Inverse determination of viscoelastic properties of human fingertip skin

Inverzno določanje viskoelastičnih lastnosti človeške kože na prstu

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Abstract: This paper presents a combined experimental-numerical procedure to determine viscoelastic properties of human skin at the tip of an index finger. The in-vivo biomechanical test was performed by a non-intrusive suction instrument Cutometer® MPA 580 (Courage-Khazaka). The measurements of the fingertip skin deflections performed at various levels of negative pressures were analysed by an inverse finite element based procedure in order to determine parameters of the Fung material model, including a non-linear elastic part and a linear viscous part represented by a five-term Prony series. The constitutive parameters of the fingertip skin are applicable for computer modeling of biophysical phenomena that govern tactile sensations as well as for setting the target viscoelastic properties for developing biomimetic materials for hand prostheses and humanoid robotics.

Izvleček: Namen članka je predstavitev kombiniranega eksperimentalno-numeričnega postopka za določanje viskoelastičnih lastnosti človeške kože na konicah prstov kazalcev. Biomehanski preizkus

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je bil narejen v živo s podtlačno napravo Cutometer® MPA 580 (Courage-Khazaka). Merili smo deformacije kože pri različnih podtlakih ter jih nato analizirali z inverzno analizo po metodi končnih elementov, kjer je bil za kožo uporabljen Fungov snovni model, ki vključuje nelinearni elastični del in linearni viskozni del, predstavljen s petimi parametri vrste Prony. Konstitutivni parametri kožnega tkiva na prstnih blazinicah so uporabni za računalniške analize biofizikalnih pojavov med dotikom, kakor tudi kot ciljne vrednosti viskoelastičnih lastnosti za biomimetične materiale, ki se uporabljajo za proteze in humanoidno robotiko.

Key words: human skin, suction test, viscoelasticity, finite element method, inverse analysis

Ključne besede: človeška koža, podtlačni preizkus, viskoelastičnost, metoda končnih elementov, inverzna analiza

INTRODUCTION

The human skin is highly nonlinear, inhomogeneous and anisotropic material which is in vivo subjected to a pre-stress. Its biomechanical behaviour may be affected by several dermatological and systemic state variables that vary with the age, body site, race, sex, mood as well as environmental conditions including temperature, humidity, chemical environment etc (WILKES et al., 1973 [1]; FUNG, 1972 [2]).

To characterize biomechanical properties of skin and understand its non-linear behaviour specially designed biomechanical tests can be performed where skin is thermo-mechanically stimulated by suction, indentation, tangential tractions and other load combinations (DIRIDOLLOU et al., 2001 [3]; ISRAELOWITZ et al., 2005 [4]). The stress-strain response of skin surface can be measured under controlled loading programme and the experimental recordings can be processed by inverse numerical methods in order to determine fundamental viscoelastic parameters for constitutive models proposed by ARRUDA & BOYCE, 1993 [5]; YEOH, 1990 [6]; FUNG, 1993 [7]; HOLZAPFEL, 1996 [8]; REES & GOVINDJEE, 1998 [9]; PERIĆ & DETTMER, 1998 [10]; DE SOUZA et al., 2008 [11]; DUPAIX & BOYCE, 2007 [12] and LUBINER, 1990 [13].

The main motivation for the research considered in this paper is to determine viscoelastic properties of human skin at the fingertip in order to derive constitutive parameters for numerical modeling of tactile sensations and related mechanotransduction phenomena than govern neural responses when exploring textured surfaces by touch (Jiyong et al., 2007 [14]; Wang & Hayward, 2007 [15]).

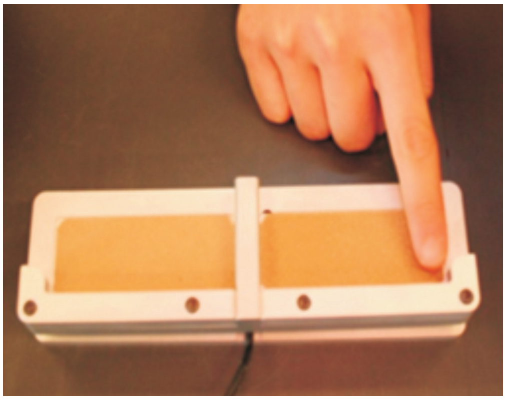

Figure 1. Tactile sensations and exploration of textured surfaces

Characterization of the Biomechanical Properties of Fingertip Skin

Characterization of the mechanical properties of skin was performed by sampling human skin at the fingertip of an index-finger. The experimental testing was performed in vivo by using a Cutometer® MPA 580 device (Courage + Khazaka Electronic GmbH, Koeln, Germany). This dermatological device is primarily used to evaluate the viscoelasticity of the skin by measuring its deflection at various levels of negative pressure. The testing principle is shown in Figure 2.

Suction test device

The device consists of a vacuum pump with a tank and a probe connected to it through a valve. Before the start of the test the pump decreases the pressure in the tank to the set value. When the test starts the skin is sucked into the probe's opening by the negative pressure created by the pump. During the test the pump generates a prescribed temporal evolution of pressure to evaluate transient deformation response of the skin.

A light sensor inside the probe measures the highest level reached by the skin sucked into the hole. The device is set to 0 mm which is the level observed when the negative pressure starts to pull the sample. This means that the possible increased level at the beginning, when the probe was pressed on the material due to the spring inside the probe, was not considered in the plotted result curve.

The probe's geometry shown in Figure 3 is composed of inner and outer concentric cylinders that are axially connected through a spring. The sensor is in the internal cylinder in which the skin is sucked.

Suction test method

In preliminary testing the identification of factors influencing the skin's mechanical response was made. The following factors were analyzed: load (pressure level imposed by the device), person (two different subjects were tested), hand (left or right) and various

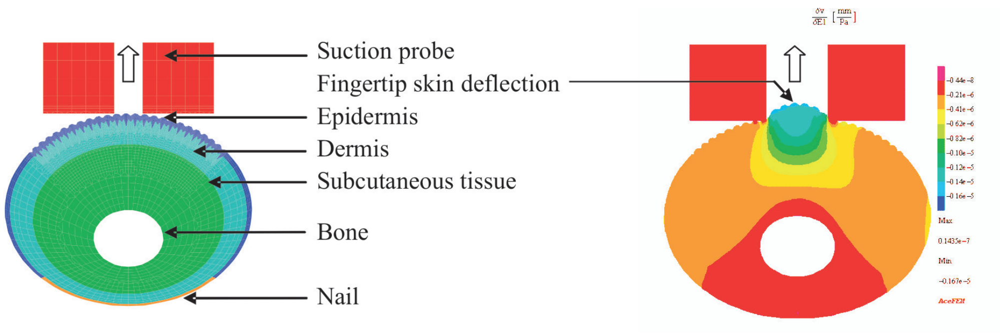

Figure 2. A crossection of a finger with different biological tissues and suction probe in undeformed (left) and deformed (right) configuration.

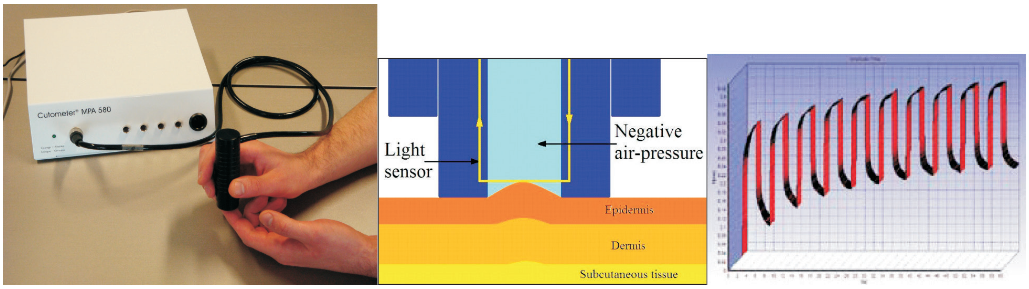

Figure 3. Experimental setup with the Cutometer® suction device

finger-tips (five fingers on each hand). The measurements for statistical analyses were acquired over a period of 15 days in both morning and afternoon sessions.

The results were analyzed using the one-way ANOVA statistical method. The results were further analyzed to compare the load, person, hand and various finger-tip results as defined by two different parameters. The first parameter was the mean value of the time interval between 49 s and 51 s for 200 data sets (MICHALERIS et al, 1994 [16]). This value is related to the response characteristics and their approximation. Note, that for the investigated stationary loading conditions, when the negative pressure was constant (see Figure 6 for loading conditions), the steady state response was not reached in the feasible time frames. The second parameter analyzed to represent how far the curve is from the steady state, was chosen as the mean value of the curve estimated derivatives in the time interval from 40 s to 50 s and represented the displacement rate at the end of the test. Statistics did not show any significant differences between left and right hand fingers. Furthermore, no significant variations were observed with respect to the sampling time over the period of 15 days.

A standard protocol of index fingertip skin characterization was defined. For the skin suction test a probe with an opening radius 2 mm was used. The probe was positioned perpendicularly to the index fingertip and fixed with the other hand of the testing subject. When holding the probe between the fingers it was loaded using just the fingers' weight so that the spring retracted minimally into the device. This handling method is shown in Figure 2. The loading curve used was a pressure step. Seven different pressures were consecutively applied (150, 200, 250, 300, 350, 400, 450) mbar each of them for 60 s followed with a 15 s break. 10 cycles of measurements were performed on both of the index fingers of two different subjects while increasing the pressure from 150 mbar to 450 mbar.

Experimental test results with Cutometer®

Test experimental results for the characterization of index finger skin are presented in Figure 4 as an average response for each pressure level. The confidence interval of the average response was statistically evaluated. The semi-length of the confidence interval (related to a p-value of 0.05) over the

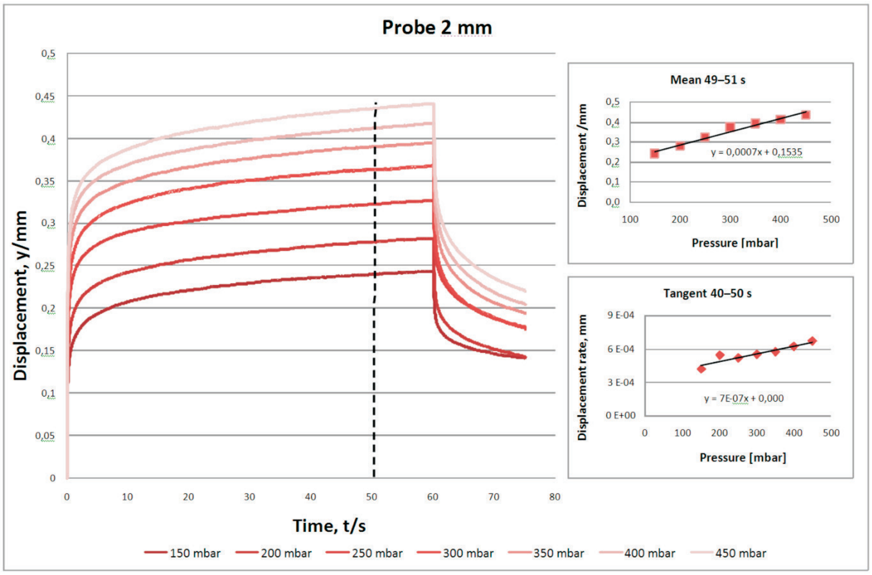

Figure 4. Average responses for each pressure levels for 2 mm hole radius of the test probe

average response is about 5 % for each test. In other word, it was affirmed with a probability of 95 %, that the estimation obtained for the average response is not more than 5 % from the real value of the estimated average response.

Finite Element model of the suction test

A finite element model of the suction tests was developed by AceGen system (KORELC, 2009 [17]) and implemented into AceFEM software (KORELC, 2009 [18]; Wolfram Research Inc Mathematica, 2008 [19]). The discretized numerical model of a finger cross-section with epidermal and dermal skin layers, subcutaneous tissue as well as bone and nail structure is presented in Figure 5. The spatial resolution of the model resembles the shape and size of fingerprint asperities. The probe was modeled by Neo-Hookian model with the stiffness much higher than the stiffness of the sample. The position of the probe was fixed. At the contact between the sample and the probe a Coulomb friction with a value of 0.35 was applied. The contact force generated due to the spring inside the probe was simulated as a pressure generated at the bottom of the bolster. The material law used for the sample was the Fung (FUNG, 1993 [7]) model.

The simulation was divided into four parts, enabling us to assign a different time step to each part, while considering both the computational time and the accuracy of the results. The loading phase was divided into two parts, the first from 0 s to 5 s and the second from 5 s to 60 s. The unloading phase was sectioned from 60 s to 65 s and from 65 s to 75 s into the test. During the loading phase both the probe's force and the negative pressure were considered. During the unloading phase the pressure load was zero and only the probe's force remained acting on the sample. The loading curve of applied negative pressure in respect to time is shown in Figure 6.

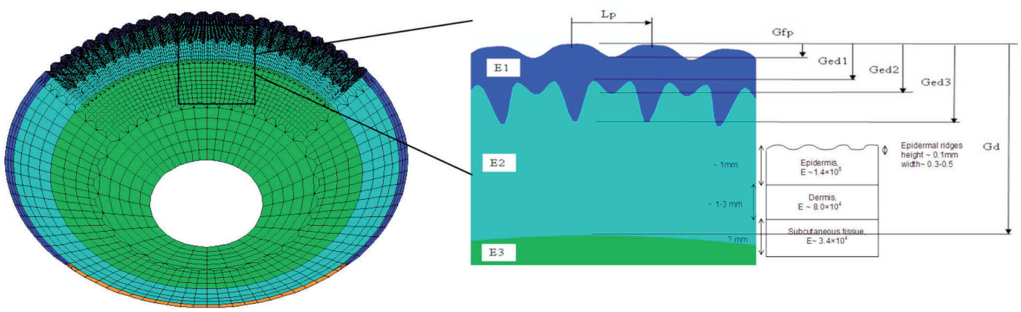

Figure 5. Finite element model of the finger cross-section and of the fingerprint asperities in skin layers

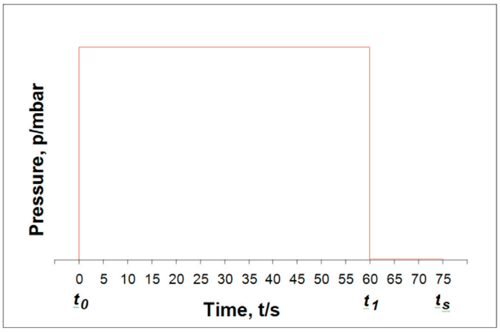

Figure 6. Loading curve used in the suction test

In order to perform the simulation of the FE model it was necessary to assign some values to the unknown parameters. The first such parameter was loading velocity, which was used to describe the function of increasing pressure load to the set value, since the implemented load was not a real step and depended on the performance of the pump, the volume that it had to move and the pressure that it had to maintain. Three different velocities (10^3, 10^5 and 10^7) mbar/s were analyzed, showing no difference between the results of the test, so the value was set as 10^5 mbar/s.

Other parameters set were the coefficients of friction between the probe and the sample and between the sample and the bolster. However, the friction coefficient does not significantly influence the response of the model and was set based on a literature search and on our preliminary research to the value 0.35. The spring force of the probe was set using balance as a 0.55 N. An error in the estimation of this value could cause different responses of the model by shifting the curves, without changing their shapes. The spring force does not influence the proportionality between the load and the displacement. Another parameter "ZeroDisp" was assigned as a displacement at the beginning of the test that represented a quantity of displacement due to the spring force. It was evaluated as an applying force to the model due to the probe. This value, obviously, shifted the response of the curves. The last parameter needed to be set was the thickness of the sample, which was the most important parameter for the characterization of the real skin. Changing the thickness of the sample minimally influenced the shape of the curve, but it had a significant impact on the maximum level reached by the sample into the probe.

Parameters sensitivity analysis

The sensitivity analyses reveal how model responses vary with model parameters. The numerical model for primal analysis was analytically differentiated by using the automatic differentiation facilities in the AceGen system (Korelc, 2009 [17]; Korelc, 2009 [20]; Korelc, 2002 [21]) in order to enable very accurate sensitivity analyses of the model and efficient inverse analysis procedure. In Figure 7 the sensitivity of the maximum fingertip deflection with respect to the elastic modulus of epidermal (a), dermal (b) and subcu-

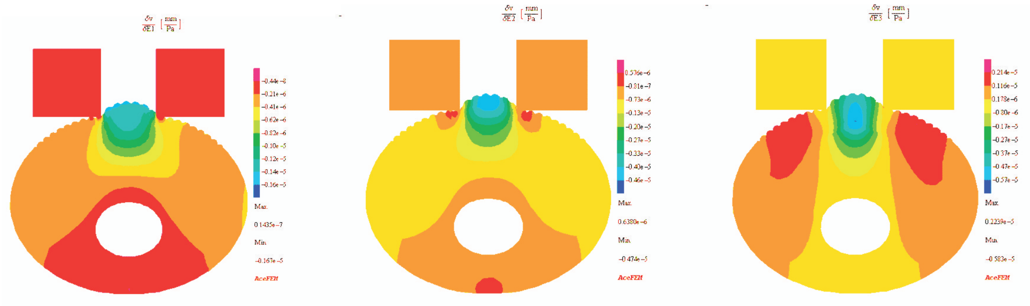

Figure 7. Sensitivity of the maximum fingertip deflection with respect to the elastic modulus of epidermal (a), dermal (b) and subcutaneous (c) tissue.

taneous (c) tissue is shown. Note that variations of the Bulk modulus shift curves up or down, while decreasing the Poisson's coefficient increased the fingertip displacement differences for a given pressure increment.

Inverse analysis procedure

Inverse analyses were used to determine unknown constitutive parameters of fingertip skin (TARANTOLA, 2004 [22]; GREŠOVNIK, 2000 [23]; C3M home page [24]). This was performed by an iterative procedure where the parameters of the model were automatically updated in such a way that the discrepancies between experimental and numerical results were minimized in a least square sense. The inverse approach combines the Finite Element Method with an optimization algorithm in order to find a set of parameters for which the fit between the numerical results and the experimental measurements is optimal (Kauer, 2001 [25]; Seshaiyer & Humprey, 2003 [26]; Kim & Srinivasan, 2005 [27]; Einstein et al., 2005 [28]). The optimum set of parameters x for the model was measured by the user-defined Objective Function $f(x)$ for $x \in R^n$. Constraint functions are defined as $c_i(x) \le 0$ for $i \in I$. The physical consistency of lower (l_k) and upper (u_k) bounds of the set of parameters x, is ensured by $l_k \le x_k \le u_k$ condition for $k = 1, 2, \ldots, n$.

Material model

The skin's behavior was described in terms of the Fung model (FUNG, 1993 [7]) as a quasi-linear viscoelastic (QLV). It was decoupled into a time-dependent elastic response and a linear viscoelastic stress relaxation response, which can be separately determined from the experimental results. The stresses in the tissues, which may be linear or nonlinear, were linearly superposed with respect to time. Rheological scheme of implemented constitutive viscoelastic model is shown in Figure 9. The three-dimensional constitutive relationship in the framework of QLV is given by

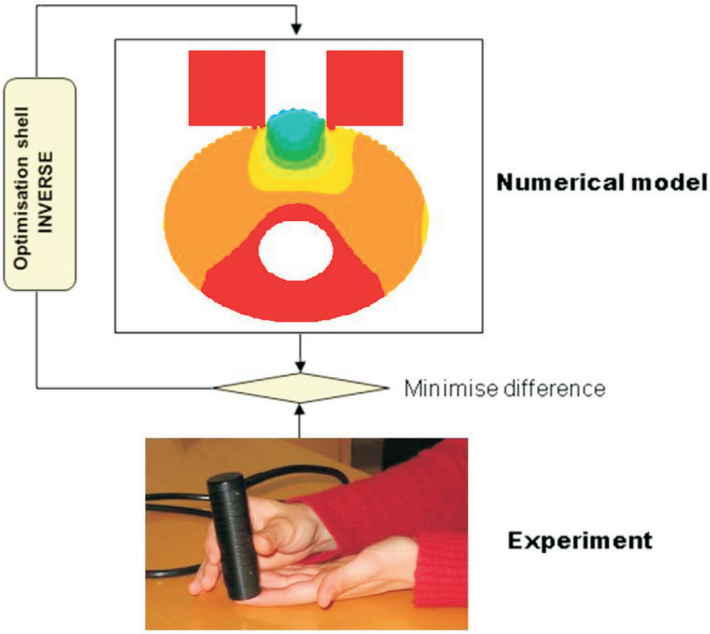

Figure 8. Inverse parameter identification concept based on iterative minimization of discrepancies between numerical results and experimental measurements.

equation:

$$S(t) = G(t)S^e(0) + \int_0^t G(t-\tau)\frac{\partial S^e(E)}{\partial \tau}d\tau \quad (1)$$

where $S(t)$ is the second Piola-Kirchhoff stress tensor, that does not change with material orientation, t is time and $G(t)$ is called the reduced relaxation function, which can be additively split in isochoric ${}^{iso}G(t)$ and volumetric ${}^{vol}G(t)$ part. $S^e(E)$, which is defined by the Green-Lagrange strain tensor E, is called the pure elastic response of the material and can be nonlinear or linear. The reduced isochoric relaxation function ${}^{iso}G(t)$ is a scalar function of time and can be often expressed by the Prony series:

$${}^{iso}G(t) = \sum_{i=0}^{n} {}^{iso}g_i \cdot e^{-t/{}^{iso}\tau_i}$$

$${}^{iso}\tau_0 = \infty \quad (2)$$

where ${}^{iso}g_i$ are the Prony series parameters (Soussou et al., 1970 [29]) and ${}^{iso}\tau_i$ are the relaxation times.

For the nonlinear elastic response a nearly-incompressible hyperelastic material representation was used, which is commonly applied for living tissues that are in general assumed to be incompressible, due to their high water content.

The material properties of the hyperelastic material can be determined by the strain energy function W. Ideally this function is defined with only as many parameters as required to make a FEM model. Many specific material models could be used, depending on how the strain energy function is approximated. In this work the Yeoh energy potential was considered, with a maximum of 5 elastic coefficients c_{i0}, because it best fitted the experimental curves; also, it is quoted in literature that this strain energy potential has often been used for the characterization of nearly incompressible hyperelastic rubber (YEOH, 1990 [6]).

$$W = \sum_{i=1}^{N} c_{i0} \cdot \left(\tilde{I}_1 - 3\right)^i + \frac{k}{2} \cdot \left(J_{el} - 1\right)^2$$

$$S^e\left(E\right) = \frac{\partial W}{\partial E} \qquad (3)$$

where c_{i0} are material elastic parameters (having units of stress) and $\tilde{I}_1$ is the first principal invariant of isochoric Cauchy-Green tensor.

Since the analytical solution considering the above material law and experimental conditions is very difficult, the simulation using FEM has been widely used. Simulation of the test using the QLV approach gives the simulated force F_s, which depends of material parameters ${}^{iso}g_i$, ${}^{iso}\tau_i$ and c_{i0} containing the viscoelasticity and nonlinear elasticity. Elastic parameters are c_{20}, c_{30}, c_{40}, c_{50}. Viscous parameters are ${}^{iso}g_1$, ${}^{iso}g_2$, ${}^{iso}g_3$, ${}^{iso}\tau_1$, ${}^{iso}\tau_2$, ${}^{iso}\tau_3$. Shear modulus was defined as $G = \mu = 2 \times c_{10}$.

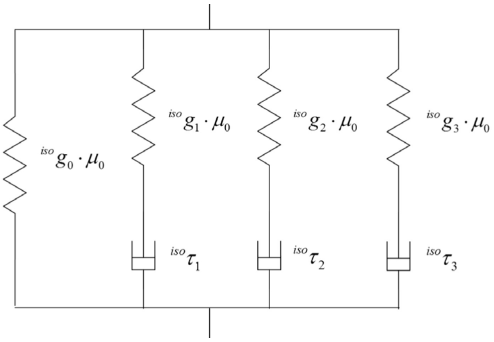

Figure 9. Rheological scheme of implemented constitutive viscoelastic model

Inverse algorithm

For inverse evaluation of skin parameters the FSQP algorithm was applied (YUUNG-HWA et al., 2006 [30]; FSQP home page [31]; FLETCHER, 1980 [32]). It is based on the concept of Feasible Sequential Quadratic Programming (FSQP). It is usually used for problems without nonlinear equality constrains. The algorithm starts with a feasible point, which is provided by the user or generated automatically and produces successive iterates that all satisfy the constraints. Algorithms in FSQP have global and two-step superlinear convergence properties. They also include a special scheme for efficiently handling problems with more objectives or constrains than variables, thus greatly reducing computational efforts.

Parameters evaluated by the inverse procedure were shear modulus G, Poisson's coefficient v (Raveh Tilleman et al., 2004 [33]), displacement at the start of the test ("ZeroDisp") and coefficients (${}^{iso}g_1$, ${}^{iso}g_2$, ${}^{iso}g_3$). Parameter "ZeroDisp" was defined as a value of 0.07 mm. Based on our previous experience of testing various biomimetic materials in which the values were in the range between 0.05 mm and 0.09 mm the average value was taken to define "ZeroDisp".

First analysis of experimental and simulated suction skin testing revealed that experimental data are more dispersed as compared to the predictions of the FEM model. A new and important aspect observed in experimental fingertip testing was the presence of a high residual displacement after unloading. The thickness of human skin sucked into the probe could not easily exit from the probe's opening after unloading, probably because of irreversible slip. The other reasons may be attributed to the deep skin layers that have lower elastic properties, and due to microvascular responses in the living skin, which change the water content of the tissues during and after testing. The last aspect that had to be taken into account was the volume of the attached tissue also mobilized during the test that is described with an index of non-elasticity of the skin.

Estimation of the viscous parameters

Three viscous parameters (${}^{iso}g_1$, ${}^{iso}g_2$, ${}^{iso}g_3$) and three deviatoric relaxation times (${}^{iso}\tau_1$, ${}^{iso}\tau_2$, ${}^{iso}\tau_3$) were inversely analyzed. Only four levels of pressures (150, 250, 350, 450) mbar were used for this first optimization. The objective function was defined as

$$ObjFunc = \frac{a_1}{N_1}\cdot\sum_{i=1}^{N_1}\left(\tilde{d}_i - d_i\right)^2 + \frac{a_2}{N_2}\cdot\sum_{i=1}^{N_2}\left(\tilde{d}_i - d_i\right)^2 + \frac{a_3}{N_3}\cdot\sum_{i=1}^{N_3}\left(\tilde{d}_i - d_i\right)^2 + \frac{a_4}{N_4}\cdot\sum_{i=1}^{N_4}\left(\tilde{d}_i - d_i\right)^2 \qquad (4)$$

where $\tilde{d}_i$ represents the simulation displacement and d_i the experimental displacement. Coefficients a_1, a_2, a_3 and a_4 represent weights for different parts of the tests (0–5 s, 5–60 s, 60–65 s, 65–75 s) that enable us to define the importance of a specific part of the test to ensure a better fit of simulation results to the experimental curve.

Estimation of the elastic parameters

The parameters that influence the elastic part of the response curve were inversely analyzed: shear modulus G and Poisson's coefficient ν. Three values of pressure (200, 300, 400) mbar were used. The differences between the displacement values recorded at the end of the loading phase (after 60 s) for FEM simulation and the experimental data were evaluated. The objective function used was the same as for the estimation of viscous parameters. The process of determination of elastic and viscous parameters was iterative and the final optimization was done simultaneously in order to get optimal values for both parameters.

RESULTS AND DISCUSSION

Comparison of experimental results and results of inverse analysis

Comparisons between experimental results and results from inverse analysis using FEM simulation are shown in Figure 10. They are shown as a correlation between displacement and time for different loads for probe with opening radius of 2 mm. The results of the simulation were fit on the experimental curves by the inverse analysis procedure.

In the tests with the probe with 2 mm of opening radius the elastic parameters are better interpolated for the lowest load pressure. An explanation of such behavior is most likely because the boundary conditions at the lowest pressure load influence the response more strongly. The interpolations in the unloading curves deviate more for the high pressure loads.

Summary of the results

During the experimental phase and also during the parameter estimation process it was confirmed that the in-vivo response of human skin is rather com-

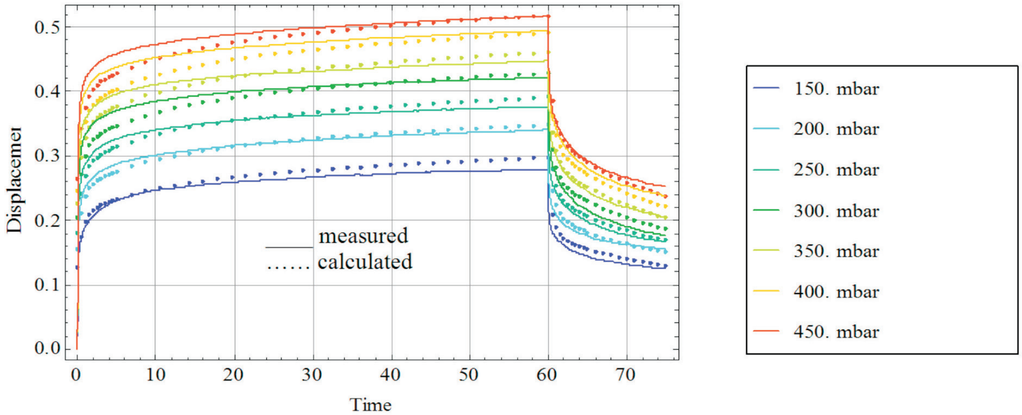

Figure 10. Results of optimization of viscous parameters for skin tested with 2 mm probe radius

plex. Our current viscoelastic model predicts the initial loading phases rather well while additional attention must be paid to the phenomena related to the unrestored energy that are currently not adequately captured.

The elastic parameters used in numerical model were Poisson's coefficient v (hypothesis of isotropic material), shear modulus G (material response to shear strains) and coefficients c_{i0}, which describe non-linearity in range of high strains.

Viscous parameters were ${}^{\mathrm{iso}}g_i$, relaxation times ${}^{\mathrm{iso}}\tau_i$ and a sum of viscous parameters $\Sigma_i {}^{\mathrm{iso}}g_i$. The sum of viscous parameters was defined on a scale from 0 to 1 and represented the unit fraction of the modulus influenced by viscosity in respect to the equilibrium level. In fact its complement to value 1, represents the level of the elastic instantaneous response effect to the equilibrium level.

With reference to the Fung model the skin's parameters which were identified are shown in the Table 1.

In regard of the viscous characteristics, our attention was focused on the parameter ${}^{\mathrm{iso}}g_3$, which represents the fastest viscous contribution in our material model. Even though it was still related to a rather long time constant, it is the most relevant parameter in reproducing tactile sensitivity within our model.

Table 1. Optimal parameters set obtained for the index fingertip skin.

ELASTIC parameters of fingertip skin	
υ	0.489
k/MPa	59.29
G/MPa	1.30398
C_{10}	0.65199
C_{20}	5.39103
C_{30}	0
C_{40}	0
C_{50}	0
${}^{\mathrm{iso}}g_0$	0.207428
VISCOUS parameters	
${}^{\mathrm{iso}}g_1$	0.131161
${}^{\mathrm{iso}}g_2$	0.137995
${}^{\mathrm{iso}}g_3$	0.523416
${}^{\mathrm{iso}}\tau_1$/s	6.73678
${}^{\mathrm{iso}}\tau_2$/s	60.073
${}^{\mathrm{iso}}\tau_3$/s	0.334463
$\Sigma_i {}^{\mathrm{iso}}g_i$	0.792572

Conclusions

This work describes a combined numerical-experimental procedure for the evaluation of the mechanical properties of the human skin at the fingertip of an index finger. In order to characterize the viscoelastic response of human skin a non-intrusive test done "in vivo" was applied by using MPA 580 Cutometer® instrument from Courage+Khazaka. To interpret the measurements in terms of biomechanical parameters an inverse

FEM based procedure was developed where skin's behavior was simulated by Fung's constitutive model. The constitutive parameters of the fingertip skin are applicable for computer modeling of tactile sensations as well as for setting the target viscoelastic properties for biomimetic materials for hand prostheses and humanoid robotics.

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