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Research Article

Keywords: porcine aorta, aortic biomechanics, regional heterogeneity, mechanical anisotropy, uniaxial tensile testing

Posted Date: October 19th, 2026

DOI: <https://doi.org/10.21203/rs.3.rs-11059027/v1>

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Additional Declarations: No competing interests reported.

Biomechanical properties of the porcine thoracic aorta: An in-depth analysis of elastic modulus, stress-strain behavior and tissue toughness with regard to anatomical location and loading orientation

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Abstract: The mechanical integrity of the aortic wall is crucial for its physiological function and susceptibility to vascular pathologies. This study provides a systematic biomechanical characterization of the porcine aorta, focusing on regional and orientation-dependent variations in stiffness, strength, and tissue toughness. A total of 480 specimens from ten porcine aortas were obtained from defined anatomical regions and subjected to uniaxial tensile testing in circumferential and longitudinal orientations. No significant differences were found regarding specimen position (anterior, posterior, left, right), whereas biomechanical properties varied along the cranial–caudal axis in a parameter- and orientation-dependent manner. The high-strain modulus, and rupture stress in longitudinal specimens, increased towards the distal aorta, with the most pronounced differences occurring between proximal and distal segments. Toughness decreased distally in circumferential specimens but increased in longitudinal specimens. Circumferential specimens exhibited higher stiffness, strength, and toughness at high strains, whereas at low strains longitudinal specimens tended to be stiffer. In longitudinal specimens, the intima consistently ruptured first, followed by the media and adventitia at higher strains; in circumferential specimens, all three layers ruptured simultaneously. These findings demonstrate distinct regional, orientation-, and strain-dependent biomechanical behavior of the porcine aorta and emphasize the importance of anatomical location and loading direction in experimental aortic biomechanics, biomechanical modeling, and the development of vascular devices.

Keywords: porcine aorta, aortic biomechanics, regional heterogeneity, mechanical anisotropy, uniaxial tensile testing

1 Introduction

Aortic dissection is an acute and life-threatening disease of the aortic wall that predominantly affects the thoracic aorta. In Western populations, its annual incidence has been estimated at approximately 3–5 cases per 100,000 individuals and may increase to approximately 35 cases per 100,000 per year in individuals aged 65–75 years (Nienaber et al., 2016). Approximately two-thirds of aortic dissections involve the ascending aorta and may propagate through the aortic arch into the descending thoracic aorta (Ince and Nienaber, 2007; Nienaber et al., 2016). Since initiation and propagation of a dissection are closely related to the mechanical response and failure of the aortic wall, a detailed characterization of thoracic aortic tissue is essential for understanding the underlying biomechanical mechanisms.

Biomechanical characterization of the aortic wall is also highly relevant for other aortic diseases, particularly abdominal aortic aneurysms (AAA). AAA represents a considerable health burden and is associated with high mortality in the event of rupture. Despite advances in treatment, overall mortality of ruptured AAA remains high, emphasizing the need for improved and more individualized methods of assessing aortic disease and rupture risk (Gawenda and Brunkwall, 2012; Kühnl et al., 2017). These outcomes emphasize the need for improved and more individualized methods of assessing aortic disease and rupture risk.

Computational approaches such as finite element method (FEM)-based models provide a means of estimating patient-specific biomechanical parameters, including aortic wall stress and rupture-related indices (Erhart et al., 2015; Lindquist Liljeqvist et al., 2017; Polzer et al., 2020). However, the predictive value of these models strongly depends on the material properties assigned to the aortic wall. The aorta is mechanically heterogeneous and anisotropic, and its properties may vary according to anatomical location, loading direction and tissue microstructure. Consequently, accurate regional material data are required to improve biomechanical models of both thoracic and abdominal aortic disease.

Systematic characterization of human aortic tissue is limited by specimen availability, ethical constraints and substantial interindividual variability. Porcine aortas are therefore frequently used as experimental models because of their anatomical, structural and biomechanical similarities to the human aorta (Martin et al., 2011; De Beaufort et al., 2018). Previous studies have investigated the anisotropic response of porcine thoracic aortic tissue (Myneni et al., 2022), its behavior under physiological and supraphysiological pressures (Rastgar Agah et al., 2017), regional and layer-specific deformation (Sokolis, 2019), relationships between microstructure and mechanical properties (Mureşan et al., 2024), and fatigue behavior (Polzer and Vitásek, 2023).

Despite these studies, a systematic characterization of mechanical properties along the entire porcine thoracic aorta remains limited. In particular, comparatively little information is available on how elastic modulus, ultimate stress, strain at failure and tissue toughness vary with longitudinal anatomical location, circumferential position and loading orientation. Furthermore, only a limited number of studies have characterized porcine aortic tissue up to complete failure, and direct comparison is complicated by differences in experimental methodology and animal characteristics. For example, (Myneni et al., 2022) did not report maximum stress, (Peña et al., 2019) investigated animals that are not directly comparable with those used in the present study, and (Alloisio et al., 2025, 2024) used compact-tension tests of notched specimens rather than uniaxial tensile testing to failure.

Therefore, the aim of the present study was to provide an in-depth biomechanical characterization of the porcine thoracic aorta and to determine the influence of anatomical location, specimen position, and loading orientation on elastic modulus, stress–strain behavior, and tissue toughness. We hypothesized that the investigated biomechanical parameters would change systematically along the cranial–caudal axis, with higher stiffness and strength in more distal segments, and biomechanical properties would differ between anatomical wall positions (posterior, anterior, left, and right) and between circumferential and longitudinal loading orientations.

2 Materials and Methods

2.1 Specimen acquisition, storage and preparation

Porcine thoracic aortas (N = 10) from pigs (170 to 210 days old and weight of approximately 110 kg to 12 kg) were obtained from a local slaughterhouse. Fat and other tissue was removed with surgical scalpels, tweezers and scissors from the aorta and stored in a freezer at -20 °C in physiological buffer (Krebs-Henseleit-Buffer – Sigma-Aldrich, St. Louis, MO, USA). Before testing, an aorta was slowly thawed in a fridge at 4 °C for approximately 12 to 24h. At the day of the experiment the specimens were prepared at room temperature ($\approx +20^{\circ}\text{C}$) and stored in well plates with Ringer's solution till the actual measurement.

While some studies suggest freezing, even up to 12 month, maintain tissue functionality and does not have significant implications on the measured Young's Modulus as well as other key mechanical properties (Stemper et al., 2007; O'Leary et al., 2014), others present a more complex picture. For example, some research indicates that cold storage can increased the average incremental elastic modulus of the aortic tissue along circumferential direction; however, the difference was not significant along axial direction (Zhao et al., 2015). Another study points out that the storage at -20 °C increases the Young's Modulus (Chow and Zhang, 2011). Regardless potential effects, which will be further discussed below, storing our tissue at -20 °C was a necessary compromise to manage the logistics of specimen acquisition and testing. However, in order to minimize the influence as much as possible, all aortas were measured within one month of being removed from the animal.

2.2 Classification and categorization of specimens

Each aorta was divided into six segments, each 5 cm long, starting at the ascending aorta near the coronary ostia. The aortas were cut lengthwise and in each segment a total of 8 patches were acquired, each then split in a longitudinal and a circumferential specimen, respectively. The initial gauge length between the clamps was set to $l_0 = 10$ mm, while the remaining tissue was used for clamping. The specimen width was 5 mm. The specimen thickness, to calculate the cross-sectional area, was measured with a caliper. To further differentiate the aorta, the four specimens from each direction were divided according to their position within the aorta. In terms of handling and preparing a total of four positions – P: Posterior, R: Right, A: Anterior, L: Left – were determined (see Fig. 1B). Thus, a clear classification of each specimen in the aorta was possible, following the rule: AortaNr. – Section – Cut – Position. For example, 3-4-L-R denotes a specimen from the third measured aorta, segment four in a longitudinal cut from the right position. In total, forty-eight specimens were taken from each aorta, resulting in 480 specimens analyzed in this study.

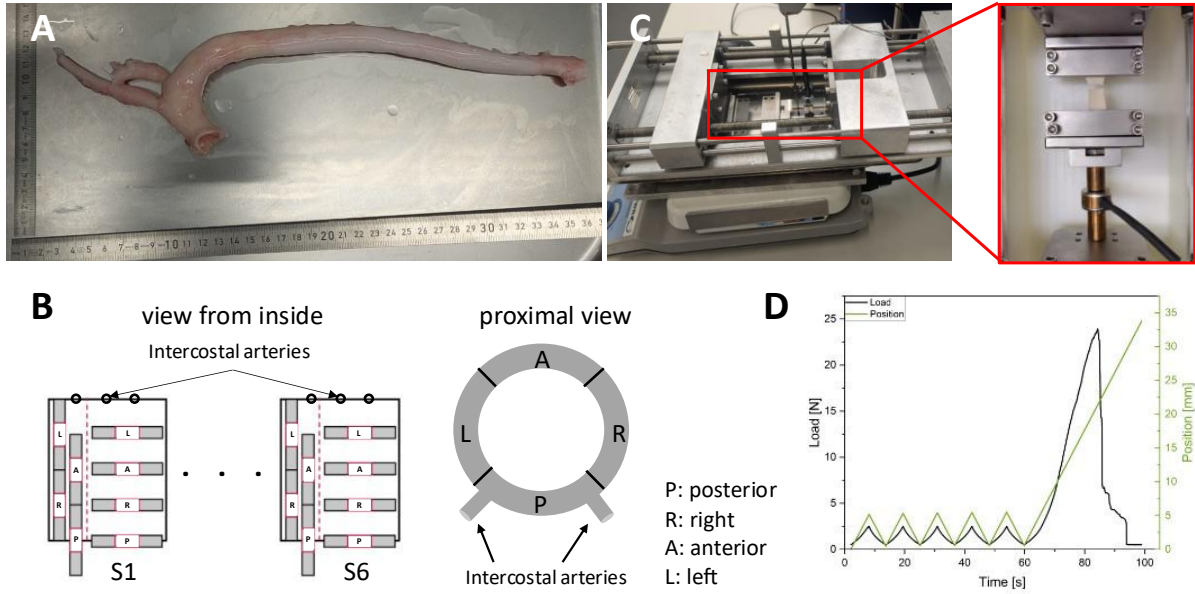


Fig 1: Overview of the classification of the specimen and the measurement setup. **A:** Porcine aorta separated from fat and other tissue, divided into six segments, each 5 cm long. Starting point was the ascending aorta near the coronary ostia. **B:** Each segment was subdivided and a total of eight patches were taken. Four specimens each were excised circumferentially and longitudinally of the aorta with a scalpel. In addition, the specimens were subdivided according to their position in the aorta. P - Posterior, L - Left, A - Anterior, R - Right. **C:** The specimens were measured with a uniaxial tensile testing unit from ADMET (10 lbs (≈ 44.5 N) loadcell) in Ringer's solution at 37°C. The tissue is clamped in the tensile testing unit using clamping jaws and pulled apart at 0.5 mm/s after pre-stretching (10 times from 70 mmHg to 140 mmHg converted into a force F using the diameter of the Segment and Laplace's law) until rupture. **D:** Position x and force F are recorded as a function of time using position sensors and load cell.

2.3 Tensile testing unit and test protocol

Uniaxial tensile tests were performed using an ADMET tensile testing system (Norwood, MA, USA) equipped with a 10-lb-capacity load cell (approximately 44.5 N; Honeywell, Morristown, NJ, USA). The specimens were mounted between two clamps. To reduce tissue damage and specimen slippage, the original metal clamps were replaced with custom-designed, 3D-printed clamps. The clamps were inspected regularly and replaced when signs of wear became apparent.

The tissue specimens and clamps were immersed in Ringer's solution throughout the experiment. The bath was maintained at 37°C using a laboratory magnetic stirrer with an integrated heating plate (Fig. 1C). To protect the load cell from contact with the bath solution, a custom 3D-printed protective chamber was installed around the sensor. Any liquid entering the chamber was continuously removed using a small peristaltic pump.

Positions sensors of the tensile testing unit and the load cell measured the position x [mm] and respective applied force F [N] as a function of time t [s]. As proposed elsewhere (Valente et al., 2024), pre-stretching cycles (10 times) were applied to precondition the specimen and to achieve a consistent and reproducible mechanical data. In contrast to using fixed force limits for all specimens, the lower and upper force thresholds were determined individually from the original aortic diameter using

Laplace's law. Target pressures of 70 and 140 mmHg were used to represent the lower and upper limits of the preconditioning cycles, respectively. The corresponding specimen-specific forces were calculated as

$$F = \frac{1}{2}pwd$$

where p is the target pressure, w is the specimen width, and d is the inner aortic diameter. Thus, specimens obtained from aortas with different diameters were preconditioned at force levels corresponding to the same pressure range rather than at identical absolute forces.

Following cyclic preconditioning, the specimens were monotonically elongated until rupture. A constant displacement rate of 0.5 mm/s was applied during all measurements. Figure 1D presents a representative force–displacement curve showing the cyclic preconditioning phase, the subsequent loading-to-failure phase, and the detected rupture peak.

2.4 Post processing of raw data

The acquired raw data was post processed using Python (version 3.11), managed via the Anaconda distribution (version 2.7.0). Scripts were developed in the Spyder IDE (version 5.4.3). First of all, the initial cross-sectional area A_0 was calculated as

$$A_0 = w * t ,$$

where w and t denote the width and thickness of the specimen in the reference configuration, respectively. For consistency, our specimens were always prepared with a width of 5 mm. Given the initial length of $l_0 = 10$ mm and the change in length Δl during the measurement, the strain

$$\varepsilon = \frac{\Delta l}{l_0}$$

was defined. As our deformation is finite, the logarithmic or true strain

$$\varepsilon_T = \ln(1 + \varepsilon) = \ln\left(\frac{l}{l_0}\right)$$

expresses more adequately tissue deformation (Faridmehr et al., 2014; Xuan et al., 2021). In addition, given the measured force F and the cross-sectional area A_0 , we compute the nominal or engineering stress (Korhonen and Saarakkala, 2011)

$$\sigma = \frac{F}{A_0}.$$

During the experiment the tissues cross-sectional area decreases as the elongation increases (Xiong et al., 2008), such that under the assumption of incompressibility, the true or Cauchy stress reads

$$\sigma_T = \sigma(1 + \varepsilon).$$

The stress–strain analysis was performed on the monotonic loading phase following cyclic preconditioning and preceding the detected rupture point. Because aortic tissue exhibits a nonlinear, strain-stiffening response, the material stiffness was characterized using both low-strain and high-strain moduli. The low-strain region was defined as 10–30% and the high-strain region as 70–90% of the strain interval between the onset of monotonic loading and the maximum strain reached before

rupture. Within each region, the modulus was determined as the slope of a linear least-squares regression,

$$E = \frac{\partial \sigma}{\partial \varepsilon} = \text{const.}$$

or for true stress and true strain,

$$E_T = \frac{\partial \sigma_T}{\partial \varepsilon_T} = \text{const.}$$

Contiguous fitting windows within the predefined strain regions were evaluated automatically. Only fits with a positive slope and sufficient linearity were accepted, and the fit with the highest coefficient of determination was selected. If the standard fitting criteria could not be fulfilled, predefined fallback criteria with progressively reduced minimum window sizes and R^2 thresholds were applied. Engineering and true low-strain and high-strain moduli were determined independently. A detailed description of the underlying continuum-mechanical principles is provided in Chapter 3 of Gasser's *Vascular Biomechanics: Concepts, Models, and Applications* (Gasser, 2021). The energy absorbed by the specimen until rupture was quantified as the area under the engineering stress–strain curve:

$$W = \int_{\varepsilon_0}^{\varepsilon_r} \sigma d\varepsilon$$

Where ε_r denotes the engineering strain at rupture. The rupture point was identified as the global maximum of the monotonic loading curve. The corresponding engineering rupture strain, engineering rupture stress, low-strain modulus, high-strain modulus, and toughness were extracted for each specimen. The same rupture criterion was applied to both circumferential and longitudinal specimens. The following lists only the parameters for the engineering parameters, the values for the true stress-strain relationship can be found in the supporting information.

2.5 Statistics

Data are presented as mean $\pm$ standard deviation. Differences between aortic segments and specimen orientations were visualized using box plots. Statistical differences between groups were assessed using analysis of variance (ANOVA), followed by Tukey's honestly significant difference post hoc test for pairwise comparisons. A p-value below 0.05 was considered statistically significant. Statistical significance is indicated as follows: *: $p < 0.05$, **: $p < 0.01$, ***: $p < 0.001$. The exact p-values can be found in the supporting information. All statistical analyses were performed using Python 3.11 within the Anaconda distribution (version 2.7.0). The analysis scripts were developed and executed in the Spyder integrated development environment (version 5.4.3).

3 Results

3.1 Size of aortas

In Figure 2 the wall thickness and inner diameter (mean and standard deviation) of the measured aortas ($N = 10$) in regard to the distance to the coronary ostia in distal direction is shown. In addition, the boundaries of the different segments are marked with thin dashed upright lines. The inner diameter was measured at the beginning and ending of each segment, while the specimen thickness (and therefore the vessel wall thickness) was averaged separately across the four circumferential and the four longitudinal specimens within each segment, yielding two thickness values per segment that correspond to the circumferential and longitudinal sub-regions from which the respective specimens

were excised (see Fig. 1). The wall thickness of the aorta decreases with increasing distance to the coronary ostia whereas the inner diameter first increases, reaches its maximum value of 18 mm in the aortic arch, and then steadily decreases. The wall thickness of the porcine aorta is about 2.5 mm in the ascending aorta and decreases to 1.3 mm at the coeliac trunk. The inner diameter of the aorta also decreases from 17 mm to 11 mm, after an increase to 18 mm in the aortic arch, from cranial to caudal direction.

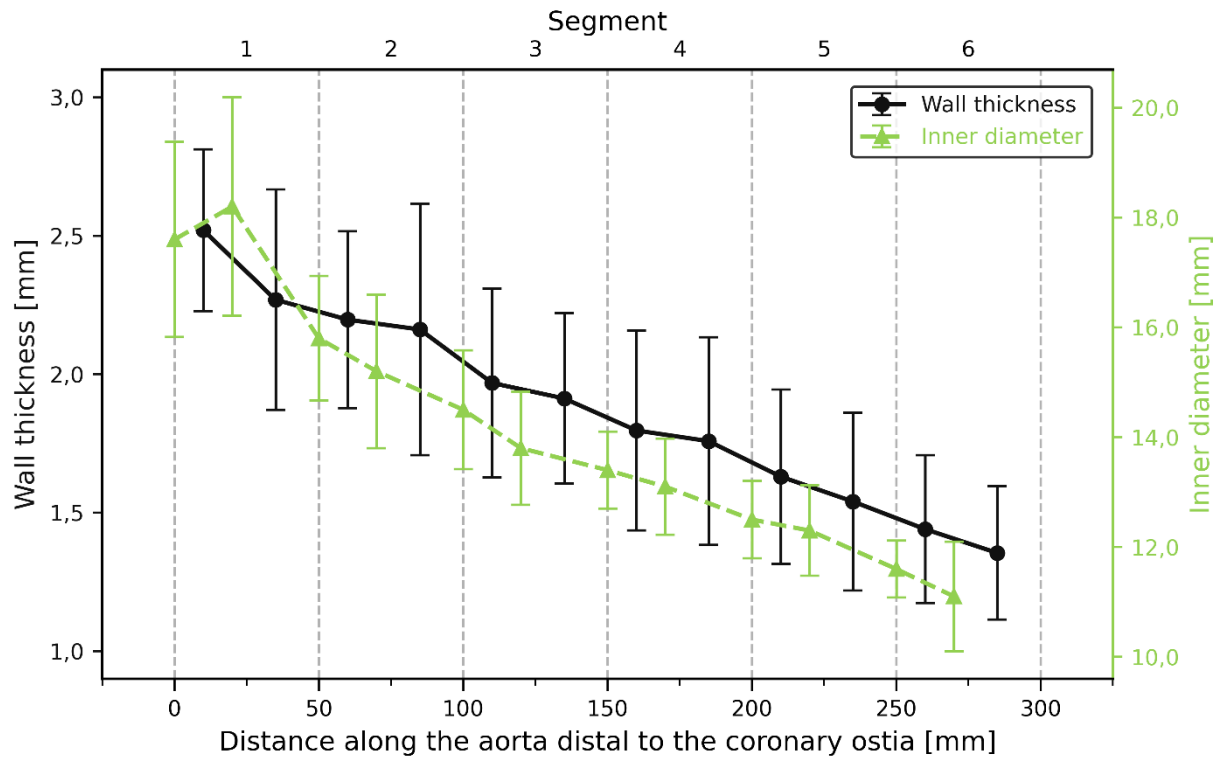


Fig. 2: Wall thickness and inner diameter of porcine aortas as function of distance from the coronary ostia (mean $\pm$ stdv, N = 10 aortas). Wall thickness values represent the mean across the four circumferential or four longitudinal specimens within each segment, respectively; inner diameter was measured at the beginning and end of each segment

3.2 Homogeneity of Young's moduli at individual anatomical positions

Figure 3 presents the Young's Moduli (low- and high-strain as mean and standard deviation) of each position (P, L, A, R) and segment. Individual segments are covered by separate plots illustrating four positions, specimen alignment and orientation, respectively. The variation in each segment and specimen is relatively low and the evaluated ANOVA indicated no significances and therefore the position was no longer taken into account for the subsequent calculations.

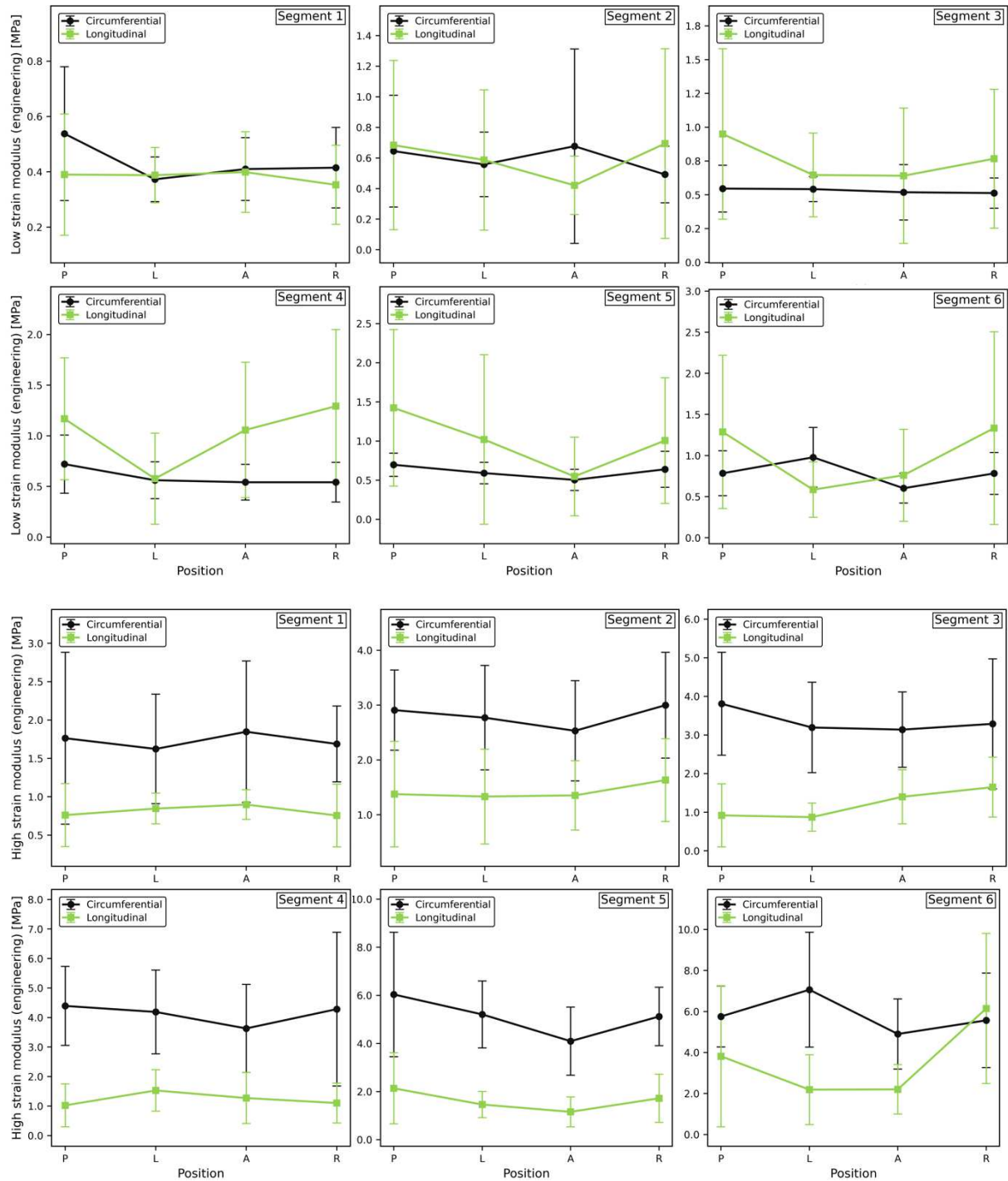


Fig. 3: Young's moduli as function of position and segment (N = 10 aortas) for longitudinal and circumferential cuts for low- and high-strain moduli. No significant differences of the shown mean values were obtained.

3.3 Regional and orientation-dependent differences in Young's moduli

To characterize the strain-dependent mechanical behavior of the aortic wall, the low-strain and high-strain moduli were compared across the six aortic segments (S1–S6) and between circumferential and longitudinal specimens (Fig. 4, Fig. 5). One-way ANOVA on the aorta-aggregated mean values (n = 10) confirmed significant main effects of segment for both the low-strain ($F(5,54) = 6.39$, $p < 0.001$, $\eta^2 = 0.372$) and high-strain modulus ($F(5,54) = 26.28$, $p < 0.001$, $\eta^2 = 0.709$), while position (anterior,

posterior, left, right) had no significant effect on either parameter (low-strain: $p = 0.063$; high-strain: $p = 0.124$). Loading direction significantly affected both moduli, but with opposite directionality: the low-strain modulus was on average higher in longitudinal specimens ($F(1,18) = 5.40$, $p = 0.032$, $\eta^2 = 0.231$), whereas the high-strain modulus was markedly higher in circumferential specimens ($F(1,18) = 78.86$, $p < 0.001$, $\eta^2 = 0.814$).

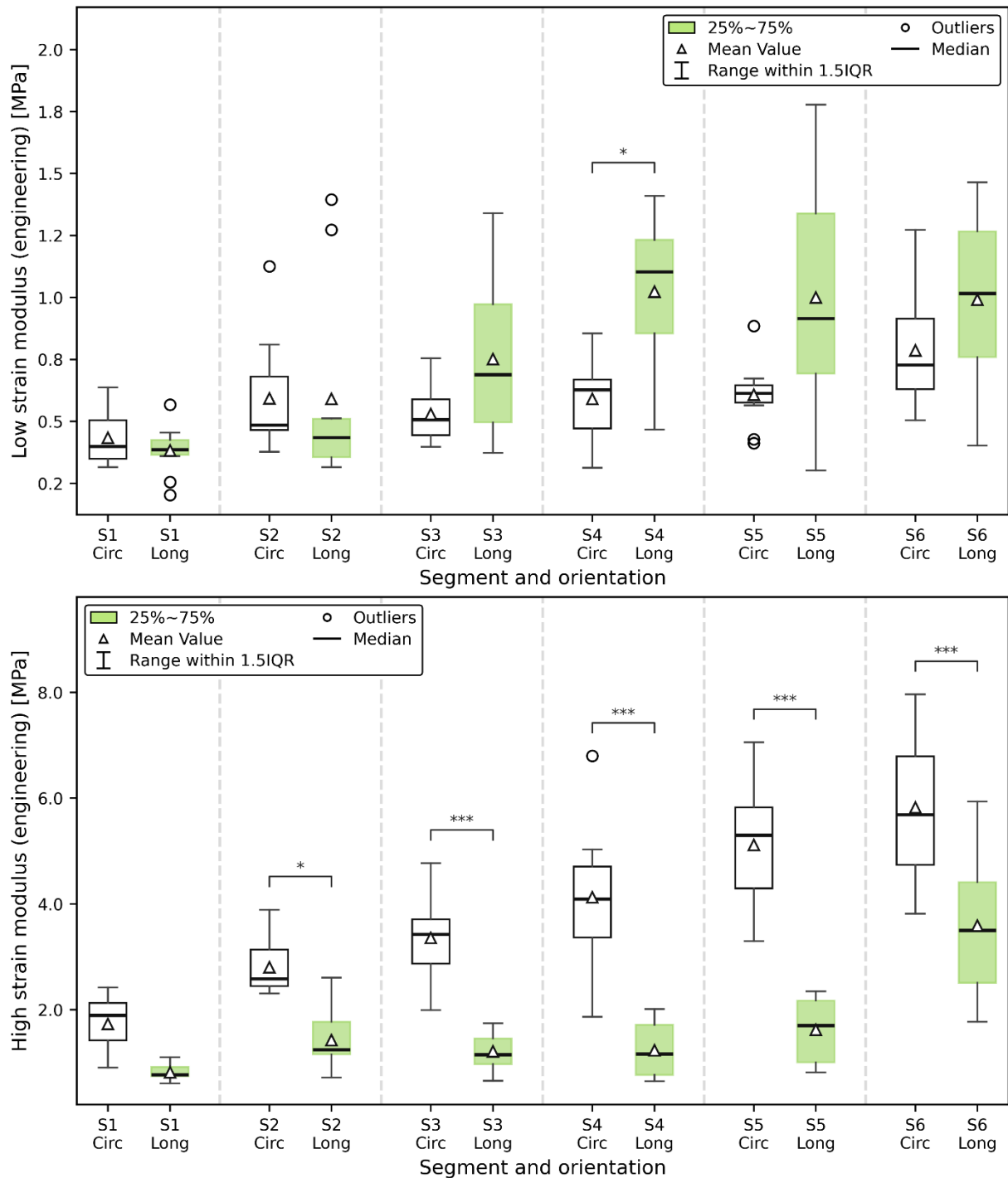


Fig. 4: Low and high-strain modulus (engineering) across aortic segments and specimen orientations. Boxplots show the engineering low-strain modulus of circumferential specimens (Circ; white) and longitudinal specimens (Long; green) for aortic segments S1–S6. Boxes represent the interquartile range from the 25th to the 75th percentile, horizontal lines indicate the median, and open triangles indicate the arithmetic mean. Whiskers extend to the most extreme observations within 1.5 times the

interquartile range, while observations beyond this range are shown as open circles. Bracket indicates significant differences between specimen orientations within individual segments (ANOVA with Tukey's post hoc test *: $p < 0.05$, **: $p < 0.01$, ***: $p < 0.001$). Exact values can be found in the supporting information.

Comparing circumferential and longitudinal specimens within each segment, the low-strain modulus differed significantly only in S4 (circumferential: 0.59 ± 0.16 MPa; longitudinal: 1.02 ± 0.34 MPa; $p = 0.030$), while no significant orientation-dependent difference was found in S1 (0.43 ± 0.11 vs. 0.38 ± 0.10 MPa), S2 (0.59 ± 0.23 vs. 0.59 ± 0.40 MPa), S3 (0.53 ± 0.12 vs. 0.75 ± 0.32 MPa), S5 (0.61 ± 0.13 vs. 1.00 ± 0.49 MPa) or S6 (0.79 ± 0.23 vs. 0.99 ± 0.35 MPa; all $p > 0.05$).

In contrast, the high-strain modulus showed a pronounced and increasingly pronounced circumferential–longitudinal difference along the aorta. No significant difference was observed in S1 (1.73 ± 0.53 vs. 0.82 ± 0.15 MPa; $p = 0.781$), but circumferential specimens were significantly stiffer than longitudinal specimens from S2 onward: S2 (2.80 ± 0.53 vs. 1.42 ± 0.58 MPa; $p = 0.038$), S3 (3.36 ± 0.79 vs. 1.21 ± 0.35 MPa; $p = 0.003$), S4 (4.13 ± 1.33 vs. 1.23 ± 0.53 MPa; $p < 0.001$), S5 (5.12 ± 1.19 vs. 1.62 ± 0.62 MPa; $p < 0.001$) and S6 (5.82 ± 1.50 vs. 3.59 ± 1.39 MPa; $p < 0.001$).

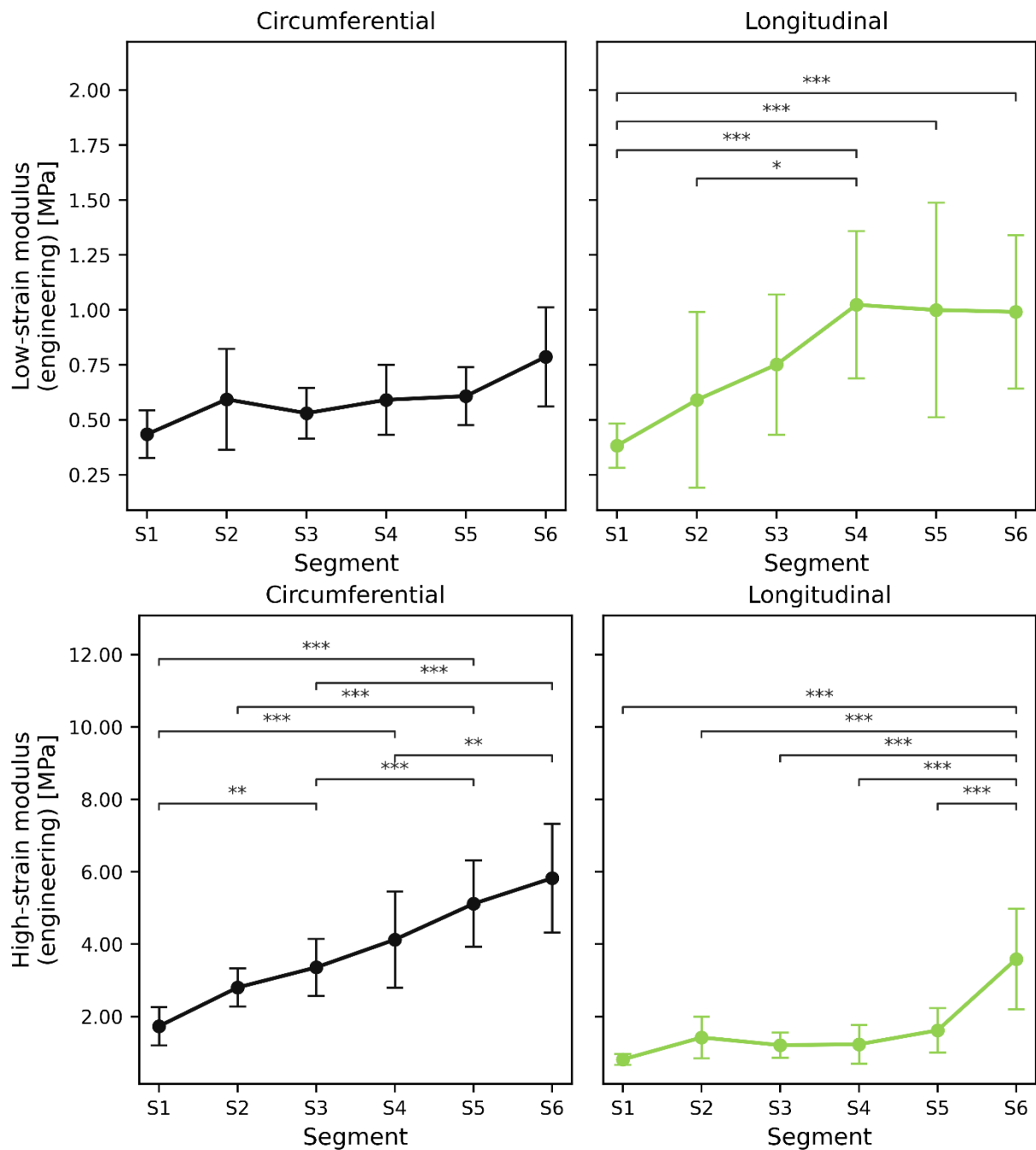


Fig. 5: Regional variation in the low- and high-strain modulus (engineering) along the porcine aorta.

Mean low-strain (top row) and high-strain moduli (bottom row) are shown for circumferential (left) and longitudinal (right) specimens across aortic segments S1–S6. Symbols represent arithmetic means and error bars indicate standard deviations. Brackets indicate significant pairwise differences between aortic segments within the respective specimen orientation (*: $p < 0.05$, **: $p < 0.01$, ***: $p < 0.001$; ANOVA followed by Tukey's post hoc test). Exact p -values are provided in the supporting information.

Within the longitudinal direction, the low-strain modulus increased from proximal to distal segments, with S1 (0.38 ± 0.10 MPa) significantly lower than S4 (1.02 ± 0.34 MPa; $p < 0.001$), S5 (1.00 ± 0.49 MPa; $p < 0.001$) and S6 (0.99 ± 0.35 MPa; $p < 0.001$), and S2 (0.59 ± 0.40 MPa) significantly lower than S4 ($p = 0.030$). No further pairwise segment comparisons reached significance for the longitudinal low-strain modulus, and no pairwise segment comparison reached significance for the circumferential low-strain modulus, indicating a comparatively stable modulus along the vessel length in this loading direction.

The high-strain modulus showed a distinct, direction-specific pattern of regional variation. In circumferential specimens, the modulus increased progressively from proximal to distal: S1 (1.73 ± 0.53 MPa) was significantly lower than S3 (3.36 ± 0.79 MPa; $p = 0.003$), S4 (4.13 ± 1.33 MPa; $p < 0.001$), S5 (5.12 ± 1.19 MPa; $p < 0.001$) and S6 (5.82 ± 1.50 MPa; $p < 0.001$); S2 (2.80 ± 0.53 MPa) was significantly lower than S5 and S6 (both $p < 0.001$); S3 was significantly lower than S5 and S6 ($p < 0.001$ and $p < 0.001$); and S4 was significantly lower than S6 ($p = 0.002$). No significant differences were found between directly neighboring segments, consistent with a gradual, cumulative stiffening along the vessel. In longitudinal specimens, by contrast, the high-strain modulus remained largely stable across S1–S5 (range: 0.82–1.62 MPa), with a marked increase confined to the most distal segment: S6 (3.59 ± 1.39 MPa) was significantly higher than all of S1–S5 (all $p < 0.001$), while no other pairwise segment comparison reached significance.

3.4 Tissue toughness

To assess the energy-absorbing capacity of the aortic wall until rupture, tissue toughness was compared between loading directions and across the six aortic segments (Fig. 6, Fig. 7). One-way ANOVA on the aorta-aggregated mean values ($n = 10$) revealed no significant main effect of segment ($F(5,54) = 0.68$, $p = 0.638$, $\eta^2 = 0.060$) and no significant effect of position ($F(3,36) = 0.57$, $p = 0.636$, $\eta^2 = 0.046$). In contrast, loading direction had a strong and significant effect on toughness, with circumferential specimens absorbing more energy until rupture than longitudinal specimens on average ($F(1,18) = 25.07$, $p < 0.001$, $\eta^2 = 0.582$).

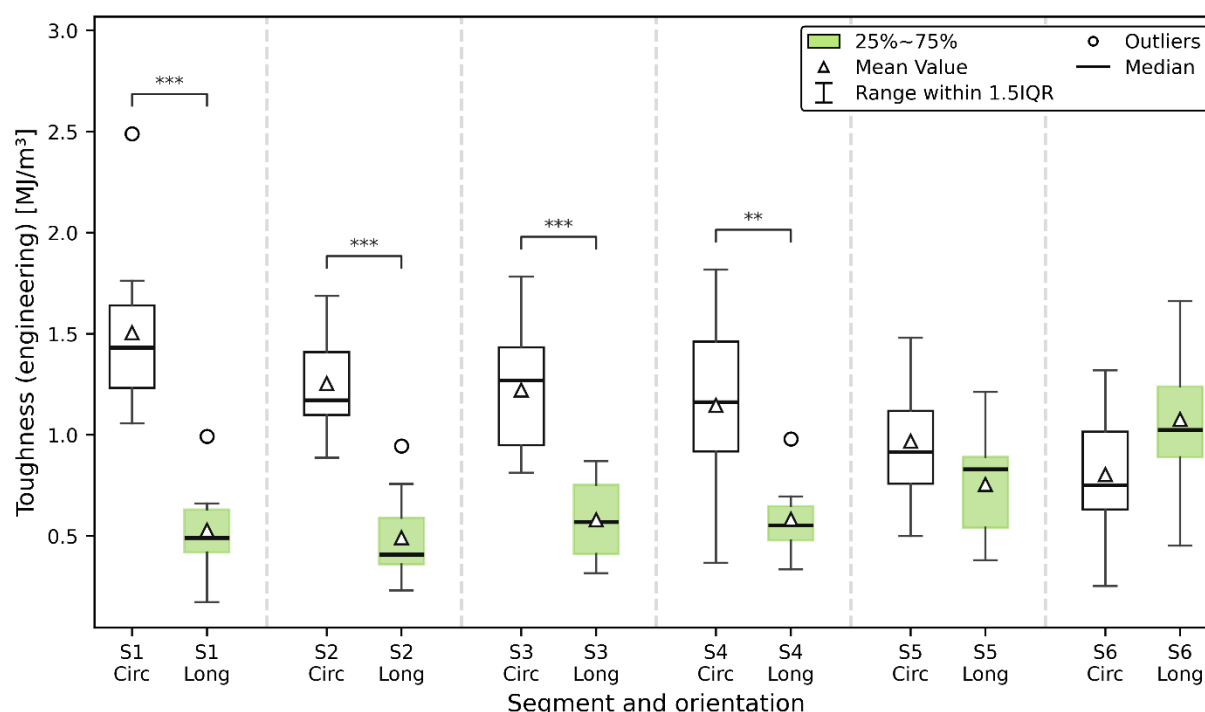


Fig. 6: Tissue toughness (engineering) across aortic segments and specimen orientations. Boxplots show the tissue toughness of circumferential (Circ) and longitudinal (Long) specimens across aortic segments S1–S6. Boxplot elements and statistical annotations are defined as in Fig. 4. Exact p -values are provided in the supporting information.

Comparing circumferential and longitudinal specimens within each segment, toughness was significantly higher in circumferential specimens in S1 (1.50 ± 0.41 vs. 0.53 ± 0.22 MJ/m³; $p < 0.001$), S2 (1.25 ± 0.26 vs. 0.49 ± 0.23 MJ/m³; $p < 0.001$), S3 (1.22 ± 0.32 vs. 0.58 ± 0.19 MJ/m³; $p < 0.001$) and S4 (1.15 ± 0.44 vs. 0.58 ± 0.18 MJ/m³; $p = 0.002$). This directional difference was no longer significant in the two most distal segments, S5 (0.97 ± 0.31 vs. 0.75 ± 0.26 MJ/m³; $p > 0.05$) and S6 (0.80 ± 0.31 vs. 1.08 ± 0.36 MJ/m³; $p > 0.05$), where the circumferential and longitudinal values converged and, on average, numerically reversed in S6.

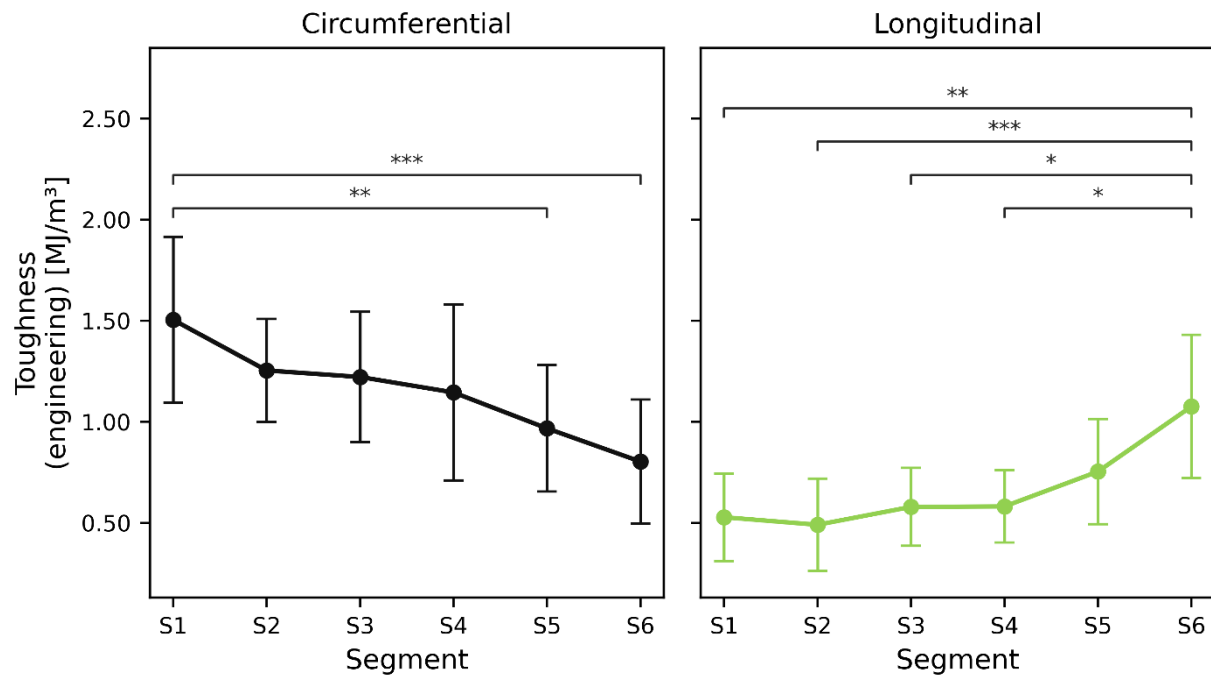


Fig. 7: Regional variation of tissue toughness (engineering) along the porcine aorta. Mean engineering toughness are shown for circumferential (left) and longitudinal (right) specimens across aortic segments S1–S6. Symbols represent arithmetic means and error bars indicate standard deviations. Brackets indicate significant pairwise differences between aortic segments within the respective specimen orientation (*: $p < 0.05$, **: $p < 0.01$, ***: $p < 0.001$; ANOVA followed by Tukey's post hoc test). Exact p -values are provided in the supporting information.

Despite the absence of a significant overall segment effect, direction- specific pairwise comparisons revealed opposing regional trends. In circumferential specimens, toughness was significantly higher in S1 (1.50 ± 0.41 MJ/m³) than in S5 (0.97 ± 0.31 MJ/m³; $p = 0.004$) and S6 (0.80 ± 0.31 MJ/m³; $p < 0.001$), consistent with a gradual decline in energy-absorbing capacity from proximal to distal; no other pairwise segment comparison reached significance in this direction. In longitudinal specimens, the pattern was reversed: toughness in S6 (1.08 ± 0.36 MJ/m³) was significantly higher than in S1 (0.53 ± 0.22 MJ/m³; $p = 0.003$), S2 (0.49 ± 0.23 MJ/m³; $p < 0.001$), S3 (0.58 ± 0.19 MJ/m³; $p = 0.014$) and S4 (0.58 ± 0.18 MJ/m³; $p = 0.015$), but not S5 (0.75 ± 0.26 MJ/m³; $p > 0.05$). Taken together, the circumferential and longitudinal toughness profiles run in opposite directions along the vessel, converging in the distal segments (S5, S6).

3.5 Rupture properties

Stress and strain at failure were determined for each specimen and compared across loading directions and aortic segments (Fig. 8, Fig. 9). One-way ANOVA on the aorta-aggregated mean values ($n = 10$) showed significant main effects of segment for both stress at failure ($F(5,54) = 10.38$, $p < 0.001$, $\eta^2 = 0.490$) and strain at failure ($F(5,54) = 29.35$, $p < 0.001$, $\eta^2 = 0.731$), while position had no significant effect on either parameter (stress: $p = 0.377$; strain: $p = 0.645$). Loading direction significantly affected both parameters: circumferential specimens failed at higher stress ($F(1,18) = 43.87$, $p < 0.001$, $\eta^2 = 0.709$) but lower strain ($F(1,18) = 6.30$, $p = 0.022$, $\eta^2 = 0.259$) than longitudinal specimens on average.

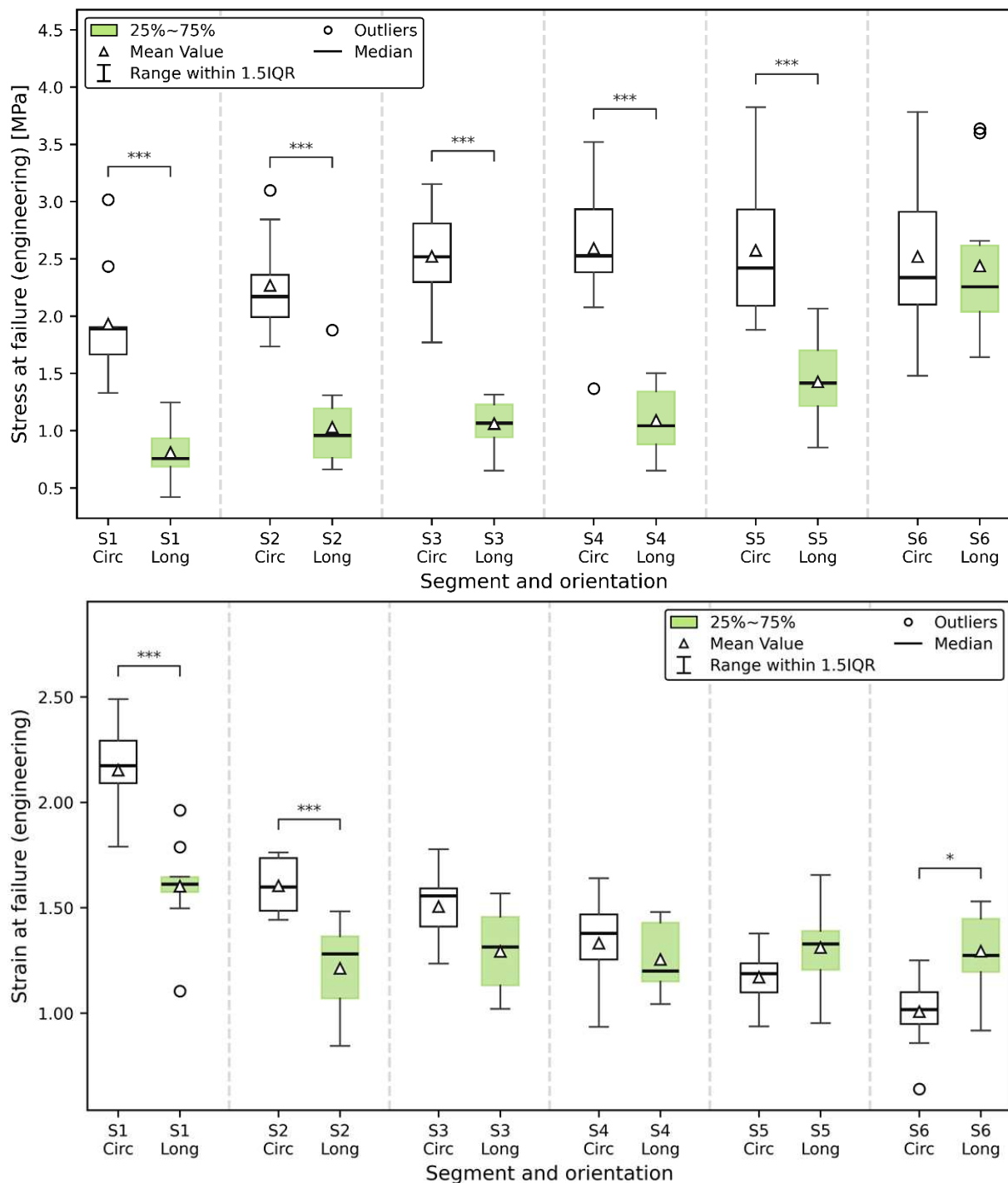


Fig. 8: Stress and strain at failure (engineering) across aortic segments and specimen orientations. Boxplots show the stress (top) and strain (bottom) at failure for circumferential (Circ) and longitudinal (Long) specimens across aortic segments S1–S6. Boxplot elements and statistical annotations are defined as in Fig. 4. Exact *p*-values are provided in the supporting information.

Comparing circumferential and longitudinal specimens within each segment, stress at failure was significantly higher in circumferential specimens in S1 (1.93 ± 0.48 vs. 0.81 ± 0.24 MPa; $p < 0.001$), S2 (2.27 ± 0.42 vs. 1.03 ± 0.37 MPa; $p < 0.001$), S3 (2.52 ± 0.43 vs. 1.06 ± 0.21 MPa; $p < 0.001$), S4 (2.59 ± 0.63 vs. 1.09 ± 0.31 MPa; $p < 0.001$) and S5 (2.58 ± 0.61 vs. 1.43 ± 0.39 MPa; $p < 0.001$). This directional difference disappeared in S6, where circumferential and longitudinal specimens failed at comparable stress levels (2.52 ± 0.71 vs. 2.44 ± 0.70 MPa; $p > 0.05$).

Strain at failure showed a more heterogeneous directional pattern: circumferential specimens failed at significantly higher strain than longitudinal specimens in S1 (2.15 ± 0.21 vs. 1.60 ± 0.22 ; $p < 0.001$) and S2 (1.60 ± 0.13 vs. 1.21 ± 0.22 ; $p < 0.001$), with no significant directional difference in S3, S4 or S5 (all $p > 0.05$). In S6, the direction effect reversed, with longitudinal specimens failing at significantly higher strain than circumferential specimens (1.29 ± 0.19 vs. 1.01 ± 0.18 ; $p = 0.039$).

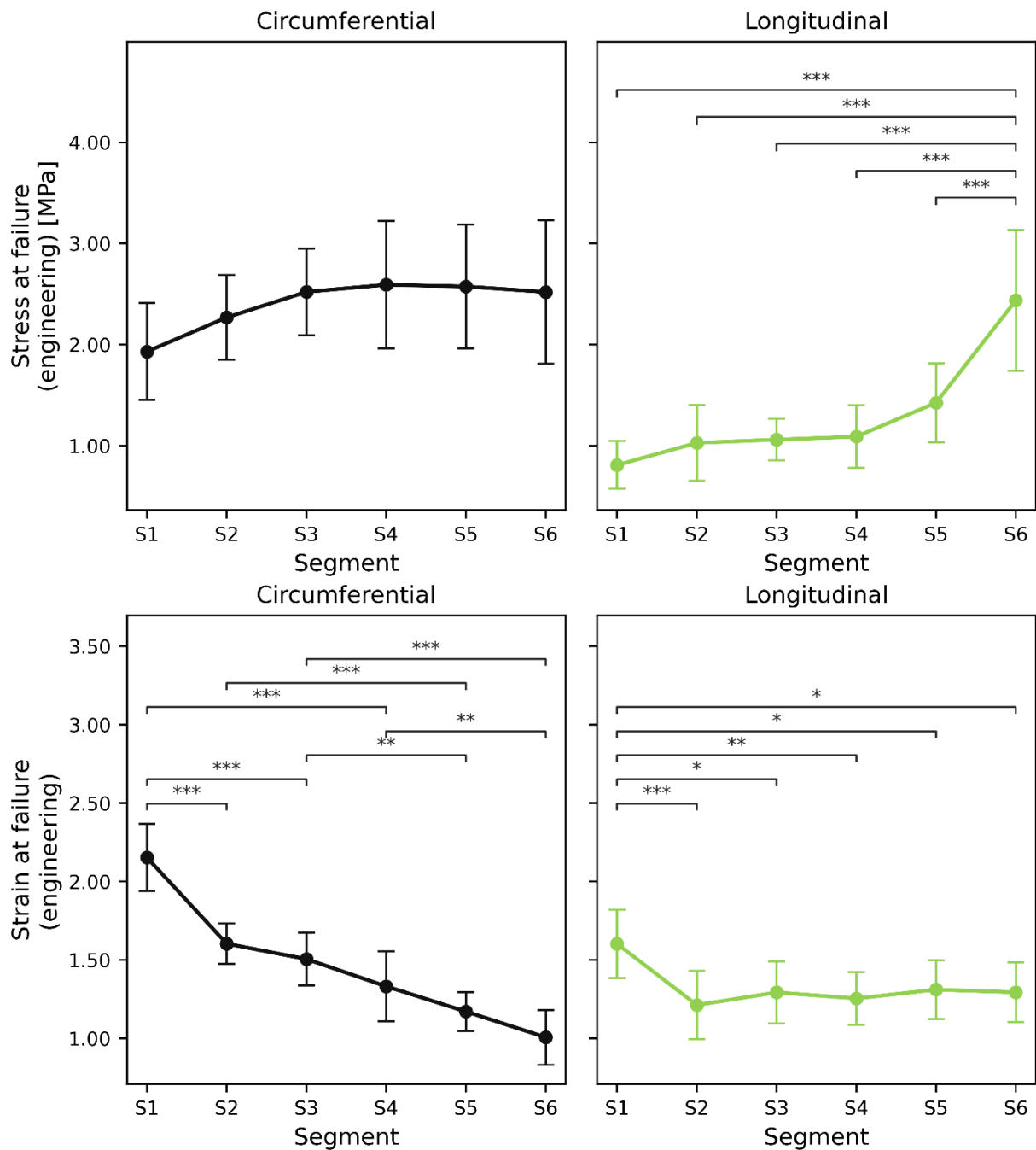


Fig. 9: Regional variation in stress and strain at failure (engineering) along the porcine aorta. Mean failure stress (top row) and failure strain (bottom row) are shown for circumferential (left) and longitudinal (right) specimens across aortic segments S1–S6. Symbols represent arithmetic means and error bars indicate standard deviations. Brackets indicate significant pairwise differences between aortic segments within the respective specimen orientation (*: $p < 0.05$, **: $p < 0.01$, ***: $p < 0.001$; ANOVA followed by Tukey's post hoc test). Exact p -values are provided in the supporting information.

For stress at failure, circumferential specimens showed no significant pairwise segment differences (all $p > 0.05$), indicating a comparatively stable failure stress along the vessel in this direction. In longitudinal specimens, by contrast, stress at failure increased sharply toward the distal aorta: S6 (2.44 ± 0.70 MPa) was significantly higher than S1 (0.81 ± 0.24 MPa; $p < 0.001$), S2 (1.03 ± 0.37 MPa; $p <$

0.001), S3 (1.06 ± 0.21 MPa; $p < 0.001$), S4 (1.09 ± 0.31 MPa; $p < 0.001$) and S5 (1.43 ± 0.39 MPa; $p < 0.001$), while no other pairwise segment comparison reached significance.

For strain at failure, circumferential specimens showed a pronounced, monotonic decline from proximal to distal: S1 (2.15 ± 0.21) was significantly higher than all other segments (S2, $p < 0.001$; S3, S4, S5, S6, all $p < 0.001$); S2 (1.60 ± 0.13) was significantly higher than S5 ($p < 0.001$) and S6 ($p < 0.001$); S3 (1.51 ± 0.17) was significantly higher than S6 ($p < 0.001$); and S4 (1.33 ± 0.22) was significantly higher than S6 ($p = 0.007$). In longitudinal specimens, the regional decline was markedly weaker: S1 (1.60 ± 0.22) was significantly higher only than S5 (1.31 ± 0.19 ; $p < 0.001$) and S6 (1.29 ± 0.19 ; $p < 0.001$), with no further pairwise segment comparisons reaching significance.

3.6 Overview of Results

Table 1 provides a summary of our findings in terms of low- and high-strain moduli, toughness and rupture properties of the thoracic porcine aorta as mean $\pm$ SD values (engineering).

Tab. 1: Overview of mean $\pm$ SD values (engineering) for all measured biomechanical parameters across aortic segments S1–S6, separated by loading direction (circumferential / longitudinal).

Parameter	Segm. 1	Segm. 2	Segm. 3	Segm. 4	Segm. 5	Segm. 6
Young's Moduli [MPa]						
Low strain (circ.)	0.43 ± 0.11	0.59 ± 0.23	0.53 ± 0.12	0.59 ± 0.16	0.61 ± 0.13	0.79 ± 0.23
Low strain (long.)	0.38 ± 0.10	0.59 ± 0.40	0.75 ± 0.32	1.02 ± 0.34	1.00 ± 0.49	0.99 ± 0.35
High strain (circ.)	1.73 ± 0.53	2.80 ± 0.53	3.36 ± 0.79	4.13 ± 1.33	5.12 ± 1.19	5.82 ± 1.50
High strain (long.)	0.82 ± 0.15	1.42 ± 0.58	1.21 ± 0.35	1.23 ± 0.53	1.62 ± 0.62	3.59 ± 1.39
Toughness [MJ/m³]						
Circumferential	1.50 ± 0.41	1.25 ± 0.26	1.22 ± 0.32	1.15 ± 0.44	0.97 ± 0.31	0.80 ± 0.31
Longitudinal	0.53 ± 0.22	0.49 ± 0.23	0.58 ± 0.19	0.58 ± 0.18	0.75 ± 0.26	1.08 ± 0.36
Rupture properties						
Stress (circ.) [MPa]	1.93 ± 0.48	2.27 ± 0.42	2.52 ± 0.43	2.59 ± 0.63	2.58 ± 0.61	2.52 ± 0.71
Stress (long.) [MPa]	0.81 ± 0.24	1.03 ± 0.37	1.06 ± 0.21	1.09 ± 0.31	1.43 ± 0.39	2.44 ± 0.70
Strain (circ.) [-]	2.15 ± 0.21	1.60 ± 0.13	1.51 ± 0.17	1.33 ± 0.22	1.17 ± 0.12	1.01 ± 0.18
Strain (long.) [-]	1.60 ± 0.22	1.21 ± 0.22	1.29 ± 0.20	1.25 ± 0.17	1.31 ± 0.19	1.29 ± 0.19

4 Discussion

The aim of this study is to provide new insights into the biomechanical heterogeneity of the porcine thoracic aorta with respect to the anatomical location and the direction of the mechanical loading. Our characterization covers Young's Modulus, strength and toughness of the aortic wall.

Interestingly, no significant differences were found for the positions (Posterior, Anterior, Left and Right) within the same segment. Regardless of intercostal arteries in the posterior position and their decrease in diameter and tissue thickness from cranial to caudal, our findings suggest that, at least in porcine tissue or more precisely, in the thoracic porcine aorta, the circumferential plane of the thoracic aorta is relatively uniform in its ability to absorb the internal forces applied to the vessel wall. Such circumferential heterogeneity likely reflects the fact that the vessel develops under a symmetric pressure field, where hemodynamic forces which are caused by the pressure are distributed evenly around the circumference, resulting in a uniform build-up of the wall. There are publications covering this topic from a theoretical (Secomb, 2016) or from an experimental (Weizsacker and Pinto, 1988) point of view, which underline our argumentation. The lack of significant circumferential differences in biomechanical properties simplifies the biomechanical characterization of porcine aortic tissue and

supports its use as a reproducible experimental model. Notably, this finding contrasts with an inflation-based study reporting significant circumferential position-dependent differences in the mechanical stiffness of the porcine thoracic aorta, with the posterior region being significantly stiffer than the anterior region (Kim and Baek, 2011). This discrepancy may reflect methodological differences between biaxial inflation testing, which probes the intact, pressurized vessel wall in its native residual-stress state, and the uniaxial strip testing used in the present study, which necessarily removes each specimen from its physiological, curved and pre-stretched configuration.

Regional heterogeneity of the aortic wall was observed along the cranial–caudal axis. Several biomechanical parameters showed systematic changes with increasing distance from the ascending aorta, generally indicating increased stiffness and strength in more distal segments. However, the magnitude and direction of these regional differences varied among the individual biomechanical parameters and between circumferential and longitudinal loading. This increase in stiffness and strength arises from a combination of geometric, compositional, and hemodynamic factors. Distally, the aorta narrows down, with a reduction of wall thickness and lumen diameter. According to Laplace's law, a smaller diameter lowers the circumferential wall tension for a given pressure. Compositional changes likely play a central role as well: proximal regions of the aorta are elastin-rich, enabling the flexibility needed for the Windkessel effect, while distal regions contain a higher collagen content (Davidson et al., 1986, 1985). This shift in the elastin-to-collagen ratio is a plausible driver of the increasingly collagen-dominated, stiffer mechanical behavior observed towards the distal aorta.

The directional dependence of the mechanical properties provides further insight, though this dependence itself changes with the strain regime and with location along the vessel. At high strains, circumferential specimens were markedly stiffer, stronger and more resistant to failure than longitudinal specimens, with the magnitude of this difference increasing towards the distal aorta before converging again in the most distal segment (S6). This pattern is consistent with the notion that the aortic wall has adapted its collagen fiber architecture to the roughly two-fold higher circumferential wall stress predicted for a pressurized thin-walled cylinder by Barlow's formula (Budynas et al., 2015; Ibrahim et al., 2015), with collagen recruitment — and with it this anisotropy — becoming dominant only once the tissue is loaded beyond the low-strain, elastin-dominated regime.

At low strains, however, this relationship reversed: longitudinal specimens were on average stiffer than circumferential specimens, although this difference reached significance only in segment 4. Because low-strain stiffness is governed primarily by the elastin network rather than by collagen, this suggests that the elastin architecture of the porcine thoracic aorta does not simply mirror the organization of the collagen network, and may be oriented preferentially along rather than around the vessel axis - a hypothesis that would benefit from direct histological or microstructural verification.

Toughness followed yet another pattern: rather than a uniform circumferential dominance, circumferential toughness declined while longitudinal toughness increased with increasing distance from the ascending aorta, such that the pronounced circumferential–longitudinal difference observed proximally (segments 1–4) had disappeared by the two most distal segments (S5, S6). Rupture stress showed a similarly direction-dependent regional trend: circumferential rupture stress remained stable across all six segments, whereas longitudinal rupture stress rose sharply only in the most distal segment. Together, these strain- and direction-dependent reversals indicate that the anisotropy of the porcine thoracic aorta is not a fixed, uniform tissue property, but instead varies systematically with

both loading regime and anatomical location. Our absolute modulus values are in good agreement with previously reported data for the porcine thoracic aorta. Synthesizing biaxial literature data at physiological pressure, a circumferential stiffness of 0.46 ± 0.03 MPa and a longitudinal stiffness of 0.44 ± 0.01 MPa were reported for the thoracic porcine aorta (De Beaufort et al., 2018). This is closely matching our low-strain modulus in the most proximal segment (S1: 0.43 ± 0.11 MPa circumferential, 0.38 ± 0.10 MPa longitudinal), despite the methodological differences between biaxial and uniaxial testing. The distal increase in longitudinal rupture stress observed in the present study is further consistent with Peña et al. (2019), who reported a higher fracture stress in the infrarenal abdominal aorta compared with the thoracic aorta in porcine tissue.

Our failure-stress data are further supported by uniaxial measurements on the porcine thoracic aorta, showing a circumferential failure stress of 2.35 ± 0.40 MPa and a longitudinal failure stress of 0.76 ± 0.12 MPa for specimens taken from the upper thoracic aorta (Myneni et al., 2022). These values closely match our own findings in the corresponding proximal region: our circumferential rupture stress ranged between 1.93 and 2.59 MPa across all six segments, and our longitudinal rupture stress in segment S1 (0.81 ± 0.24 MPa) is nearly identical to their longitudinal value. Anisotropic mechanical behavior has further been observed in the porcine aorta even at small stretches, within the strain range thought to be dominated by elastin (Myneni et al., 2022). This is broadly consistent with the directional difference we observed in the low-strain modulus, although their study did not resolve which direction was stiffer at low strain specifically.

Using inflation-extension testing across the physiological to supra-physiological pressure range, the porcine thoracic aorta was similarly found to be stiffer in the circumferential than in the longitudinal direction, based on the deviation of the circumferential-longitudinal stretch relationship from the isotropic prediction (Rastgar Agah et al., 2017). This biaxial evidence is consistent with the circumferential dominance we observed at high strains. However, because their metric pools the full physiological to supra-physiological pressure range rather than separating an elastin-dominated low-strain regime from a collagen-dominated high-strain regime, it cannot directly confirm or refute the reversal we observed at low strains - a distinction that our low- and high-strain modulus framework is able to resolve, but single-metric biaxial approaches such as theirs are not.

The failure pattern also differs between orientations. The layers (intima, media, adventitia) of the aorta ruptures simultaneously for the circumferential specimens, indicating uniform fiber orientation and density across the layers. In contrast, the longitudinal specimens consistently show a sequential rupture, with a failure of the intima first, followed by a delayed rupture of media and adventitia. This behavior indicates, that the intima is the weakest layer, while the adventitia acts as the barrier upon rupture. From a mechanical point of view, this sequential failure resembles the initiation of an aortic dissection in humans, where a fissure of the intima precedes the spread into deeper layers of the vessel (Gawinecka et al., 2017; Ince and Nienaber, 2007; Nienaber et al., 2016).

The mechanical testing of vascular soft tissue, particularly the aorta, remains essential for understanding disease mechanisms like aortic dissections, aortic dilatations or aneurysms. Furthermore, the data of these experiments are required to improve computational models and guiding clinical interventions. Besides the uniaxial tensile test, which was used in this study, there are also different experimental protocols like biaxial tensile tests, inflation tests or compact tension tests to characterize tissue strength, stiffness, anisotropy and failure properties.

According to the systematic review about experimental protocols to test soft tissues (Valente et al., 2024) human soft tissue remains the gold standard for acquiring clinically relevant data, but several challenges persist when testing human tissue. These include limited specimen availability, ethical constraints, patient heterogeneity (e.g., age, sex, pathology, degeneration) and difficulties in preserving mechanical properties during handling and storage. For these reasons animal models continue to play a crucial role in the completion of human studies. Porcine tissue in particular is frequently used because of its close anatomical and biomechanical similarity. Smaller animal models like rat, mouse or rabbit also contribute by enabling controlled experiments and genetic modifications, though their scaling to human physiology is limited.

This study provides a valuable, high-resolution characterization of the biomechanical properties of the porcine aorta, but it is not without limitations. Two of the most relevant are the choice of the specimen storage protocol and the use of a uniaxial tensile testing unit. As explained in the chapter about materials and methods there are several studies which indicate oppositional views on the impact of freezing on the biomechanical parameters. This is a known source of potential error in vascular biomechanics and the decision to freeze the tissue at -20°C was a necessary compromise to manage the logistics of specimen acquisition and testing. It will be very challenging, but it would be best to always measure a fresh aorta one after another in further experiments.

Our study utilized a uniaxial tensile testing unit, which is a known limitation when it comes to characterizing complex, anisotropic biomaterials like the aorta. In a physiological state, the aortic wall undergoes a biaxial stress from both, the internal pressure and axial stretch. Uniaxial testing cannot fully reproduce these *in vivo* conditions or provide a complete characterization of the material behavior under such complex loadings. Further research should aim to overcome this limitation by implementing biaxial testing for better and more robust results which can be used in FEM simulations.

5 Conclusion

Altogether, these findings underline that the thoracic aorta cannot be treated as a mechanically uniform structure. Instead, its material properties vary significantly both along its length and between loading orientations. Despite a reduction of the anisotropic behavior of AAA tissue compared to a non-aneurysmal abdominal aortic tissue (Vande Geest et al., 2006), a region-specific and still anisotropic material should be used for computational modeling and FEM analysis of aneurysm rupture risk rather than relying on homogenous assumptions. The sequential failure pattern of longitudinal specimens further suggests that models should be layer-specific, particularly when simulating dissection-prone conditions where the rupture of the intima is the initiating event. Clinically, the observed anisotropy and cranio-caudal heterogeneity of the aortic wall may help explain why aneurysms preferentially develop in the abdominal aorta. The distal thoracic aorta, approaching this transition, was already found to be stiffer and less compliant than the proximal aorta in the present data. If this trend continues into the abdominal aorta, abdominal regions may be similarly stiffer and less compliant than the proximal aorta and are less able to buffer pulsatile loads. This increased stiffness would raise local wall stress and may promote progressive dilation, which contributes to a higher prevalence of abdominal aortic aneurysm.

By providing a detailed characterization of regional biomechanical heterogeneity along the porcine aorta, this study contributes to the refinement of biomechanical models and may support the development of more accurate predictive tools for vascular disease progression and rupture risk.

6 Acknowledgments

C.W. would like to acknowledge funding by the DFG (INST 94/135 FUGG 508235635). Moreover, all authors would like to acknowledge funding for the project “Analysis of the elasticity and tensile strength of the aneurysmatic infrarenal aortic wall in tensile testing” by the University of Augsburg. The authors thank Anja Meißner for fruitful discussion.

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