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Mechanical characterisation of porcine rectus sheath under 34 uniaxial and biaxial tension

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ABSTRACT

Incisional hernia development is a significant complication after laparoscopic abdominal surgery. Intra-abdominal pressure (IAP) is known to initiate the extrusion of intestines through the abdominal wall, but there is limited data on the mechanics of IAP generation and the structural properties of rectus sheath. This paper presents an explanation of the mechanics of IAP development, a study of the uniaxial and biaxial tensile properties of porcine rectus sheath, and a simple computational investigation of the tissue. Analysis using Laplace's law showed a circumferential stress in the abdominal wall of approx. 1.1 MPa due to an IAP of 11 kPa, commonly seen during coughing. Uniaxial and biaxial tensile tests were conducted on samples of porcine rectus sheath to characterise the stress–stretch responses of the tissue. Under uniaxial tension, fibre direction samples failed on average at a stress of 4.5 MPa at a stretch of 1.07 while cross-fibre samples failed at a stress of 1.6 MPa under a stretch of 1.29. Under equi-biaxial tension, failure occurred at 1.6 MPa with the fibre direction stretching to only 1.02 while the cross-fibre direction stretched to 1.13. Uniaxial and biaxial stress–stretch plots are presented allowing detailed modelling of the tissue either in silico or in a surrogate material. An FeBio computational model of the tissue is presented using a combination of an Ogden and an exponential power law model to represent the matrix and fibres respectively. The structural properties of porcine rectus sheath have been characterised and add to the small set of human data in the literature with which it may be possible to develop methods to reduce the incidence of incisional hernia development.

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1. Introduction

Hernia development at wound sites after laparoscopic surgery has a prevalence rate of 1–3% (Boldo et al., 2006; Bowrey et al., 2001; Comajuncos et al., 2011; Kadar et al., 1993; Tonouchi et al., 2004). The biomechanical driver behind hernia formation is evidently elevated intra-abdominal pressure (IAP), though this has not been proven. The rectus sheath, a fibrous layer encompassing the aponeurosis of the lateral abdominal muscles and enveloping the rectus abdominis muscle, is often implicated in hernia formation. However, there is surprisingly no literature on the mechanical environment that generates IAP and thus no fundamental understanding of the stress states during hernia formation in the rectus sheath. There is also limited data on the structural properties of rectus sheath, with varying protocols and

conflicting results (Ben Abdelounis et al., 2013; Martins et al., 2012; Rath et al., 1997). Rath et al. studied human rectus sheath in uniaxial tension but only reported failure stress and elongation. Martins et al. measured uniaxial stress–strain behaviour in human tissue in both fibre and cross-fibre directions, but reported large scatter with stresses between 2.5 and 20 MPa and failure stretches up to 2.6 which seem doubtful. Statistical tests compared fibre and cross-fibre orientations, effects of BMI, and gender, but the sample sizes were small. Ben Abdelounis et al. recently examined the effect of loading rate on the fibre direction response of human tissue in uniaxial tension. These authors also used an image based strain measure to address slippage at the tissue grip interface which may account for some of the findings of Martins et al. However, their sample size was also small. The aims of this paper are therefore to analyse the loading environment in the abdominal wall, to characterise the tensile structural properties of the porcine rectus sheath using appropriate mechanical tests and to evaluate whether a fibre reinforced computational material model can adequately replicate the observed experimental behaviour.

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2. Methods

2.1. The loading environment in the abdominal wall under IAP

IAP is generated through the action of the diaphragm and the abdominal muscles. During quiet breathing, the abdominal muscles are inactive (Campbell and Green, 1953) and contraction of the diaphragm reduces the craniocaudal diameter of the abdominal cavity and the IAP causes the dorsoventral diameter to increase passively (Fig. 1(a)). This results in a minimal variation of IAP during normal breathing of ca. 1 kPa (Malbrain et al., 2004; Sanchez et al., 2001; Sugerman et al., 1997). From Eq. (1), modelling the abdomen as a cylinder and assuming a radius of 200 mm and an abdominal wall thickness of 2 cm (Sandler et al., 2010) (Fig. 1(a)) the circumferential stress in the abdominal wall during breathing would be around 10 kPa.

$$\sigma = \frac{Pr}{t} \quad (1)$$

In contrast, in an abdominal straining manoeuvre the muscles of the abdominal wall contract and the diaphragm is then lowered, significantly reducing the volume of the abdominal cavity. Approximating the cavity as an elliptical hemi-cylinder, the volume is

$$\frac{1}{2} L \pi r_1 r_2 \quad (2)$$

Using CT scanning, Duez et al. (2009) found the dimensions of the abdomen as length: 341 mm, R1: 282 mm and R2: 104 mm which equates to a volume of 15.7 L using Eq. (2). Talasz et al. (2011) used MRI to evaluate the change in these dimensions during coughing and found the length decreased to 304 mm, R1 remained the same and R2 reduced to 94.5 mm giving a new, reduced volume of 12.7 L.

Applying Boyle's law ($PV=k$), assuming the abdomen was sealed and gas filled, a reduction in volume of 3 L (19%) would create a 19% increase in pressure. Given a resting IAP of 1 kPa this would generate an IAP of only 1.19 kPa during coughing,

much less than the average of 11 kPa reported (Cobb et al., 2005). However, with only 115 ml of gas in the abdomen at rest (Bedell et al., 1956), blood must be expelled from the abdomen via the venous and arterial networks to permit this volume reduction. Central venous pressure is approximately 2 kPa (Egan et al., 2009) but blood can only flow proximally through the venous network due to valves. Diastolic blood pressure is approximately 10.5 kPa (Kshirsagar et al., 2006), and thus to expel blood via the arterial network, the IAP must exceed this.

Since contracting the abdominal muscles reduces the abdominal circumference, the transversalis fascia and peritoneum should not be under tension as they do not act in series with the muscles (Fig. 1(b)). However, contraction of the internal and external oblique muscles and the transverse abdominis would result load the anterior and posterior rectus sheath and the linea alba (Fig. 1(b)). Application of Eq. (1) shows that an IAP of 11 kPa in an abdominal cavity of radius 200 mm and thickness 2 mm (Martins et al., 2012; Song et al., 2006) would yield an average circumferential stress in the abdominal wall of 1.1 MPa. Defects, caused by abdominal surgery, would be particularly stressed in this configuration.

This preliminary analysis has shown that the biomechanics of abdominal wall loading during high IAP is complex. For future physical and computational modelling of the abdominal wall for wound closure analysis, it is of interest to establish the stress-strain relationship of the rectus sheath for stresses of the order of megapascals. Furthermore, during periods of increased IAP the stress in the cross-fibre direction is likely to be non-negligible, and the biaxial tensile behaviour of the rectus sheath is therefore also of interest. Accordingly, mechanical testing in these modes was performed and is reported next.

2.2. Physical tests

Twenty porcine abdominal walls were sourced from a local abattoir. All pigs were aged 26–28 weeks and all females were nulligravida. Animals were slaughtered and dissected in the abattoir as per their procedures where the abdominal walls were harvested. They were subsequently kept frozen at -20°C . Prior to testing, abdominal walls were defrosted at $5 \pm 1^\circ\text{C}$ for 40 h and dissected to isolate

Fig. 1. Dimensions of the abdominal cavity, from Duez et al. (2009), (b) layers of the abdominal wall from Netter (2006) and (c) posterior view of a porcine belly showing fibre and cross-fibre directions.

Fig. 2. (a) Uniaxial gripping jig and (b) uniaxial gripped sample in Zwick Machine. Matlab dot tracking software identifies regions A, B, C and D in the direction of the applied stretch and regions a, b and c in the orthogonal direction. Stretch was calculated in each of these regions and averaged to give global vertical (applied) and horizontal (response) stretch values.

the anterior rectus sheath. In all tests, a high definition Samsung HMX-QF20 camera (Samsung Electronics Co., Gyeonggi-Do, Korea) recorded the sample deformation at 2 fps.

2.2.1. Uniaxial

Tissue was dissected into fibre direction and cross-fibre direction specimens (Fig. 1(c)). Rectangular specimens 12–20 mm width and 16–66 mm in length were cut. Samples were labelled and stored in phosphate buffered saline solution (PBS) until immediately prior to tensile testing. Sixteen fibre direction and 16 cross-fibre direction samples were tested.

A jig was developed to improve alignment of specimens in the grips (Fig. 2). Grips were covered in grade P60 sandpaper. To prevent tissue damage during gripping, grip bolts were tightened using a torque wrench to 0.2 N m, as this was found empirically to minimise both slippage (grips too loose) and tissue damage (grips too tight).

The width, thickness and mass of each sample were recorded prior to gripping. After gripping, the grip to grip distance (characteristic length) of the tissue was measured and recorded. A grid of 6 dot markers was applied to the centre of the sample for strain analysis (Fig. 2(b)). Samples were tested under displacement controlled uniaxial tensile loading in a Zwick Roell Z005 machine (Zwick/Roell GmbH Ulm, Germany). A preload equal to the weight of the sample was applied and samples were stretched until failure (> 20% drop in applied force) at 10% of the characteristic length per minute to minimise viscoelastic effects.

Custom Matlab code (Takaza et al., 2013a) was used to calculate strain from the marker motion. The engineering stress was calculated from the applied force and the original CSA. Average stress/stretch curves preserving the overall shape of individual tests for fibre and cross-fibre samples were generated.

- (1) Each curve was divided into the same predetermined number of discrete points.
- (2) Points were numbered from 1 to n , with shorter curves having a higher density of points.
- (3) The stress and stretch at each point was averaged over all of the curves.

An additional 0.1 MPa preload to reduce the scatter in the results was applied to overcome challenges in identifying the unstrained sample length. This preload represents only 2–5% of the maximum stress applied to the samples.

2.2.2. Biaxial

A custom bladed punch was used to cut square samples 39.5 mm × 39.5 mm, and a custom gripping jig was used to ensure equal placement of four fish hooks (at 11 mm spacing) to grip along each edge of the sample (Fig. 3(a) and (b)).

A custom designed equi-force biaxial testing rig (Prendergast et al., 2003) (Fig. 3(c)) was used to apply equal tensile force to each of the four sides of the

square specimen using a simple wire and pulley system connected to a Zwick Roell Z005 (Zwick/Roell GmbH Ulm, Germany) testing machine. A pre-load was applied to straighten the sample (approx. 3 N). Crosshead movement was displacement controlled, at a rate of 5 mm/min, and the custom rig moved to maintain equal force on each side of the specimen despite different strains. A speckle pattern of drawing ink was applied to the tissue for subsequent strain analysis using digital image correlation (DIC) (Fig. 3(c)).

DIC was conducted using freely available MATLAB code (Eberl, 2010). This method of strain analysis was used as the fish hooks caused local deformation of the tissue and also left corner regions unstressed. Force measurements were recorded by the Zwick machine. Average stress/stretch plots were created for the fibre and cross-fibre direction in the same way as for uniaxial samples.

2.3. Statistical analysis

Time history control charts with 3σ limits were used to prevent time bias affecting the results (operator experience etc.). Normal probability plots were generated for the average tissue stretch and stress at failure to ensure the assumption of normality applied. ANOVAs were generated to ensure that the variation between pig bellies was not significantly different from samples within each belly. A Student's T test was conducted to compare the stress/stretch response of fibre and cross-fibre specimens.

2.4. Computational modelling

Computational modelling of the uniaxial and biaxial tests was performed using FeBio (Maas et al., 2012) to assess the ability of a fibre reinforced hyper-elastic model to reproduce the observed experimental results. A first order Ogden model represented the tissue matrix and an exponential power law represented the fibres (Maas and Weiss, 2013), with the following strain energies (Ψ).

Ogden

$$\Psi = \sum_{i=1}^N \frac{C_i}{m_i} (\lambda_1^{m_i} + \lambda_2^{m_i} + \lambda_3^{m_i} - 3) + U(J) \quad (3)$$

where λ_i are the deviatoric stretches, c and m are material constants and $U(J)$ represents the volumetric component, where J is the determinant of the deformation gradient tensor. A first order Ogden model was used. The volumetric component is computed based on the bulk modulus.

Exponential – power law

$$\Psi = \frac{\xi}{\alpha\beta} (e^{\alpha(I_n - 1)^\beta} - 1) \quad (4)$$

Fig. 3. Biaxial gripping jig: (a) one edge of the jig with hooks, (b) the complete jig and (c) the gripped sample in the equi-force biaxial testing rig.

where α and β are material constants and ξ is a measure of the fibre modulus and $\bar{l}_n$ is the square of the fibre stretch.

A custom Matlab code (Moerman et al., 2013; Takaza et al., 2013b) was used to find the optimum material parameters for these models in both the fibre and cross-fibre directions for uniaxial testing. The optimisation algorithm used the Matlab `fminsearch` function to minimise the sum of squared difference between the experimental and predicted force displacement data with a tolerance of 0.0001. A nominal bulk modulus of 10 was used in all cases, with a higher value having minimal effect on results but yielding longer solution times.

Two uniaxial simulations were performed for each of the fibre and cross-fibre directions to assess the influence of tissue gripping on the mid-sample stretch-stress relationship. Ideal geometries matching the average geometries of fibre and cross-fibre specimens respectively were created and meshed using 3200 hexahedral elements. In the simple uniaxial case, nodes at one end of the sample were constrained from translating in the z direction (see Fig. 6 for coordinate system) and from rotating. Nodes at the other end of the sample were also constrained from rotating and prescribed a vertical displacement at 10% of specimen length per minute to stretch the sample, similar to the physical tests. The stress-stretch response of a central element was exported. In a second simulation, nodes at both ends of the sample, in the region under the grips, were constrained from translating in the z direction and from rotating. Equal compression of 20% in the y direction was applied to both sides of the tissue in the region of the grips to simulate gripping. This represented the approx. 40% reduction in thickness measured in the region of the grips in the experimental samples. The final configuration after gripping was taken as the initial configuration for subsequent uniaxial tension. Nodes at one end, in the region of the bottom grip were constrained from translating in the z direction and from rotating. Nodes at the other end, in the region of the top grip, were constrained from rotating and a

prescribed displacement of 10% per minute was applied in the z direction. The stress-stretch response of the same central element was exported for comparison purposes.

Subsequently, a biaxial simulation was conducted using the average material parameters from the individual simulations of each of the uniaxial tests and the results were compared to the average experimental biaxial results.

3. Results

3.1. Uniaxial

Fig. 4 shows the fibre and cross-fibre direction stress-stretch curves and the average curve for each direction.

3.2. Biaxial

Fig. 5 shows biaxial stress-stretch plots for all the samples.

3.3. Computational modelling

The average and the ranges of the predicted model parameters from fitting to the uniaxial data are presented in Table 1.

Fig. 4. Stress-stretch plots for all fibre direction ($N=16$) and cross-fibre direction ($N=16$) samples in uniaxial tension.

Fig. 5. Biaxial stress-stretch plots for all samples ($N=14$).

Table 1

FeBio material model parameters and average goodness of fit of computational data to experimental data for all samples.

First order Ogden model			
C		M	
Mean	Range	Mean	Range
0.057	0.016–0.119	8.203	1–14.01
Exponential power law model		Ksi	
Beta		Mean	Range
2	2–2	0.3424	0.162–0.864
Longitudinal samples		Transverse samples	
R^2	SSQD	R^2	SSQD
98.1%	0.391	98.1%	0.391

The goodness of fit of the computational stress-stretch data to the experimental data is shown in Table 1 in terms of R^2 and sum of squared difference. Plots of the fit for a fibre and cross-fibre uniaxial sample and a biaxial sample are shown in Fig. 6 along with images of undeformed and deformed samples.

4. Discussion

4.1. Fundamental pressure analysis

The stress state in the abdominal wall during periods of raised IAP such as coughing is not discussed in the literature. Using

Fig. 6. Typical fit of FeBio stress-stretch results to experimental uniaxial stress-stretch results in the (a) cross-fibre, (b) fibre directions and (c) biaxial tensile simulations. (d) Typical undeformed and deformed sample for an ungripped (left) and gripped (right) case.

simple mechanics, some geometric measurements and preliminary assumptions, the preliminary analysis presented here has been used to determine an estimated stress in the rectus sheath as a result of intra-abdominal pressure. For coughing, the circumferential stress is likely to be of the order of 1 MPa. The cross-fibre stress is more difficult to estimate, but is likely to be non-negligible, and the biaxial behaviour of rectus sheath is therefore of interest. The skin and fat are not considered in this analysis. While this may be appropriate due to the compliant and elastic nature of the skin and the very low toughness and strength of the fat, experimental verification is still required. Furthermore, biological variations could significantly affect the stress state. These variations could be natural, or as a result of surgery or other co-morbidities. These results provide a useful context for hernia modelling, but should be supported by future experimental or finite element modelling.

This biomechanical analysis shows stresses in the order of megapascals can be expected in the rectus sheath during abdominal straining manoeuvres such as coughing, and the biaxial as well as the uniaxial behaviour are of interest.

4.2. Physical tests

Despite the importance of the rectus sheath in IAP loading, there is uncertainty regarding the structural properties of rectus sheath and this was the motivation to characterise the stress-stretch response of the rectus sheath in uniaxial and biaxial tension. Uniaxial tensile tests (16 cross-fibre and 16 along fibre) and biaxial tests (14) were conducted on porcine rectus sheath samples. There was noticeable variation in the stretch in each region of the samples, and the average graphical stretch was found to be less than the machine stress in all cases indicating that, despite best efforts, there was some slippage at the grips. This further highlights the necessity to use graphical strain analysis methods, unlike what was done previously (Martins et al.,

2012; Rath et al., 1997). The stress-stretch response for each sample in both directions is shown in Fig. 4. The variability can mostly be attributed to biological variation between the samples. ANOVA showed that the variation between samples was more significant than the variation between pig bellies used ($P=0.18$). It is clear that there is a large difference between the fibre and cross-fibre direction samples, and this was statistically significant at the 99.9% level ($P<0.001$). Breaking stretch values for the cross-fibre samples were much larger (1.3) than for the fibre direction samples (1.07) and the respective ultimate stress values were much lower (1.6 MPa vs. 4.5 MPa).

Comparisons with the data from Rath et al. and Martins et al. are shown in Fig. 7(a). Rath et al. only reported failure stress and strain, thus straight lines for their data are shown. Results from Martins et al. seem doubtful due to the significant variation, the very high stress and the large elongation reported. Given the complications experienced in the current study with sample gripping, necessitating the use of a torque wrench and the use of image-based strain measurement, it seems likely that slippage occurred in the Martins et al. tests. Both Rath et al. and Martins et al. relied on the grip to grip separation to determine strain, which is known to be problematic (Takaza et al., 2013a). Furthermore, the variation in reported stiffness from the three authors, which is likely to be partially due to variations in age, BMI and parity, makes direct comparison difficult with the presented results from healthy, young, nulligravid pigs.

Nonetheless, Fig. 7(b) compares current results with recent data from Ben Abdelounis et al. evaluating the uniaxial stress-strain response of human rectus sheath in the cross-fibre direction. Ben Abdelounis et al. also used image based strain and, coupled with the similarity in the average responses observed, the comparison increases confidence in the current results. Furthermore, the comparison suggests that the uniaxial behaviour of porcine and human rectus sheath is similar.

Fig. 7. Comparison between the current study and (a) Martins et al. (2012) and Rath et al. (1997) and (b) Ben Abdelounis et al. (2013).

Fig. 8. Predicted stretch for a linear elastic transversely isotropic model in cross-fibre (left group) and fibre (right group) directions for range of Poisson's ratios and Young's moduli under biaxial tensile loading of 1.5 MPa. Changing stiffness values are represented by colours and changing Poisson's ratio values are shown by each point in a colour bar. In an idealised case of large stiffness and Poisson's ratio differences between fibre and cross-fibre directions, there is a large difference in the stretch observed in each direction respectively. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Due to the unusual behaviour of the tissue, it was not possible to quantify the Poisson's ratio in our experiments. During gripping, the tissue in contact with the grips spread in the horizontal direction under the compression of the grips.

Under biaxial loading, the samples stretched uniformly across their width in both directions. Fig. 5 shows the fibre direction was significantly stiffer than the cross-fibre direction. In some cases contraction occurred in the fibre direction while elongating in the cross-fibre direction under biaxial tensile loading (see left-hand side of Fig. 5). It is believed this is due to a large difference between the fibre and cross-fibre direction stiffnesses, and this is corroborated by an analysis based on linear elastic transverse isotropy where the stress-strain relationship is

$$\begin{bmatrix} \varepsilon_L \\ \varepsilon_T \\ \varepsilon_{LT} \\ \varepsilon_{LT} \\ \varepsilon_{LT} \\ \varepsilon_{TT} \end{bmatrix} = \begin{bmatrix} \frac{1}{E_L} & -\frac{\partial_{TL}}{E_L} & -\frac{\partial_{TL}}{E_T} & 0 & 0 & 0 \\ -\frac{\partial_{TL}}{E_L} & \frac{1}{E_T} & -\frac{\partial_{TL}}{E_T} & 0 & 0 & 0 \\ -\frac{\partial_{TL}}{E_L} & -\frac{\partial_{TL}}{E_T} & \frac{1}{E_T} & 0 & 0 & 0 \\ 0 & 0 & 0 & \frac{1}{G_{TT}} & 0 & 0 \\ 0 & 0 & 0 & 0 & \frac{1}{G_{LT}} & 0 \\ 0 & 0 & 0 & 0 & 0 & \frac{1}{G_{LT}} \end{bmatrix} \begin{bmatrix} \sigma_L \\ \sigma_T \\ \sigma_T \\ \sigma_{LT} \\ \sigma_{LT} \\ \sigma_{TT} \end{bmatrix} \quad (5)$$

Eq. (5) was used to predict strains under equibiaxial tensile loading for a range of fibre and cross-fibre direction stiffnesses

(E_T and E_L respectively) derived from fitting to the experimental data (Fig. 4), in which high fibre direction stiffness was paired with low cross-fibre direction stiffness and vice-versa (Fig. 8) to represent extreme experimental cases. A range of estimated Poisson's ratios were used (Fig. 8), since these could not be derived directly from the experimental data. There are limitations to this analytical approach, particularly that rectus sheath is non-linear and closer to a fibre reinforced composite than a transversely isotropic material. However, this approach yields contractile stretches in the stiffer direction in almost all cases except low ∂_{LT} coupled with low E_T and high E_L (Fig. 8). These results can be interpreted by considering energy minimisation concepts: under biaxial loading, when significant anisotropy is present, stretch in the stiffer direction will may be contractile to reach a state of overall minimum potential energy.

As expected, Fig. 5 shows that the stiffness of the tissue in biaxial tension is higher than in uniaxial loading, but the anisotropy is less pronounced. In the uniaxial case, the fibre direction is approximately $11 \times$ stiffer, on average, than the cross-fibre direction while in the biaxial case, the fibre direction is $7 \times$ stiffer on average. This may be because constraining in the cross-fibre direction has a greater influence on effective stiffness than constraining in the fibre direction.

Figs. 4 and 5 show that a stress of 1.6 MPa seems to be a limiting factor in the cross-fibre direction for both uniaxial testing and in biaxial tests. This may reflect the matrix strength and reveals the

Fig. 9. FeBio stress-stretch plots for a (a) cross-fibre and (b) fibre direction sample in a gripping and non-gripping scenario.

importance of the fibre reinforcement in withstanding the stresses greater than 1.6 MPa created by IAP.

Statistical tests showed no anomalies in the testing procedure. Control charts showed no time bias affecting the results and normal probability plots showed normality in the results. ANOVAs revealed that variation between pig bellies was not statistically significantly different from variation within each belly.

4.3. Computational modelling

The Ogden model was a very good fit to the cross-fibre direction uniaxial experimental data, with high R^2 (98%) coupled with a low SSQD (0.391). The model for the fibre direction, which includes an uncoupled matrix represented by the Ogden model and an exponential power law representing the fibres also had a high R^2 (96%) and a SSQD of 18.1 (Fig. 6 and Table 1).

The fit for the biaxial simulation (Fig. 6) was not as good as for the uniaxial optimisation, but it still captures the essential difference in stiffness between the two directions. The material properties were obtained from uniaxial optimisation and biaxial tension was applied. This did not account for any additional interactions between the two load directions, and suggests a more complex model might be required.

As noted previously, during gripping the tissue under the grips spread, creating a somewhat dog-bone shaped sample from a sample that was originally rectangular. Upon application of uniaxial tension the tissue initially bulged in the lateral direction to re-create a rectangular specimen before beginning to narrow as would be expected. As a result of this unusual movement of the tissue, accurate measurement of the horizontal stretch was not possible in most cases, thus the Poisson's ratio could not be quantified. It is hypothesised that in the fibre direction case, the fibres become curved as a result of gripping. When tensile loading is initiated, these fibres straighten out, causing the sample to bulge outwards. Once the fibres are straight, the tissue then can behave as normal and contract in the unloaded direction. The FeBio simulation with the average model parameters was used to assess how this influenced the presented stress-stretch results. Fig. 9 shows that gripping the tissue did not significantly affect the stress at the centre of the specimen, with differences less than 5% in both load directions.

The fibre reinforced material model showed good potential to capture the uniaxial elastic behaviour of the tissue. In the biaxial simulations, the slopes the experimental curves is similar to the FeBio data, but the toe region is not well matched. This is possibly due to differences in preload application in the experimental and numerical simulations.

To the authors' knowledge, this study is the first to discuss the relationship between IAP and the loading of the abdominal wall.

The response of porcine rectus sheath to uniaxial and equi-biaxial tension has been characterised and has been shown to be similar to the response of human rectus sheath in the uniaxial case. Further work will see the development of a surrogate material to replicate rectus sheath in an experimental environment for development of surgical devices.

Conflict of interest statement

The authors declare no conflict of interest.

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