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Research paper

Mechanical characterization and constitutive modelling of the damage process in rectus sheath

P. Martins[a], *E. Peña*[b,c,*], *R.M. Natal Jorge*[a], *A. Santos*[d,e,f], *L. Santos*[d,e,f], *T. Mascarenhas*[f,g], *B. Calvo*[b,c]

[a] *IDMEC-Polo FEUP, Faculty of Engineering, University of Porto, Portugal*
[b] *Aragón Institute of Engineering Research, University of Zaragoza, Spain*
[c] *CIBER de Bioingeniería, Biomateriales y Nanomedicina (CIBER-BBN), Spain*
[d] *INML - Instituto Nacional de Medicina Legal, North Branch, Portugal*
[e] *CENCIFOR - Centro de Ciências Forenses, Portugal*
[f] *Faculty of Medicine, University of Porto, Portugal*
[g] *Department of Gynaecology/Obstetrics - Hospital de S. João - Porto, Portugal*

A R T I C L E I N F O

Article history:
Received 2 August 2011
Received in revised form
26 November 2011
Accepted 16 December 2011
Published online 24 December 2011

Keywords:
Large deformation
Damage mechanics
Hyperelastic
Anterior rectus sheath

A B S T R A C T

The aim of this study is to characterize and model the damage process in the anterior rectus abdominal aponeurosis (anterior rectus sheath) undergoing finite deformations. The resistance of the anterolateral abdominal aponeuroses is important when planning the surgical repair of incisional hernias, among other medical procedures. Previous experiments in prolapsed vaginal tissue revealed that a softening process occurs before tissue rupture. This nonlinear damage behaviour requires a continuum damage theory commonly used to describe the softening behaviour of soft tissues under large deformations. The structural model presented here was built within the framework of non-linear continuum mechanics. Tissue damage was simulated considering different damage behaviours for the matrix and the collagen fibres. The model parameters were fit to the experimental data obtained from anterior rectus sheath samples undergoing finite deformations in uniaxial tension tests. The tests were carried out with samples cut along the direction of the collagen fibres, and transversal to the fibres. Longitudinal and transverse mechanical properties of human anterior rectus sheath are significantly different.

The damage model was able to predict the stress–strain behaviour and the damage process accurately. The error estimations pointed to an excellent agreement between experimental results and model fittings. For all the fitted data, the normalized RMS error ε presented very low values and the coefficient of determination R^2 was close to 1. The present work constitutes the first attempt (as far as the authors know) to present a damage model for the human rectus sheath.

* *Corresponding author at:* Aragón Institute of Engineering Research, University of Zaragoza, Agustín de Betancourt BuildingMaría de Luna, 3, 50018 Zaragoza, Spain. Tel.: +34 876555233; fax: +34 976762578.
E-mail address: fany@unizar.es (E. Peña).

doi:10.1016/j.jmbbm.2011.12.005

1. Introduction

The rectus abdominal aponeurosis (rectus sheath) is a strong connective tissue made of collagen. It forms a sheath of dense white fibrous tissue that encloses the rectus abdominis muscles. These muscles are a superficial pair of muscles extending along the entire length of the anterior abdominal wall, from the thoracic cage to the pelvis. They have multiple functions, such as abdominal compression to increase the intra-abdominal pressure (parturition, defecation, etc.), support the abdominal contents and flexion. The rectus sheath enables the rectus abdominis muscles to slide through neighbouring structures and also protects them. Other lateral abdominal muscles (external oblique, internal oblique and transversus abdominis) become aponeurotic near the linea alba and contribute with fibres to the rectus sheath (Standring, 2004). These bundles of collagen fibres become parallel (highly oriented) and are visible to the naked eye.

The available literature on the mechanical behaviour of the rectus sheath is limited. However, there have been some important research with potential impact in fields such as plastic surgery (Silveira et al., 2010) or herniation processes (Ozdogan et al., 2006; Kureshi et al., 2008; Hernández et al., 2011) and hernia repair (Rath and Chevrel, 1997; Pans et al., 1997; Hollinsky and Sandberg, 2007). The resistance of the anterolateral abdominal aponeuroses is important when planning the surgical repair of incisional hernias (Rath and Chevrel, 1997). It also provides relevant data for mathematical modelling and model tissue engineering on collagen tissue herniation (Kureshi et al., 2008). Another research front investigates the collagen and elastic fibre content of the abdominal wall layers rich in connective tissue, such as the skin, the rectus sheath, the transversalis fascia and the peritoneum (Ozdogan et al., 2006). The scientific data correlates inguinal hernia with (local) connective tissue disorders. A significant step towards a better understanding of the morphology and biomechanics of the rectus sheath was carried out by Axer et al. (2001a,b). These authors provided information on rectus sheath fibre orientations, fibre diameter and distribution (Axer et al., 2001a), as well as an interpretation for the biomechanical role of different fibre groups (Axer et al., 2001b).

The basis for the research in this work, relies on uniaxial tension tests of rectus-sheath samples. The authors have already carried out work on the mechanical behaviour of prolapsed vaginal tissue (Calvo et al., 2009; Martins et al., 2010a; Peña et al., 2010b, 2011). During these investigations non-linear damage behaviour was observed before tissue rupture. To correctly model these phenomena requires a continuum damage theory commonly used to describe the softening behaviour of soft tissues under large deformations (Calvo et al., 2007; Peña, 2011c). The authors proposed a model for the damage process in vaginal tissue (Calvo et al., 2009), followed by a study of tissue softening behaviour (Peña et al., 2011). There are several constitutive models able to describe the failure of soft tissues (Tanaka and Yamada, 1990; Emery et al., 1997; Hokanson and Yazdami, 1997; Hurschler et al., 1997; Liao and Belkoff, 1999; Gasser and Holzapfel, 2002; Natali et al., 2005; Wulandana and Robertson, 2005; Balzani et al., 2006; Ionescu et al., 2006; Rodríguez et al., 2006; Volokh, 2007; Vita and Slaughter, 2007; Li and Robertson, 2009; Ciarletta and Ben-Amar, 2009; Forsell and Gasser, 2011; Gasser, 2011). Following our previous experience, in this paper we decide to use the directional damage model proposed by Calvo et al. (2007, 2009) due to its simplicity.

The medical community has started to acknowledge the importance of numerical simulation tools as an aid (Petros, 2007) to enhance medical knowledge of the physiology and pathophysiology of tissues, organs and systems. Along this lines this work addresses a major issue of the finite element simulation process, which is the link between the theoretical modelling and the biological material specificity provided by experimental measurement and the subsequent determination of the material model parameters.

The mechanical behaviour of the anterior rectus sheath evidences non-linear damage. Therefore, the aim of this paper is the characterization and the development of a damage model for the non-linear behaviour of the anterior rectus sheath.

Constitutive modelling and the characterization of the mechanical properties of the anterior rectus sheath, are fundamental to achieve an accurate modelling and simulation of the abdominal and pelvic structures of women, in particular when using the finite element method. The structural model presented here was built within the framework of non-linear continuum mechanics. Tissue damage was simulated considering different damage behaviours for the matrix and the collagen fibres, following the considerations presented in (Peña et al., 2008). Despite some recent efforts to study the properties and mechanics of the abdominal fasciae (Chi et al., 2010; Kureshi et al., 2008), none of the consulted bibliographies proposed a model for the non-linear soft damage of the anterior rectus sheath, in this sense, this work is the first.

2. Experimental data

The experimental component of this work relies on human tissue experimentation. The samples are exclusively of female tissue samples, since the current study lies under the scope of a broader investigation, aiming to characterize the mechanical properties of the soft tissues from the female pelvic cavity (Calvo et al., 2009; Peña et al., 2010b,a, 2011; Martins et al., 2008, 2010a,b, 2011; da Silva-Filho et al., 2010a,b). All procedures were done in compliance with the ethics guidelines regarding human tissue experimentation, following a protocol approved by the Board of the Forensic Pathology Service of the *North Branch of INML, I.P.* For this investigation, tissue samples of the left anterior rectus sheath were harvested from 12 fresh (4 ↔ 24 hours) female cadavers with ages ranging from 18 to 65 years old (46.08 ± 12.30), Table 1. Fig. 1 shows the anatomic site where anterior rectus sheath samples were excised from.

Using the experimental data presented here, the authors propose and validate a hyperelastic constitutive model able to characterize the mechanical behaviour of the anterior rectus sheath, including the deformation range where damage occurs.

Table 1 – Individual information used for the experimental tests. (-) non-available data. BMI – Body Mass Index. Individual numbers in bold mark the contributors to the parameter fitting.

Individual	Age (years)	Height (m)	Weight (kg)	BMI (kg/m^2)	Parity	Samples **L**	Samples **T**
I	51	1.58	62.4	25.00	1	2	0
II	51	1.56	55.0	22.60	1	1	1
III	47	1.68	91.0	32.24	1	1	2
IV	43	1.59	80.0	31.64	2	2	1
V	36	1.54	58.0	24.46	0	2	0
VI	18	1.79	104.2	32.52	0	1	1
VII	38	1.63	89.0	33.50	1	2	0
VIII	59	1.65	67.5	24.80	–	1	1
IX	40	1.58	53.0	21.23	1	2	0
X	51	1.60	66.0	25.78	2	2	0
XI	54	1.51	–	–	–	1	0
XII	65	1.59	65.0	25.71	–	2	1

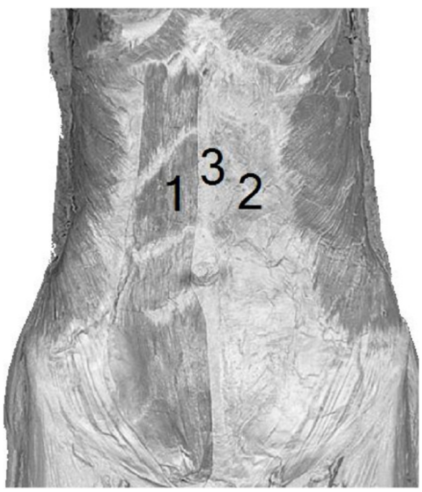


Fig. 1 – Anatomic site where samples were excised from, adapted from Parkin et al. (2007). 1 Rectus abdominis; 2 anterior rectus sheath; 3 Linea alba.

Based on the available patient information, it was decided to include a statistical study which investigated the potential influence of age, BMI and parity in the tissue mechanics. Ageing, overweight and parity have been linked to significant changes in the mechanical properties of pelvic tissues and linked with known female pelvic pathologies such as prolapse (Abramowitch et al., 2009; Alperin and Moalli, 2006; Jelovsek et al., 2007). Along with these individual characteristics, the fibre directions were also considered since they affect the mechanical properties. This investigation required all available data from the individuals I to XII in Table 1.

2.1. *Test specimens*

All mechanical tests were developed at the University of Porto (Portugal) in the biomechanics laboratory (LBM-IDMEC). The test specimens were cut along the direction of the collagen fibres (**L**-Longitudinal) and transverse (**T**-transverse) to the direction of the fibres, Fig. 2. **The specimens, with a rectangular shape were prepared avoiding tissue regions where cuts or apparent damage was noticed. The thickness of the specimens was 1.00 ± 0.17 mm and the typical length to width ratio was 1:4**. During sample collection it retraction of the tissue patch resected, i.e., the effect of residual strains found in many soft tissues (Fung, 1993) was observed. This effect was thoroughly described and quantified in a recent work by Hernández et al. (2011). The human cadavers used in the present work were dead for at least 4 h before any of the samples were collected. The fundamental difference to the work referred (Hernández et al., 2011) lies in the fact that the samples were collected immediately after the animals (rabbits), used in the tests were sacrificed. In their work the retraction of the tissue was clearly observed after 15 min, a time interval much inferior to the 4 h interval of the present work.

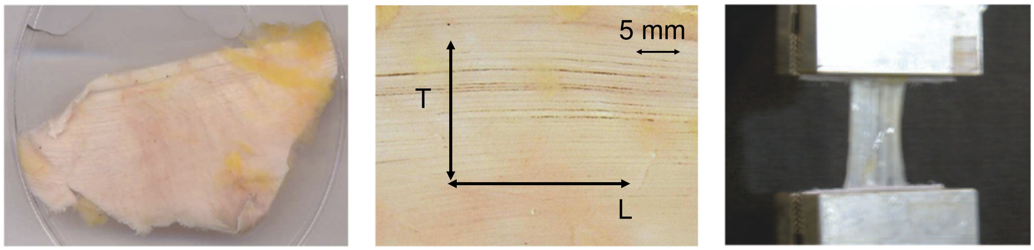
2.2. *Test protocol*

The mechanical tests, consisting of simple tension tests performed at 5 mm/min along the **collagen fibres** (longitudinal) and perpendicular (transverse) to the fibres (see Fig. 2) were conducted on a drive system previously validated for testing biological materials (Martins et al., 2006).

The clamp-tissue fixation was prepared using Velcro tape glued to sandpaper. The sandpaper promotes friction with the metal clamp, while the Velcro tape (Velcro hooks side) stays in contact with the tissue sample. Four pieces of sandpaper-Velcro were used to fix each specimen, as shown in Fig. 3. To ensure that no sliding occurred during the tests all the mechanical tests were recorded on video and posteriorly visually validated. The experimental data used was exclusively from validated mechanical tests.

Of a total of 12 valid tests, only 6 were performed in both directions (**L** and **T**) due to the small size of some samples. To characterize damage behaviour, the specimens were stretched until full rupture. **Given the composition of the human rectus sheath and despite preconditioning (in principle) allows for a better repeatability of the stress–strain response of the sample under the same loads, the specimens were not preconditioned prior to the mechanical test** (Axer et al., 2001a,b).

The load (N) and displacement (mm) data were acquired using a load cell ($F_{max} = 200$ N) and a displacement sensor. All the tests were performed at room temperature (≈20 °C). No significant dehydration effects were observed during the tests



(a) Rectus sample. **(b) Direction of collagen fibres.** **(c) Rectus sample assem-bled.**

Fig. 2 – Schematic showing (a) and anterior rectus sheath sample (b) the direction of the fibres in a sample (c) sample assembled in the testing grips. L-longitudinal or, along the collagen fibres; T-transverse, or perpendicular to the collagen fibres.

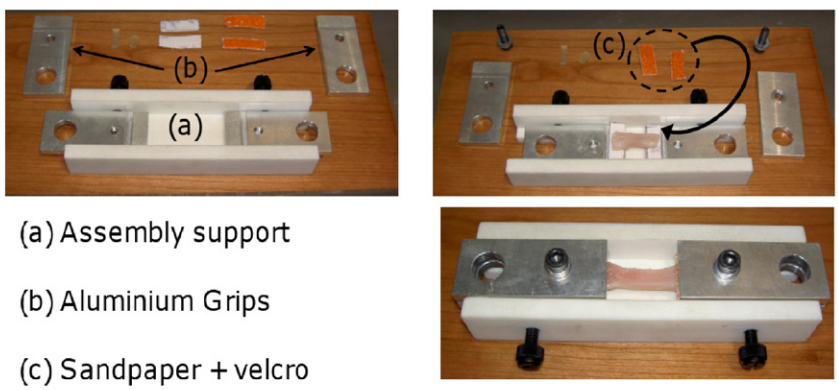


Fig. 3 – Tissue-clamp fixation.

(7 ↔ 10 min) (Rubod et al., 2007). The criteria for the thickness variations in this work was to take the initial thickness at middle length. The stretch of the sample as $\lambda = \frac{L_0+\Delta L}{L_0}$, where L_0 is the initial length between the reference and ΔL is the displacement, and the Cauchy stress was obtained using the expression $\sigma = \lambda \frac{F}{A_0}$, where F is the load applied during the test and A_0 is the area of the sample at the beginning of the test.

2.3. Results

The main aim of this research is to study the mechanical behaviour of the human rectus sheath. This includes building a constitutive model (including damage) proposed in Section 3 and a statistical analysis of the experimental data. A detailed characterization of the experimental data may support the model predictions enabling a better understanding the mechanical behaviour of the human rectus sheath.

The results of different subjects were compared. Fig. 4, shows that, the curves have a high dispersion among different individuals. Transverse data (**T**) has higher dispersion than longitudinal (**L**) data.

The stress–strain curves show the strong non-linearity observable in the aponeurosis tissue. To gain some insight into the influence of specific individual characteristics on the mechanical properties of the anterior rectus sheath, the maximum values of stress and stretch, σ_R and λ_R respectively,

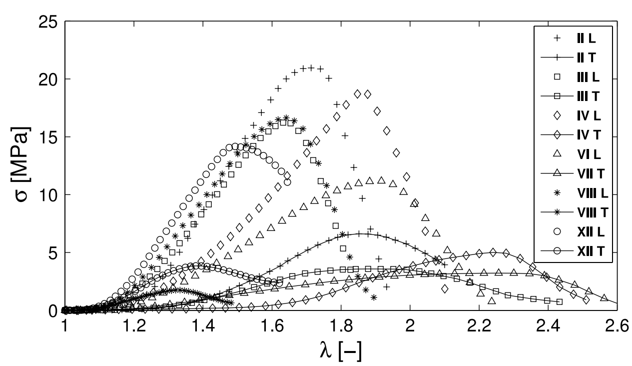


Fig. 4 – *In vitro* experimental data, the curves with a continuous line correspond to transverse T test samples and the curves without lines to the longitudinal L samples.

and the modulus in the linear portion of the stress–stretch curve that precedes the damage region $E_t = \frac{\Delta\sigma}{\Delta\lambda}$, for both **L** and **T** samples were analysed. For the statistical analysis the Statistical Package for the Social Sciences (SPSS) software package version 16.0 (SPSS Inc., Chicago, IL, USA) was used. In Table 2(a) all quantities were presented as the mean ± SD (Standard deviation). The normal distribution of the data was verified using the Kolmogorov–Smirnov and Shapiro–Wilk tests. The mechanical properties were compared for the

Table 2 – Mechanical properties, σ_R, λ_R, E_t for L and T samples; (a) Comparison between L and T samples using Student's paired-samples t test applied to subjects II, III, IV, VI, VIII, XII; (b) Univariate analysis using Student's independent-samples t test to evaluate the influence of Age, Parity or BMI on the mechanical properties of the anterior rectus sheath. The statistically significant differences (p < 0.05) are indicated in bold type.

(a)

	n	σ_R (MPa)	λ_R	E_t (MPa)
L	6	12.84 ± 3.74	1.71 ± 0.15	30.30 ± 10.49
T	6	3.38 ± 1.67	1.75 ± 0.42	10.07 ± 5.32
p (two-tailed)		<**0.001**	0.793	<**0.001**

(b)

Group	Longitudinal sample (L) n	σ_R (MPa)	λ_R	E_t (MPa)	Transverse sample (T) n	σ_R (MPa)	λ_R	E_t (MPa)
Age								
≥50	6	14.62 ± 6.25	1.64 ± 0.13	36.19 ± 12.55	3	2.27 ± 1.40	1.40 ± 0.08	10.32 ± 6.78
<50	6	10.81 ± 3.91	1.67 ± 0.13	25.98 ± 8.44	3	4.50 ± 1.12	2.10 ± 0.26	9.81 ± 4.96
p (two-tailed)		0.235	0.715	0.129		0.097	**0.012**	0.921
BMI								
≥25 kg/m^2	7	12.83 ± 4.83	1.67 ± 0.13	31.11 ± 11.01	4	4.34 ± 0.97	1.92 ± 0.42	11.79 ± 5.65
<25 kg/m^2	4	10.15 ± 4.47	1.65 ± 0.14	26.32 ± 9.68	2	1.47 ± 0.39	1.41 ± 0.11	6.64 ± 3.20
p (two-tailed)		0.390	0.860	0.489		**0.019**	0.181	0.312
Parity								
>0	7	11.79 ± 5.47	1.67 ± 0.11	28.91 ± 11.43	3	3.83 ± 2.28	1.85 ± 0.39	9.78 ± 5.01
=0	2	9.56 ± 2.30	1.72 ± 0.25	22.10 ± 4.16	1	3.22	2.25	4.47
p (two-tailed)		0.606	0.672	0.453		–	–	–

following groups,

(a) **L** and **T** samples.

The statistical question was: is there a statistically significant difference between **L** and **T** mechanical properties?

The Student's paired t test with a significance level of $p < 0.05$ was used. Only subjects II, III, IV, VI, VIII and XII were considered since both **L** and **T** data was available.

(b) **Age**- Older/Younger; **BMI**- Heavier/Lighter; **Parity**- No Child/Child(ren).

The statistical questions regarding the mechanical properties of anterior rectus sheath were.

Does ageing affect the mechanical properties significantly?

Does obesity affect the mechanical properties significantly?

Does childbirth affect the mechanical properties significantly?

In all 3 cases the Student's unpaired t test with a significance level set of $p < 0.05$ was used.

A statistically significant ($p < 0.001$) difference between **L** and **T** samples for σ_R and E_t was observed as shown in Table 2.3(a). This difference can also be seen in the stress–stretch results in Fig. 4. Generally the anterior rectus sheath is stiffer along the longitudinal direction and more flexible on the transverse one.

The influence of ageing, obesity (BMI) and parity on the biomechanical properties could not be clearly established. The transverse sample shows two significant results: λ_R regarding Age ($p = 0.012$) and σ_R regarding BMI ($p = 0.019$). In both cases, caution should be exercised due to the reduced number of elements of the groups in question ($2 \le n \le 4$). However, there is some sense in expecting the obese subjects (BMI $\ge$ 25 Kg/m^2) to have stronger anterior rectus sheaths, since an increase in the intra-abdominal pressure affecting the abdominal wall can be expected (Frezza et al., 2007). No statistically significant results were seen for the longitudinal samples.

Correlations between the mechanical properties were investigated. A linear tendency was noted in the graphical analysis (scatter plots) of correlations for both L and T samples. Therefore, the statistical correlations were calculated using the Pearson correlation coefficient. The significance level was set to $p < 0.05$. Fig. 5 shows a strong positive correlation (0.935) between E_t and σ_R for the L sample, ($p \ll 0.001$).

The Pearson correlation coefficient in this case is very close to 1 meaning that an increase in the observed maximum stress (σ_R) is associated with an increase in stiffness (E_t). This data shows that the longitudinal direction (fibre direction) of the anterior rectus sheath is not only stronger but is also stiffer than the transverse direction.

3. Constitutive modelling

Following the histological and morphological results of Axer et al. (2001a,b), the tissue was assumed to behave as a transversely isotropic hyperelastic material. The authors used confocal laser scanning microscopy to perform a three-dimensional reconstruction of ventral and dorsal segments (Axer et al., 2001b) of the rectus sheath which evidences its transverse isotropy.

To validate the non-linear damage model proposed, the model results (theoretical) were compared with experimental data from individuals II, III, IV, VI, VIII, XII (Table 1) where both longitudinal **L** and transverse **T** data were available.

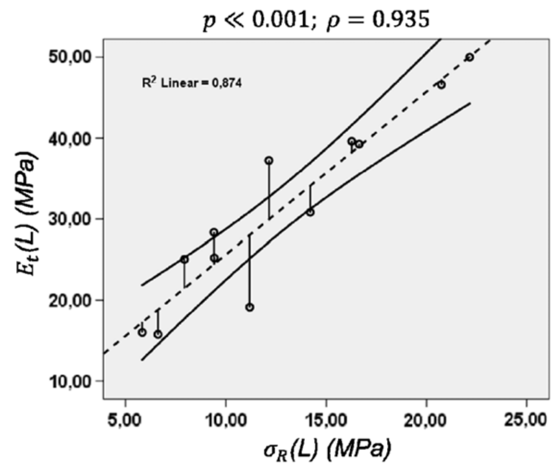


Fig. 5 – Correlation between E_t and σ_R for the longitudinal sample. The continuous line indicates the 95% confidence region. The correlation coefficient (R^2) of the tendency line (dashed) evidences the linear correlation between the two variables.

3.1. *Anisotropic hyperelastic response of soft tissues*

Let $\mathbf{x} = \chi(\mathbf{X}, t) : \Omega_0 \times \mathbb{R} \to \mathbb{R}^3$ denote the motion mapping and let $\mathbf{F}$ be the associated deformation gradient. Here $\mathbf{X}$ and $\mathbf{x}$ define the respective positions of a particle in the reference Ω_0 and current Ω configurations so that the mapping φ represents a motion of the body B that establishes the trajectory of a given point when moving from its reference position $\mathbf{X}$ to $\mathbf{x}$. The two-point deformation gradient tensor is defined as $\mathbf{F}(\mathbf{X}, t) := \nabla_{\mathbf{X}}\varphi(\mathbf{X}, t)$. Further, let $J \equiv \det\mathbf{F}$ be the Jacobian of the motion. To properly define volumetric and deviatoric responses in the nonlinear range, we introduced the following kinematic decomposition $\bar{\mathbf{F}} = J^{-\frac{1}{3}}\mathbf{F}$ and $\bar{\mathbf{C}} = \bar{\mathbf{F}}^T\bar{\mathbf{F}}$ (Flory, 1961). The term $J^{\frac{1}{3}}\mathbf{I}$ is associated with volume-changing deformations, while $\bar{\mathbf{F}}$ is related to volume-preserving deformations. We shall call $\bar{\mathbf{F}}$ and $\bar{\mathbf{C}}$ the modified deformation gradient and the modified right Cauchy–Green tensors, respectively. For transversely isotropic hyperelastic materials, it is necessary to introduce the unit vectors $\mathbf{m}_0$ describing the anisotropy direction. The collagen fibre moves with the material points of the continuum body, so the stretch λ_m of the fibres is defined as the ratio between its lengths at the deformed and reference configurations (Spencer, 1971).

To characterize isothermal processes, we postulated the existence of a unique decoupled representation of the strain-energy density function Ψ (Simo and Taylor, 1985) that explicitly depends on both the right Cauchy–Green tensor $\mathbf{C}$ and the fibres direction $\mathbf{m}_0$

$$\Psi(\mathbf{C}, \mathbf{M}) = \Psi_{\text{vol}}(J) + \bar{\Psi}(\bar{\mathbf{C}}, \mathbf{M}) = \Psi_{\text{vol}}(J) + \bar{\Psi}(\bar{I}_1, \bar{I}_2, \bar{I}_4), \tag{1}$$

where $\Psi_{\text{vol}}(J)$ and $\bar{\Psi}$ are given scalar-valued functions of J, $\bar{\mathbf{C}}$, $\mathbf{M} = \mathbf{m}_0 \otimes \mathbf{m}_0$ respectively that describe the volumetric and isochoric responses of the material (Weiss et al., 1996), $\bar{I}_1$ and $\bar{I}_2$ are the first and second modified strain invariants of the symmetric modified Cauchy–Green tensor $\bar{\mathbf{C}}$. Finally, the invariant $\bar{I}_4 = \mathbf{C} : \mathbf{M} = \lambda_m^2$ characterizes the constitutive response of the fibres (Spencer, 1971).

From the Clausius–Planck inequality, the constitutive equation for compressible hyperelastic materials can be defined as

$$\mathbf{S} = 2\frac{\partial\Psi(\mathbf{C}, \mathbf{M})}{\partial\mathbf{C}} = Jp\mathbf{C}^{-1} + 2\sum_{j=1,2,4}\frac{\partial\bar{\Psi}}{\partial\bar{I}_j}\frac{\partial\bar{I}_j}{\partial\bar{\mathbf{C}}} : \frac{\partial\bar{\mathbf{C}}}{\partial\mathbf{C}}$$
$$= \mathbf{S}_{\text{vol}} + \bar{\mathbf{S}} = Jp\mathbf{C}^{-1} + \bar{\mathbf{S}}, \tag{2}$$

where the second Piola–Kirchhoff stress $\mathbf{S}$ consists of a purely volumetric contribution $\mathbf{S}_{\text{vol}}$, a purely isochoric one $\bar{\mathbf{S}}$ and $p = \frac{d\Psi_{\text{vol}}(J)}{dJ}$ is the hydrostatic pressure.

The Cauchy stress tensor $\boldsymbol{\sigma}$ is $1/J$ times the push-forward of $\mathbf{S}$ ($\boldsymbol{\sigma} = J^{-1}\chi_*(\mathbf{S})$)

$$\boldsymbol{\sigma} = p\mathbf{1} + \frac{2}{J}\left[\left(\frac{\partial\bar{\Psi}}{\partial\bar{I}_1} + \bar{I}_1\frac{\partial\bar{\Psi}}{\partial\bar{I}_2}\right)\bar{\mathbf{b}} - \frac{\partial\bar{\Psi}}{\partial\bar{I}_2}\bar{\mathbf{b}}^2 + \bar{I}_4\frac{\partial\bar{\Psi}}{\partial\bar{I}_4}\mathbf{m}\otimes\mathbf{m}\right.$$
$$\left.- \frac{1}{3}\left(\frac{\partial\bar{\Psi}}{\partial\bar{I}_1}\bar{I}_1 + 2\frac{\partial\bar{\Psi}}{\partial\bar{I}_2}\bar{I}_2 + \frac{\partial\bar{\Psi}}{\partial\bar{I}_4}\bar{I}_4\right)\mathbf{1}\right], \tag{3}$$

with $\mathbf{1}$ the second-order identity tensor and $\bar{\mathbf{b}} = \bar{\mathbf{F}}\bar{\mathbf{F}}^T$ the modified left Cauchy–Green tensor.

3.2. *Modelling of the damage process*

In order to reproduce the damage process in the human rectus sheath, we considered the directional damage model proposed by Calvo et al. (2007). The damage phenomenon is assumed to affect only the isochoric elastic part of the deformation, as proposed by Simo (1987). The free energy density can then be written in a decoupled form, such as

$$\Psi(\mathbf{C}, \mathbf{M}, D_m, D_f) = \Psi_{\text{vol}}(J) + (1 - D_m)\,\bar{\Psi}_0^m(\bar{\mathbf{C}})$$
$$+ (1 - D_f)\,\bar{\Psi}_0^f(\bar{\mathbf{C}}, \mathbf{M}, \mathbf{N}), \tag{4}$$

where $\bar{\Psi}_0^m$ denotes the isochoric effective strain energy density of the undamaged material, which describes the elastic response of the matrix, and $\bar{\Psi}_0^f$ denotes the isochoric effective strain energy of the undamaged material, which describes the isochoric elastic response of the collagen fibres. The factors $(1 - D_m)$ and $(1 - D_f)$ are known as the reduction factors (Simo and Ju, 1987), where the internal variables $D_m \in [0, 1]$ and $D_f \in [0, 1]$ are normalized scalars referred to as the damage variables for the matrix and fibres respectively.

From (2) and (4)the following can be obtained:

$$\mathbf{S} = \mathbf{S}_{\text{vol}} + (1 - D_m)\bar{\mathbf{S}}_0^m + (1 - D_f)\bar{\mathbf{S}}_0^f. \tag{5}$$

The evolution of the damage parameters D_m and D_f is characterized by an irreversible equation of evolution as follows. Let Ξ_s^m, Ξ_s^f be defined by the expression (Simo, 1987)

$$\Xi_s^m = \sqrt{2\bar{\Psi}_0^m(\bar{\mathbf{C}}(s))} \quad \text{and} \quad \Xi_s^f = \sqrt{2\bar{\Psi}_0^f(\bar{\mathbf{C}}(s))}, \tag{6}$$

where $\bar{\mathbf{C}}(s)$ is the modified right Cauchy–Green tensor at time s. Now, let Ξ_t^m, Ξ_t^f be the maximum values of Ξ_s^m, Ξ_s^f over the past history up to the current time t, that is (Simo, 1987)

$$\Xi_t^m = \max_{s\in(-\infty,t)}\sqrt{2\bar{\Psi}_0^m(\bar{\mathbf{C}}(s))} \quad \text{and}$$
$$\Xi_t^f = \max_{s\in(-\infty,t)}\sqrt{2\bar{\Psi}_0^f(\bar{\mathbf{C}}(s))}. \tag{7}$$

A damage criterion for the ground substance or matrix in the strain space is defined bythe condition that, at any time t of the loading process, the following expression is fulfilled (Simo, 1987)

$$\Phi^k(\bar{\mathbf{C}}(t), \Xi_t^k) = \sqrt{2\bar{\Psi}_0^k(\bar{\mathbf{C}}(t))} - \Xi_t^k \leq 0, \quad (8)$$

when the condition $\Phi^k(\bar{\mathbf{C}}(t), \Xi_t^k) = 0$ is fulfilled, the damage increases and $k = m, f$.

The damage evolution functions in (4) for the matrix and collagen fibres used here correspond to the expressions proposed by Peña et al. (2008)

$$D_k(\Xi_{k_t}) = \begin{cases} 0 & \text{if } \Xi_t^k < \Xi_{\min}^{0_k} \\ \xi^2[1 - \beta_k(\xi^2 - 1)] & \text{if } \Xi_{\min_k}^{0_k} \leq \Xi_t^k \leq \Xi_{\max}^{0_k}, \\ 1 & \text{if } \Xi_t^k > \Xi_{\max}^{0_k} \end{cases} \quad (9)$$

where $\xi = \frac{\Xi_t^k - \Xi_{\min}^{0_k}}{\Xi_{\max}^{0_k} - \Xi_{\min}^{0_k}}$ is a dimensionless variable and $\Xi_{\min}^{0_k}$ are the variables (8) associated to the strain energies at initial damage for matrix and collagen fibres respectively; $\Xi_{\max}^{0_k}$ are the variables (8) associated to the strain energy at total damage for matrix and collagen fibres, and β_k is model exponential parameters. To maintain a monotonic increasing function of D_k with Ξ_{k_t} (9), it is required that $\beta_k \in [-1.0, 1.0]$ (Calvo et al., 2009; Peña et al., 2009; Peña, 2011b).

It should be pointed out that the model defined by Eqs. (4)–(9) is phenomenological. The damage parameters estimated by the applied damage model cannot be interpreted since the constitutive model lacks a clear physical meaning.

3.3. *Parameter estimation*

The experimental data presented in Section 2 and depicted in Fig. 4 was used to evaluate the ability of the constitutive model presented in Section 3.2 to predict the damage process in the anterior rectus sheath. The elastic parameters were determined from the elastic region of the curves delimited by $\Xi_{\min_k}$ while the damage parameters were fitted from the softening region of the curves delimited by $\Xi_{\min_k}$ and $\Xi_{\max_k}$.

The experimental data showed that the specimens experienced finite strains for small loads and a strongly marked non-linearity was found in the tissue. The tissue was assumed as incompressible, that is $I_3 = J^2 = 1$, so $\bar{I}_i = I_i$ with $i = 1, 2, 4, 6$, and $\Psi = (1 - D_m)\hat{\Psi}_0^m(\mathbf{C}) + (1 - D_f)\hat{\Psi}_0^f(\mathbf{C}, \mathbf{M}, \mathbf{N})$ (Ogden, 1996; Bonet and Kulasegaram, 2000).

If we consider a simple tension test in the direction of the collagen fibres with $\mathbf{n}_0 = \mathbf{e}_z$, then $\lambda_z = \lambda$, $\lambda_x = \lambda_y = \lambda^{-\frac{1}{2}}$ and $I_4 = \lambda^2$, the Cauchy stress tensor $\boldsymbol{\sigma}_\mathbf{e}$ becomes diagonal with $\sigma_{zz} = \sigma$, $\sigma_{xx} = \sigma_{yy} = 0$. The isotropic response was modelled by means of the Demiray 1988's SEF (Demiray et al., 1988) while the fibre response was represented by Calvo's SEF (Calvo et al., 2009),

$$\begin{aligned} &\Psi = \hat{\Psi}_{\text{iso}}^m + \hat{\Psi}_{ani}^f \\ &\hat{\Psi}_{\text{iso}}^m = \frac{c_1}{c_2}\left(\exp^{\frac{c_2}{2}(I_1 - 3)} - 1\right) \\ &\hat{\Psi}_{ani}^f = 0 \qquad I_4 < I_{4_0} \\ &\hat{\Psi}_{ani}^f = \frac{c_3}{c_4}[\exp(c_4[I_4 - I_{4_0}]) - c_4[I_4 - I_{4_0}] - 1] \\ &\quad I_4 > I_{4_0} \text{ and } I_4 < I_{4_{ref}} \\ &\hat{\Psi}_{ani}^f = c_5\sqrt{I_4 - I_{4_0}} + c_6 \ln(I_4 - I_{4_0}) + c_7 \quad I_4 > I_{4_{ref}}. \end{aligned} \quad (10)$$

$I_{4_{ref}}$ characterizes the stretch at which collagen fibres start to be straightened. It was assumed that the strain energy corresponding to the anisotropic terms only contribute to the global mechanical response of the tissue when stretched, that is $I_4 > I_{4_0}$. For simplicity of the strain energy function (SEF), it is usual in biomechanics not to include the invariants I_2 in the SEF. In Eq. (10), $c_1 > 0$, $c_3 > 0$, $c_5 > 0$ and $c_6 > 0$ are stress-like parameters, $c_2 > 0$ and $c_4 > 0$ are dimensionless parameters while c_7 is an energy-like parameter. Note that c_5, c_6 and c_7 are not independent parameters as they enforce strain, stress and stress derivative's continuity.

The fitting of the experimental data was developed by using a Levenberg–Marquardt type minimization algorithm (Marquardt, 1963), that is commonly used for experimental data fitting. The fittings were performed multiple times starting with randomized parameters due to problems with global minima (Fung, 1993). The results presented are the parameters that gave minimal values for χ^2 (11)

$$\chi^2 = \sum_{i=1}^{n}[(\sigma^{\exp} - \sigma^{\Psi})_{iL}^2 + (\sigma^{\exp} - \sigma^{\Psi})_{iT}^2], \quad (11)$$

where $\sigma_i^{\exp}$ and σ_i^{Ψ} represent the measured and the fitted stress values for the ith data point, respectively. L and T subscript means the direction of the test, longitudinal and transverse respectively.

The quality of the fitting was measured by computing the coefficient of determination R^2 and the normalized mean square root error $\varepsilon = \frac{\sqrt{\frac{\chi^2}{n-q}}}{\mu}$, where q is the number of parameters of the SEF, $n - q$ is the number of degrees of freedom, and μ is the mean stress defined as $\mu = \frac{1}{n}\sum_{i=1}^{n}[\sigma]_i$.

3.4. *Results*

The fitting results for each specimen are shown in Table 3. The values of R^2 are close to 1 and ε values are low for all the fitted data, indicating the quality of the fittings. The stretch and energy associated with the initial damages of the matrix $\Xi_{\min_m}$ and the collagen fibres $\Xi_{\min_f}$ also resulted in differences between specimens. The same occurred for the strain energy at total damage for matrix $\Xi_{\max_m}$ and collagen fibres $\Xi_{\max_f}$. β_k that depends strongly on these parameters and the size of the softening region of the curve.

The stress–stretch plots in Fig. 6 show the mechanical response, i.e., Cauchy stress (σ) vs. stretch (λ), of the different fascia tissues for subjects II, III, IV, VI, VIII, XII. All tissues exhibited a pronounced non-linear mechanical response with a significant softening during the damage process. As can be seen in the plots of Fig. 6, the correlation between the experimental data and the model response was satisfactory. This indicates a good agreement between the model and the experimental data for aponeurosis tissue for each sample, thus supporting the use of this constitutive model.

Some remarks are needed regarding the fittings of the transverse samples. Fig. 6 evidences a degradation of the results in the transverse direction when compared to the longitudinal one. This is a critical issue in specimens VIII and XII where the isotropic model is unable to fit the damage part accurately. This phenomenon is due to the small number of

Table 3 – Material model parameters obtained for the tissue *in vitro* tests. Numbers II to XII indicate the sets of parameters of the experimental data fitting.

	C_1 (MPa)	C_2 (-)	C_3 (MPa)	C_4 (-)	C_5 (MPa)	C_6 (MPa)	C_7 (MPa)	I_{4_0} (-)
II	0.3380	1.0511	1.3366	2.0000	43.2414	−53.2830	−42.1348	1.4400
III	0.2434	0.8000	0.0064	9.6300	31.8214	−36.9188	−31.4118	1.0000
IV	0.1062	0.8000	0.1859	2.2881	25.9590	−32.2022	−25.1737	1.1025
VI	0.2500	0.2500	0.3884	1.3765	16.7155	−20.7564	−16.2061	1.2200
VIII	0.6029	0.8000	0.0217	9.0502	30.9531	−35.8784	−30.5612	1.1025
XII	0.5000	1.7500	0.5000	3.1001	41.0746	−48.2455	−40.4341	1.2100
	β_m (-)	β_f (-)	$\Xi_{\min_m}$ (MPa$^{1/2}$)	$\Xi_{\max_m}$ (MPa$^{1/2}$)	$\Xi_{\min_f}$ (MPa$^{1/2}$)	$\Xi_{\max_f}$ (MPa$^{1/2}$)	R^2	ε
II	0.0000	1.0000	1.0897	2.5195	2.0040	3.6337	0.9670	0.1706
III	0.7000	1.0000	0.9459	2.3429	2.0316	3.0808	0.9862	0.1483
IV	0.6965	1.0000	1.1468	2.1691	2.4658	3.3009	0.9737	0.2045
VI	1.0000	1.0000	1.6744	2.6522	1.5175	3.0237	0.9640	0.1545
VIII	−1.0000	1.0000	1.3102	1.6976	2.0498	3.2314	0.9667	0.1774
XII	−1.0000	1.0000	0.9232	1.9219	1.2939	2.7540	0.9243	0.3179

parameters of the SEF's elastic component when compared to the anisotropic one. In some cases, it cannot reproduce the mechanical behaviour, in particular when damage appears at higher strains, see Fig. 6.f where $\varepsilon = 0.317955$.

4. Discussion

The present study investigated the mechanical damage process in the rectus abdominis aponeurosis (rectus sheath) undergoing finite deformations. The resistance of the anterolateral abdominal aponeuroses is important when planning the surgical repair of incisional hernias, among other medical procedures.

The experimental data presents evidence of non-linear mechanical behaviour; therefore the non-linear SEF for the elastic part of the tissue presented by Calvo et al. (2009) was used in this work. We also examined the ability of an uncoupled damage model previously used to model vaginal tissue (Calvo et al., 2009) to capture the non-linear and anisotropic damage response of the anterior rectus sheath in the large strain domain. The predictions of the model were examined in detail and compared with those from the *in vitro* experiments.

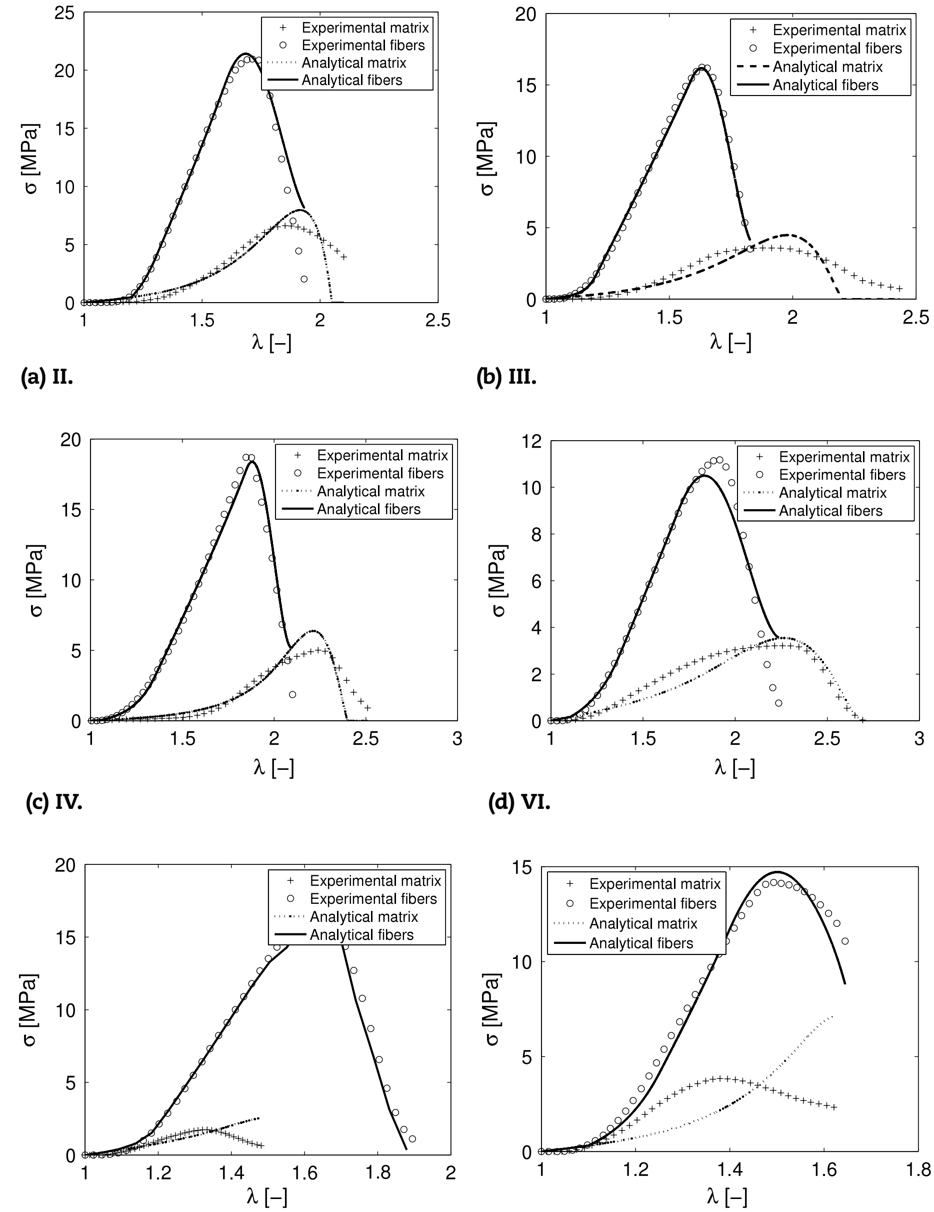
The mechanical properties of the anterior rectus sheath were extracted from the experimental data presented in Section 2. The ultimate stress (σ_R), stretch (λ_R) and the tangent modulus (E_t) were obtained for the longitudinal **L** ($n = 12$) and transverse **T** samples ($n = 6$). The normal distribution of the data was verified using the Kolmogorov–Smirnov and Shapiro–Wilk tests, therefore parametric statistical techniques were chosen for the analysis. To compare patient groups and investigate the influence of factors such as ageing, obesity and parity on the mechanical properties of the anterior rectus sheath, the Student's *t* test (significance level set at $p < 0.05$) was used. The statistical correlations were evaluated using Pearson's correlation coefficient, ranging from −1 (absolute negative correlation) to +1 (absolute positive correlation).

The experimental result analysis suggested that in soft tissues, the elastic properties and damage behaviour were both characterized by anisotropy. The elastic behaviour of the tested specimens was noticeably non-linear and anisotropic to high strains. Fig. 4 shows the elastic and damage effects for both longitudinal and transverse directions, however these phenomena are different, depending on the test direction. Longitudinally tested specimens are stiffer and have higher maximal stresses than specimens tested in the transverse direction. There is a statistically significant difference ($p < 0.001$ for Student's paired-samples *t* test) between longitudinal and transverse samples in respect to σ_R and E_t (Table 2(a)), which confirms unequivocally the differences between directions.

A close observation of the experimental data revealed a damage process with significant variations, even among specimens from the same sample. The damage phenomenon is different for the different directions. The longitudinal samples present a narrow damage zone (of the stress–stretch curve) which indicates a rapid damage process, however the transverse samples suffered a soft damage process identifiable by the smooth round-shaped damage zone, Fig. 4.

Longitudinal data did not reveal any influence of ageing, obesity or parity on the mechanical properties of the anterior rectus sheath, Table 2(b). The traversal data however, showed two significant results, λ_R regarding ageing ($p = 0.012$) and σ_R regarding obesity ($p = 0.019$). As shown in Fig. 4 the transverse sample evidenced a strong dispersion of λ_R which cast some doubt on this result. The effect of obesity in the ultimate stress σ_R, despite sharing the same dispersion issue, may be interpreted as an obesity (BMI $\geq$ 25 Kg/m^2) induced strengthening of the anterior rectus sheath.

Table 2(b) showed that σ_R results are higher for the obese group in both longitudinal (12.83 ± 4.83 vs 10.15 ± 4.47) and transverse data (4.34 ± 0.97 vs 1.47 ± 0.39). Obesity i.e., increased BMI, has been correlated with increased intra-abdominal pressure in women, (Frezza et al., 2007). In women, the intra-abdominal pressure increase may affect the abdominal wall, leading to stronger and stiffer (E_t: longitudinal 31.11 ± 11.01 vs 26.32 ± 9.68; transverse 11.79 ± 5.65 vs 6.64 ± 3.20) anterior rectus sheaths. These conclusions should be validated with biomechanical studies considering a larger population to minimize statistical effects associated to a small *n*.



(a) II. (b) III. (c) IV. (d) VI. (e) VIII. (f) XII.

Fig. 6 – Experimental *versus* analytical stress–stretch relationships for each individual.

A strong positive correlation (Pearson coefficient, ρ = 0.935, $p \ll 0.001$) between E_t and σ_R for the longitudinal sample was observed. Fig. 5 shows that this correlation is linear (R^2 = 0.874). This relation means that in the longitudinal direction, stronger anterior rectus sheaths are stiffer as well. This correlation was not observed in the transverse data.

Regarding the mathematical model, the specificity of the experimental data led to the use of the damage model presented by Calvo et al. (2009). For the elastic region, the SEF proposed by Martins et al. (2010a) was used (10). It takes into account the mechanically distinctive regions (exponential and linear) and was previously able to fit the experimental data accurately in vaginal tissue (Martins et al., 2010a), abdominal tissue (Hernández et al., 2011) and skeletal muscle (Calvo et al., 2011; Grasa et al., 2011). For the damage region, the damage functions used for the matrix and collagen fibres correspond to the expressions (9) proposed by Peña et al. (2008). The values obtained for R^2 are very close to 1 ($R^2 = 0.9637 \pm 0.0209$) and ε values are very low ($\varepsilon = 0.19556 \pm 0.0631$) for the fitted data showing good fitting quality. The model contains eleven parameters, five of which are related to elastic behaviour (c_1, c_3, c_4, $\bar{I}_{4_0}$ and $\bar{I}_{4_{ref}}$) and six to the damage mechanical properties (β_m, β_f, $\Xi_{\min_m}$, $\Xi_{\max_m}$, $\Xi_{\min_f}$ and $\Xi_{\max_f}$). The model is well suited for predicting abrupt and prolonged failure regions of highly stretched samples. It must be noted that the model was also able to describe the deformation behaviour appropriately over the entire experimental loading range.

The mechanics of the human anterior rectus sheath and other abdominal aponeurotic tissues, has been studied and modelled by several authors. The works of Axer et al. (2001a,b) investigated with great detail the structure and morphology of the rectus sheath. They constitute the basis for the symmetry considerations (transverse isotropy) used in the damage model proposed.

The results available in literature, agree with the significant difference observed ($p > 0.001$) regarding the ultimate stress of anterior rectus sheath in the transverse and longitudinal directions (Table 2(a)). Hollinsky and Sandberg (2007) reported maximum stresses of $\sigma_R = 8.5 \pm 2.1$ MPa (longitudinal) and $\sigma_R = 3.4 \pm 2.0$ MPa (transversal) for the hypogastric anterior rectus sheath. These values are close to the results shown in Table 2(a), $\sigma_R = 12.84 \pm 3.74$ MPa (longitudinal) and $\sigma_R = 3.38 \pm 1.67$ MPa (transversal). Although σ_R results for transverse direction coincide, they differ for the longitudinal one. Some differences in the experimental protocol may have contributed to this result. Pans et al. (1997) observed a viscoelastic behaviour in the rectus sheath, although via an indirect non-destructive measurement. Hollinsky and Sandberg (2007) used a test velocity of 10 mm/min, 2 times faster than the 5 mm/min used in this work, possibly inducing a response in the longitudinal direction. In accordance with the results reported in Table 2(b), Hollinsky and Sandberg (2007) did not encounter a significant influence of Age and BMI on the biomechanical properties.

Rath and Chevrel (1997) studied the ultimate stress, σ_R (7.57 MPa) and stretch, λ_R (1.37) of the infra-arcuate line portion of the anterior rectus sheath. Comparing these results with results for the longitudinal sample in Table 2(a), ($\sigma_R =$ 12.84; $\lambda_R = 1.71$) there is a considerable difference. This difference may be explained by the technique used to re-establish the hydration of the dead tissue, which consisted in specimen collection within 72 h of death and a saline bath preservation for 24 h. According with Tanaka and Yamada (1990) the state of mechanical stability achieved with this procedure induces an underestimation of σ_R and a stabilization of λ_R. This is consistent with the differences verified for σ_R, 12.84 vs 7.57 MPa and may explain the differences in λ_R since it shows a significant dispersion, Fig. 4.

Nevertheless, the present study has some important limitations. One is the number of subjects available, since only 12 subjects were used and only 6 provided experimental data in both directions. A study with a larger n would be able to produce more statistically robust results. Another limitation lies in the fitting procedure. The damage model proposed depends on many parameters, therefore to obtain them with greater accuracy would require a suitable experimental plan. The uniaxial tension test (simple tension), was the experimental technique used in this work. However, it is theoretically impossible to determine the mechanical properties of a three-dimensional material with such techniques. Further information from other kinds of tests (e.g., biaxial tests) could provide useful additional information for the rectus abdominis aponeurosis (rectus sheath) tissue characterization. Moreover, biaxial tests may reproduce the physiological deformation and loading conditions of the tissue more accurately.

In terms of the damage model, it is now well known that there are problems of uniqueness, well-posedness and numerical convergence associated with the apparent strain-softening due to the loss of the strong ellipticity of the material. As a result, finite element computations exhibit localization and spurious mesh sensitivity when the mesh size goes to infinitesimal. These numerical difficulties may be overcome by means of the nonlocal damage theory (Bazant and Jirasek, 2002), by cohesive models (Oliver, 1996) or viscous damage models (Ju, 1989; Peña, 2011a).

It should be mentioned that the inelastic model presented is purely phenomenological and, therefore, it is not possible to associate the internal variables to any micro-structural change or process. A structural model that integrates structural information of the damage and the softening processes is thought to enrich the predictability of purely phenomenological models (Gasser, 2011). For example using the Small-angle light scattering technique to detect strain-directed collagen degradation and rupture during the test (Robitaille et al., 2011). However for us this technique was not available. Finally, two important phenomena associated with damage and rupture of biological tissues should be pointed out. First, exposing biological soft tissue to supraphysiological mechanical loads produces irreversible deformations (Oktay et al., 1991; Ridge and Wright, 1967). Second, there is a failure behaviour resulting from fibre rupture and matrix disruption associated with material damage and fracture due to the micro-cracks that are produced in the tissue during the load process. As mentioned before, we did not have experimental data to know how this would effect our experiments. We think that both phenomena occur at the same time. In our model, we decided to use the classical continuum damage mechanics to model the softening of the rectus sheath neglecting the plastic deformations. Obviously, this hypothesis is related to the phenomenological character of the mathematical model. In addition, the softening behaviour of biological tissues is also related to viscoelastic effects (Peña et al., 2010a). Neglecting viscosity when modelling damage in soft tissue might be a non-admissible overestimation (Forsell and Gasser, 2011). Given the restricted sample size due to the logistic constraints inherent to cadaver research, the work reported in this paper did not include an investigation on the strain rate effects. Following our previous experience on characterization of pelvic tissue (another similar tissue), the viscosity effect at 5 mm/min would be irrelevant (Peña et al., 2010b).

This work constitutes the first attempt (as far as the authors know) to present a damage model for the human rectus sheath. This model may be used with finite element simulations to study complex biological processes under various loading conditions. It can have a potential impact in fields such as plastic surgery (Silveira et al., 2010) or herniation process (Ozdogan et al., 2006; Kureshi et al., 2008; Hernández et al., 2011) where damage is expected to play a significant role. The model was able to capture the mechanical behaviour in both longitudinal and transverse directions (more accurately in the longitudinal direction). This opens up its use in hernia repair (Rath and Chevrel, 1997; Pans et al., 1997; Hollinsky and Sandberg, 2007), especially for the improvement of prostheses (meshes) and to study the mechanics of tissue-mesh interaction. The realistic simulation of complex and extreme physiologic mechanisms such as a cough, a sneeze or jumping which involve a change in the intra-abdominal pressure, may also benefit from this contribution.

Acknowledgements

Financial support for this research was provided by the Spanish Ministry of Science and Technology through the research projects DPI2010-20746-C03-01 and DPI2011-27939-C02-01 and the Instituto de Salud Carlos III (ISCIII) through the CIBER initiative. CIBER-BBN is an initiative funded by the VI National R&D&i Plan 2008–2011, Iniciativa Ingenio 2010, Consolider Program, CIBER Actions and financed by the Instituto de Salud Carlos III with assistance from the European Regional Development Fund. Funding by Ministério da Ciência, Tecnologia e Ensino Superior, FCT, Portugal, under grant SFRH/BPD/71080/2010 and project PTDC/EEA-CRO/103320/2008 are also gratefully acknowledged.

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