Is there any objective and independent characterization and modeling of soft biological tissue?

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Abstract

The characterization of soft tissue raises several difficulties. Indeed, soft biological tissues usually shrink when dissected from their *in vivo* location. This shrinkage is characteristic of the release of residual stresses, since soft tissues are indeed often pre-stressed in their physiological configuration.

During experimental loading, large extension at very low level of force are expected and assumed to be related to the progressive recruitment and stretching of fibers. However, the first phase of the mechanical test is also aiming at recovering the pre-stressed *in vivo* behavior. As a consequence, the initial phase, corresponding to the recovering of prestress and/or recruitment of fibers, is questionable and frequently removed. One of the preferred methods to erase it consists in applying a preforce or prestress to the sample: this allows to easily get rid of the sample retensioning range. However this operation can impact the interpretation of the identified mechanical parameters.

This study presents an evaluation of the impact of the data processing on the mechanical properties of a numerically defined material. For this purpose, a finite element simulation was performed to replicate a uniaxial tensile test on a biological soft tissue sample. The influence of different pre-stretches on the mechanical parameters of a second order Yeoh model was investigated.

The Yeoh mechanical parameters, or any other strain energy density, depend strongly on any pre- and post-processing choices: they adapt to compensate the error made when choosing an arbitrary level of prestretch or prestress. This observation spreads to any modeling approach used in soft tissues. Mechanical parameters are indeed naturally bound to the choice of the pre-stretch (or pre-stress) through the elongation and the constitutive law. Regardless of the model, it would therefore be pointless to compare mechanical parameters if the conditions for the processing of experimental raw data are not fully documented.

Keywords: mechanical modeling, experimental protocol, soft tissue, initial strain, identification

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1. Introduction

The mechanical characterization and modeling of biological tissues is a key tool toward a better understanding of physiological and pathological behaviors. A greater knowledge of the mechanical properties of the tissue should also help to develop and improve medical and surgical treatments, through the design of adequate implants or patient-specific oriented models.

Biological animal or human tissues are mainly divided in two categories: hard and soft tissues. If mechanical characterization methods are globally similar, mechanical modeling differs substantially. Hard tissues, like bones are generally considered to have a linear anisotropic elastic response behavior [1] while soft biological tissues, such as skin, abdominal wall or pelvic tissues, present a non-linear mechanical behavior. In this case, hyperelastic constitutive models [2, 3] help to model and identify the mechanical properties of soft tissues from experimental observations.

To identify the mechanical behavior of soft tissues, the literature reveals numerous characterization protocols. Biological soft tissues are tested under uniaxial or biaxial traction [4–11], inflation test [12–14], ball-burst test [15] etc. Sample size is often a major limitation in the study of biological tissues. Whether during an animal study or the collection of cadavers tissues, samples are rare, quite difficult to obtain and often small in size. Even if it may not be the more physiological loading, the uniaxial tension test remains, above all, the less consuming in terms of material, since the minimal sample sizes are generally in the order of a few millimeters wide by about ten millimeters long [16–19]. Many studies [5, 17, 19–22] performed this type of test for the characterization of biological tissues, even if it lacks physiological representativity.

Most connective tissues are not uniaxially but rather multi-axially stressed: e.g. pelvic tissues, abdominal sheathing tissues and therefore shrink when they are dissected. Multi-axial tests would undoubtedly make it possible to provide richer information. However, uniaxial tensile tests show many advantages given the constraints imposed by the study of biological tissues. It is easy to set up and analyze from a mechanical perspective. Computations of stress and strain are obtained directly from the dimensions and force measures for an homogeneous material, without any numerical simulation or any other assumptions than homogeneous deformations. Considering an anisotropic material, this loading case gives the lowest error induced by the approximation of the stress and strain fields in comparison to other ones.

Even within the studies in uniaxial tension, differences in terms of measurement equipment, loading command and data treatment occur. There is actually no consensus in the community about a universal method either to test or to analyze and understand the mechanical behavior of soft tissues. Following Fung [23] pioneering research, numerous tests are performed with a preconditioning phase, where cycles are performed. This preconditioning phase include cycles at different level of deformation, force or stress [21, 24–27]. Numerous studies perform a pre-load [12, 15, 28–32] or pre-stress [22, 33, 34] threshold to compensate the retraction of the sample and remove any buckling of the sample prior to testing. The choice of a prestress strongly depends on the operator and mechanical testing means. Pre-force use is even more questionable as load is not an objective measure and fully relies on the sample size. Those thresholds can be used during the testing or in post-processing: the latter should be preferred to the former since it does not remove definitely a part of the measured information. It can also happen that the data processing for the identification of the mechanical properties of soft tissues is not always extensively described.

Whether it is in post-processing or during testing, the postprocessing methods leading to the results often lack details and could have an effect on the identified mechanical properties of the studied material. If these problems are widely known in the community of soft tissues biomechanics, no previous work, to our knowledge, directly addresses the question and no consensus is existing.

In a first part, the context and methods of this study are recalled. A simple isotropic hyperelastic law is chosen to represent the stress-strain response of soft tissues. A numerical simulation of a uniaxial traction test is performed to mimic a real test on a biological sample. Different prestress levels are applied to the numerical stress response to assess their impact on the mechanical properties studied. **Finally, some guidelines are proposed to biomechanical researchers to make and share the best of their experimental data.**

2. Context of the study: experimental observation and mechanical framework

2.1. Toward an *ex vivo* zero-strain and zero-stress state

When soft tissue specimens are collected, the dissection causes in general a more or less significant retraction of the sample [23, 35, 36]. This phenomenon is due to the release of residual stresses [37, 38]. According to their function in the body, soft tissues are often stressed in their physiological state [23]. Resection thus leads to the relaxation of the sample. The *ex vivo* zero-stress state does not match with the *in vivo* unloaded state. There is no simple way to evaluate the *in vivo* stress configuration, or to estimate the residual stress: to our knowledge, the simplest approach consists in measuring the shrinkage of the sample [36]. More advanced techniques involving digital image correlation [39] or sonomicrometry [40] have also been developed: they are more difficult and longer to set-up in controlled environment like biomechanical labs.

Easier techniques to compensate the relaxation of the sample are often applied *a posteriori*. Usually, the first phase of a uniaxial tension test shows a plateau where stresses are very low and assumed as negligible, as shown in Figure 1. The two main reasons for that plateau might be the retensioning of the sample to compensate its retraction and any buckling of the sample due to the placement of the specimen between the clamps (in order to avoid applying any uncontrolled pre-stress to the sample). An additional interpretation of the low-force plateau comes from the very nature of the tissues. From an histological point of view, soft tissues results from the interweaving of elastin and collagen fibers [41–45]. Due to the release of the physiological stresses during the resection the fibers within the tissues are entangled and crimped at the local scale [46]. During the first phase, no fiber is fully stretched and recruited within the sample, they remain loose within the extra-cellular matrix. Gradually the fibers realign with the direction of traction and contribute to the overall stiffness of the material [27, 32].

In an *ex vivo* uniaxial tension, because of the free edges of the mechanical sample, the shrinkage in the transverse direction cannot be compensated during the uniaxial loading as it is the case in an *in vivo* configuration. That could even lengthen the flat initial plateau, corresponding to the realignment of the fibers. The *in vivo* loading is rarely purely uniaxial since the edges are not free. It actually consists, even in the case of tension, in a combination of tension and shear loadings.

Figure 1: Typical phases of the mechanical response during a uniaxial tensile test on a rat abdominal wall sample and their mechanical interpretations. a) Re-tensioning phase b) Low rigidity c) Large deformations and rigidity d) First signs of damage and rupture.

The first phase of the test corresponds to the re-tensioning of the sample at the global and local scales. According to this last assumption, during the first phase, the measured deformations are therefore not characteristic of the material's mechanical response but of its re-tensioning and un-buckling, and should consequently be excluded from the characterization and modeling of the material. Once this phase is over, the material's non linear mechanical response is observable, in Figure 1, with a first low rigidity region while

fibers are partially realigned and stretched in the traction direction; followed by a large deformations and higher rigidity phase once all the fibers are recruited.

The raw experimental data therefore require a processing phase before the identification of a mechanical model: filtering the experimental noises, cutting the area after the first signs of damage and above all removing the beginning of the test corresponding to the re-tensioning phase. The range of the re-tensioning phase is not trivial to discriminate from the region of low rigidity in small deformations. In the close-up view in Figure 1, one can notice that the transition between the retensioning and low stiffness regions remains smooth because of the monotony and constant convexity of the experimental response. A drastic change of convexity (inflection point), for example, would be easier to detect. The end of the deformation range, i.e. where the data are not anymore considered for the identification of mechanical parameters, is also rarely described in the literature. Because of clear sign of damages (change in the slope, drops in the force signal etc.), it seems easier to define, even visually, and will not be investigated here.

In the literature, when described, the most common method is a thresholding method using a pre-stress or a pre-force [11, 21, 22, 36, 47, 48]. This thresholding can be performed during the test or in post-processing. This method is very effective but its use remains questionable: the level of force or stress is very dependent on the material or the sample dimension, and setting such a threshold can be tricky. The value of the pre-stress, if too high, will remove a large number of experimental measurements representative of the effective mechanical behavior of the material; if it is too low, may erroneously consider the tensioning phase of the uniaxial test as a part of the response of the biological material. In both cases, this will impact the identification of mechanical parameters, regardless of the modeling choices.

2.2. Mechanical framework

This work aims at studying the impact of the retensioning zone and its identification on the mechanical properties of connective tissues. The first step consists in fixing a modeling approach. In the literature, a lot of nonlinear models coexist for the modeling of soft tissues [2, 3], that could be equally valid.

Over the years, our research team characterized various soft tissues, from animal or human origin: rat abdominal wall [49], abdominal aponeuroses [50] and sheathing tissues [11], ewe [17] and human pelvic tissues [6, 51–53]. Mechanical characterization occurred under uniaxial tension to optimize the use of collected sample and build statistically significant databases. This isotropic law was applied to isotropic and anisotropic tissues as well [11, 50]. For an anisotropic material, the identification of all the mechanical properties would require, in uniaxial tension, a large number of tests in different loading directions to reach a complete characterization of the material mechanical parameters. When only one direction of the material is uniaxially stretched, it is not possible to properly fit an anisotropic model. Therefore, the isotropy assumption prevails for the sake of simplicity under similar conditions.

To model their experimental response and quantify their mechanical properties, a simple energy density was first chosen, with a second order polynomial Yeoh law [54]. Since Fukahori and Seki [55] showed that the contribution in I_2 is negligible in uniaxial tension, the selected energy density depends only on I_1 :

$$W = C_0(I_1 - 3) + C_1(I_1 - 3)^2 \quad (1)$$

where $I_1 = Tr(\mathbf{F})$, with Tr the trace operator and $\mathbf{F}$ the deformation tensor, and C_0 and C_1 two mechanical constants expressed in MPa. The Yeoh model, in its reduced or complete form was also frequently used, thanks to its simple polynomial form for the study of aorta tissue [56, 57], skin [58], breast tissue [59] etc.

Usually soft connective tissues are considered incompressible due to their high water content. Under these assumptions, and in uniaxial tension, the deformation tensor $\underline{\mathbf{F}}$ takes the form $\underline{\mathbf{F}} = \text{diag}(\lambda, \frac{1}{\sqrt{\lambda}}, \frac{1}{\sqrt{\lambda}})$,

with the stretch $\lambda = \frac{l_0+x}{l_0}$, where l_0 is the initial length and x the displacement.

Considering an isotropic material, the principal deformations and stresses directions coincide. The first Piola-Kirchhoff tensor (PK1 tensor) takes the diagonal form $\underline{\boldsymbol{\tau}} = \text{diag}(\tau, 0, 0)$, where $\tau = \frac{F}{S_0}$ with F and S_0 the force and initial cross-section respectively. Under the incompressibility assumption and with the I_1 -dependent energy density, the longitudinal PK1 stress τ is modeled with the following expression by virtue

of the chain rule:

$$\tau = 2 \left(\lambda - \frac{1}{\lambda^2} \right) (C_0 + 2C_1(I_1 - 3)) \quad (2)$$

The asymptotic behavior provides a straightforward interpretation of the mechanical parameters. C_0 and C_1 in the Yeoh model can be interpreted as rigidities under small and large deformations respectively. The longitudinal stress is indeed equivalent to $\tau \sim 6C_0(\lambda - 1)$ under small deformations and $\tau \sim 4C_1\lambda^3$ under large ones.

The Yeoh model provides an information about the state of the retensioning: Figure 2 shows that the best fit comes with a thermodynamically non admissible value of C_0 ($C_0 < 0$). The model gives a strong hint that the sample is actually not in tension at the beginning of the test. Models involving the exponential function [35, 60, 61] or polynomials of the second invariant [62] present a null tangent for $\lambda \approx 1$: the retensioning range is more likely to be interpreted as a material response with such laws.

Figure 2: Optimization of two 2nd-order Yeoh model on an unprocessed dataset

The Yeoh model offers a simple modeling approach, with a clear mechanical meaning of the parameters. It is also "foolproof": negative C_0 highlights even better than the correlation coefficient the presence of the retensioning zone and the necessity of data-processing before model optimization. In what follows, we will investigate the influence of processing methods on the mechanical parameters in a simplified framework.

3. Material and method

3.1. Numerical simulation

To ensure objective and controlled investigations, a numerical test has been simulated to explore the effect of the identification of the re-tensioning. Unlike an experimental study, a numerical one makes it possible to overcome the inherent biases and deviations related to the characterization of soft biological tissues. It is moreover able to fully control the mechanical parameters and to know exactly where the retensioning zone ends. The prior knowledge of the material parameters allows to quantify the efficiency and uncertainties caused by the choice of experimental data analysis.

For the purposes of this study, a uniaxial tensile test is simulated using Abaqus/CAE 6.12-2 software (Dassault Systèmes Simulia Corp.). The dimensions and mechanical properties of the sample are consistent with the experimental protocol set-up for the observation of implant-tissue composites and results obtained during rat abdominal wall characterization [49].

The material specimen is represented by a shell, 20 mm wide and 0.05 mm thick. This thickness is not chosen to correspond to the reality of connective tissue but to impose a very low flexion stiffness to the material and to complete simulation under the plane stress assumption. At one end, the specimen is fixed

on 10 mm to simulate the clamps of the specimen. Symmetrically, a displacement is applied at 10 mm from the other end. The sample is meshed with quadrilateral shell elements (1480 S4R elements).

To reproduce the tensioning phase, the specimen is deliberately placed under "extreme" initial conditions: the plate is corrugated, the actual length of the specimen being much greater, 65 mm, than the length between the two clamps, fixed at 50 mm. The initial configuration is illustrated in Figure 3.

Figure 3: Specimen configuration at the beginning of the numerical test: a) transverse view, b) meshed sample in the initial configuration.

The following modeling choices are applied to the specimen:

- an isotropic and quasi-incompressible constitutive law derived from a second order Yeoh law, in the following form:

$$W = C_0(I_1 - 3) + C_1(I_1 - 3)^2 + \frac{1}{D_1}(J - 1)^2 + \frac{1}{D_2}(J - 1)^4 + \frac{1}{D_3}(J - 1)^6$$

with $I_1 = \text{Tr}({}^t\mathbf{F} \cdot \mathbf{F})$,

$$C_0 = 0.003 \text{ MPa}, C_1 = 0.03 \text{ MPa}, D_1 = 10^{-4}, D_2 = 10^{-3}, \text{ and } D_3 = 10^{-5},$$

the parameters are chosen to correspond to values identified on the rat abdominal wall without a prosthesis [49] and to numerically manage the incompressibility;

- a specimen configuration with an unfolded length of 65.11 mm; this value imposes a significant buckling on the plate, which is not at all representative of the experimental situation but will allow to simulate the tensioning phase.

Once the simulation is completed, the following data are calculated to obtain the stress-stretch response:

- the value of the displacement and the associated longitudinal deformation $\epsilon = \frac{\Delta l}{l_0}$,
- the sum of the reaction forces at the application nodes of the displacement,
- the longitudinal stress over the entire specimen $\tau = \frac{F}{S_0}$ with F the force and S_0 the initial cross-section, $S_0 = 1 \text{ mm}^2$.

This numerical simulation of the traction test gives a non linear stress-stretch curve, shown in Figure 4. One can observe three main phases:

- a first flat plateau; the test is ideal, efforts are exactly equal to zero as long as the distance between the clamps has not reached the real length of the sample;
- a second phase corresponding to the low stiffness; in small deformations the Taylor expansion of Eq. (2) gives an approximation of the stress: $\tau \sim 6C_0(\lambda - 1)$;

Figure 4: Displacement force curve obtained after simulation of a uniaxial tensile test on a specimen of length 65.11 mm. When the value of the displacement exceeds 15.11 mm, the specimen is completely unfolded and the loading of the material begins.

– finally a zone of high rigidity associated with large deformations.

The stress and strain values from the numerical evaluation are flawless: unlike an experimental measurement, there is no noise or bias. To provide a better representation of the experimental reality, the numerical stress data are first "noised" by adding to the stress signal a random noise of variable amplitude generated by a Pareto distribution and a power distribution. Figure 5 presents the mechanical response of the numerical test alone and with the addition of a random noise. This was performed in Python with the `random.pareto` and `random.power` functions of the Numpy library [63].

Figure 5: Stress and stretch curve rate of a numerical test with and without the addition of a random noise.

3.2. Method

To our knowledge, there is no preconceived method to set the end of the tensioning zone. Given the material simulation, the present section describes the impact of the threshold on the identified mechanical

properties.

Once the experimental test is performed, three parameters remain unknown: the range of the retensioning phase, marked with λ_{app} and the hyperelastic parameters C_0 and C_1 . Thanks to the numerical mechanical test, we first studied how each of these parameters impacts the accuracy of the modeled mechanical response.

For this purpose, a design of experiments is prepared to assess the evolution of the sum of squared residuals between the experimental (simulated) and modeled behavior for various values of λ_{app} , C_0 and C_1 . This quantity compares the values of the numerical test after they have been shifted according to λ_{app} and the modeled behavior. It is evaluated for various values λ_{app} , C_0 and C_1 , chosen around the exact quantities, respectively 1.3, 0.003 MPa and 0.03 MPa.

The experimental measures are first shifted according to the elongation λ_{app} at which the sample is considered to be stretched, and τ_{app} its corresponding stress level, in order to ensure a zero-stretch and zero-stress initial configuration in an undeformed state. Measures before the threshold λ_{app} are thus removed and the following ones updated (or shifted) according to:

$$\begin{cases} \tau_{update}^i = \tau^i - \tau_{app} \\ \lambda_{update}^i = \frac{\lambda^i}{\lambda_{app}} \end{cases} \quad (3)$$

where τ^i and λ^i are the i^{th} value of the experimental stress and stretch respectively.

Figure 6 illustrates the shifting method for the numerical data obtained in Section 3.1 and three values of λ_{app} .

Figure 6: Stress-stretch response of the numerical tension test before and after shifting with different values of λ_{app} , according to Eq. (3).

The sum of squared residuals (SSR) is then estimated with:

$$SSR = \sum_i \left(\tau_{upd}^i - 2\left(\lambda_i - \frac{1}{\lambda_{upd}^i}\right) \left(C_0 + 2C_1(\lambda_{upd}^i)^2 + \frac{2}{\lambda_{upd}^i} - 3 \right) \right)^2, \quad (4)$$

with $\begin{cases} 0.001 \leq C_0 \leq 0.008 \text{ MPa} \\ 0.015 \leq C_1 \leq 0.045 \text{ MPa} \\ 1.2 \leq \lambda_{app} \leq 1.4 \end{cases}$

4. Results and discussion

Figure 7 shows some iso-contours of the error values in the space $(\lambda_{app}, C_0, C_1)$. The SSR strongly depends on the values of the different parameters of the law. The values of C_0 and λ_{app} are naturally interdependent, as shown by the very large minimization surfaces in this plane (see Figure 7a). The problem looks moreover ill-conditioned: the minimization planes, highlighted in Figure 7b, do not follow closed contours.

Figure 7: Cartography of the error SSR as a function of $0.001 < C_0 < 0.008$, $0.015 < C_1 < 0.045$ and $1.2 < \lambda_{app} < 1.4$. (a) 3D view, (b) In the (C_0, C_1) plane.

The SSR is now computed at fixed $C_1 = 0.03$ MPa. Only the values of λ_{app} and C_0 vary. Figure 8 describes the evolution of the residual according to the choice of λ_{app} with two fixed values of C_0 (in black and red) and an optimized C_0 for each λ_{app} (in blue).

Since the deformation measure is not related to the material before λ_0 , the choice of the C_0 parameter in small deformations has barely no influence on the SSR, thus on the location of the re-tensioning zone. The minimum error is reached at close values of λ_{app} for the two values of C_0 tested. The exact λ_{app} , noted λ_0 in Figure 8 is computed from the numerical simulation conditions, $\lambda_0 = 1.30$, since the exact length of the sample and initial length between grips are set in the simulation. For both values of C_0 , the apparent elongation is taken where the SSR is minimal, ie $\lambda_{app} = 1.28$ and 1.30 for C_0 equal 0 and 0.003 MPa respectively.

In a second step, with $C_1 = 0.03$ MPa fixed, the coefficient C_0 is optimized for each re-tensioning range λ_{app} . In Figure 8, this method drastically reduces the SSR beyond the tensioning area. Outside the non-thermodynamically permissible zone, i.e. leading to negative C_0 coefficient, the optimization of the rigidity factor in small deformations corrects the error made when selecting λ_{app} . Compared to fixed C_0 configurations, the maximum SSR, with admissible C_0 is divided by 17. C_0 coefficient neutralizes the error made when arbitrarily choosing the stretching range of the sample.

In a similar way, the SSR is calculated at fixed values of $C_0 = 0.003$ MPa and varying values of λ_0 and C_1 . Its evolution is shown in Figure 9. With a wrong value of C_1 moves the apparent elongation value λ_0 . When the C_1 coefficient is optimized for any values of λ_0 , the SSR drops significantly even in the retensioning range with admissible C_1 values.

Since C_0 characterizes the rigidity in small strain, and is therefore supposed to be strictly positive, its evolution with the choice of the λ_{app} provides a useful insight about the material state of stretch. However it remains difficult to identify the end of the retensioning area with C_0 : the apparent elongation does not come with a clear change of convexity of the mechanical response. The boundary between the "structure" behavior of the retensioning zone and the true material behavior remains thus blurry. The coefficient in large deformation cannot provide any information to decide: it will always adapt to minimize the SSR. It appears

Figure 8: SSR evolution as a function of λ_{app} with $C_0 = 0.003$ MPa, $C_1 = 0.03$ MPa (in black), $C_0 = 0.0$ MPa, $C_1 = 0.03$ MPa (in blue) fixed and with $C_1 = 0.03$ MPa fixed and optimized C_0 (in red). The true realignment range is delimited on the graph with λ_0 .

Figure 9: SSR evolution as a function of λ_{app} with $C_0 = 0.003$ MPa, $C_1 = 0.03$ MPa (in black), $C_0 = 0.003$ MPa, $C_1 = 0.1$ MPa (in blue) fixed and with $C_0 = 0.003$ MPa fixed and optimized C_1 (in red). The true realignment range is delimited on the graph with λ_0 .

that C_0 can balance the choice of the apparent elongation, the C_1 coefficient can then be optimized to minimize the error along the whole deformation range and fit the stress-stretch response. These mechanical parameters are therefore highly dependent on the experimental and processing conditions. These conclusions would be similar for any other models. The apparent elongation λ_0 impacts indeed the elongation value through Eq. (3) and thus the constitutive law.

There are two possibilities to process uniaxial tensile tests on biological tissues. In this study, the apparent elongation λ_{app} was considered. This method is simple to perform on numerical data. In practice, however, experimenters would rather use a "well chosen" apparent stress, or pre-stress, τ_{app} that determines the value of λ_{app} . Choosing a pre-stress threshold looks more relevant and easier to implement since a stretch threshold would be too dependent on the nature and initial configuration of the sample, the accuracy of the test operator and the testing conditions. For example, the impact of the strain rate was not investigated in this work, as the experimental testing was supposed quasi-static in order to neglect any viscous phenomenon in the modeling framework. Several works in the literature [26, 64, 65] tend however to suggest that the loading rate could impact the toe region, and therefore the retensioning zone. The pre-stress τ_{app} would not be always associated with the same value of the apparent elongation λ_{app} , which could also in a lesser extent cause a certain dispersion of the results. Modeling parameters are therefore always determined with

respect to the processing method.

When it comes to the mechanical characterization of biological tissues, specially in non-physiological conditions, good practice should thus rule to provide not only the identified mechanical parameters of a constitutive law but also the raw experimental data and to detail as much as possible the experimental conditions (thickness measurement for the cross-section calculation, measure of deformation etc.) and post-processing method that both cause bias. This way, it would be possible to perform comparisons from one study to the other and reliably build functional database of mechanical parameters for any considered tissues.

Conclusion

In the present study, we investigated the impact of the data processing on the identification of mechanical parameters of soft biological tissues. With a simplified and ideal material model, we showed that the identified mechanical parameters might not represent reliably the mechanical behavior of the tested sample. Biological soft tissue characterization is a complex task in light of the constraints it imposes: rarity and small size of the samples. When the testing configuration cannot be representative of the physiological loading conditions, special attention should be given to the considered reference configuration and zero-stress/zero-strain state. In any case, one should be careful about the identification and interpretation of the mechanical parameters. Future studies should provide all of the raw experimental data that were collected during the experiments in addition to the traditional method section, describing thoroughly the test conditions and data-processing choices. That would allow to apply similar post-processing methods and thus compare mechanical parameters in a consistent manner.

Acknowledgment

The authors thanks the French National Research Agency (ANR-13-TECS-0003-01) for the financial funding of this study. This work is also part of PROBIOMESH, an Interreg France-Wallonie-Vlaanderen project supported by the European Regional Development Fund.

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Declaration of interests

X The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

☐The authors declare the following financial interests/personal relationships which may be considered as potential competing interests:

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A.Morch : Conceptualization, Methodology, Investigation, Writing

L.Astruc : Investigation, Writing – Review and editing

O.Mayeur : Software, Investigation, Writing Review and editing

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