Deformation and distensibility distribution along the abdominal aorta in the presence of aneurysmal dilatation

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Funding: The research project was partially supported by the Action «Supporting Postdoctoral Researchers» of the Operational Program "Education and Lifelong Learning" (Action’s Beneficiary: General Secretariat for Research and Technology), and is co-financed by the European Social Fund (ESF) and the Greek State.

This study was presented at the 62st ESCVS congress (11-13 April 2013, Regensburg-Germany) and was honored with the Young Surgeon Award Silver Prize.

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ABSTRACT

Aim: In order to evaluate the elastic behavior of the abdominal aortic aneurysm (AAA), the distribution of aortic deformation during the cardiac cycle is measured. Moreover, the distensibility of the AAA composite structure consisting of the AAA wall and the intraluminal thrombus (ILT), as well as that of the adjacent non-aneurysmal aortic segment (NAA), are calculated.

Methods: Ten patients underwent electrocardiographically-gated computed tomography. 3D-surfaces of aortic wall and lumen were reconstructed during peak-systole and end-diastole and cross-sections perpendicular to the centerline were extracted 1mm apart. Comparison of cross-sectional areas between peak-systole and end-diastole provided the relative area change (RAC). Mean values were calculated for NAA (RAC_{NAA}), aneurysmal wall (RAC_{Wall}), and aneurysmal lumen (RAC_{Lumen}). Distensibility of aneurysmal and unaffected aorta was calculated using brachial pressure measurements (D_{AAA} and D_{NAA} respectively). Normalized distensibility (D_{NORM}) of the AAA was calculated with respect to normal aortic segment distensibility and related to aneurysm size and thrombus content.

Results: A map of aortic deformation during the cardiac cycle was obtained. Differences between RAC_{Wall} (median=0.7%, range=0.3-2.1%) and RAC_{NAA} (median=2.8, range=0.9-4.8%), as well as between RAC_{Wall} and RAC_{Lumen} (median=1.8%, range=0.5-3.4%) were statistically significant. D_{AAA} (median=0.30\cdot10^{-5} \text{ Pa}^{-1}, \text{ range}= 0.05-0.64\cdot10^{-5} \text{ Pa}^{-1}) and D_{NAA} (median=0.43\cdot10^{-5} \text{ Pa}^{-1}, \text{ range}= 0.16-0.83\cdot10^{-5} \text{ Pa}^{-1}) were not significantly different. Median D_{NORM} was 0.73 (range=0.1-3.1) and presented a significant positive correlation with AAA size and thrombus content.
Conclusion: Aneurysmal wall deforms significantly less than non-aneurysmal wall and aneurysmal lumen, due to altered elastic properties and reduced loading. In large AAAs with larger amounts of ILT, the lumen deformation is comparable or even exceeds that of NAA and subsequently so does the distensibility of the Wall-ILT composite, an observation suggesting a thrombus cushioning effect. $D_{NORM}$ may provide insight in the estimation of AAA evolution and assist in rupture risk assessment.

Key Words: AAA deformation, distensibility, ECG-gated CT, cushioning effect
INTRODUCTION

Abdominal aortic aneurysm (AAA) represents a major health problem that affects increasing number of individuals with aging of the population.\textsuperscript{1,2} Its major complication is rupture followed by a striking overall mortality of \textit{80-90\%}.\textsuperscript{3,4} All the relevant screening and therapeutic protocols attempt to prevent this catastrophic outcome indicating immediate surgical repair or watchful waiting and intervention when this is considered appropriate. According to current guidelines elective repair of AAAs is based on size and growth rate with surgical intervention recommended at maximum diameter $\geq 5.5\text{cm}$ or enlargement $\geq 1\text{ cm/year}$ respectively.\textsuperscript{5} However it is well established that such universal criteria can not accurately predict AAA evolution on a patient specific basis. Even small AAAs, well under the threshold for surgical repair, can rupture, while almost 50\% of larger aneurysm will never proceed to rupture.\textsuperscript{6,7}

In an effort to identify additional aneurysm rupture risk markers that will assist in the management of these patients, there is increasing interest in the role of AAA biomechanical parameters. Aneurysmal disease is mainly a degenerative process affecting the arterial wall leading to loss of structural integrity. Therefore reduced elastin and collagen that accompanies AAA formation and evolution have been shown to be related to altered elastic properties as well as reduced strength of aortic wall.\textsuperscript{8,9} Moreover results of \textit{ex vivo} mechanical testing indicate a positive correlation of aneurysmal wall stiffness with its strength suggesting that reduced stiffness may foretell a higher rupture risk.\textsuperscript{10,11} It has therefore become apparent that additional knowledge on AAA wall mechanical properties could provide insight in the regional loss of wall integrity and lead to more sensitive, patient-specific rupture risk estimations.

Recent advances in medical imaging and specifically the use of electrocardiographically-
gated computed tomography (ECG-gated CT) allowed the documentation of aortic wall motion throughout the cardiac cycle. Using advanced imaging techniques, the aortic wall regional elastic properties, could directly be estimated in vivo.\textsuperscript{12} Since previous imaging ultrasound based studies indicate that an increase in aneurysm distensibility over time is related to a higher rupture risk profile, the exploitation of such modern imaging techniques is expected to assist in rupture risk assessment.\textsuperscript{13} Various research groups that estimate elastic wall properties, non-invasively using imaging modalities, either record the change of vessel cross-sectional area during the cardiac cycle, taking into account predefined CT slices, or the volume of the vessel, and relate it to the arterial pressure.\textsuperscript{14-16} However, both recordings provide only a global estimation of wall properties not taking into account regional variations that may be present along the vessel.\textsuperscript{17}

In the current study we use ECG-gated CT to capture aortic deformation throughout the cardiac cycle and estimate the distribution of aortic wall and intraluminal thrombus (ILT) distensibility along the normal aorta and AAA in an effort to detect changes that accompany aneurysm formation and evolution. Moreover we attempt to compare elastic properties of the aneurysmal and the unaffected aortic segment and provide an intra-patient measure of differences which could be of value for rupture risk estimations.

**MATERIALS AND METHODS**

**Study Population**

Ten patients with AAA were included in this study, which was approved by the Institutional Review Board of the University Hospital of Crete, and all subjects gave informed consent. The patients were under surveillance and underwent CT imaging for follow-up. After CT imaging, 3
patients underwent elective repair, based on the current guidelines of AAA management. Mean age was 71.7 years ranging from 59 to 82 years. The majority was of male gender with a male:female ratio of 9:1. Maximum diameter of the studied AAAs ranged from 32 mm to 68 mm with an average of 49 mm. Demographics and clinical characteristics of patients, as well as aneurysm size, are presented in Table 1.

Data acquisition

All patients underwent Multi-Detector ECG-gated CT. Image acquisitions were performed with a Somatom Definition Flash, Dual source-Dual energy CT scanner (Siemens, Erlangen, Germany), before and after contrast media administration with retrospectively ECG gated spiral acquisition. Non-ionic contrast media was used. Slice thickness was 0.625 mm and image matrix size 512x512. The temporal resolution was 83 ms and in plane spatial resolution 0.33 mm. The total effective dose was 5.5 mSv at 80 bpm. Two ECG-gated series of axial images were reconstructed, at peak-systole and end-diastole during the R-R interval. Non-invasive blood pressure measurements at the time of the CT scan were also obtained.

Image post processing

Segmentation and three dimensional (3D) reconstruction

Segmentation and 3D surface reconstruction of the aortic wall and lumen was performed using the software ITK-SNAP. Briefly, outlines of the outer surface of the AAA (“wall”) and the luminal surface (“lumen”) were manually obtained slice by slice and the 3D surfaces were reconstructed from the stack of contours. This was performed for peak-systole and end-diastole. During segmentation, each slice of one phase (systole or diastole) was placed side-by-side with the respective slice of the other phase (blinded) in order to exclude minor artifacts mainly caused
by branching vasculature or wall calcification. The segmentation included the visceral segment of the abdominal aorta to obtain information of a seemingly healthy vessel, to the last slice just before the aortic lumen splits to the common iliac arteries. Due to contrast enhancement, lumen displacement between peak-systole and end-diastole could be detected for all cases. For aortic wall, however, because of poor discrimination from surrounding tissues, displacement could be accurately detected in 6 out of 10 cases while there was significant uncertainty in 4 cases. Therefore, three 3D surfaces were reconstructed for each case: diastolic-lumen, systolic-lumen, and diastolic-wall surfaces. Systolic-wall surfaces were reconstructed for those 6 cases that allowed confident wall boundary detection.

**Extraction of lumen centerline and estimation of geometric parameters**

The reconstructed 3D lumen and wall surfaces of each case were processed using the vascular modeling tool kit (VMTK) software. After smoothing the surfaces, the centerline of the lumen was created, and was used to extract cross-sections of the 4 AAA surfaces (lumen and wall during systole and diastole) at 1 mm intervals along the centerline, and assign them the same number for direct comparison of their cross-sectional area (Figure 1).

With regard to the wall surfaces, the maximum diameter of each section was also recorded. The ILT cross-sectional area was calculated by subtracting the section area of the lumen surface from that of the wall.

**Identification of the two aortic regions of interest**

Two aortic regions were defined based on the diameter change. The first was the non-aneurysmal aortic (NAA) region, where the diameter did not change with distance monotonically, had minor fluctuations, and the vessel was not tortuous appearing overall healthy
from the CT images. The second was the AAA region, which started where the diameter increased with distance and ended at the aortic bifurcation.

Deformation and Elasticity quantification

To provide a map of aortic deformation along the healthy and aneurysmal vessel for both aortic wall and lumen, the relative area change (RAC) was calculated for every cross-section of the AAA wall and lumen as well as the NAA (Eq. 1):

\[
RAC = \frac{\text{Area}_{\text{Systole}} - \text{Area}_{\text{Diastole}}}{\text{Area}_{\text{Diastole}}} \tag{Equation 1}
\]

For each case under examination mean values of \(RAC_{\text{Wall}}\), \(RAC_{\text{Lumen}}\) and \(RAC_{\text{NAA}}\) were recorded.

To override the uncertainty in aortic wall displacement detection for those cases with poor demarcation of AAA wall boundary from surrounding tissues, aortic lumen displacement was used for the quantification of vessel elastic properties. Specifically, by assuming ILT incompressibility, the area change of the lumen closely represents the area change of the wall (Eq. 2):

\[
T_s = T_d \Rightarrow W_s - L_s = W_d - L_d \Rightarrow W_s - W_d = L_s - L_d \tag{Equation 2}
\]

\(T =\) thrombus area, \(W =\) wall area, \(L =\) lumen area, \(s =\) systole, \(d =\) diastole

Thrombus incompressibility has been proposed by previous mechanical and imaging studies and is widely used in computational models that estimate wall stress.\textsuperscript{20-22} This was also verified in our study population by recording ILT systolic and diastolic areas in those cases where both lumen and aortic wall displacement could be recorded.\textsuperscript{23}

Moreover, by using brachial blood pressure measurements the distensibility \(D\) was calculated.
This is defined as the relative change in the vessel cross-sectional area during the cardiac cycle divided by the corresponding change in blood pressure (Eq. 3).24,25

\[
D = \frac{\text{Area}_{\text{Systole}} - \text{Area}_{\text{Diastole}}}{\text{Area}_{\text{Diastole}} \Delta P} = \frac{RAC}{\Delta P} \quad \text{Equation 3}
\]

Since pulse pressure exerted on the arterial wall of the NAA and luminal surface of ILT represents the loading that causes aortic deformation, we used the observed expansion data along with blood pressure measurements to calculate distensibility that regarded the NAA and the whole AAA structure consisting of the wall-ILT composite (\(D_{\text{NAA}}\) and \(D_{\text{AAA}}\) respectively). On the other hand the loading exerted in the outer AAA wall is reduced due the presence of ILT and therefore its distensibility cannot be determined using pulse pressure.

Furthermore the normalized distensibility \((D_{\text{NORM}})\), defined as the ratio of distensibility between AAA and normal aorta is calculated to reduce uncertainty due to non-invasive blood pressure measurements. Specifically it has been proposed that the brachial blood pressure can underestimate intra-aortic pressure to almost 15% but \(D_{\text{NORM}}\) as defined here is not sensitive to this error (Eq. 4).16

\[
D_{\text{NORM}} = \frac{D_{\text{AAA}}}{D_{\text{NAA}}} = \frac{\frac{\text{RAC}_{\text{Lumen}}}{\Delta P}}{\frac{\text{RAC}_{\text{NAA}}}{\Delta P}} = \frac{\text{RAC}_{\text{Lumen}}}{\text{RAC}_{\text{NAA}}} \quad \text{Equation 4}
\]

Statistical analysis

For \(\text{RAC}_{\text{NAA}}, \text{RAC}_{\text{Wall}}\) and \(\text{RAC}_{\text{Lumen}}\) as well as \(D_{\text{NAA}}\) and \(D_{\text{AAA}}\) and \(D_{\text{NORM}}\), median and range are reported. Statistical significance of differences was tested using Wilcoxon-rank sum test. Correlation between \(D_{\text{NORM}}\) and AAA maximum diameter as well as average ILT cross-sectional area was assessed by Spearman’s rho test. Correlation coefficient is reported.
RESULTS

Determination of the distribution of aortic deformation throughout the cardiac cycle is feasible using ECG-gated CT scan. Two representative cases are presented in Figure 2. All cases presented a reduced deformation of AAA wall compared to that of NAA while AAA lumen expansion approximated that of the non-aneurysmal segment. Specifically $RAC_{Wall}$ was 0.7% (0.3-2.1%), $RAC_{Lumen}$ was 1.8% (0.5-3.4%) and $RAC_{NAA}$ was 2.8% (0.9-4.8%). There were statistical significant differences between $RAC_{Wall}$ and $RAC_{Lumen}$ ($P=0.005$) as well as $RAC_{Wall}$ and $RAC_{NAA}$ ($P=0.007$). Differences between $RAC_{NAA}$ and $RAC_{Lumen}$ were not statistically significant ($P=0.12$).

In some cases there was a consistent finding of negative values in $RAC_{Wall}$ and $RAC_{Lumen}$ in the AAA shoulder region (at the entrance of the aneurysm expansion where the aortic diameter increases) that was verified by independent observers. This could be explained by the AAA expansion pattern during the cardiac cycle which is both circumferential and longitudinal. Subsequently in regions of steep change of vessel size, such as the shoulder region, cross-sections during systole may be compared to slightly distal parts of the vessel, which are anatomically larger during diastole and therefore present negative values of area change.

$D_{NAA}$ was 0.43 (0.16 - 0.83)$\cdot 10^{-5}$ Pa$^{-1}$ while $D_{AAA}$ was 0.30 (0.05 - 0.64)$\cdot 10^{-5}$ Pa$^{-1}$. Differences between values were not statistically significant ($P=0.12$). $D_{NORM}$ was 0.73 (0.1-3.1). To identify if this index correlates with maximum diameter and thrombus content, it was presented versus the aforementioned parameters (Figure 3). There was a strong positive correlation of $D_{NORM}$ with aneurysm size (Spearman’s rho 0.65, $P=0.04$) and average thrombus cross-sectional area (Spearman’s rho 0.81, $P=0.005$).
The values of the indexes representing aortic deformation and distensibility ($RAC_{NAA}$, $RAC_{Lumen}$, $RAC_{Wall}$, $D_{NAA}$, $D_{AAA}$, $D_{NORM}$) for all cases under examination as well as statistical significance of differences are presented in Table 2.

DISCUSSION

The current study utilizes modern imaging techniques to record vessel deformation during the cardiac cycle along the normal and aneurysmal abdominal aorta. Such methodology provides data for the in vivo determination of aortic elastic properties and their regional variations through direct measurements from non-invasive imaging techniques.

Our results indicate that AAA wall expansion during the cardiac cycle is significantly less than that of the adjacent non-aneurysmal segment and the aneurysmal lumen. Two factors may be implicated to this phenomenon. Firstly, the altered mechanical properties of the aneurysmal wall compared to the NAA wall, due to the degeneration that aneurysmal disease causes, could have significantly contributed to the reduced deformation of the former. Mechanical studies indicate that AAA wall becomes stiffer than that of the unaffected aortic segment and this is consistent with the present findings.\textsuperscript{8,10,11} Secondly, the loading that is exerted to the aneurysmal wall causing its deformation may be reduced compared to that acting on the NAA wall and the luminal surface of ILT, due to the cushioning effect of the latter. Specifically the pulse pressure represents the loading force that causes deformation of the non-aneurysmal wall as well as the whole AAA structure consisting of the Wall-ILT composite. Many computational studies indicate a cushioning effect of ILT that significantly reduces stress exerted on the aneurysmal wall which
also could have contributed to the observed reduced deformation.\textsuperscript{26,27} Therefore, the use of measured arterial pulse pressure to calculate distensibility of the aneurysmal wall, when ILT is present, which is the case in the majority of AAAs, although extensively used in the literature, may represent an oversimplification leading to inaccurate conclusions.\textsuperscript{14,15,26,27} Subsequently, without knowing the actual loading exerted on the aneurysmal wall, which could be possibly evaluated using finite element analysis, the determination of AAA wall elasticity using deformation data acquired \textit{in vivo} and recordings of pulse pressure, could be misleading. On the other hand, pulse pressure can be used to calculate the distensibility of NAA and aneurysmal Wall-ILT composite. Our results indicate a reduced AAA distensibility compared to the unaffected segment but differences were not statistical significant. The wide range of $D_{\text{AAA}}$ values indicates the presence of large differences between cases.

Since each patient’s baseline aortic distensibility may be different due to several factors such as age or comorbidities, in order to obtain a measure of distensibility change due to AAA disease, $D_{\text{NORM}}$ was introduced, which is a measure of the relative distensibility of the AAA structure compared to that of the unaffected aorta.\textsuperscript{28,29} Such an intra-patient normalization eliminates possible inaccuracies introduced by non-invasive blood pressure measurements which can deviate from actual intra-aortic pressure.\textsuperscript{16} Furthermore, differences in $D_{\text{NORM}}$ found between patients, can possibly be attributed to the extent of degeneration caused by AAA disease, since the baseline of aortic distensibility for each patient has been taken into account.

There was a wide range of $D_{\text{NORM}}$ values in our study population. Moreover a strong positive correlation of this parameter with aneurysm size and more importantly with ILT content, as measured by average ILT cross-sectional area, was established. Subsequently the whole AAA
structure consisting of the Wall-ILT composite is significantly more distensible for large AAAs with larger amount of thrombus while smaller AAAs with less or no ILT appear stiffer. This indicates the possible cushioning effect of ILT which, despite the fact of the previously mentioned globally reduced AAA wall deformation, allows for the increased lumen expansion possibly deforming and absorbing part of the pulse pressure loading which otherwise would be exerted on the weakened aneurysmal wall.

The increased lumen expansion and subsequent ILT deformation does not contradict ILT incompressibility, which is suggested by others and also confirmed in our study population.\textsuperscript{20-23} Specifically, since aortic wall and lumen both expand during the cardiac cycle and ILT is incompressible thus retaining its area, there must be a thinning of ILT from diastole to systole as also suggested by others.\textsuperscript{20} Therefore our results, in agreement with previous studies, indicate a mechanical cushioning effect of ILT on the AAA wall.

Previous ultrasound based studies suggest a positive predictive role of AAA distensibility for rupture risk estimation with higher baseline values and an increase of distensibility over time being associated with a higher rupture rate.\textsuperscript{13} Moreover mechanical testing suggests a significant negative correlation of stiffness with aneurysmal wall strength (the higher the distensibility the lower the strength).\textsuperscript{10,11} Taking into account these findings and the significant positive correlation of $D_{NORM}$ with aneurysm size found here, a higher rupture risk could be expected for AAAs with increased $D_{NORM}$. Our study population presented a wide range of $D_{NORM}$ values even between same sized AAAs. The possible value of this index in assessing susceptibility to rupture should be explored through larger scale studies that will also examine follow-up of such patients.
CONCLUSION

Recording of abdominal aortic deformation during the cardiac cycle for both the unaffected and aneurysmal aortic segments is feasible using current non-invasive imaging techniques. Our results indicate that AAA wall is less deformable than both adjacent non-aneurysmal aortic segment and aneurysmal lumen. This could be due to both altered mechanical properties of the aneurysmal wall and reduced loading due to the presence of ILT. Larger AAAs with higher thrombus accumulation are significantly more distensible than those of smaller size which is in agreement with previous studies indicating an ILT cushioning effect. $D_{NORM}$ may be of value in assessing susceptibility to rupture but this should be verified by larger scale studies.
REFERENCES


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FIGURE-TABLE LEGENDS

Table 1: Demographic and clinical information of patients under examination. Maximum diameter of AAAs and blood pressure measurements at the time of CT scan are also presented.

Table 2: The variables that represent deformation ($RAC_{NAA}$, $RAC_{Wall}$, and $RAC_{Lumen}$) as well as distensibility of the vessel ($D_{NAA}$, $D_{AAA}$ and $D_{NORM}$) are presented for all cases under examination. Statistical significance of differences is also presented.

Figure 1: A. 3D-reconstruction of AAA models from 2D CT images and creation of centerlines.

  B. Perpendicular cross-sections 1mm apart for both aortic wall and lumen at end-diastole and peak-systole. Direct comparison of cross-sectional areas provides a map of aortic deformation.

  C. A representative cross-section and its area change during the cardiac cycle. Such information was recorded for all cross-sections.

Figure 2: $RAC$ of cross sections along the aorta for two representative AAA cases. A. Mapping of $RAC_{Lumen}$ on the 3D reconstruction of the aorta (light opaque grey surface) where the renal arteries can be located, and the lumen is shown with a yellow, darker surface. $RAC_{Lumen}$ is represented with a colored scale on the centerline of the lumen. B. Graph of $RAC_{NAA}$ (green line), $RAC_{Wall}$ (red line) and $RAC_{Lumen}$ (blue line) with distance along the centerline, where the starting point is above the renal arteries and last at the aortic bifurcation. Data of maximum diameter (black, continuous line) are presented.

Figure 3: Correlation of AAA $D_{NORM}$ with A. aneurysm size (maximum diameter) and B. thrombus content (average ILT cross-sectional area). A strong positive correlation was found for both
variables which is compatible with an ILT cushioning effect as explained in the text.
Fig. 1A
Fig. 1C
Fig. 2
**Fig. 3**

- **Top graph**:
  - $D_{NORM}$ vs. Maximum Diameter (mm)
  - Spearman's rho 0.65
  - $P = 0.04$

- **Bottom graph**:
  - $D_{NORM}$ vs. ILT Cross-sectional Area (cm$^2$)
  - Spearman's rho 0.81
  - $P = 0.005$
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**Average** 71.7  **49.4**
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$P=0.12$  $P=0.007$  $P=0.12$  $P=0.005$