In-vivo human skin elastography : a preliminary study

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Abstract **- This study is devoted to the estimation of the potential of the high frequency imaging (20 MHz) as a tool to better understand the behaviour of the skin under static stress. In the context of the static elastography technique we have used 2 methods to estimate axial strain field : the gradient method and the scaling factor method. In this paper we evaluate the performances of these two methods in several ways : simulations of layered media by means of 2D finite element model and *FIELD II* software, measurements of strain field in several cryogel phantoms and in vivo measurements in the skin of the forearm.**

I. Introduction

Many studies have focused on *in vitro* and *in vivo* measurements of mechanical properties of the skin. Nevertheless, most of these studies have considered the skin in its entirety. The interest of the elastography is to perform images of mechanical displacements and/or strains within soft tissues and so, to present information about the elasticity of internal structures.

In a first part, this study evaluated methods and estimators performances with 2D finite element simulation which were associated to the acoustic simulator *Field II* software. In a second part, these methods were applied to explore strain field in cryogel acoustic phantoms with an ultrasound scanner developed in our laboratory. Finally, we coupled this ultrasound scanner with a mechanical device called *Extensiometer* and developed in the LMARC laboratory (Besançon-FRANCE), and used this new elastography system for *in-vivo* preliminary measurements of the skin.

I. Methods

Static elastography is based on the detection of the maximum of correlation between local RF signals acquired before and after the application of the stress. This maximum of correlation leads to estimate small displacements or compressions of the tissue. In this approach, two methods are currently used to estimate the deformation.

The first one, the gradient method [1], considers the signal (S'), after stress, as a time-delay (δt) of signal (S) before physical deformation. In this case, local displacement of material is a simple shift of acoustical signatures of the respective pre and post-compressed material regions.

$$S'(t) = S(t + \delta t)$$

Once displacement field has been calculated, the strain field is then computed as the displacement derivative.

The second method, the scaling factor [2], considers the deformed signal as a time-delay (δt) and scaling replica (α) of the signal before deformation.

$$S'(t) = S(\frac{t}{\alpha} + \delta t)$$

With this method, the deformed signal is locally stretched and the strain is directly estimated from the scaling factor corresponding to the maximum of correlation between correlated kernels.

$$strain = 1 - \alpha$$

II. Simulation

In order to evaluate the performance of elastographic estimators, we tested our algorithms using a 2D finite element model . We choose three-layers model, which can be considered as a near geometry of the structure of the skin (epidermis, dermis and hypodermis). We considered that epidermis is four times harder than dermis and dermis is two times harder than hypodemis. In this simulation, the thickness of the epidermis, dermis and hypodermis were 100 µm, 2 mm and 1 mm , respectively (the first two values are commonly found in skin).

Regarding boundary conditions we affected Neumann condition (free) to all borders except borders of the supports : for these borders, we affected dirichlet condition (fixed).

The stress was axially and uniformly applied with a free compressor posed on the surface of the phantom (Figure 1).

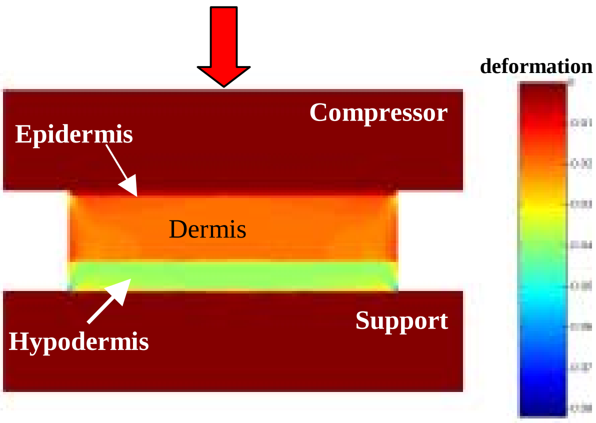


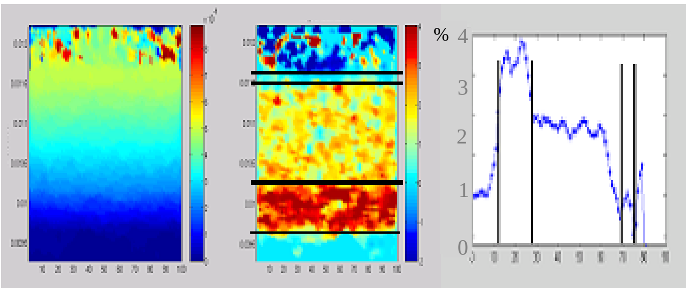
Figure 1 : Finite element simulation of three-layers model: Epidermis (0.5%) Dermis (2%) and Hypodermis (4%).

Mechanical parameters used in this simulation correspond to cryogel material used in the part III of this paper :

Young modulus (E) = 8.10^3 Pa
Poisson coefficient (ν) = 0.5
Density (ρ) = 1030 g/m^3

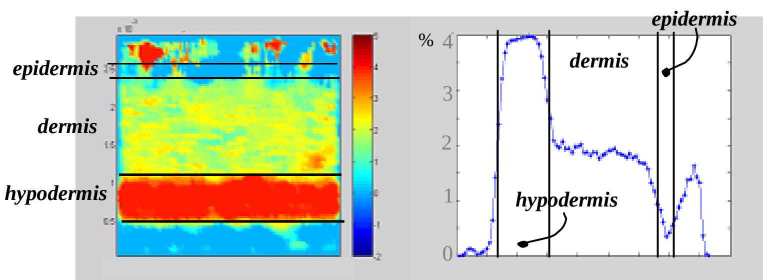
The numerical acoustic phantom was simulated with the *Field II* software [3]. The displacement of the scatterers in the post compressed media were calculated by considering the displacements of the nodes in the finite element model. Moreover, the ultrasonic parameters fixed in this simulation were identical to those in the experiment described below, except for the focal depth (center frequency of the probe = 20MHz, diameter of the transducer = 3mm, depth of the focal zone = 10mm (the surface of the phantom was placed at 9 mm from the transducer), axial resolution = 80 µm, radial resolution = 280 µm, sampling frequency = 100 MHz and 256 rf-lines).

Deformations applied were 4% for the hypodermis, 2% for the dermis and 0.5% for the epidermis. The results of the estimations were obtained with 2λ-temporal window large, 50% overlap and (5-5) median filter. For the gradient method, we implemented parabolic interpolation of the maximum of correlation. With these parameters, we obtained high correlation coefficients (> 0,97) to estimate correctly the strain field (Figures 2 and 3).



a/ b/ c/

Figure 2 : Displacement and strain images (a/, b/), average strains (c/) obtained with the gradient method in the simulated media .



a/ b/

Figure 3 :Strain image(a/) and average strains (b/) obtained with the scaling factor method in the simulated media.

The average strains were computed by considering all the local values of the strain with all the rf-lines and at each depth.

Gradient method is faster but less robust than scaling factor method. As one can see in figures 2 and 3, the 2% and 4 % deformations were better estimated with the scaling factor method from the point of view of the bias of the estimated deformation (the values are under-estimated with the gradient method, especially for 4% deformation) and the regularity of the estimated deformation.

One can also consider the epidermis deformation. In this case, the thickness of the layer was about two times larger than the radial resolution and ,in these conditions, the estimation was difficult. However, with the scaling factor method, one can note that a very thick layer was detected with a deformation of 0.5 %.

III. Experimental study with cryogel phantom

Experimental studies were conducted with a real time ultrasound scanner (Dermcup 2020 developped in LUSSI) with a 20 MHz center frequency probe (R_{axial}=80µm, $R_{lateral}$=280µm and focal depth = 3 mm) The cryogel was chosen as an acoustic material, because cyogel and soft tissues have acoustical and mechanical properties very close[4]. Elasticity of the cryogel is controlled by freeze-thaw cycles and it becomes harder with the number of freeze-thaw cycles. Moreover, at a frequency of 20 MHz, the echogenicity of this material is enough without adding scatterers.

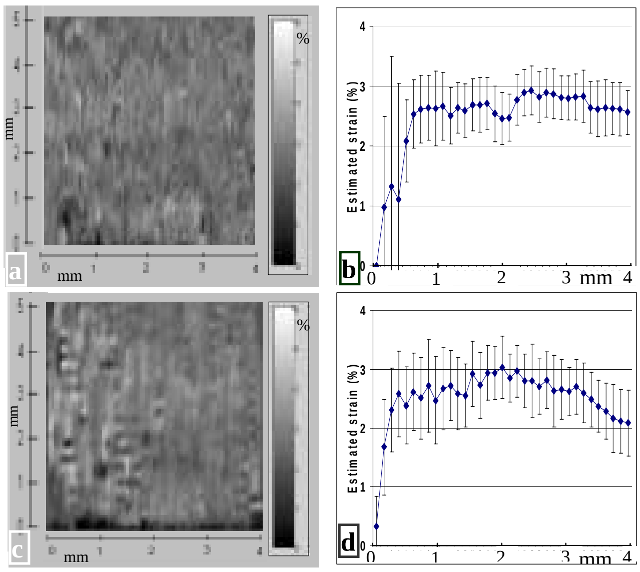


Figure 4 : Strain images (a/ c/) and average strains (b/ d/) obtained with the gradient (a/ b/) scaling factor methods(c/ d/) for an homogeneous phantom and a uniform compression of 3%.

In a first experimentation, we built a parallepipedic homogeneous phantom (thikness: 5mm) of cryogel.

The echographic probe were used as a static and uniform compressor. The parameters were : correlation kernel = 0.3 mm, overlap = 50% and (5,5) median filter.

Figure 4 and 5 show results (strain image and average strains) for a compression of 3% and 5%, respectively, with the gradient and scaling factor method. For the two values of compression, the estimated average strains were less biased with the scaling factor method. The dispersion of the values were similar for the two methods and could be attributed both to the estimators and to the local variations of elasticity of the cryogel.

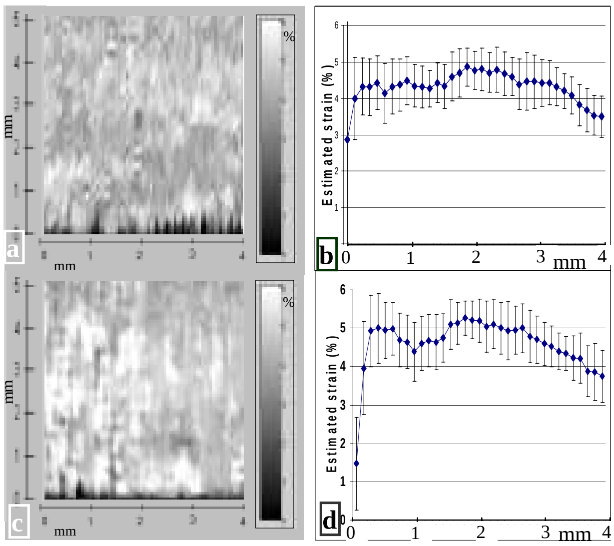


Figure 5 : Strain images (a/ c/) and average strains (b/ d/) obtained with the gradient (a/ b/) scaling factor methods(c/ d/) for an homogeneous phantom and a uniform compression of 5%.

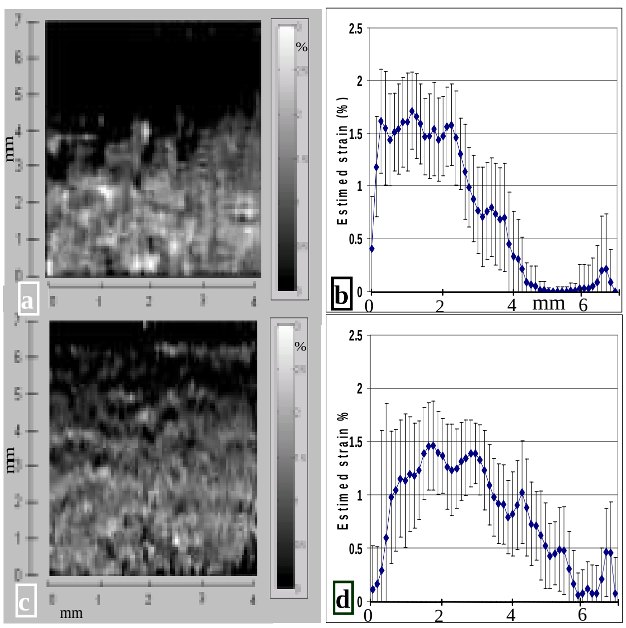


Figure 6 : Strain images (a/, c/) and average strains (b/, d/) obtained with the gradient (a/, b/) scaling factor methods(c/, d/) for a layered phantom and a uniform compression of 1.5%.

We built a second parallepipedic phantom with two layers.

The first layer had a thickness of 4 mm, and the second layer a thickness of 3 mm. The elasticity of these layers were different : soft for the first layer and very hard for the second. Consequently, we could consider that the hard layer was very little strained. The global uniform and axial compression was 1.5 %. Results are shown in figure 6. The two layers are clearly detected with the two methods. Nevertheless, the hard layer was very distinguished from the soft layer with the scaling factor method.

IV. *In-vivo* study

The goal of this part was to propose an experimental technique for *in-vivo* elastography which must allow calibrated constraint of the skin. In order to reach this purpose, we have combined the ultrasound scanner Dermcup 2020 with a device developed by the LMARC Laboratory from Besançon University [5]. In this device, two elements had both two functions : they allowed the skin to be attached to these elements by means of a low suction and stressed the skin by a translation of these two elements in the same plane that the skin, as shown in figure 7. The mechanical probe was placed between these two elements and the image was performed transversally to the stress.

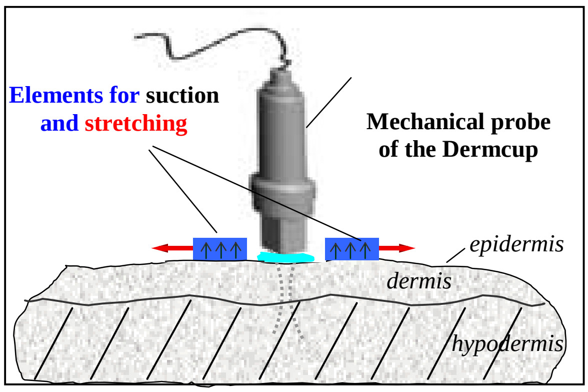


Figure 7 : Method of measurement for in-vivo experiment: sucking in first(black arrows) and stretching in second (red arrows) by means of two blue elements.

In figure 9, one can see a RF image performed in the skin of the forearm. In the following of this part, the displacements and strains images were only estimated by using the gradient method and in the ROI defined in this figure.

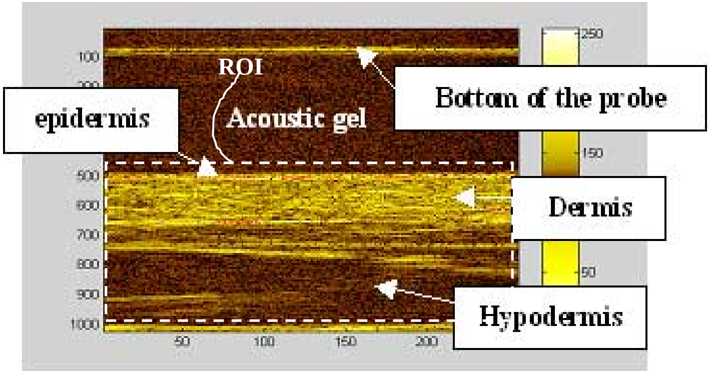


Figure 9 : RF image of the skin of the forearm with the Dermcup 2020 sonographer

The following image (figure 10) show us the image of correlation estimated with the previous RF image. One can clearly see (high values of correlation = white color) the bottom of the probe, the dermis and several large structures in the hypodermis.

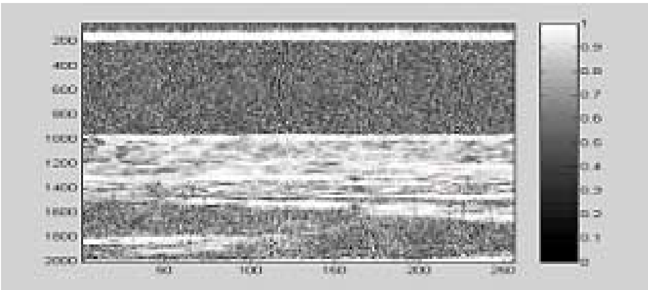


Figure 10 : Image of correlation obtained from the pre and post compression the rf images of the skin.

Images of displacements and strains are shown in figures 11 and 12. Theses images were calculated with gradient method with the following parameters : kernel = 2λ, overlap = 60% and (5,5) median filter.

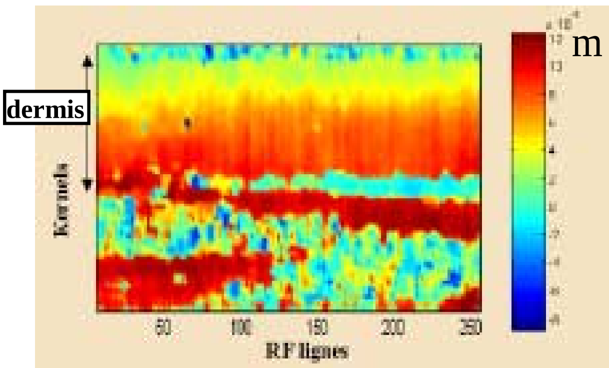


Figure 11 : Image of displacements in vivo estimated in the ROI defined in figure 9.

Displacements appeared as progressive in dermis (low in the top and high in the bottom),while the two large structures of the hypodermis were uniformly translated (no compression).

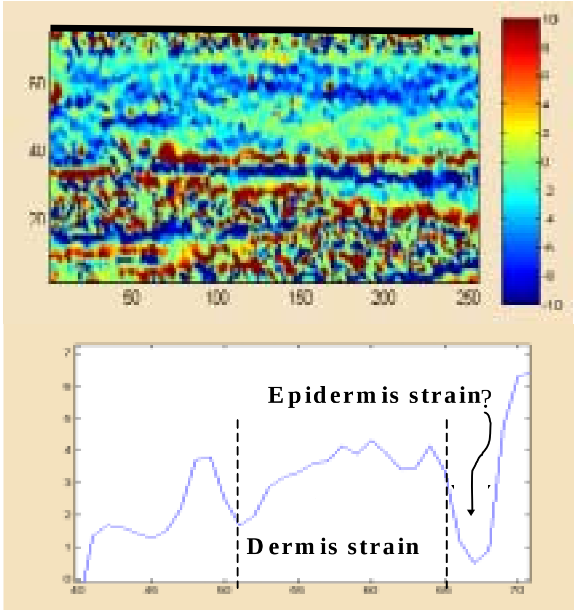


Figure 12 : Images of strains(a/) and average strains(b/) estimated in the ROI defined in figure 9

Average strains showed clearly variations of strain in depth. Regarding the dermis area, values of deformation varied between 3.5% and 1.5%. A value of deformation of 0.5 % could correspond to the epidermis. The inhomogeneous structure of dermis and the kind of stress applied on the skin could explain the variation of the deformation which became smaller in the deep dermis.

V. Conclusion

Preliminary results have shown the feasibility of the displacement and strain measurements in the skin by means of elastography technique at 20 MHz. This technique could be a useful tool to better understand the mechanical properties and behaviour of the skin.

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This work was supported by the P. Fabre Institute of Research (Toulouse, France).