Three-dimensional sensitivity assessment of thoracic aortic aneurysm wall stress: a probabilistic finite-element study†

Simona Celiab,* and Sergio Bertib

a Scuola Superiore Sant’Anna, Heart Hospital ‘G. Pasquiniucci’, Massa, Italy
b Department of adult cardiology, Fondazione G. Monasterio, CNR-Regione Toscana, Heart Hospital ‘G. Pasquiniucci’, Massa, Italy

Abstract

OBJECTIVES: In clinical practice, maximum diameter is used as a criterion to estimate aneurysm-rupture risk; however, it is only a general indicator and its value becomes difficult to estimate in the thoracic segment. Improved understanding of aortic aneurysm complexity and biomechanics is needed to achieve advancements in surgical repair techniques. The objective of this study was to determine the maximum wall stress by using imaging-derived data and a specific probabilistic design integrated into finite element (FE) analysis.

METHODS: Computed tomography images of thoracic aortic aneurysms from our database were analysed and the main morphological features were identified by means of a specific automatic routine. Morphological data were used to develop an idealized finite element library of thoracic aortic arch models. Sensitivity analyses were performed by using the geometrical parameters as input variables for a statistical wall stress assessment. Numerical results were compared with those obtained from deterministic analysis on patient-specific three-dimensional reconstructions.

RESULTS: The results showed that in small aneurysms, wall stress values similar to those of large aneurysms can be obtained if a significant eccentricity is achieved. In small aneurysms, the peak stress is primarily affected by the eccentricity of the bulge [correlation coefficient (CC) = 0.86], while for diameters in the range of 50–60 mm, the CC is 0.43 for the eccentricity and 0.72 for the maximum diameter.

CONCLUSIONS: The stress distribution in small aneurysms may contribute to the pathogenesis of aortic rupture and dissections. Our method can provide a novel and efficient procedure for generating computational models to estimate the wall stress in a comparative multivariate manner.

Keywords: Thoracic aorta • Aneurysm • Biomechanics • Imaging • Computer-based model

INTRODUCTION

In clinical practice, maximum diameter is used as a criterion to estimate aneurysm-rupture risk. Nonetheless, there is evidence that small aneurysms can also rupture or dissect abruptly, while many aneurysms can become very large without rupturing [1]. For this reason, aneurysm correction should be guided not by transversal diameter (Dmax) measurements but by more reliable parameters associated with the actual risk of rupture of the specific vessel. Improved understanding of aortic aneurysm complexity and biomechanics is needed to achieve advancements in surgical repair techniques. Several studies in the last decades have pointed out that wall stress analyses can have a higher discriminatory value than the currently used Dmax [2]. In this context, the computational finite element (FE) approach pointed out that the risks of rupture and aneurysm size are not closely correlated and that the rupture phenomena are due to the concomitant influence of many factors, among them the characteristics of the wall material and the shape of the entire geometry. Regarding the morphological properties, the effect of the eccentricity of the lumen has been extensively investigated in several studies [3] on abdominal aneurysms, while no particular attention has been focused on aneurysms affecting the thoracic aorta or the aortic arch. This lack of information is the consequence of a more complex morphological shape and the very limited amount of material data regarding this region of the aorta. The aim of the present work was, therefore, to analyse the mechanical stress associated with different geometric parameters of the thoracic aorta by applying multivariate computational simulations to capture the parameters that mostly influence the wall stress value. To perform this, we have developed a specific probabilistic technique based on an integrated image process and FE simulation to develop a library based on computed tomography (CT) patient-specific images. Specific attention was focused on the arch segment.

†Parts presented at the 2nd International Meeting on Aortic Diseases 30 September–2 October 2010, Liège, Belgium and at the European Society of Cardiology Congress (ESC), 29 August–2 September, Barcelona, Spain.
MATERIALS AND METHODS

FE analysis is a computer-based technique used to solve the governing field equations in continuum mechanics such as the equations of elasticity in case of complex configurations. The computational tool developed in this study combines commercially available software with custom scripting; in particular, our platform was developed by linking image process, FE simulation and probabilistic modelling.

The basic principles of our procedure are to (i) segment CT patient-specific images to obtain patient-specific models for deterministic FE simulations (ii) extract specific morphological parameters (iii) and to build an idealized thoracic aortic aneurysm (TAA) geometry for probabilistic FE simulations. Figure 1 depicts the design data flow of our methodology.

CT image acquisition and database

A total of 18 electrocardiographic (ECG)-gated CT datasets of human TAA without thrombus that were being evaluated for thoracic endovascular aortic repair (TEVAR) were retrospectively selected for this study (Database: ‘G. Pasquintucci’ Heart Hospital, CU of Cardiology, ‘G. Monasterio’ Foundation CNR Tuscany Region, Massa, Italy). Each individual underwent ECG-gated CT scanning as part of their care and not for the purposes of this study. The TAA geometries include the ascending, the arch and the descending segments. These CT scans were performed at diastolic pressure, according to an approved institutional protocol, with 0.5 mm-thick helical images acquired following the administration of 80 ml of intravenous Iopromide 370 contrast at 4 ml/s. The average age of patients was 75.8 ± 5.83 years, and the male-to-female ratio was 2.6; the mean systolic and diastolic blood pressures were 141.7 ± 21.7 and 74.2 ± 9.17 mmHg, respectively. The location of the aneurysms was classified in accordance with the landing zone (LZ) map described in [4]. Three groups were identified based on the LZ involved: Zone 0–2 (n = 5), Zone 1–3 (n = 10) and Zone 3–3 (n = 3).

Image processing, data extraction and 3D reconstruction

The image process was performed with a specific semi-automatic in-house code developed in Matlab (Mathworks, Inc., Natick, MA, USA). The code is aimed at (i) obtaining 3D reconstructions of patient-specific models used directly for FE analyses and (ii) at extracting and collecting the main geometrical features. For each patient, four measurements were collected: the lesion extension (L), the proximal diameter (Dprox), the maximum transverse diameter (Dmax) and the distal diameter (Ddist). In Fig. 2, two examples of aneurysms in Zone 1–3 are reported with indication of the collected data for a small (A–B) and large (D–E) aneurysms for clarification.

Morphological parameter identification and model definition

Starting from the data collected during the image process phase, three morphological indices were identified to describe the TAA geometry: the maximum diameter ratio (Dr), the lesion extension ratio (LEr), the eccentricity ratio (ecc) and the lesion position along the thoracic arch (p). All measures have been normalized with respect to a linear dimension.

The maximum diameter ratio is defined as the ratio between the maximum transverse diameter (Dmax) and the mean value between the proximal (Dprox) and distal (Ddist) diameters:

$$D_r = \frac{D_{\text{max}}}{\text{mean}(D_{\text{prox}}, D_{\text{dist}})}$$  \hspace{1cm} (1)

The lesion extension ratio defined the ratio between the lesion extension (L) and the maximum transverse diameter (Dmax):

$$\text{LE}_r = \frac{L}{D_{\text{max}}}$$  \hspace{1cm} (2)

The eccentricity ratio (ecc) establishes the dimension of the lumen eccentricity with respect the healthy geometry, and it has been calculated as the ratio between the maximum eccentricity value (eccmax) and the maximum diameter:

$$\text{ecc} = \frac{\text{ecc}_{\text{max}}}{D_{\text{max}}}$$  \hspace{1cm} (3)

The eccmax value represents the linear measure of the distance between the centreline of the thoracic arc without the aneurysmatic bulge and the centre of the maximum diameter itself. Figure 2C and F depicts, as example, the results of the image process for the definition of the eccentricity for the TAA of Fig. 2A and D, respectively. In Fig. 2B and E, the 3D reconstruction obtained with the OsiriX Imaging software is reported for clarification.

The parameter p simply defines the position of the centre of the bulge with respect to the four zones previously identified in
accordance with the TEVAR procedure. In addition to the $p$ parameter, the relative position ($p_r$) with respect to the centre of the aneurysmatic zones was also calculated for each model. The extension ratio value ($LE_r$) provides some information regarding the shape of the aneurysm: the closer $LE_r$ is to 1, the more spherical is the shape of the bulge. The diameter ratio ($1.36 < Dr < 2.53$), the eccentricity ratio ($0.68 < ecc < 1.33$) and the relative position ($0.34 < p_r < 0.53$) parameters were used to develop a parameterized virtual model.

Deterministic finite element models

The 18 patient-specific CT geometries were analysed by means of deterministic FE simulations.

Given the current limitations in the experimental data on regional variations in vascular mechanical properties and thickness, models were endowed with constant and uniform wall properties.

The material properties of the aortic wall were modelled as isotropic and hyperelastic models using the two-parameter constitutive relation with material coefficients equal to $c_1 = 100$ kPa and $c_2 = 530$ kPa as proposed in [5]. In addition, all materials were defined as nearly incompressible ($\eta = 0.495$). The wall thickness was considered uniform at 1.8 mm. All the ends of the aortic vessel as well as of the left subclavian artery, of the left common carotid artery and of the brachiocephalicartery were simulated by constraining the displacements to zero in axial and circumferential directions. Because we were interested in evaluating the maximal wall stress during the cardiac cycle, the mean value of the collected peak systolic blood pressures, 18.6 kPa (140 mmHg), was applied to the inner surface and simulations were performed in a quasistatic condition.

Probabilistic finite element models

Probabilistic analysis is a method that allows uncertainty and variations to be incorporated into a FE model [6]. The basic principle of the method is that the input parameters of the model are defined not by a single value, but by means of a set of values with a specific statistical distribution. For the purpose of the study, the three previously described parameters ($D_{max}$, $ecc_{max}$, and $p_r$) were selected as variable inputs for the probabilistic FE simulations. All the geometrical parameters were described by following the variability of the morphological parameters obtained during the image processing. The maximum von Mises wall stress was calculated and assumed as a control output variable. Additionally, to measure the strength of the relationship between morphological parameters and stress value, the Pearson correlation coefficient [7] is calculated. Due to the low number of input random parameters involved in the simulation, the maximum number of 200 runs (50 samples on 4 strata)
was sufficient to reach convergence with a tolerance equal to 1 and 2% on the mean value and the standard deviation (SD) of the output stress value, respectively.

RESULTS

In order to validate the image process code, a comparison between the obtained automatic measures and those extracted manually by an expert radiologist was performed. Results were in good agreement with respect to our automatic data-extraction process: the maximum error of $\approx 1.8\%$ was founded on the value of the maximum cross-section diameter due to the different meaning of this parameter. In fact, while in current clinical practice the distance between the anterior and posterior wall portion is assumed as the $D_{\text{max}}$ measure, in our study this measure was split into the maximum diameter calculated by means of a best-fit process on a circumference and the eccentricity measure. The manual and the automatic procedure are closely correlated when the bulge involved both anterior and posterior vessel, as reported in (Figs. 2C and F). By analysing the results of the morphometric procedure, we can observe that the mean value of the $D_{r}$ is about equal to 2: this value is a consequence of the fact that all the patients analysed were scheduled for an endovascular or hybrid elective repair.

Deterministic finite element analyses

Figure 3 depicts the von Mises stress for two 3D patient-specific models. Although a decreased overall stress can be observed between the two models (from 0.382 to 0.681 MPa), no significant variations were recorded in the location of the peak stress.

By observing the stress distributions, it can be noted that the region of maximum wall stress occurred at the junction region between the arch segment and the bulge and not in the region of maximum diameter. In particular, the maximum stress was located in the inflection points of the anterolateral location in which the local curvature changed from concave outward to concave inward.

Probabilistic finite element analyses

The convergence was reached after 100 runs. The sensitivity analysis shows that the mean value of the maximum stress is about 0.48 MPa (SD 0.047 MPa) (Fig. 4A). The box-percentile plot (Fig. 4B) summarizes the distribution of the output value with respect to the samples. Results show that the median value within the box (red line) is equidistant from the hinges with good approximation and consequently, the data are not skewed.

The multivariate approach showed that the maximum wall stress is not significantly affected by the position ratio parameter ($CC = 0.03$). Indeed, as reported in Fig. 5, a positive correlation was found, as expected, for the $D_{r}$, ecc, inputs.

However, correlation analysis indicated that the most important factor for peak stress is different in the case of small and large bulges. For diameters less than 50 mm (Fig. 5A), peak stress is primarily affected by the eccentricity (ecc) of the bulge ($CC = 0.86$), while for diameters in the range of 50–60 mm (Fig. 5B) the CC is 0.43 for the ecc and 0.72 for the $D_{\text{max}}$. Figure 5C and D shows the plot of the best-fit surface in the whole variable space in case of small ($D_{r} < 1.25$) and large bulges ($D_{r} > 1.25$).

This surface explains the results of the correlation coefficient reported in Fig. 5A and B, respectively, by showing the variation of the maximum stress with respect to the two input variables in all the ranges and points out the non-linear relation between the eccentricity and the maximum diameter with respect to the maximum stress. Other interesting results are the percentile distributions of the output parameter considering the four different combinations that can be obtained from the two input variables, Table 1.
As overall considerations, they indicate that a mean value of \( \approx 0.5 \) MPa can also be obtained in the case of a small diameter if a significant eccentricity is achieved (Case 3) and that the combined effect of large bulge diameter and eccentricity (Case 4) produced an increase of \( \approx 27\% \) with respect to the first case.

Figure 6 depicts the box-percentiles plot for the abovementioned four cases. Additionally, apart from the last combined configuration, all the cases present a symmetric position of the mean value with respect to the interquartile range. The correlations between the maximum diameter and the eccentricity with respect to the maximum wall stress for each subclass are better explained in Fig. 7, where the best-fit surfaces are reported.

For each surface, a different correlation between the input and the output variables can be observed as reported in Table 1. In particular, the non-linear relation between the input and output variable is shown for the Case 4. Note that the non-symmetric position of the mean value with respect to the interquartile range for the Case 4 arises from the inflection of the surface for \( D_r = 1.4 \).
complexity of the structure and the high computational cost required by patient-specific models, sensitivity analyses have not been performed on 3D real geometries and only univariate investigations have been performed on idealized shapes to estimate the influence of a single parameter on the whole stress map [9]. Computer simulations based on patient-specific geometry are difficult to translate in clinical practice and to include in clinical decision-making for an evaluation in 'real-time', due to the high computational cost of the traditional FE approach (1 patient-specific simulation can require from several hours to some days, depending on the rate of complexity of the system and power of the computational centre). Moreover, even if on the one hand the patient-specific geometries are more attractive, on the other hand some questions remain open, e.g. how much smoothing makes the reconstructed anatomy 'smooth enough'? How is the reconstruction process operator-dependent and how does this dependence affect numerical modelling? [10]. Many of these questions are left either unanswered or as an end user’s judgment call, but can be addressed by quantifying interoperator and intraoperator variations in the repeatability of 3D reconstructions. Despite the improved computational power and techniques, the applicability of computer simulations in the current clinical analysis is limited, mainly due to the difficulty in reproducing the inter-patient variability of the input parameters [11].

In case of the computational analysis of the bio-structure, the mean values of the properties are considered for estimating the response of the system to the applied load. However, since the biological systems are variable in nature, the results obtained based on the mean value analysis of the system may not be sufficient to extract general information from a statistical perspective.

In view of this, the probabilistic approach with multivariate analyses is thought to be useful to predict the distribution of the desired response parameter. The basic concept presented here is the possibility of including the results of the numerical simulations in the decision-making. The term 'real-time' consequently is based on the possibility of accessing a database in 'real-time', in which not only the clinical data and results from experimental activities, but also the results of numerical simulations, reported in a statistical meaning, are included. The application of computational modelling with FE analysis by using probabilistic assessment to solve the clinical question of wall stress represents a novel approach. To our knowledge, no prior work on this exists in the literature.

In this work, we have focused our attention on the thoracic arch segment, proposing an imaging toolbox able to extract the morphological aspects from CT images in an automatic and operator-independent manner. Moreover, by processing CT images, we have also presented a novel interpretation of maximum vessel dilatation by considering both the maximum transversal diameter and the eccentricity of the bulge in the arch.

In the literature, great effort has been devoted to predicting stress distribution and rupture events in the abdominal district, while only a few papers have focused on the thoracic regions [12, 13]. More recently, Nathan et al. [14] simulated 17 patient-specific aneurysms from CT, by using linear elastic FE models. They observed that in saccular geometries, the maximum stress was greater than that of the fusiform and that the correlation between wall peak stress and maximal diameter was more robust for fusiform aneurysms than for saccular. However, the adopted linear constitutive relation was unable to take into account the non-linearity of the soft biological tissue and they did not establish any additional correlation with other morphological parameters.

### Table 1: Results of the 4 cases of the sensitivity analysis

<table>
<thead>
<tr>
<th>Case</th>
<th>Condition</th>
<th>Number of samples</th>
<th>Mean peak stress (MPa)</th>
<th>CC Dr</th>
<th>CC ecc</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>$D_r^+$ and ecc&lt;sub&gt;+&lt;/sub&gt;</td>
<td>25</td>
<td>0.39</td>
<td>0.0487</td>
<td>0.8669</td>
</tr>
<tr>
<td>2</td>
<td>$D_r^+$ and ecc&lt;sub&gt;++&lt;/sub&gt;</td>
<td>25</td>
<td>0.50</td>
<td>0.9025</td>
<td>0.1307</td>
</tr>
<tr>
<td>3</td>
<td>$D_r^+$ and ecc&lt;sub&gt;-&lt;/sub&gt;</td>
<td>29</td>
<td>0.48</td>
<td>0.6131</td>
<td>0.6854</td>
</tr>
<tr>
<td>4</td>
<td>$D_r^+$ and ecc&lt;sub&gt;+&lt;/sub&gt;</td>
<td>21</td>
<td>0.54</td>
<td>0.2511</td>
<td>0.8595</td>
</tr>
</tbody>
</table>

Four different geometrical combinations of $D_r$ and ecc (where $D_r^+$: $D_r<1.25$; $D_r^+$: $D_r>1.25$; ecc<sub>+</sub>: ecc<sub>-</sub>; ecc<sub>-</sub>: ecc<sub>+</sub>): The CC for each case is reported for both input variables.

This phenomenon is more significant for small eccentricity (ecc<sub>-</sub> < 0.35).

### DISCUSSION

The criterion of the maximum transverse diameter to estimate aneurysm rupture risk is based on the law of Laplace that establishes a linear relationship between diameter and wall stress. However, the law of Laplace is simply based on cylindrical geometries, whereas aneurysms are complex structures and therefore the law fails to predict realistic wall stresses. Moreover, focussing on the TAA, measuring $D_{max}$ becomes difficult due to the geometrical complexity of the three-dimensional (3D) sharp curve of this segment [8]. Numerical simulation has the potentiality to predict the behaviour and response of complex systems, particularly when considerations of non-linear effects, multiple physics or complex geometry are required. In view of this, mathematical and numerical modelling has been widely used in the literature to obtain more accurate wall stress values, and many biomechanical models have been developed for clinical applications with the objective of improving the diagnosis and treatment of diseases. However, to date, the applicability of computer simulations in current clinical analysis still remains limited.

Studies, to date, have typically used FE models starting from 3D geometries constructed from computer tomography and magnetic resonance (MR) images or have used simplified morphologies. However, both approaches present some limitations. Due to the...
Similar to previous studies on TAA [14, 15], in our study we have not considered patients with intraluminal thrombus (ILT). This working decision arises from the observation that, while abdominal aortic aneurysms are usually associated with the development of ILT [16], in case of TAA the incidence of ILT is significantly reduced [17].

Although simplifications have been introduced to make the model tractable, it is believed that the results of the stress analysis are meaningful in a comparative sense. In particular, our multivariate sensitivity analyses have pointed out that the eccentricity, in junction with the maximum diameter, is able to provide additional information on the entire geometry with a direct consequence on the identification of potential critical zones. Our FE-based calculations depict (Fig. 7E and F) the non-linear relationship between the maximum wall stress and the $D_r$ and $ecc_r$ parameters. In particular, Fig. 7F shows an opposite behavior for the two eccentricity

Figure 7: Best-fit surface for the trend of the maximum wall stress (MPa) in the whole input variables space for each of the four subclasses reported in Table 1: case 1 (A), 2 (B), 3 (C) and 4 (D), respectively. Maximum wall stress (MPa) plots as function of the eccentricity (E) for two different diameter ratios ($D_r = 1.3$ and $D_r = 1.5$) and with diameter ratio (F) for two different eccentricity values ($ecc_r = 0.3$ and $ecc_r = 0.4$). These trends refer to plot (D).
values: a negative trend of the wall stress is observed for \( \text{ecc}_r = 0.4 \) by increasing the \( D_f \). This phenomenon can be justified considering an increase of the radius of curvature between the bulge and the vessel, and it is the combination of the local transverse dimension and local curvature that governs the wall stress distribution under the assumption that the wall thickness is uniform. These results corroborate that the stresses acting on the wall are not evenly distributed and cannot be adequately described by the law of Laplace due to the missed significant contributions of local complex wall surface shapes.

In describing the results from our FE analysis, the emphasis was placed on the mechanical stress; however, the deformations of the model were shown to be in the physiological range. The strain measurements, in fact, are in agreement with those obtained by Morrison et al. [18]. In fact, by considering two different cross sections, distal and proximal to the left coronary artery, we recorded a maximum circumferential strain value of \( \sim 10 \) and \( 13\% \), respectively, at 18.6 kPa (140 mmHg) of inner pressure. The distribution of the stress map plays a fundamental role due to its importance in rupture phenomena and for its correlations with the inflammatory activity of the artery wall [14]. Obviously not all aneurysms rupture at the same stress. This is due to differences in wall strength; however, the increase in stress has been associated with a higher growth rate [19] in the remodelling processes of the aortic wall aimed to restore the homeostatic stress value. Xu et al. [20] made a comparative study of FE simulation and positron emission tomography–computed tomography (PET-CT) scan. They found that predicted high wall stress regions co-localized with areas of positive 18F-fluoro-2-deoxyglucose (18F-FDG) and in particular, that the locations of rupture corresponded well with regions of elevated metabolic activity. Changes in glycolytic activity and the aortic aneurysmatic (AA) wall are associated with inflammation, macrophage infiltration and MMP2 and MMP9 activity [21], suggesting a relationship of aortic biomechanics with underlying histopathological changes detected by FDG-PET/CT.

**Study limitations.** Despite encouraging results with clinical evidence, some limitations still remain, and our models may still be improved, in particular the absence of a non-uniform wall thickness distribution and the description of the artery wall as an isotropic material. Regarding the first aspect, it is known that the realistic aneurysm has a non-uniform wall thickness and that, in particular, reduced wall thickness is found in correspondence with the bulge due to the collagen network breakdown. Results show that the thickness in the anterior area was less than that of the dorsal one by \( \sim 50\% \) and FE simulations showed an increase of the peak stress of \( \sim 40\% \) in correspondence with the wall-thickness reduction [22]. A recent study [23] performed on idealized abdominal aortic aneurysm (AAA) geometries has pointed out that the wall-thickness reduction co-participates with wall stress; however, the bending effects at the junctions of vessel segments still remain a critical point due to the role of the geometrical curvature.

However, the measurement and implementation of the patient-specific non-uniform thickness of the arterial wall are complex tasks, due primarily to the limitations of the current technology for measuring this parameter non-invasively [10]. Nonetheless, estimation of wall thickness on the basis of anatomical variables (such as age and gender) and in correlation with histomechanical investigation might further improve the ability of the analysis to distinguish high-risk aneurysms. Along with these limitations in computing wall thickness, further anisotropic constitutive material models are also needed. However, regarding the thoracic arch, mechanical tests are reported only in [24] and consequently the material descriptions are less developed. In vivo material property characterizations require in vivo intraluminal pressure measurement and 3D strain estimation. In this regard, 4D CT/MR images have been used to assess blood-vessel deformation [18]. The analysis of harvested surgical specimens as well as ex vivo models will improve the capability to test the fully anisotropic properties of the artery wall.

Finally, it is worth pointing out that in case of patient-specific models, geometries are acquired in vivo at diastolic blood pressure and consequently in a deformed and not load-free state.

Ideally, the boundary condition should be applied to either the unloaded geometry or a prestressed geometry to obtain physiologically realistic stresses. In recent years, several authors have proposed approaches to recover the undeformed/unloaded state from the image-based deformed configuration for AAA. These conclude that not accounting for the zero pressure configuration may lead to an overestimation of the maximum peak wall stress. However, finding a physically meaningful prestressed configuration in a finite deformation problem, which that is, is non-trivial due to the non-uniqueness of the obtained state and severe buckling phenomena [25], and it requires a priori assumptions of the material properties. Consequently, to date, it still remains an open field of research rather than a limitation of this study.

Prescribing physiologically realistic conditions is both difficult and critical in any computational model, and the associated results are not yet expected to be fully physiological. As with any model, more experimental data are needed to improve the simulations.

**CONCLUSIONS**

The present study demonstrated a novel approach to estimate the wall stress. It corroborates findings reported by other groups that diameter is not sufficient to estimate the potential rupture risk, particularly for eccentric geometries. These results provide a theoretical support for the repair of eccentric bulges even if at smaller diameters.

The aneurysmal disease is a complex phenomenon and requires the causative links between several fields of investigation: there is an urgent need for tractable multidisciplinary approaches to improve our current view of the physiology and pathology of vascular diseases. In this context, appealing aspects of the potential clinical application of this technology are the integration with biohumoral markers analysis and additional imaging procedures for a more extensive and multidisciplinary study of this pathology.

We point out that the results presented here are devoted to illustrating the eventual impact of additional parameters that currently are not measured in clinical practice. In the present study, we focused on establishing a 3D computational simulation of TAA, which involved additional geometrical parameters adopting commonly used configurations. For this reason, future work will be focused on the quantification of additional parameters such as the regional wall thickness and material variation.

In this perspective, the additional appealing aspects of potential clinical applications of this technology consist of the collection of data before and after rupture events as well as from other centres worldwide. The opportunity of having a central network would
REFERENCES