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Determination of the WYPiWYG strain energy density of skin through finite element analysis of the experiments on circular specimens

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ABSTRACT

Skin is a biological material which mechanical behavior has large variations depending on the individual and the location of the specimen in that individual, among other factors. Large differences are also encountered in measurements between *in vivo* and *in vitro* specimens. Then, optimal characterization of the skin for simulation (for example) of surgical procedures requires that all experiments to characterize the material behavior be performed on the same specimen and *in vivo* if possible. Recent experiments on circular discs (Groves et al., 2013 [16]) permit this characterization using a single specimen as we show in this paper, and may constitute a good starting point for ulterior characterization *in vivo*. However, because in these tests deformations are not homogeneous, the determination of the material behavior is not as direct as with tensile or biaxial tests, so finite element analysis is needed to propose a procedure to determine the material behavior. In this work we perform an analysis of the experiments using finite elements obtaining an insight which permits a very simple iterative procedure to determine the stress-strain behavior of the material and, thereafter, the corresponding What-You-Prescribe-Is-What-You-Get (WYPiWYG) stored energy densities.

1. Introduction

Skin is the largest organ of the human body, the one with most contact with the environment and which accounts to about the 15% of the body weight [15]. Skin is a very complex organ that has three well-defined and interconnected structural and functional units, namely: epidermis, dermis and hypodermis (from the external surface to the inner surface). In all the units of skin tissues there are cells and extracellular matrix (ECM), mainly collagen, elastin and proteoglycans. Additionally, other specific structures such as blood vessels, nerves, and glands are may be present in certain units. From a mechanical point of view the dermis, and more precisely, the ECM is the component mostly responsible for the observed mechanical behavior.

Mechanical properties of skin have been measured *in vitro* following diverse techniques, see for example [8,3,37,45,17,34,44,50,51,23]. However, even though freezing specimens for conservation does not affect mechanical properties [13], the mechanical behavior of skin has a large variability which strongly depends on the individual and on the location in the body [15,3,17], as well as in the presence of skin pathologies [14]. Mechanical properties also change with the length scale [18], with sex [2], humidity [54,41,18], temperature [41,25,55], overall health condition [15], environmental damage [35], etc. Therefore, it is apparent that for obtaining reliable sets of experimental curves that can be used in the characterization of constitutive models, experiments should be performed in the same specimen or otherwise in the same individual in very nearby locations. Averaging or mixing experimental data from different specimens may result in an unphysical behavior [43] because in fact, the result does not correspond to that of any real, existing material. To this end, *in vivo* testing in preferred. Several techniques have been applied as indentation tests in [40,39], surface waves in [33], air pressure in [7], suction in [19], mechanical movement in [12] also added to digital image correlation in [11], and even in *in vivo* but boundary-free configurations obtained by surgery in [6]. Obviously, although more difficult and more expensive, true biaxial tests are to be preferred in characterizing biological tissues because it allows for the determination of coupling terms in the stored energy function. However, because of the complex anisotropic structure of the skin, all these experiments are difficult to use in developing finite element simulations to model completely the mechanical response of the skin in a general configuration and in a wide range of possible deformations, which is the purpose if finite element simulations of surgery in organs are to be performed.

The experiments of [16] are of special relevance because, even being *ex vivo*, several relatively simple tests are performed on the same specimen. This is thanks to the circular shape of the specimen which allows to test the material in a similar way as it is done for tensile tests but

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in different directions. We will see herein that despite being nonhomogeneous tests, they can be assimilated to equivalent homogeneous tests, and furthermore, the tissue outside the gauge section has little influence. Thus, an extrapolation of the methodology for *in vivo* testing and then address patient-related simulations seems to be natural and promising. However, the hypothesis of homogeneity of the stress field typically assumed in tensile tests is, in principle, questionable for the tests at hand. In fact, Groves et al. [16] did not consider a homogeneous stress field and performed finite element analysis to develop a complex inverse analysis to determine the parameters of their constitutive model. The use of optimization algorithms and of inverse analyses using finite element meshes are frequent in the biomechanics literature, see for example [5,11,24,52,22] and [1]. In this work we analyze some experiments in Ref. [16] in order to establish a simple methodology which allows us to perform accurate predictions on the behavior observed in the experiments without employing optimization algorithms or material parameters. In contrast to the work of [16] and accepting the current difficulty in developing a structure-based model which accurately accounts for the explained complex multilayer structure of the skin and is still efficient for finite element simulations, we employ herein the new What-You-Prescribe-is-What-You-Get (WYPiWYG) methodology [47,27,28], which has been employed successfully in capturing the behavior of other soft biological materials, see [27,32] and [31]. The procedure herein introduced is intuitive and uses a nonlinear iterative procedure to determine the effective length and effective area from which stress-strain curves may be obtained. These stress-strain curves are captured by the WYPiWYG procedure. Then, the finite element analysis of the experiments predict the load-displacement curves with excellent accuracy. To the best of our knowledge, this type of simple deterministic analysis has not been done in the literature for obtaining the material behavior in soft tissues from nonhomogeneous tests, where the use of costly optimizations, giving non-unique material parameters, is usual.

The rest of the paper is organized as follows. We first briefly review and comment the experiments of Groves et al. [16] to be analyzed herein. Then, in order to explain the ideas in a simple context, we analyze the experimental setting under an isotropic small strains model and an isotropic Ogden model with parameters typical for skin. Thereafter we review the WYPiWYG model and introduce an improvement in the computational algorithm. Afterwards, we explain the iterative process for inverse analysis, perform stress-strain predictions and obtain the resulting load-displacement curves which are compared to those obtained experimentally by Groves et al. [16]. Finally we discuss the approach and make some conclusions.

2. Experiments and parameter-fitting procedure from Groves et al. [16]

As mentioned, in [16] a new experimental procedure to characterize soft biological tissues is introduced, and used therein to characterize human and murine skin. The procedure was based on three different tensile tests on circular skin specimens. The use of circular specimens allowed them to conduct the test in the same material, provided that no damage nor permanent deformation is introduced to the specimen.

Human skin samples were obtained from two different donors. Murine samples were obtained from eight donors. For the murine samples equivalent orientations were recorded whilst for human samples these orientations were not recorded. For each specimen, three tensile tests were conducted in directions corresponding to 0°, 45°, and 90°with a common reference, which is the centerline of the back, see Fig. 1. Hence 0^o naturally corresponds to a symmetry plane. The corresponding load-displacement curves for each test were obtained. An example of this set of curves is shown in Fig. 2. Special grips were designed in order to hold the skin correctly in the tensile tests and applying a constant pressure in the jaw faces during the test. The maximum load in the tests was limited to ensure that no permanent damage occurred to the samples.

A further study on the mechanical properties of the skin was conducted by the authors based on the data retrieved from the tests. An anisotropic structure-based hyperelastic model was chosen to characterize the hyperelastic response of the skin. The material behavior was modeled using three layers of transversely isotropic hyperelastic material with a different family of fibers for each layer. For each layer the strain energy function was written as a function of the classical invariants, the stretch in the fibers and the jacobian determinant, as

$$\Psi = F_1(I_1, I_2) + F_2(\lambda) + \frac{K}{2}(\ln J)^2 \tag{1}$$

where $F_1(I_1, I_2)$ corresponds to the isotropic matrix modeled by Veronda and Westmann [49]. The families of fibers were modeled based on the stretch of the fibers λ as indicated by the contribution $F_2(\lambda)$ proposed by Weiss et al [53].

The parameters were fitted for each specimen using optimization to reproduce as close as possible the experimental results. A finite element model with the material was implemented in order to reproduce the tests. Few computational details are given about the actual minimization procedure followed during the simulations. They report that the optimization procedure followed the Simplex algorithm connected to finite element analysis and that the maximum number of iterations allowed were 1500. The authors report that the optimization procedure presented great sensibility to the changes in the parameters and that only a local minimum is obtained, a frequent observation found in other works employing optimization procedures for determination of material parameters in soft materials, and that will not be present in our analysis. The authors include a volumetric term, but the associated bulk modulus is not given nor included in the optimization procedure, probably because the material is assumed quasi-incompressible and, hence the specific value is usually irrelevant for the purpose. The finite element mesh was made of standard solid elements.

The disc dimensions were assumed constant. The diameter of the disc was 31 mm and the thicknesses were assumed 1.86mm and 0.265mm for the human and murine skin, respectively. Clamps were modeled using contact in an area of $15 \times 2\text{mm}^2$. Even though the constitutive model may result in a strongly anisotropic behavior, the authors used only one quarter of the disc in order to save computational time in the probably very time consuming optimization procedure. Because in our work the computational times are small, we do not need to take this simplification and, hence, we have considered the full disc. However, despite that simplification (justified below by computational results) Groves et al [16] report excellent fittings for all three tests in skin from all donors.

3. Comparison of rectangular specimens and circular specimens

The circular specimens introduced in the tests [16] differ from the rectangular specimens used to characterize the constitutive behavior in uniaxial tension of tissues. In uniaxial tests of metals it is typical the use of specimens with gauge length-to-width ratios of 4, see for example 4:1 for rectangular specimens in the US ASTM:E8 standard [4] and in the ISO [20] one. Grip distance-to-width ratios are even larger, orders of 7:1 are usual [20]. These aspect ratios are required in order to be able to consider uniaxial test boundary conditions and uniform deformations in the cross section. An experimental study on the strain distributions in metals along the gauge length [42] showed that for a 4:1 ratio this distribution was uniform over an 80% of the width in the central section of the gauge length. However, for the circular specimens used [16] the gauge section presents a ratio of 1:1, i.e. the ratio of the distance between clamps (15 mm) to the width of the clamps (15 mm). Within the small strain regime, one should not expect a uniaxial stress state in the central cross-section of circular specimens, but in the nonlinear regime the situation may change. Indeed, the particular nonlinear response that skin exhibits gives as a result a transverse behavior