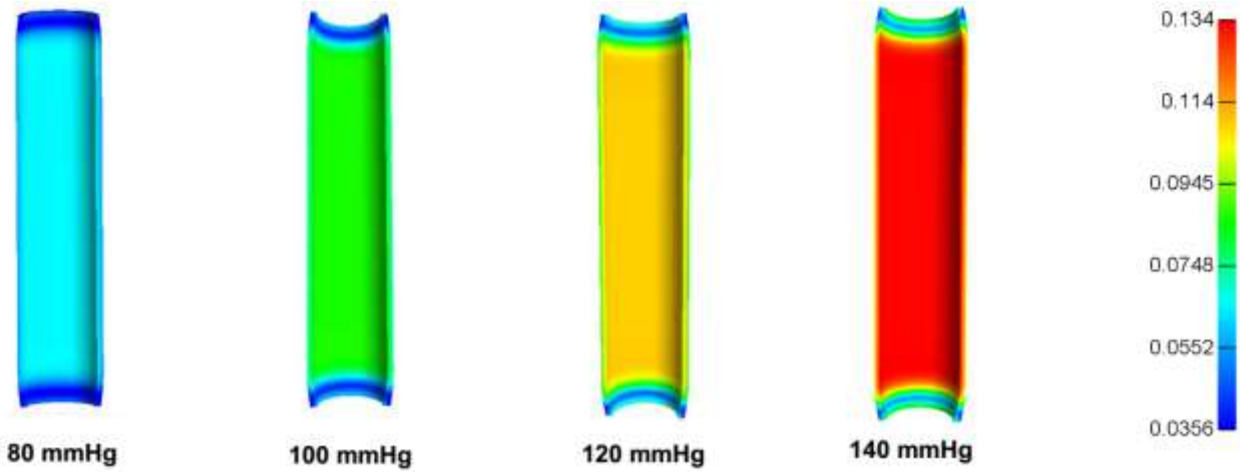

Figure 6. The simulated mechanical wall stress of normal AA for various peak systolic pressures.

Here, we have examined the influences of the geometry of AAA which could change the magnitude and the distribution of the peak wall stress which is a good agreement with a study of Vorp et al. (1998) who have been the first to investigate the geometry effect on AAA. They also found that the diameter is also found to play an important role in increasing peak wall stress. In recent study by Rodriguez et al. (2008) investigated the influence of diameter, asymmetry and material anisotropy in AAA model. They suggested that diameter is not the only geometric parameter that influence the severity and rupture risk of an AAA, but also the asymmetry should be considered in a rupture risk decision criterion. In addition, they also found that the anisotropy can scale up the stress magnitude and give different distribution and stress gradient which was not considered in our study.

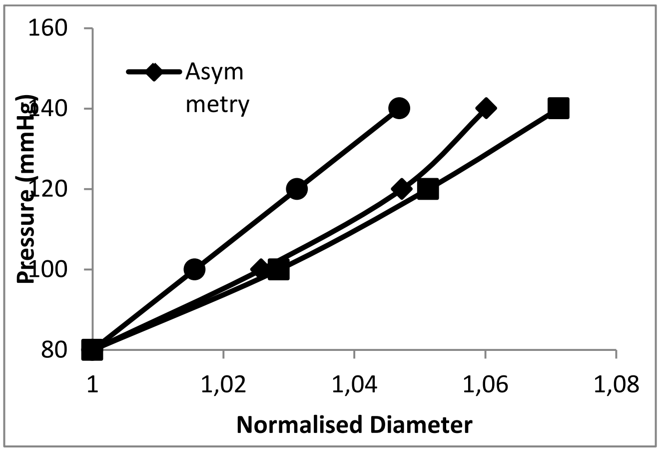

Figure 7. Pressure-diameter curve for asymmetry, symmetry of AAA and normal AA models.

Further, our study also revealed that the geometry effect can influence the distensibility of the AAA wall (fig. 7). It was also demonstrated that the peak wall stress increases as the blood pressure increases which relates to higher risk rupture for an AAA patient with hypertension. Experimental study carried out by Gadowski et al (1993) has demonstrated that hypertension can increase the growth rate of AAA. However, a finding by Raghavan et al. (2000) showed that a hypertension condition is not always related to higher rupture risk of AAA. They compared a patient with normal blood pressure and a patient with hypertension that had the same diameter of AAA and demonstrated that the normal one had higher rupture risk than the one with hypertension. They suggested that the shape effect could be the factor influencing the rupture risk.

The selection of experimental data could also influence the simulated peak wall stress when the uniaxial data is fitted to an isotropic constitutive model. In this case the stress distribution remains the same but only the magnitude which vary. Thus, we suggested that isotropic constitutive model is not reliable to be used for AAA simulation when the response shows anisotropic behaviour as it may lead to inaccurate prediction.

Table 1. The magnitude of peak wall stress (MPa) for various magnitudes of the material parameters.

	Asymmetric model	Symmetric model
Circumferential	0.413	0.379
Longitudinal	0.337	0.306
Average	0.365	0.334

It is also noteworthy to notice some limitations in our study. All the geometry were idealised as the real aorta can have a non-smooth surface which can lead to different stress gradient and distribution. Thus, many researchers have employed an advanced technique of image processing obtained from CT scan to provide a patient specific geometry. Latter, the ILT was not considered in our study whereas the ILT can reduce the peak wall stress in AAA (Mower et al. 1997).

As for future work, we plan to incorporate anisotropy in the model and employ several anisotropic constitutive models since it may yield to different responses where a study of Polzer et al. (2013) and Di Achille et al. (2011) has shown that the selection of constitutive model is very crucial in predicting AAA response even though they have been fitted to the same datasets. The effect of axial pre-stretch will be considered as well.

4. Conclusions

The influence of geometry and curve-fitting experimental data has been examined. The three dimensional model of AAAs were generated and implemented in the finite element software. The results revealed that the geometry can affect the stress distribution and magnitude of AAA wall stress. The input of material parameters can also influence the magnitude of wall stress. Since the datasets have been fitted to the isotropic constitutive model, it can lead to different material input as the tissue behaves anisotropically. We suggest that an anisotropic model would provide a more accurate stress prediction. Therefore, care should be taken when simulating peak wall stress since some factors including geometry and magnitude of material parameters can affect the results significantly.

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