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1 **A non-invasive methodology for ATAA rupture risk estimation**

2
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41

42 **ABSTRACT**

43

44 Ascending thoracic aortic aneurysms (ATAA) are a life-threatening pathology provoking an
45 irreversible dilation with a high associated risk of aortic rupture or dissection and death of the
46 patient. Rupture or dissection of ATAAs remains unpredictable and has been documented to
47 occur at diameters less than 4.5 cm for nearly 60% of patients. Other factors than the
48 aneurysm diameter may highly affect the predisposition to rupture. In order to have a better
49 insight in rupture risk prediction, a bulge inflation bench was developed to test ATAAs
50 samples collected on patients during surgical interventions. Preoperative dynamic CT scans
51 on a cohort of 13 patients were analyzed to estimate volumetric and cross-sectional
52 distensibility. A failure criteria based on in vitro ultimate stretch showed a significant
53 correlation with the aortic membrane stiffness deduced from in vivo distensibility. These
54 results reinforce the significance of stretch-based rupture criteria and their possible non-
55 invasive prediction in clinical practice.

56

57 **Keywords:** material property, ascending thoracic aortic aneurysm, brittleness, risk of rupture.

58

59 Word count: 3357

60

61 INTRODUCTION

62

63 Ascending thoracic aortic aneurysm (ATAA) is generally an asymptomatic localized
64 enlargement of the aortic diameter which makes it difficult to detect. Untreated or undetected
65 ATAAs can lead to instantaneous death caused by dissection or rupture of the aneurysm
66 (Johanson et al., 1995, Ramanath et al., 2009). In population-based studies, the annual
67 incidence of aortic dissection ranged from 6 cases per 100,000 in a British study, to 9.1 per
68 100,000 for women and 16.3 per 100,000 for men in a Swedish study (Goldfinger et al.,
69 2014).

70 Concerning ATAA, timely surgery is required; it consists in replacing the diseased aortic
71 segment with a synthetic graft. Elective surgical repair of ATAAs is recommended for
72 diameters larger than 5.5 cm or for fast growing aneurysms (>1 cm per year) for patients
73 without any familial disorders such as Marfan syndrome (Elefteriades et al., 2010). The
74 diameter of 5.5 cm as a criterion for deciding surgical repair is widely acknowledged as
75 insufficient. For instance, the International Registry of Acute Aortic Dissection (IRAD)
76 reported that among 591 type “A” aortic dissections, 59% had a diameter below 5.5 cm (Pape
77 et al., 2007). Several studies dedicated to abdominal aortic aneurysms indicate that
78 biomechanical factors may better predict the risk of rupture than the diameter criterion
79 (Fillinger et al., 2003, McGloughlin, 2011, Leemans et al., 2016).

80 Biomechanical studies have also been achieved to have better insights in ATAAs rupture or
81 dissection (Vorp et al., 2003, Pasta et al., 2016, 2012) and to elucidate the risk profile of the
82 thoracic aorta (Martufi et al., 2016, Trabelsi et al., 2015). Recently, Trabelsi et al. (2016)
83 developed an approach to identify the patient-specific material properties of ATAAs by
84 minimization of the difference between model predictions and gated CT images. Moreover,
85 they characterized the mechanical properties of ATAA on collected samples of patients
86 undergoing surgical repair. Our research group (Duprey et al, 2016) also defined a rupture
87 risk indicator based on the brittleness of the tissue (the rupture criterion is reached when the
88 stretch applied to the tissue is greater than its maximum extensibility or distensibility) and
89 showed a strong correlation between this rupture risk criterion and the physiological elastic
90 modulus of ATAAs estimated within a bulge inflation test. The elastic properties of the aorta,
91 responsible of the Windkessel effect, may change significantly with age or in pathological
92 conditions. The determination of these properties may give an insight toward abnormalities
93 undetectable on aortogram or echocardiogram and may predict the evolution of some diseases
94 (Stefanadis et al., 1990).

95 Aortic distensibility is an accurate and reproducible parameter closely related to the bio-
96 elastic function of the aorta, and can serve as a marker to identify early cardiovascular
97 diseases (Voges et al., 2012, Redheuil et al., 2010). It can be measured from changes in aortic
98 diameter and pulse pressure. The aortic diameter can be measured non-invasively using gated
99 cine-MRI and echocardiographic or invasively using angiographic techniques (Cavalcante et
100 al., 2011, Wilson et al., 2003, Stefanadis et al., 1990). This parameter has been presented and
101 analyzed in several previous studies. It has been studied for healthy subjects (Voges et al.,
102 2012, Stefanadis et al., 1990), patients with Marfan’s syndrome (Adams et al., 1995) or
103 coronary artery diseases (Stefanadis et al., 1990), ascending/descending thoracic aortic
104 aneurysms (Koullias et al., 2005, Cavalcante et al., 2011, Redheuil et al., 2010, Stefanadis et
105 al., 1990), and infrarenal abdominal aortic aneurysms (Wilson et al., 2003). It has been
106 considered by Redheuil et al (2010), as the most sensitive and specific marker of age-related
107 arterial stiffness and dysfunction of large artery in individuals less than 50-year-old.

108 All cited studies have reported that the distensibility can be a reliable measurement related to
109 changes in aortic wall structure, aneurysm growth and rupture.

110 The main objective of this study is to show that the rupture risk of ATAAs could be predicted
111 non-invasively using preoperative dynamic imaging. After acquiring gated CT images on 13
112 patients and after using them to estimate the aortic distensibility, we show that the aortic
113 membrane stiffness deduced from this distensibility is well correlated with a recently
114 proposed stretch-based rupture risk.

115

116 MATERIAL AND METHOD

117

118 This study involves a cohort of 13 patients who underwent elective surgery for ATAA repair
119 at the University Hospital of Saint-Etienne (CHU-SE). Patient records were reviewed to
120 obtain demographic data, medical history, and blood pressure information (Table 1).

121 Neither diabetes nor infections were present in the cohort. One of the patients (Patient 9) had
122 the Marfan syndrome.

123 Following IRB agreement and after informed consent, a pre-operative ECG gated dynamic
124 CT scan was acquired for each patient. The scans were processed to reconstruct the thoracic
125 aorta and aneurysm geometries during the cardiac cycle, including diastole and systole. For
126 each patient, CHU-SE supplied DICOM images of 10 phases throughout the cardiac cycle
127 (resolution: 512x512, slice thickness = 0.5 mm). The lumen of the aneurysm was clearly
128 visible in the DICOM files, but detection of the aneurysm surface was not possible
129 automatically.

130 A non-automatic segmentation of the CT image slices was performed using MIMICS (v.
131 10.01, Materialise NV). The three-dimensional surface of the aorta in each phase was
132 identified and the aneurysm recognized. The three-dimensional (3D) surface of aorta was
133 generated for each phase and exported in STL format. Smoothing factor for all phases was
134 assumed identical. To recognize the systolic and diastolic phases, the luminal volumes of all
135 phases were calculated. The systolic scan was defined as the one with the largest volume and
136 the diastolic scan as the one with the smallest volume. Diastolic and systolic peripheral
137 arterial pressure was obtained by cuff sphygmomanometry before the CT scan.

138

139 The reconstructed surfaces were imported in Rhinoceros (v.4.0, Robert McNeel & Associates)
140 to measure the changes of cross-sectional areas of the ascending and descending thoracic
141 aorta, and the changes of volume of the aortic lumen across the whole ascending aorta,
142 throughout the cardiac cycle. The lumen area was determined by defining a cross section
143 perpendicular to the centerline and measuring its intersection with the segmented aorta using
144 the commercial package MIMICS. The volume was determined as the total volume occupied
145 by luminal voxels located between the cross section taken right after the right coronary artery
146 and the one taken right before the brachiocephalic artery.

147

148 The cross-sectional distensibility D_A ($mmHg^{-1}$) was then estimated using the following
149 formula:

150

$$151 \quad D_A = \frac{A_{max} - A_{min}}{A_{min} * (P_{sys} - P_{dia})} = \frac{\Delta A}{\Delta P} \quad (\text{eq.1})$$

152 where A_{max} and A_{min} represent the maximal and minimal cross-sectional area of the aortic
153 geometry (mm^2), and P_{sys} and P_{dia} represent the systolic and diastolic blood pressure
154 ($mmHg$), respectively.

155 We also introduced the segmental volume distensibility D_V ($mmHg^{-1}$),

156
$$D_V = \frac{V_{max} - V_{min}}{V_{min} * (P_{sys} - P_{dia})} = \frac{\Delta V}{V \Delta P} \text{ (eq. 2)}$$

157 where V_{max} and V_{min} represent the maximal and minimal volumes of the ascending or
 158 descending aortic lumen (mm^3). This value was estimated only for the aneurysmal part of the
 159 ascending aorta.

160 After assessing aortic distensibility, we used the Laplace law to define the tangent membrane
 161 stiffness in the circumferential direction, named $E_{in vivo}$, as:

162
$$E_{in vivo} = \frac{\phi}{D} \text{ (eq. 4)}$$

164 where ϕ is the diameter of the aneurysm measured from the CT scans. D is the distensibility.
 165 Although D is commonly defined using the cross-sectional area variation (in this case denoted
 166 D_A), volume variation were used to estimate $E_{in vivo}$ in the present study (in this case denoted
 167 D_V). A statistical study was performed to compare (D_A) and (D_V). A parametric inference was
 168 assumed comparing the relative volume variation $\Delta V/V$ and the relative cross-sectional area
 169 variation for the ascending and descending aorta $\Delta A/A$. The normal distribution was verified
 170 by the Anderson-Darling test. The analysis of variance (ANOVA) was implemented to
 171 evaluate the influence percentage of these factors. Those statistical methods were performed
 172 using Minitab®.

173 As a membrane stiffness, $E_{in vivo}$ has the dimension of a stress per unit thickness. As there is
 174 no means of measuring accurately the thickness of the aortic wall in vivo, membrane stiffness
 175 was assessed instead of elastic or Young modulus.

177 Additionally to the ECG gated dynamic CT scans, tissue samples were obtained from the 13
 178 patients after ATAA surgical repair. The samples were stored in saline solution at 4°C until
 179 mechanical testing within 24 hours. Immediately before testing, the thickness of each ATAA
 180 was measured at 10 locations; the average thickness h_0 for each patient is reported in Table 2.
 181 All the collected samples were tested in bulge inflation, during which images were recorded
 182 and processed using a stereo Digital Image Correlation (sDIC) system to reconstruct the 3D
 183 shape and deformations of the inflated samples. The experimental set-up and the bulge
 184 inflation tests were explained in details in previous publications (Trabelsi et al., 2015, Romo
 185 et al., 2014, Davis et al., 2015).

186 We defined a tangent membrane stiffness, named $E_{in vitro}$, as:

187
$$E_{in vitro} = \lambda_{1,physio} \frac{d\tau_{1,physio}}{d\lambda_{1,physio}} \text{ (eq.5)}$$

188 Where:

- 190 - τ_1 is the circumferential component of the Kirchhoff stress tensor (the Kirchhoff stress
 191 is the Cauchy stress times the thickness, also named the tension by engineers);
- 192 - λ_1 is the circumferential component of the stretch tensor;
- 193 - $\frac{d\tau_1}{d\lambda_1}$ is the slope of the circumferential stress/stretch curve;
- 194 - $\tau_{1,physio} = (\tau_{1,sys} + \tau_{1,dia})/2$, where $\tau_{1,dia}$ is the value taken by τ_1 at diastole and
 195 $\tau_{1,sys}$ is the value taken by τ_1 at systole;

196 - $\lambda_{1,physio}$ is the value taken by λ_1 when $\tau_1 = \tau_{1,physio}$. $\lambda_{1,physio}$ is the circumferential
 197 stretch occurring in the aorta for a deformation between $P = 0$ and $P = P_{physio}$;
 198 where: $P_{physio} = \frac{P_{sys} + P_{dias}}{2}$.

199
 200 $\tau_{1,dia}$ and $\tau_{1,sys}$ were determined using the following equation based on the Laplace's law:
 201

$$202 \tau_{1,dias/sys} = \frac{P_{dia/sys} * \phi}{2} \quad (\text{eq.6})$$

203
 204 From the bulge inflation tests, we also deduced the circumferential stretch at burst (rupture),
 205 named $\lambda_{1,burst}$. The risk of rupture, denoted $\gamma_{stretch}$, was retrospectively derived for each
 206 patient as the ratio between $\lambda_{1,physio}$ and $\lambda_{1,burst}$. When $\gamma_{stretch}$ is close to one, the specimen
 207 is close to rupture. All the samples had a rupture stretch larger than the stretch at systole
 208 (Duprey et al., 2016).

209
 210 Relationships among different ways of estimating the membrane stiffness (volumetric
 211 distensibility, cross section distensibility, bulge inflation tests), was assessed using Pearson
 212 correlation analysis. A normality test was performed using Anderson-Darling test for the
 213 parametric inference, to determine whether the membrane stiffness of all patients estimated
 214 using each technique followed a normal distribution.

215 One-Way ANOVA was performed to define the statistical differences between methods. The
 216 results are presented as mean \pm typical error. The statistical significance was set at $p < 0.05$.
 217 The applied statistical methods were calculated using Minitab®.

218
 219 The objective of the present study is to evaluate the correlation between $\gamma_{stretch}$, $E_{in vitro}$ and
 220 $E_{in vivo}$.

221 222 RESULTS

223
 224 A statistical analysis revealed that $E_{in vitro}$, $E_{in vivo}$ estimated using D_V or D_A , and $\gamma_{stretch}$,
 225 have a normal distribution for a pre-determined level of significance of 95% ($p \geq 0.05$)
 226 ($p(E_{in vitro})=0.103$; $p(E_{in vivo}$ using $D_V)=0.081$; $p(E_{in vivo}$ using $D_A)=0.332$;
 227 $p(\gamma_{stretch})=0.395$).

228 It also showed a non-significant difference between all membrane stiffness values, whether
 229 estimated in vitro ($E_{in vitro}$) or estimated in vivo using the distensibility ($E_{in vivo}$) ($p=0.05$, S
 230 $= 2.577$, $R^2= 16.69\%$). This means that the membrane stiffness of the ATAA is independent
 231 from the way of estimating it.

232
 233 Distensibility values were calculated and analyzed for both ascending and descending thoracic
 234 aorta, using volume, and cross section variations of the aorta during the cardiac cycle (Figure
 235 1).

236 To verify tendencies published in the literature (Koullias et al., 2005, Voges et al., 2012,
 237 Redheuil et al., 2010), distensibility variation with age and aneurysm diameter were plotted.
 238 Figure (2) confirms these tendencies and shows a decrease of the distensibility when age or
 239 maximal diameter of the aneurysm increases.

240 A plot of $E_{in vitro}$ versus $\gamma_{stretch}$ for the 13 patients of this study is displayed in Figure 3,
 241 showing that an exponential fit can give a coefficient of determination of 0.88.

242 By analyzing the volume and the cross-sectional area variation of the ascending aorta of all
243 patients, it was shown that the values of distensibility estimated from the cross section area is
244 generally smaller compared to distensibility estimated from volume variation (Figure 4-A).
245 Moreover, by comparing the cross section area variation, it can be seen that it is always lower
246 for the ascending aorta than for the descending one. Only patient 9 showed a different trend.
247 In addition to BAV, patient 9 had a coarctation in the descending thoracic aorta (Figure 7-B).
248 By comparing the ascending and the descending aorta, we showed that the ATA is stiffer than
249 the DTA which is a direct consequence of the aneurysm formation.

250
251 Statistical study and data analysis showed that estimating the in vivo tangent physiological
252 membrane stiffness by estimating the distensibility (D_V), or by estimating the distensibility
253 (D_A) of the ascending aorta between systole and diastole, makes no significant differences.
254 The statistical analyses showed that a normal distribution was satisfied with a p-value larger
255 than 0.05 and that there is no significant difference between $\Delta V/V$ and $\Delta A/A$ in the ascending
256 thoracic aorta (Figure 4-bis). Moreover, the data distribution of $\Delta V/V$ is less dispersed than
257 the one of $\Delta A/A$ and it does not present any outlier (Figure 4-bis). For these reasons, D_V was
258 preferred to D_A in the stiffness analysis. Results also showed that no difference exists in $\Delta A/A$
259 between the ascending and the descending thoracic aorta (Figure 4).

260 Results from patient 9 (with Marfan syndrome) were systematically excluded from the
261 statistical study.

262
263 Figure 5 shows a linear correlation with a slope close to 1 between the in vitro tangent
264 membrane stiffness and the in vivo membrane stiffness obtained using D_V .

265 This results leads to a direct relationship between the in vivo tangent membrane stiffness and
266 the proposed index of rupture $\gamma_{stretch}$ (Figure 6). The intersection of the correlation between
267 D_V and $E_{in vivo}$ is around 3 MPa.

268 If we compare Figure 3 and Figure 6, an exponential tendency between $\gamma_{stretch}$ and both in
269 vivo and in vitro tangential membrane stiffness of the aorta, with very similar fitting curves is
270 shown. The coefficient of determination is smaller in Figure 6 than in Figure 3, with a
271 coefficient of determination $R^2=44\%$ and $R^2=88\%$, respectively.

272

273 DISCUSSION

274 Our research group (Duprey et al., 2016) recently defined a novel rupture risk
275 indicator, $\gamma_{stretch}$, based on the brittleness of the tissue (the rupture criterion is reached when
276 the stretch applied to the tissue is greater than its maximum extensibility or distensibility) and
277 showed a significant correlation between this rupture risk criterion and the physiological
278 elastic modulus of ATAAs estimated within a bulge inflation test. In the present analysis, we
279 show interestingly that $\gamma_{stretch}$ was also strongly correlated to the membrane stiffness of the
280 ATAA that we assessed non-invasively from dynamic CT-scans by estimating the local aortic
281 distensibility. This is a result with major significance as it means that the local aortic
282 distensibility is a relevant measure to deduce the risk of rupture of ATAA. Other saying, the
283 risk of rupture could be predicted in advance using measures of aortic distensibility.

284

285 Aortic distensibilities were reported by several authors, although none had shown previously
286 that deducing the membrane stiffness could be relevant to evaluate the risk of rupture of
287 ATAA. It is worth mentioning that aortic distensibilities estimated for the 13 patients of our
288 study, whether based on cross section area or volume variations, showed similar tendencies as

289 what was previously found in the literature. In fact, the distensibility decreases with age (see
290 Figure 2), as demonstrated by Voges et al. (2012) and Redheuil et al. (2010). Distensibility
291 decreases when aneurysmal diameter increases as published by Koullias et al. (2005) (see
292 Figure 2).

293
294 The aorta, through its elastic properties, actively participates into the propulsion of the blood
295 and interplays with the left ventricle in a “game of catch” with the stroke volume. A properly
296 functioning aorta actually unloads the left ventricle (Koullias et al., 2005). Therefore, the loss
297 of elasticity of the ascending thoracic aorta means a loss of function which may induce many
298 clinical consequences such as diastolic LV dysfunction with dyspnea, predisposition to
299 angina, and heart failure, and small vessel degeneration in brain and kidney with intellectual
300 deterioration and renal failure (O’rourke et al., 2007).

301
302 We claim here that the loss of elasticity also increases the risk of rupture. To better understand
303 this, rupture must be defined in terms of maximum stretch. Rupture of the aortic tissue is
304 traditionally defined when the maximum stress that the tissue can withstand is reached.
305 However, when we derived the maximum stress ratio between the stress applied to the tissue
306 in vivo and its strength, we noted that most of the collected ATAA samples were far from
307 rupture (Duprey et al., 2016). The stretch-based definition of rupture, which is equivalent,
308 states that rupture occurs when the stretch applied to the tissue exceeds its maximum
309 extensibility or distensibility. This way of defining rupture can be more physiologically
310 meaningful as it is reported that aneurysm rupture or dissections often occur at a time of
311 severe emotional stress or physical exertion (Martin et al., 2013). Such emotionally or
312 physically stressful situations can induce significant changes of blood volumes in the aorta,
313 making more compliant aneurysms less prone to rupture as they can sustain such changes of
314 volume. Based on this analysis, Martin et al. also defined a similar criterion, named the
315 diameter risk, which is the ratio between the current diameter of the aneurysm and the rupture
316 diameter (Martin et al., 2013). Like us, they also showed that the diameter risk increases
317 significantly with the physiological elastic modulus of the artery. Indeed, if the aortic wall is
318 stiff, a rather large increase of pressure can be induced by a small increase of volume.

319
320 The main results of this study is accordingly the tendency found between $\gamma_{stretch}$ and
321 $E_{in vivo}$, but this tendency shows a low R^2 . This result is still insufficient to use $\gamma_{stretch}$ in a
322 clinical setting and propose it as a new criterion for aneurysm rupture prediction. Further
323 exploration of rupture index based on stretch is needed before possible clinical application.
324 This is probably due to several limitations that should be addressed. The major one is the fact
325 that $E_{in vivo}$ was estimated globally and homogenously across the ascending thoracic aorta
326 through its volume variations. There was a higher R^2 for the tendency between $\gamma_{stretch}$ and
327 $E_{in vitro}$, but $E_{in vitro}$ and $\gamma_{stretch}$ were both estimated locally on the small 30x30mm²
328 sample tested in the bulge inflation device. Our group is currently working on an inverse
329 method to deduce the local membrane stiffness in vivo using dynamic CT scan (Bersi et al.,
330 2016).

331 Another difference to notice between in vivo and in vitro behavior is that, despite the
332 significant correlation between $E_{in vivo}$ and $E_{in vitro}$, $E_{in vivo}$ averagely overestimated
333 $E_{in vitro}$ (Figure 6). A possible explanation is the contribution of the perivascular environment
334 in vivo, which was not taken into account here to derive the in vivo membrane stiffness.

335 Possible inter-individual variations of the perivascular effects may also contribute to lower R^2
336 tendency between $\gamma_{stretch}$ and $E_{in vivo}$.

337 We mostly focused on the circumferential membrane stiffness, derived from aortic
338 distensibility. The longitudinal extensibility of the ascending thoracic aorta would also be
339 interesting for characterizing the elastic properties of ATAAs. The longitudinal extensibility
340 of the ascending thoracic aorta is caused by the heart movement during the cardiac cycle
341 (Beller et al., 2004, García-Herrera et al., 2013). The change of length could be measured by
342 tracking the position of anatomical reference such as the sinotubular junction and the
343 brachiocephalic artery at every phase throughout the cardiac cycle (see Figure 7).

344 Finally, our rupture index is the ratio between the maximum hoop stretch and in vivo hoop
345 stretch of the aorta. We need to investigate further what would be the effect of the presence of
346 residual stresses on such rupture index based on stretch.

347 **CONCLUSION**

348 In this paper, ATAA rupture was defined as the time when the stretch applied to the tissue
349 exceeds its maximum extensibility or distensibility. We showed interestingly that the
350 membrane stiffness of ATAA, assessable non-invasively from dynamic CT-scans by
351 estimating the local aortic distensibility, is a relevant measure to deduce this stretch-based risk
352 of rupture. Other saying, the risk of rupture of an ATAA could be predicted in advance using
353 measures of aortic distensibility. Despite the major interest of this result, it would be
354 interesting in future studies to consider also the longitudinal extensibility of the ascending
355 thoracic aorta to better characterize the elastic properties of ATAAs.

356

357 **CONFLICT OF INTEREST**

358

359 The authors have no conflicts of Interest to declare concerning the contents of this manuscript.

360

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470

471 **Tables**

472 Table 1. Patient clinical information

Patient Id	Sex/ Valve	Age (year)	ΔP (kPa)	Aneurysm diameter (mm)
1	M/BAV	58	8.86	65
2	M/TAV	78	5.34	51
3	M/BAV	61	11.5	52
4	M/TAV	69	5.34	50
5	M/BAV	70	5.8	51
6	M/TAV	81	5.34	50
7	M/TAV	84	13.4	55
8	M/TAV	74	10.68	51
9*	M/BAV	27	8.66	50
10	M/TAV	42	5.34	55
11	M/TAV	81	5.34	52
12	F/TAV	78	5.34	65
13	M/BAV	57	5.34	55
Mean		66.2	7.4	54
STD		16.9	2.9	5.2

473 BAV: Bicuspid aortic valve, TAV: Tricuspid aortic valve

474 *Patient with genetic disorder (Marfan syndrome)

Table 2: Geometric and biomechanical parameters measured for the 13 patients

Patient	h_0 (mm)	$E_{in\ vitro}$ (mm*MPa)	$\gamma_{stretch}$	$\Delta A/A(asc)$	$\Delta V/V(asc)$	$\Delta A/A (desc)$	D_V (mmHg ⁻¹)	D_A (mmHg ⁻¹)	$E_{in\ vivo} (D_V)$ (mm*MPa)	$E_{in\ vivo} (D_A)$ (mm*MPa)
1	2.10	1.216	0.87	0.080	0.08	0.160	0.055	0.001	7.199	7.199
2	1.98	2.202	0.92	0.100	0.11	0.110	0.062	0.002	2.476	2.723
3	2.74	0.922	0.88	0.110	0.13	0.120	0.043	0.001	4.600	5.436
4	1.86	*0.866	0.84	0.160	0.21	0.290	0.060	0.004	1.271	1.669
5	2.50	3.996	0.95	0.110	0.06	0.110	0.051	0.003	4.930	2.689
6	2.72	1.237	0.84	0.090	0.10	0.190	0.041	0.002	2.670	2.967
7	1.79	3.314	0.94	0.090	0.12	0.160	0.070	0.001	6.142	8.189
8	1.90	3.437	0.97	0.059	0.13	0.103	0.068	0.001	4.190	9.185
9	1.77	*0.569	0.81	0.210	0.22	0.170	0.061	0.003	1.968	2.062
10	1.59	1.026	0.88	0.054	0.16	0.086	0.074	0.001	1.836	5.396
11	1.76	5.752	0.96	0.039	0.02	0.075	0.073	0.001	11.570	7.193
12	2.41	4.124	0.95	0.034	0.02	0.034	0.053	0.001	**17.355	10.089
13	2.35	1.271	0.89	0.073	0.11	0.098	0.050	0.002	2.557	3.999
Mean	2.11	2.591	0.90	0.093	0.11	0.131	0.059	0.002	4.284	5.292
STD	0.39	1.625	0.05	0.049	0.06	0.064	0.011	0.001	2.936	2.856

*Atypical value (observation) **Statistical outlier

Figures

Figure 1: Method for estimating aortic distensibility using cross-section area and volume variation between systole and diastole. A- Ascending and descending cross-section area variation between systole and diastole. B- Ascending volume variation between systole and diastole.

Figure 2: Sectional and volumetric ascending aorta distensibility variation respect to (A) age and (B) maximal aneurysmal diameter. A- Distensibility variation with age. B- Distensibility variation with aneurysm diameter.

Figure 3: Correlation between the index of $\gamma_{stretch}$ and the tangent elastic modulus $E_{in\ vitro}$

Figure 4: Distensibility estimation using volume or cross section area variation between systole and diastole. A- Ascending aorta volume and cross section area. B- Cross-sectional area variation between the ascending and the descending aorta.

Figure 4-bis: A statistical comparison between the relative volume variation for the ascending aorta $\Delta V/V(asc)$ and the relative cross-sectional area variation for the ascending and descending aorta $\Delta A/A(asc)$ and $\Delta A/A(desc)$

Figure 5: Correlation between the in vivo and in vitro tangent membrane stiffness.

Figure 6: Correlation between the in vivo tangent membrane stiffness and the index of rupture $\gamma_{stretch}$.

Figure 7: Movement of the aorta in a healthy subject and in an ATAA patient. A- Healthy aorta: Movement of the carotids and left subclavian artery between the systolic and diastolic phase. B-Marfan patient (Patient 9): Movement of the right coronary between the systolic (2) and diastolic (1) phase. A stenosis in the descending aorta is visible.















