A computational wall mechanics study of an ascending thoracic aortic aneurysm under hypertensive conditions

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Abstract. A wall mechanics study is performed on two human aortic models, reconstructed from computed tomography (CT) image data using the Materialise Mimics software. The first model represents a rare ascending thoracic aortic aneurysm (aTAA) case with an excessive aortic ballooning that has displaced the cardiac cavities, and the second one a normal case free of cardiovascular diseases. Special attention was paid for the reconstruction of realistic models that do not deviate from the original data. The study presents the workflow from medical imaging data to structural simulation with the use of various software, aiming to examine the stress state of a normal aorta and an aneurysmal one (both patient specific) under a range of systolic blood pressure loads. Using the FEBio software, the effective (Lagrange) strain and the effective stress (von Mises) distributions are calculated for assessing the risk of rupture or dissection of the aorta.

Keywords: aorta wall mechanics; ascending aortic aneurysm; hypertension; aorta rupture risk

1. Introduction

Ascending thoracic aortic aneurysms (aTAs) are defined as the pathological bulging of the proximal aortic section where blood ejects through the arterial valve from the left ventricle of the heart. The dilatation of the aortic wall is a consequence of an irreversible gradual weakening caused by a degenerative process. \cite{1} aTAA is a silent, though fatal, disease that can lead to the rupture or dissection of the arterial wall. \cite{2} Most patients do not experience symptoms. It is even more striking that half of the symptomatic cases, emergently presenting at a hospital, had an undiagnosed aTAA according to post-mortem studies.\cite{3} The estimation of the aTAA rupture or dissection risk traditionally depends on computed tomography (CT) or ultrasound (US)-based measurements of maximum diameter and growth rate. As the aortic size
does not account for the regional vessel wall heterogeneity, which is evident especially in aneurysmal segments, the diameter-based complication risk prediction is possibly underinformed thus less trustworthy. There are strong advocates of wall stress and strength as more reliable risk predictors than aTAA diameter and growth rate. [4] As it is impossible to measure wall stresses in vivo, researchers have turned to the computational approach for a solution. The Finite Element Analysis (FEA) is a validated and widely used technique but, still, the calculation of arterial wall stresses requires prior knowledge of the aortic tissue mechanical behaviour. The review by Martufi et al. [1] delineates how a multidisciplinary approach involving biomechanics and medical imaging can provide accurate predictions of aTAA behaviour, based on structural and fluid mechanics studies.

The most common application of biomechanical modelling is the evaluation of wall stresses in abdominal aortic aneurysms (AAAs), while literature is limited in aTAAs. Recently, Wang et al. [5] determined the wall stress distribution in aTAAs with diameter <5.0 cm and >5.0 cm, which is the threshold for elective surgical repair according to the medical guidelines. The patient-specific aneurysmal geometries were obtained from echocardiogram-gated CT scans and the structural response was determined by the LS-DYNA FEA software (LSTC Inc, Livermore, California, USA). The peak wall stress was found greater in aTAAs > 5.0 cm but the correlation of size and stress in aTAAs < 5.0 cm was poor.

A considerable percentage of aTAAs patients (15%) also suffer from Bicuspid Arterial Valve (BAV), which is a common congenital aortic valve defect [6]. The aortic size as an index ignores other comorbidities, such as BAV, that might play a role in the risk of aTAA rupture or dissection. In a biomechanics study of 17 BAV patients and 19 subjects with a physiological Tricuspid Aortic Valve (TAV), the circumferential and longitudinal stresses were found greater in BAV- than TAV- aTAAs, suggesting that patient-specific aneurysm wall stress analysis is essential for accurate complication risk prediction. [7] Besides BAV, in plenty of cases, a large brachiocephalic trunk gives rise to both subclavian arteries and a bicaudal trunk, termed as the Bovine Aortic Arch (BAA) congenital variation. Comparing aTAA patients with BAV, BAA and none of the two, Martin et al. [8] concluded that the aortic size index was insufficient for delineating between patients at moderate and high risk, while no correlation existed between BAV and BAA aTAA rupture risk. The study also involved tissue mechanical testing data thus combining experiments with computational simulations. Avanzini et al. [9] provide a comprehensive analysis of the experimental methodologies for the determination of aTAAs biomechanical properties.

The current pilot study is an application of computational biomechanics in a rare aTAA case with an excessively developed aneurysmal bulge that has displaced the cardiac cavities. The modelling procedure is thoroughly described and the response of the aTAA is studied under various pressure conditions in comparison to a normal case that is free of cardiovascular diseases. A presurgical biomechanical analysis, especially in difficult cases, can enhance the clinical decision-making process.

2. Materials and methods

2.1. Material

The biomechanical analysis relies on the computed tomography (CT) scans of a 67 y.o. male patient and a 69 y.o. normal male subject who gave informed written consent for the use of their medical imaging data. The patient suffered from aTAA and underwent a successful aortic repair at “Hippocration” General Hospital of Athens in 2013. The CT data was made available for further processing by the Attending Physician (cardiac surgeon) at the Cardiac Surgery Department of “Hippocration” General Hospital of Athens.

2.2. CT imaging protocol

The CT scan was performed on a 16-slice multidetector CT (MDCT) scanner (Activion 16, Model TSX-031A, Toshiba). A protocol of 1 mm slice thickness was used with 120 kV, 60 mA tube current, 330 mm field of view for reconstruction and a sharp reconstruction kernel. The CT was performed in the craniocaudal direction, from the aortic arch until the diaphragm level in a single breath-hold. Fig. 1
shows the cross-section in the CT data featuring the maximum aTAA diameter and Fig. 2 the 3D volume rendering of the whole CT series, clearly disclosing the big aTAA bulge.

**Figure 1.** CT image annotated with the maximum aTAA diameter value.

**Figure 2.** 3D volume rendering of CT data featuring the big aneurysmal bulge.
2.3. Segmentation & Reconstruction

The Mimics software (Materialise inc.) is an advanced software with sophisticated tools for the segmentation and reconstruction of medical imaging data. The aorta was initially isolated from the surrounding bones and organs using the “Dynamic region growing” tool that is based on the gray value connectivity. The aorta was further restricted to the desired limits (the aortic arch upwards and the descending aorta downwards) by batch-removing the out-of-bounds segmented cross-sections. The segmentation of the aorta was further optimized, with a special focus on the aortic wall calcifications (not included in the final reconstruction), via a slice-by-slice manual process for the removal of any leftover segmented artifacts. After completing the segmentation process, the 2D images were transformed into a 3D model. The initial reconstructed models feature rough artificial edges on their surfaces due to the resolution of the medical images and the slice thickness. Further refinement is required prior to the use of the models for computational simulations.

Smoothing is an iterative process for the reduction of the surface roughness, but special care should be given to minimize the deviation of the post-smoothing model from the original data. During the smoothing process, it was observed that the cross-sectional area (and the volume of the aorta in extend) depended on the number of smoothing iterations and the smooth factor. More iterations were leading to a more natural result, while the smooth factor had a heavy impact on the shape of the aorta. Conclusively, it was decided to perform several iterations (500) and keep the smooth factor to a minimum value (0.1).

It is also worth noting that during the segmentation process, the segmentation mask was slightly expanded to compensate for the volume shrinkage due to the smoothing, ending up with a volume-preserving models (with respect to the original data).

To identify the aortic valve, the left ventricle of the heart was also segmented and reconstructed. Specifically, a hole, representing the aortic valve without its cusps, was opened on the aortic surface by removing the intersected area between the left ventricle and the aortic surfaces using the Visualization Toolkit (VTK) library. The post-reconstruction procedure leading eventually to the production of the simulation-ready aTAA model is illustrated in Figure 3.

![Figure 3](image_url)

**Figure 3.** The procedure after the initial 3D reconstruction involving the smoothing, the left ventricle reconstruction, the opening of the aortic valve, the extrusion of the surface and the generation of the computational mesh.
2.4. Mesh generation
The aTAA model after the segmentation and reconstruction processes, is represented by an open surface without thickness, in STL format. Towards performing the structural simulations, a solid model with a given wall thickness and the corresponding computational mesh need to be generated. In this direction, a multi-step workflow involving open-source software and Python libraries, was put in place. The aTAA model (STL file) was initially imported into the free Autodesk Meshmixer (Autodesk Inc.) software where the surface was extruded outwards in the normal direction with an offset of 1.526 mm corresponding to the wall thickness of a typical aTAA, as mentioned in [10]. The extruded surface, also in STL format, was subsequently imported into the SimVascular software for the generation of the computational mesh. SimVascular is a fully open-source software package that provides a complete pipeline from medical image data segmentation to patient-specific blood flow simulation and analysis [11]. Upon importing the model, SimVascular identifies the parts of the model exterior where boundary conditions are applied, i.e. the inner and the outer surfaces of the wall and the rings taking shape at the inlet (aortic valve) and outlet (descending aorta) sections upon extrusion (Fig. 4). The aTAA model was finally discretized into a computational mesh of 180K elements that was converted from VTU (unstructured grid format used by VTK) to INP format which is operational in the FEBio software that was the FEA package of choice in the current study (more information in the next section).

![Figure 4. The parts of the aTAA exterior surface where boundary conditions are applied. Specifically, pressure is prescribed at the inner surface of the wall, while the aortic valve and the descending aorta rings are fixed in all directions.](image-url)
2.5. Material model and structural simulations

The material models for the aTAA and the normal case were based on the study by Pasta et al. [12] who pursued the mechanical response of tissue specimens combining image-derived parameters and tensile testing data. Specifically, they examined the collagen fiber dispersion and alignment in experimentally dissected halves of the aTAA and normal tissue specimens, mechanically characterizing both the intimal and the adventitial arterial wall layers. The fiber-reinforced structural model introduced by Pasta et al. [12] is based on the widely known Gasser-Ogden-Holzapfel model [13], prescribed with the strain energy function,

\[ W = \frac{C}{2}(I_1 - 3) + k_1 \frac{k_2}{k_1} \left\{ \exp \left[ k_2 (k_1 + (1 - 3k)I_4) - 1 \right] \right\} - 1 \]  

where \( C \), \( k_1 \) and \( k_2 \) are the material parameters, \( k \) and \( \gamma \) are the structural parameters, \( I_1 = \lambda^2_{\text{CIRC}} + \lambda^2_{\text{LONG}} \) is the first invariant of the right Cauchy–Green strain tensor, and \( I_{41} = I_{42} = \lambda^2_{\text{CIRC}} \sin^2 \gamma + \lambda^2_{\text{LONG}} \cos^2 \gamma \) is a tensor invariant assuming that the two fiber families (\( i=1,2 \)) for each layer (intimal and adventitial) were mechanically equivalent. In the current study, the arterial wall was modelled by a single-layered fiber-reinforced material using the mean value of the parameters provided by Pasta et al. for the intimal and adventitial layers, as reported in Table 1 for the aTAA and the normal case. Moreover, the density was considered to be \( \rho = 1.07 \times 10^{-6} \text{Kg/mm}^3 \), [14] and the Bulk modulus was set to 12.4 MPa [15] both for the aneurysmal and the non-aneurysmal case. 

Table 1. The parameters of the single-layered fiber-reinforced material model used in the structural simulations for the aTAA and normal case.

|        | C (kPa) | \( k_1 \) (kPa) | \( k_2 \) | \( \gamma \) (deg) | \( k \) |
|--------|---------|----------------|---------|----------------|------|
| aTAA   | 4.8     | 135.8          | 14.8    | 32.8           | 0.25 |
| CONTROL| 6.85    | 324.8          | 19.3    | 13.9           | 0.22 |

The FEBio Studio software was used for the performance of the structural simulations. FEBio is a nonlinear finite element solver specifically designed for biomechanical applications. It offers modelling scenarios, constitutive models and boundary conditions that are relevant to numerous research areas in biomechanics. Development, distribution and support is a joint effort between Jeff Weiss's lab at the University of Utah and Gerard Atleshian's lab at Columbia University. [16] The boundary conditions for the solution of the structural problem were defined as follows: a range of systolic pressure values (100-300mmHg with a step of 20mmHg) applied at the inner surface of the wall, and fixation in all directions of the aortic valve and descending aorta rings, as depicted in Fig. 4. 

The post-processing analysis involves the calculation of the effective (Lagrange) strain and the effective stress (Von Mises). The effective strain is defined in terms of the deviatoric part of the strain tensor. If \( E \) is the (Lagrange) strain tensor, then the deviatoric strain is defined by

\[ E_{\text{dev}} = E - \frac{1}{3} \text{tr}(E)I \]  

and the effective strain by,

\[ E_{\text{ef}} = \frac{2}{\sqrt{3}} E_{\text{dev}} : E_{\text{dev}} \]  

while the effective stress accordingly by,
\[ \sigma_{ef} = \sqrt{\frac{3}{2} \sigma_{dev}^2 \sigma_{dev}} \] (4)

3. Results and discussion

The results are outlined by the 3D contour rendering of the effective strain and stress distributions (Figs. 5 and 6) and the plotting of their maximum values with respect to systolic pressure (Fig. 7) both for the aTAA and the control case. The purpose of the post-processing is to highlight the areas on the surface of the aTAA bulge where high stress and strain occurs, in comparison with a normal aorta, and to identify any differences in the peak strain and stress - pressure curves between the two cases. The effective stress is a measure of rupture risk, while the effective strain describes the increase in the size of the aTAA bulge under the effect of an elevated arterial pressure condition.

The contours in Figs. 6 and 7 highlights that the risk areas of the aTAA are concentrated around the shoulder of the aneurysm and are clearly extended when applying a high systolic pressure boundary condition (300 mmHg compared to 100 mmHg). It is therefore evident that a rupture event could take place at the area where the dilation fades and the normal aortic diameter is restored, and not at the maximum dilated part of the aTAA bulge as one could expect from the clinical perspective. Similarly, the area of maximum effective strain indicates the place where the dissection of the aorta could initiate which is indeed the expected one, as found in most aortic dissection cases (the initial part of the aortic arch). On the other hand, both under 100mmHg and 300mmHg blood pressure, the control case is on a more alleviated stress state compared to the aneurysmal one, with the normal tendency of attracting stresses at areas where surface curvature changes. The effective strain of the normal case under both pressure conditions is comparable to the effective strain of the aTAA case under 100mmHg, suggesting that an aTAA patient under controlled and low blood pressure might have lower risk of dissection and rupture.
The plots of effective strain and stress with systolic pressure clearly indicate that the peak values of both variables are increasing with a different slope between the two cases under increasing blood pressure. Specifically, both are higher in the aTAA case while the plateau in their values is reached earlier in the normal case (around at 200 mmHg, while at 240 mmHg for the aTAA case). It is therefore suggested that an aTAA patient should avoid circumstances of abrupt blood pressure escalation as the stress state experienced by the aorta gets more intense while a normal aorta seems more adaptable to the increase of pressure.

A hypertensive state is common between athletes that lift weights where maximum systolic pressure can reach as high as 360/370 mmHg. [17] Such measurements were recorded in a study of 10 athletes where blood pressure was continuously being monitored through a catheter in the radial artery. Every athlete in the study was asked to perform a series of activities and physical exercises such as double-leg press at 85% and 100% of max capacity. Of course, such an elevated blood pressure requires intense physical activity but is achievable.

![Figure 6. Contours of the effective stress distribution for the aTAA and the control case under a systolic blood pressure of 100 and 300 mmHg.](image)

Besides athletes, there are rare cases of patients where blood pressure had elevated to 300 mmHg without performing any physical activities. Moreover, there are references in medical forums about cases with systolic pressure higher than 250 mmHg. Most of the subjects under high blood pressure, experience intense medical conditions, such as excessive headaches, dizziness, even loss of consciousness. In a case of a patient at an end stage renal disease, a maximum of 290 mmHg blood pressure has been recorded during the hospitalization at the intensive care unit. [18] For the above reasons, a range of pressure from 100 mmHg to 300 mmHg has been considered in the current study, covering all possible stress states experienced by a human aorta.
Figure 7. Plots of max effective strain and stress with respect to systolic pressure for the aTAA and control case.

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