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**Linear elastic properties of the facial soft tissues using an aspiration device:  
towards patient specific characterization**

V. Luboz<sup>1</sup>, E. Promayon<sup>1</sup>, Y. Payan<sup>1\*</sup>

<sup>1</sup> UJF-Grenoble1/CNRS/TIMC-IMAG UMR 5525, Grenoble, F-38041, France, {vincent.luboz,  
Emmanuel.Promayon, ypayan}@imag.fr;

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\*Corresponding author:

Yohan Payan

Equipe GMCAO - Laboratoire TIMC-IMAG

Université Joseph Fourier - CNRS UMR 5525

Pavillon Taillefer

Faculté de Médecine - 38706 La Tronche cedex - France

Tel: +33 (0)4 56 52 00 01 - Fax: +33 (0)4 56 52 00 55

email: [Yohan.Payan@imag.fr](mailto:Yohan.Payan@imag.fr)

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## 22    **Abstract**

23    Biomechanical modeling of the facial soft tissue behavior is needed in aesthetic or maxillo-  
24    facial surgeries where the simulation of the bone displacements cannot accurately predict the  
25    visible outcome on the patient's face. Because these tissues have different nature and elastic  
26    properties across the face, depending on their thickness, and their content in fat or muscle,  
27    individualizing their mechanical parameters could increase the simulation accuracy. Using a  
28    specifically designed aspiration device, the facial soft tissues deformation is measured at four  
29    different locations (cheek, cheekbone, forehead, and lower lip) on 16 young subjects. The  
30    stiffness is estimated from the deformations generated by a set of negative pressures using an  
31    inverse analysis based on a Neo Hookean model. The initial Young's modulus of the cheek,  
32    cheekbone, forehead, and lower lip are respectively estimated to be  $31.0 \text{ kPa} \pm 4.6$ ,  $34.9 \text{ kPa} \pm$   
33     $6.6$ ,  $17.3 \text{ kPa} \pm 4.1$ , and  $33.7 \text{ kPa} \pm 7.3$ . Significant intra-subject differences in tissue stiffness  
34    are highlighted by these estimations. They also show important inter-subject variability for  
35    some locations even when mean stiffness values show no statistical difference. This study  
36    stresses the importance of using a measurement device capable of evaluating the patient  
37    specific tissue stiffness during an intervention.

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39    Key terms: Soft tissues; Face; Mechanical parameters; Aspiration; Finite element method.

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## 42 **1. Introduction**

43

44 In aesthetic and maxillo-facial surgery, most of the interventions are related to the  
45 bony structures and aim either at repairing functionalities of the oro-facial structures or  
46 improving their shape. To predict the final shape of the face after the displacement of the  
47 bones, most of the surgeons rely on their experience. Nevertheless, because of anatomical  
48 variations in the facial soft tissues, *i.e.*, in the amount of fat between the muscles, the  
49 thickness of the skin, or their material parameters, predictions may not always be accurate.  
50 Several research groups have presented simulators to help in the evaluation of the outcomes  
51 of aesthetic and/or maxillo-facial surgeries. For example, Chabanas et al.<sup>6</sup> presented a Finite  
52 Element (FE) model of the skull and face used as an atlas which can be deformed to fit the  
53 patient's anatomy. In this study, the soft tissues were modeled as a homogeneous, linear  
54 elastic material and its material parameters were chosen to fit a clinical case where pre- and  
55 post-operative CT scans were available. It led to a Young's modulus of 15 kPa, and a  
56 Poisson's ratio  $\nu$  of 0.49. Other finite element models also based on a linear modeling of the  
57 soft tissues have been presented in the literature<sup>3, 12, 14, 22</sup>. In these articles, the mechanical  
58 parameters were chosen either by comparing their simulation results with imaging data  
59 acquired from patients or were values from the soft tissue literature, although not specifically  
60 measured on the facial soft tissues. The validation of the predictions given by these simulators  
61 were therefore complicated by the fact that these mechanical parameters were either only  
62 representative of a single patient, or were not directly related to the facial soft tissues, or were

63 extracted from *ex vivo* measurements (and therefore might be different from *in vivo* because  
64 of the lack of perfusion and the difference of temperature<sup>13</sup>).

65 To improve the evaluation of these FE models, it seems important to be able to  
66 determine the material parameters of the facial soft tissues for a specific patient, or at least to  
67 give an accurate cartography of the face mechanical properties for an average patient, if  
68 possible. In a first attempt to reach this aim, an aspiration device, called the Cutometer  
69 (<http://www.courage-khazaka.de>)<sup>7, 18</sup>, was used to define several specific parameters such as  
70 immediate distension, delayed distension, immediate retraction, or final deformation of the  
71 face skin. These studies showed that viscoelastic properties are significantly influenced by  
72 aging. Unfortunately, in both studies, no clear link can be established between the Cutometer  
73 specific parameters and the more classical parameters used in mechanical constitutive laws.  
74 This device is mainly aimed at helping dermatologists in their need to quantify the elasticity  
75 in an aging trend more than for mechanical simulation purpose. Sonographic elastography has  
76 proven its ability to estimate the stiffness of soft tissues for maxillo-facial applications<sup>2</sup>. It  
77 provides a map of Young moduli for tissues' superficial layers which values can only be used  
78 for simulations assuming a small deformations framework. Other studies<sup>4, 5, 9, 15, 17</sup> tried to  
79 mechanically quantify the skin stiffness either by direct measurements or by using inverse  
80 methods to match deformations visualized on medical images. Unfortunately the initial  
81 Young's modulus (i.e., the initial slope of the stress-strain curve) and Poisson's ratio reported  
82 in these studies had fairly important ranges. For example, the Young's modulus was estimated  
83 by Bickel et al.<sup>5</sup> to 78 kPa and the Poisson's ratio to 0.47 using a Neo Hookean model. The  
84 Young's modulus was found to vary between 4 kPa and 18.8 MPa in *in vivo* and *in vitro*  
85 measurements according to Lapeer et al.<sup>15</sup>. Another study<sup>17</sup> also evaluated the initial Young's

modulus to 15 kPa using a Mooney Rivlin model (where  $C_{10} = 2.5$  kPa). In a previous study<sup>9</sup>, we evaluated the *in vitro* stiffness of the cheek of a fresh cadaver to 15 kPa. This work was done in the context of a maxillo-facial simulator. The skin initial Young's modulus was evaluated to 22.8 kPa using a Mooney Rivlin model (where  $C_{10} = 3.8$  kPa) and measurements on different locations on the face of one subject by Barbarino et al.<sup>4</sup>. In a recent study, a micro-robotic device was used by Flynn et al.<sup>8</sup> to record the force-displacement response of the cheek of five volunteers. Facial skin exhibits a non-linear, anisotropic, and viscoelastic force-displacement response. When modeled using an Ogden FE model, the skin initial Young's modulus was found to be between 15.9 kPa and 89.4 kPa. Flynn et al.<sup>8</sup> also showed variations in stiffness between different locations on the face for one subject. Such a paper stems questions about patient specific variations of the stiffness of the facial soft tissues and the stiffness variation depending on the location for different subjects.

In the present paper, we aim at clarifying these patient specific variations in order to improve the planning of different maxillo-facial surgeries using FE model. The objective is to provide a tool to evaluate the facial soft tissue stiffness (characterized here with the Young's modulus) while being compatible with the constraints of the operating room. The goal is to evaluate the *in vivo* initial Young's modulus for the facial soft tissues at four representative locations, namely the cheek, the cheekbone, the forehead, and the lower lip, and for a panel of subjects of different ages and body mass indexes. For the maxillo-facial application, a quasi-static framework can be assumed since boundary conditions corresponding to bone displacements are applied and simulated without taking into account the dynamic visco-elastic behavior of the soft tissues.

## 2. Materials and Methods

### 2.1 Aspiration technique

A device characterizing the mechanical behavior of the soft tissues was used to determine their initial Young's modulus. This device, called LASTIC (for Light Aspiration device for *in vivo* Soft Tissue Characterization), is based on the aspiration technique and was first introduced by Schiavone et al.<sup>19</sup> while quantifying the brain behavior<sup>20</sup>. This technique has also been used by Hollenstein et al.<sup>11</sup> for other applications such as the uterus or the liver. It also has been used on various *in vivo* tissues such as the forearm skin and the tongue<sup>21</sup>. In its current version, Figure 1, LASTIC is a 33 mm x 34 mm metal cylinder composed of two compartments. The lower one is an airtight chamber, open at the bottom by a 12 mm diameter circular aperture and closed at the top by a glass window. The upper compartment holds the electronic part consisting of a miniature 2 megapixel digital camera and a LED used as a light source. The aspiration chamber is connected to a programmable syringe pump that can generate a negative pressure (measured by a manometer) which deforms the tissues on which LASTIC is laid on. This deformation is imaged by the camera via a 45 degree inclined mirror which provides a view of the tissue from the side. The height of the tissue deformation is segmented on the recorded image. A basic camera calibration is performed to determine the pixel size. On average, the pixel size is around 0.01 mm. Measuring the deformation height corresponding to several steps of increasing negative pressures can give an estimation of the behavior of the tissues. LASTIC is fully sterilizable and can consequently be used inside the sterile fields of operating rooms.

The tissue measurements are then processed through an inverse analysis to estimate the tissue mechanical behavior. This analysis consists of matching the measured deformation/pressure curve to a pre-computed library of displacement heights determined by a FE Analysis of the aspiration experiment using a Neo-Hookean constitutive law<sup>1</sup>. Such a material is fairly stable and shows a behavior similar to the one simulated by Yeoh or Mooney-Rivlin materials at strains levels observed for such aspiration experiments. The Neo-Hookean equation is written as:

$$W = C_{10}(I_1 - 3) + (J - 1)^2/D \quad (1)$$

where  $I_1$  is the first invariant of the left Cauchy-Green deformation tensor,  $C_{10}$  is a material parameter,  $J$  is the determinant of the deformation gradient  $F$ ,  $D$  is a material incompressibility parameter (with  $D = (1 - 2\nu)/C_{10}$ ), and  $W$  is the strain energy. As in Chabanas et al.<sup>6</sup>, the skin is assumed to be nearly incompressible and a Poisson's ratio  $\nu$  of 0.49 was chosen. Note that, for small extensions (*i.e.*, in the linear elastic domain when  $I_1$  is close to 3) the initial Young's modulus, *i.e.*, the initial slope of the stress-strain curve, can be approximated by  $E = 6C_{10}$ .

The facial soft tissues are modeled by a thick circular slice while LASTIC is described by a rigid hollow cylinder (Figure 2). Taking advantage of the axisymmetric geometry of our model, the mechanical study is reduced to a two-dimensional structural analysis. The sample is meshed with approximately 2,000 linear quad elements. The mesh is refined in the neighborhood of the aspirated region, where highest deformation occurs, in order to increase the accuracy of the computed solution. The interface between LASTIC and the sample is specifically meshed with contact elements in order to ensure that the aspirated skin tissue slides without friction inside the LASTIC hole.



Given the different thicknesses of the studied facial tissues, an estimation of their variation was performed on a CT scan of a head (courtesy of subject #4). It presented a thickness of 5 to 6 mm for the forehead skin and of 15 mm or more for the tissues of the lip, cheek and cheekbone. To take this variation into account, two different axisymmetric models were used: one with a tissue thickness of 5 mm and one with a thickness of 15 mm (Figure 2). There was no need to create a model for each location over 15 mm because the maximal negative pressure created by LASTIC does not influence layers over 13 mm, see Figure 2. On the other hand, for the model with a tissue thickness of 5 mm, the maximal negative pressure created by LASTIC leads to a deformation of the tissues that is influenced by the tissue thickness. In this case, the maximum deformation measured in the tissues is around 13 % (assuming that we do not consider the specific region of contacts between the tissues and the LASTIC cylinder). For a tissue thickness of 15 mm, the maximum deformation is 17 %. These two values show that the material deformations are not too large; using a Neo Hookean model (thus neglecting the non-linearity due to very large deformations) is therefore relevant here.

The pre-computed library of displacements was generated using a wide range of  $C_{10}$  and applied negative pressure for the two different thicknesses. Matching the measurements with the corresponding library (the 5 mm model for the forehead and the 15 mm model for the lip, cheek, and cheekbone), using a least-square minimization method, leads to an estimation of the  $C_{10}$  value corresponding to the tissue stiffness. This minimization is performed in less than a second.

The device has been validated in Luboz et al.<sup>16</sup> on several types of samples including silicone rubbers, with stiffness ranging from 10 kPa to 90 kPa, and compared to tensile tests.

This validation showed that LASTIC overestimates the stiffness by 16 % on average with a standard deviation of 9.5 %. This overestimation is mainly due to errors generated during the acquisition, namely due to the manometer precision and the camera calibration, which decreases the precision of the image segmentation to evaluate the tissue deformation.

## 2.2 Cartography of the face stiffness

To be able to improve the planning of the outcome of an aesthetic or maxillo-facial surgery, it is necessary to estimate the stiffness of the facial soft tissues in several places. Four locations with presumably different tissue thicknesses and different amounts of fat and muscle were consequently chosen: the cheek, the cheekbone, the forehead, and the lower lip. To study the possible variations between subjects, the stiffness estimation was performed on a group of 16 healthy subjects, eight males and eight females, of different ages and body mass indexes (mean age =  $29.2 \pm 6.3$ , mean BMI =  $21.5 \pm 2.1$ ). Subjects gave their informed consent to the experimental procedure as required by the Helsinki declaration (1964) and the local Ethics Committee (study agreement CERNI n°2013-11-19-30). LASTIC was used to estimate the stiffness of the soft tissues at these four locations for these 16 subjects, see Figure 3. Five measurements were performed for each location. In order to prevent the experimenter from applying too much pressure on the subject's skin with LASTIC and therefore creating a discomfort and a possible bias in the measurement, each subject was asked to place and to maintain LASTIC him/herself. The experimenter helped the subjects to reposition LASTIC as closely as possible to the previous location. The range of negative pressures applied at each

location varied for each subject. The initial negative pressure was 0 kPa while the maximal negative pressure was up to 7.3 kPa (= 73 mbar), depending on the subject and location.

To avoid any leaks at the interface between LASTIC and the tissues, a water wet gauze compress was used to wipe the skin before each measurement. This set up left a slight amount of water facilitating the suction and reducing the viscosity. Furthermore, LASTIC was very slightly pressed on the tissue by the subjects to ensure that the bottom compartment entire surface lies on the skin. The first three measured values were not used during the minimization process in order to compensate the initial load applied by the positioning of LASTIC, which could be observed as a bump on the video screen.

For each location, five measurements were made repeatedly: the first measurement was performed successively on the cheek, the cheekbone, the forehead, and the lower lip; then, the second measurement was performed successively on the cheek, the cheekbone, the forehead, and the lower lip, etc... until the fifth measurements was performed the same way. This repetitive cycle allowed the tissue of each location to relax between each measurement. A single measurement took about three minutes which means that overall, each subject's 20 measurements were performed in about one hour.

During all the measurements, the subjects were asked to stay as relaxed as possible in order to keep the activations of the underlying muscles of the face as low as possible and consequently to reduce the impact of tissue anisotropy and initial tension on the measurements.

### **3. Results**

The stiffnesses of the facial soft tissues (characterized here with the initial Young's modulus) measured with LASTIC on 16 healthy voluntary subjects at four locations, as well as the sex, age and BMI of each subject, are shown in Table 1. The cheek, cheekbone, forehead, and lower lip mean stiffnesses of each subject, as well as the standard deviation resulting from the five measurements at each location, are given in the four last columns. The overall subjects' mean stiffness for each of the four locations is also given on the bottom row. The corresponding initial Young's modulus are  $E_c = 31.0 \text{ kPa} \pm 4.6$  for the cheek,  $E_{cb} = 34.9 \text{ kPa} \pm 6.6$  for the cheekbone,  $E_f = 17.3 \text{ kPa} \pm 4.1$  for the forehead, and  $E_l = 33.7 \text{ kPa} \pm 7.3$  for the lower lip. Neither tissue stiffening nor softening is observed for the repeated measurements; we therefore assume that there is probably little influence of the pre conditioning.

The complete measurements are given in the supplementary material. Figure 4 presents a Whisker box plot showing the mean, minimal and maximal initial Young's modulus (in kPa) for the cheek, cheekbone, forehead and lower lip. The result of the bilateral paired Welch's T-test between the forehead and the three other locations is also presented. This test is an adaptation of the Student's t-test intended for use with two samples having possibly unequal variances<sup>23</sup>.

Table 2 shows the results of a bilateral paired Welch's T-test between each measurement location. The p-value resulting from this test demonstrates that the difference between the mean stiffness of the forehead  $E_f$  and the cheek  $E_c$  ( $p=2.5E-5$ ) is statistically significant ( $p \leq 0.05$ ). This is also significant for the difference between  $E_f$  and  $E_l$  ( $p=7E-6$ ) and for  $E_f$  and  $E_{cb}$  ( $p=1E-6$ ). On the other hand, there are no statistical differences between  $E_c$  and  $E_l$ , between  $E_c$  and  $E_{ch}$ , and between  $E_l$  and  $E_{ch}$ .

It is to be noted that the evaluated mean stiffnesses do not show any dependence neither with the age, BMI or sex.

Nevertheless, subject to subject differences can be observed for each location even if there is no overall statistical difference. It is the case for example for the cheekbone and lower lip stiffness of subjects #5 and #15:  $E_{cb}(5) = 22.1$  kPa and  $E_l(5) = 33.3$  kPa while  $E_{cb}(15) = 35.1$  kPa and  $E_l(15) = 22.3$  kPa. Another example of the disparity can be seen between subjects #10 and #12 for the cheek and lower lip stiffness:  $E_c(10) = 37.1$  kPa and  $E_l(10) = 28.0$  kPa while  $E_c(12) = 31.6$  kPa and  $E_l(12) = 45.7$  kPa. These two subjects have completely different stiffness values compared to the mean stiffness  $E_l$  for the lower lip (it can be observed that  $E_l(10) = E_l \times 83\%$  while  $E_l(12) = E_l \times 135\%$ ), see Figure 5, and for the cheek while overall there is no statistical difference between these two locations ( $p=0.424$ ). For all these values, the standard deviations for the five measurements used to obtain these mean values are relatively low (between 4.0 and 7.9 kPa).

#### 4. Discussion

The stiffness values presented in Table 1 and Figure 4 and outlined in the previous paragraph show fairly important variations between subjects, even if there is no statistical difference over the whole subject pool. The cheek stiffness measured on all the 16 subjects fall within the range given by Flynn et al.<sup>8</sup> (which is based on cheek measurements on five volunteers): from 15.9 kPa to 89.4 kPa.

As shown in Table 2, the stiffnesses of the forehead and the cheek, of the forehead and the lower lip, and of the forehead and the cheekbone are statistically different. It can be

explained by the smaller tissue thickness (and different boundary conditions) for the forehead as compared to the cheek, the lower lip, and the cheekbone. Precisely measuring tissue thicknesses at different locations (with MRI or US imaging) should help in understanding more deeply these differences. Measuring and estimating variations in tissue types or muscles activations should also provide some clues to explain the observed differences.

A first estimation of facial tissue stiffness can nonetheless be given with our LASTIC measurements. For a given location, the inter-subject variability can be estimated by the standard deviation  $sd(E_i/E)$ , where  $E_i$  is the normalized Young moduli of subject  $i$  and  $E$  is the average Young modulus of all  $E_i$ . The inter-subject variability  $sd(E_i/E)$  is equal to 27 %, 29 %, 29 %, and 27 %, respectively for the forehead, cheek, lower lip, and cheekbone. For a given location, the intra-subject variability can be estimated by the average standard deviation  $mean(sd(E_{i,j}/E_i))$ , where  $E_{i,j}$  is the resulting measured Young modulus for the measurement  $j$  of subject  $i$ . The intra-subject variability  $mean(sd(E_{i,j}/E_i))$  is equal to 22 %, 16 %, 21 %, 19 %, respectively for the forehead, cheek, lower lip, and cheekbone. This shows that for a given location, the intra-subject variability (i.e., the standard deviation per subject) is smaller than the inter-subject variability (i.e., the standard deviation per location). Consequently the small variation of position due to the repositioning of the device between two measurements for the same location has less effect on the estimation than the change of stiffness from a subject to another. Furthermore, we could question the fact that muscle activation could play a role in the estimation of the stiffness since two locations include muscles that can be voluntarily activated (i.e., lip and forehead) while the other two include muscles that are more difficult to activate voluntarily (i.e., cheek and cheekbone). The lower lip and forehead have indeed higher inter-subject variability than the cheek and cheekbone. Finally, it is important to note

that the maximal difference between location variability being 6 % (between forehead and cheek), these variations could also be explained by a change of device positioning, or a measurement error (see our previous work<sup>15</sup> for an estimation of LASTIC errors).

The stiffness differences pointed out in the last part of the results section illustrate the fact that even if there is no statistical difference over the whole subject pool between some locations, for example between cheekbone and lower lip, there are fairly important variations of the stiffness between subjects. For example, it would be inaccurate to take the mean value  $E_{cb} = 34.9$  kPa for the cheekbone for subjects #4 and #13 while the measured stiffness values were evaluated to  $E_{cb}(4) = 50.7$  kPa and  $E_{cb}(13) = 51.6$  kPa, which would correspond to an underestimation of 30%. The same observation can be made for the stiffness of the lower lip in subject #8: the mean value  $E_l = 33.7$  kPa while the subject's stiffness value was estimated to  $E_l(8) = 17.9$  kPa; this would lead to an over-estimation of 88%. It therefore seems essential in the context of FE model for surgical planning to take these inter-subject variations into account.

Two types of limitations should be considered for this study: the first one concerns the experimental set up, and the second one concerns the mechanical set up. The experimental set up has four main limitations: the fact that muscle activation is not recorded during the experiments, the assumption that the facial soft tissues are homogeneous, the assumption that those tissues exhibit a linear stress-strain response, and the low variation of the subject pool in terms of age and BMI. The main limitation is relative to muscle activation. Even though subjects were asked to be as relaxed as possible, we cannot guarantee that their muscles were not activated at all, which would consequently have biased the measurements by stiffening the tissues. Because this activation could not be measured without using invasive EMG sensors, it

is likely that the measurements with high standard deviation are the consequence of involuntary muscle activation.

Assuming that the facial soft tissues are homogeneous is also erroneous: they are composed of several layers of skin (including epidermis, dermis and hypodermis), muscles and fat and can be more than a centimeter thick. Given the level of aspiration generated by our LASTIC device (never higher than 7.3 kPa and a maximum tissue bump of about 5.1 mm, with a maximal pre-load of 3.2 mm, leading to an actual deformation directly due to the suction of 1.9 mm), the obtained characterization is mainly limited to the superficial layers of the facial tissue (epidermis and dermis), few millimeters below the skin at most. Assuming that these tissues are homogeneous consequently results in a non-completely accurate stiffness estimation. The stiffness estimation could be improved by considering a heterogeneous model with several layers of tissues with different mechanical properties. For instance, a two layer model including a thin superficial layer for the skin and a thicker one for the underlying tissues could be considered as a better approximation of the face tissues. Measuring the facial soft tissue stiffness in their full thickness would also need a higher level of negative pressure, which would probably have been refused by the Ethics Committee because of the risk for generating pain or damaging the tissues.

Assuming that the facial soft tissues exhibit a linear stress-strain response is also inaccurate for large deformations: because of their heterogeneity, the tissues mainly have a nonlinear response<sup>9</sup>. Estimating the Young's modulus corresponding to a Neo Hookean constitutive law is consequently accurate only for the initial low strain stiffness. Using a more complex constitutive law, such as one derived from a Mooney Rivlin or Ogden formulation, would probably improve the accuracy of the stiffness estimation with LASTIC.



Finally, despite some individual differences, the relatively small variations of the stiffness measured on our group of subjects can be explained by its low variance in age and in BMI. The subjects being all young or relatively young (between 23 and 44) their skin is likely to be in good shape and fairly elastic. As for the subject's BMI, it ranges between 16.8 and 26, and only four subjects are outside the standard deviation range [19.4; 23.6]. Most of the subjects therefore have probably a small amount of fat tissues under their face skin. The studied group is consequently not extremely representative of the world population neither in age nor in BMI.

Three main limitations can be listed concerning the mechanical set up: the low level of negative pressure generated by LASTIC, the initial load that may have been applied on the soft tissues, and the possible inaccuracy or non-reproducibility of the position of the measurements on each subject.

As stated above, the level of aspiration generated by LASTIC is never higher than 7.3 kPa which therefore limits the estimation of the stiffness to only superficial tissues. Using higher negative pressures could help to quantify deeper tissues but could also injure the subject; this was consequently not performed in our study.

Another source of inaccuracy in the stiffness estimation is due to the fact that the initial load applied on the soft tissues at the beginning of each measurement (to avoid any leaks) is not simulated in the FE model nor used for the inverse analysis. Removing the first three measured values seems a reasonable approximation as it is assumed that the pressure compensates the initial load and the consequent deformation of the tissues<sup>15</sup>. The initial load is always kept below the precision threshold of the manometer by checking the measured pressure on the manometer and the camera image. The degree of this initial pressure applied

by the subjects is controlled during and after the experiments. This initial load creates an initial deformation of the tissues and a light aspiration has consequently no influence on them. Once the pressure is strong enough to aspirate the tissues, the tissue deformation is visible on the camera image. This threshold was reached around the third pressure step. Nevertheless, this compensation is not accurate and might lead to a deviation of the stiffness value.

The inaccuracy of the location of the measurements for each subject could also be responsible for increasing the resulting variance. The experimenter visually estimates the positioning error to a maximum of 5 mm; we therefore assume that this has a minimal effect on the stiffness evaluation.

## **5. Conclusion**

A map of the stiffness of the facial soft tissues is presented in this paper as measured by LASTIC, a device based on the aspiration technique. Using an inverse analysis with a FE Neo Hookean behavior, it provides an estimation of the stiffness of the tissues at four locations: the cheek, the cheekbone, the forehead and the lower lip. On average, the stiffness coefficient of the soft tissues composing the cheek, the cheekbone, the forehead, and the lower lip are respectively estimated to be 31.0 kPa, 34.9 kPa, 17.3 kPa, and 33.7 kPa. Considering all measurements, it seems difficult to obtain and use values describing an average patient. With statistical differences between the forehead and the cheek ( $p=2.5E-5$ ), between the forehead and the lower lip ( $p=7E-6$ ) and between the forehead and the cheekbone ( $p=1E-6$ ), this study shows that the tissue thickness as well as the amount and nature of fat tissues and muscles below the skin probably play a role in the stiffness. Although no

dependences can be exhibited between the mean tissue stiffnesses and neither the age, the BMI nor the sex of the subjects, this study gives an interesting first insight in the variation of the stiffness between subjects at different locations on the human face. The inter-subject variations appearing in our measurements are pointing out that *in vivo* patient specific measurements are essential to accurately model the facial soft tissues and that an average stiffness value is not adequate for a patient specific model. Specifically designed small aspiration devices are a convenient and practical way for doing so. As LASTIC can be sterilized, it could be used routinely during clinical practice in order to assess rapidly the patient specific tissue stiffness during the planning of aesthetic or maxillo-facial surgeries.

Future works will aim at correcting the limitations of this study. Firstly, the range of the population tested with LASTIC should be broaden by measuring the stiffness of the tissues on younger and older subjects and with a larger variation in BMI. Secondly, the accuracy and reproducibility of the device positioning during the measurements could also be improved, for instance by marking precisely onto the skin the aspiration locations. Increasing the level of negative pressure that LASTIC can generate in order to quantify the stiffness of deeper tissues is also possible but can only be done after evaluating the risk of damaging the skin. Quantifying the properties of the different skin layers and underlying tissues could also be implemented by using different sizes of orifice for the suction similarly to the work of Hendricks et al.<sup>10</sup>. Another improvement to be addressed is the fact that following only the deformation of one point at the top of the tissue aspired dome is not enough to evaluate soft tissues' anisotropy. We therefore plan to segment the whole deformation dome from the acquired images and to measure its possible asymmetries which could lead to study the tissues' anisotropy. Lastly, monitoring the muscle activation might be possible using surface

EMG, even though it might be difficult to avoid the electrodes placed close to LASTIC, thus interfering with the aspiration device.

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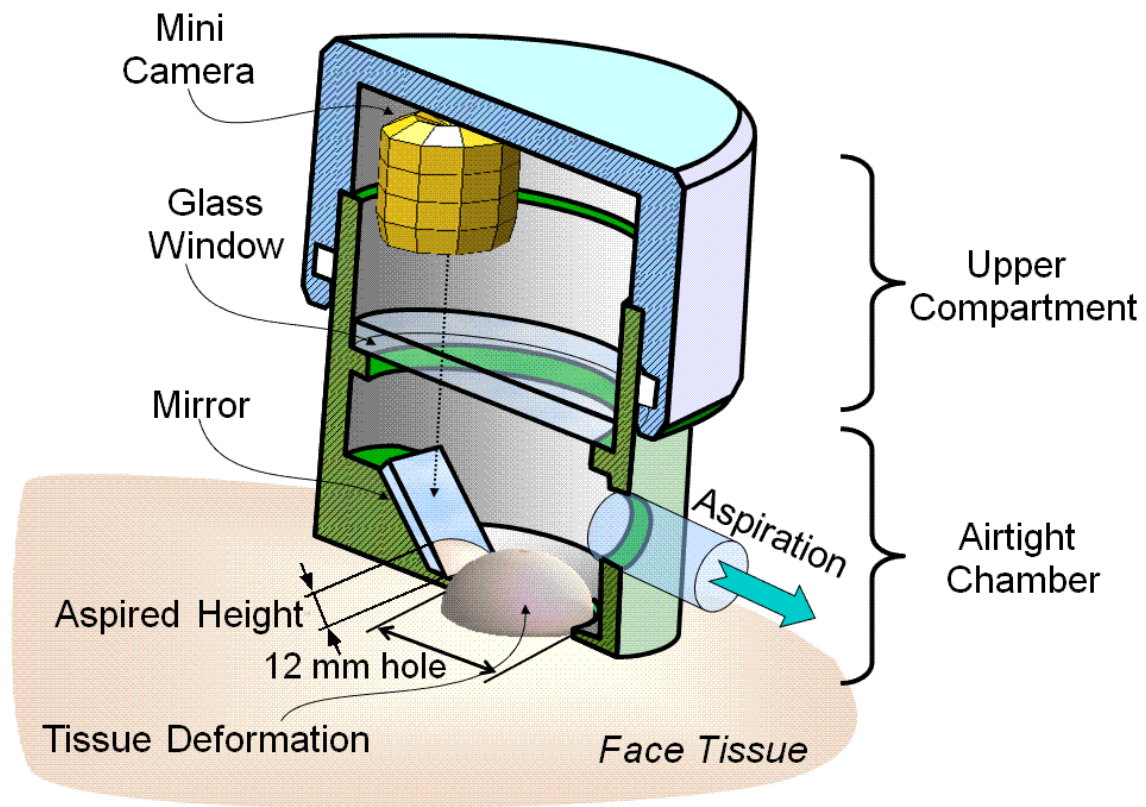
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485 Figures and tables

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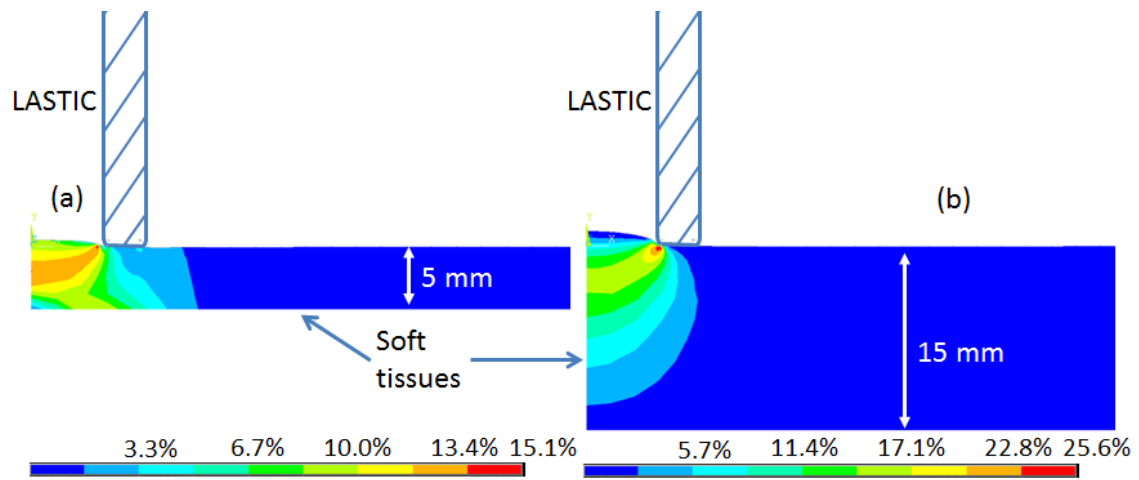
487 Figure 1 Luboz



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489 Figure 2 Luboz



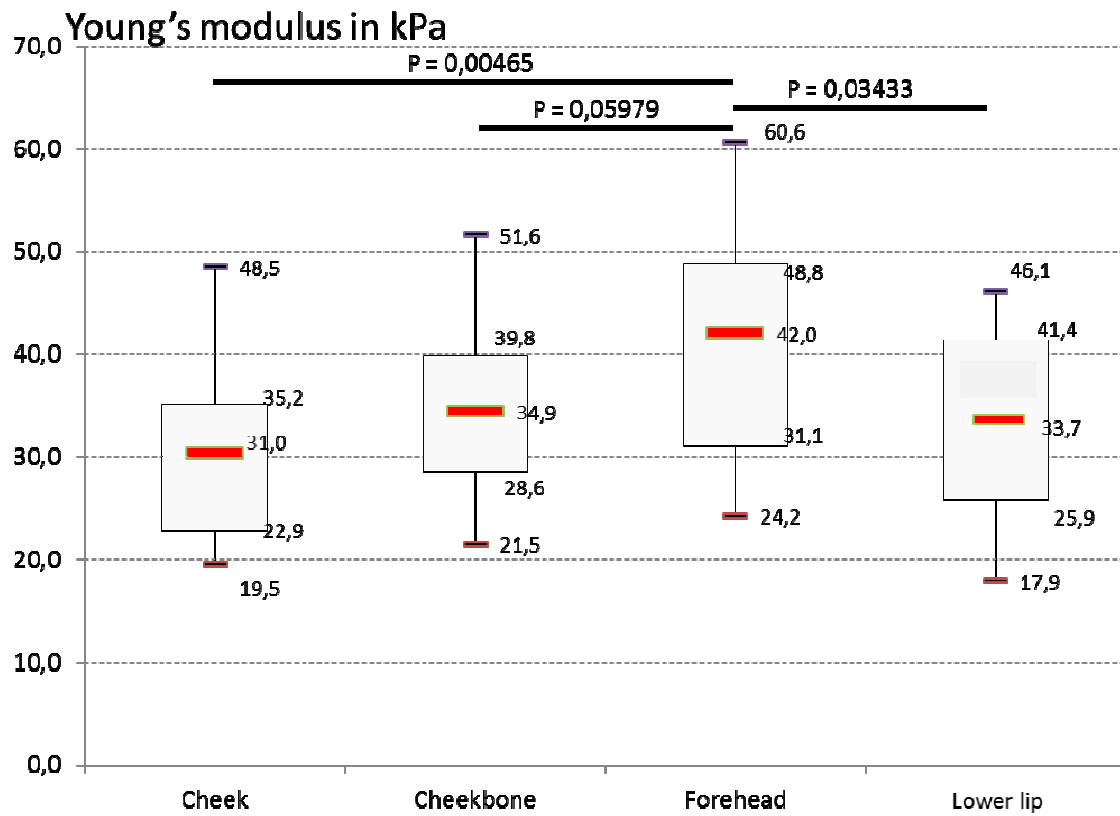
492 Figure 3 Luboz

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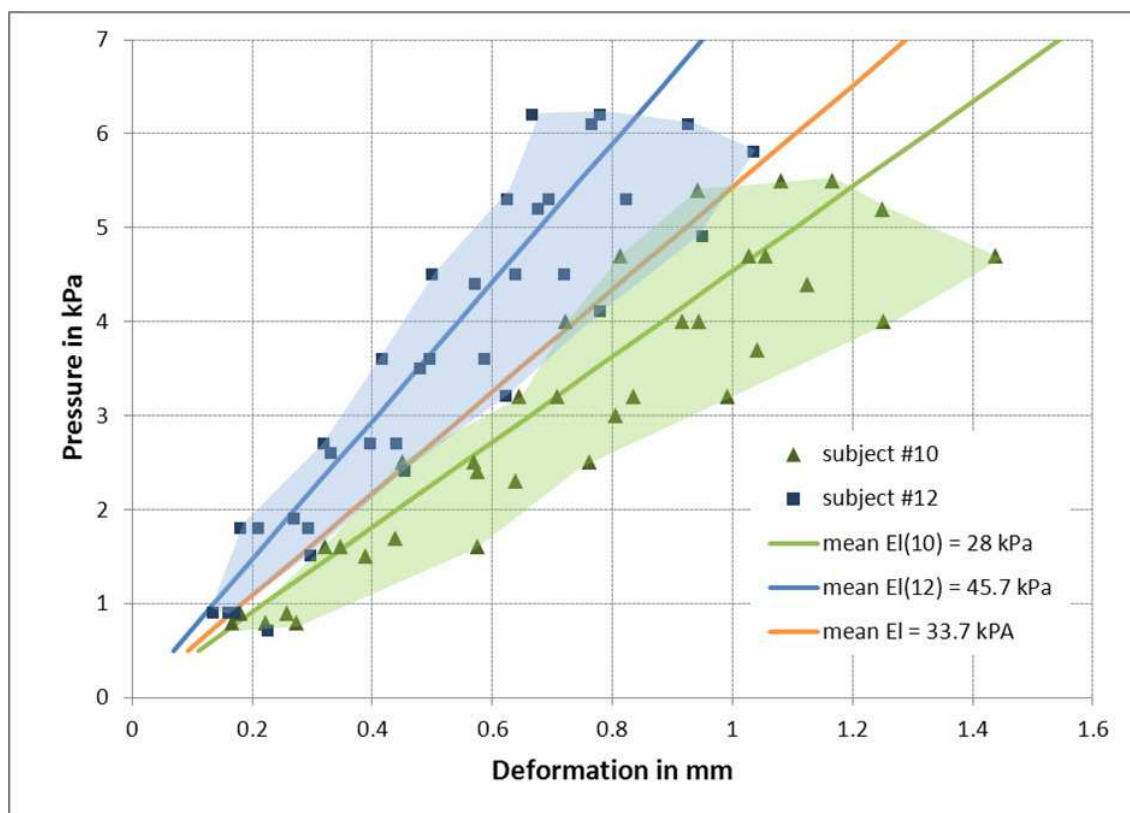
495 Figure 4 Luboz



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498 Figure 5 Luboz



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500 Table 1 Luboz

Subject # / sex	age (years)	BMI (kg.m <sup>-2</sup> )	Cheek E <sub>c</sub> (i) ± std (kPa)	Cheekbone E <sub>cb</sub> (i) ± std (kPa)	Forehead E <sub>f</sub> (i) ± std (kPa)	Lower lip E <sub>l</sub> (i) ± std (kPa)
1 / M	44	22.0	21.3±3.7	21.5±2.1	10.6±1.6	19.1±3.4
2 / M	35	21.2	20.7±4.5	23.1±3.4	9.9±2.4	20.1±4.1
3 / F	26	20.7	23.4±7.4	29.0±11.8	16.0±2.7	46.1±12.7
4 / F	35	22.8	48.5±9.9	50.7±5.0	24.2±7.4	41.2±7.2
5 / F	26	19.2	21.1±5.0	22.1±2.5	10.5±2.9	33.3±8.2
6 / M	23	21.1	28.9±3.3	32.1±2.8	13.0±2.9	27.1±7.2
7 / F	24	20.3	34.5±4.6	39.3±13.8	30.2±7.9	36.5±7.9
8 / M	23	21.9	19.5±2.3	27.3±6.6	11.7±3.5	17.9±4.6
9 / M	39	26.0	31.2±4.0	41.4±5.8	18.9±3.8	39.8±11.4
10 / F	24	16.8	37.1±4.9	39.3±10.9	19.5±5.1	28.0±5.9
11 / M	27	25.1	29.1±3.3	39.1±10.1	22.8±1.8	38.0±11.9
12 / M	27	19.9	31.6±4.0	30.5±8.1	24.8±6.6	45.7±7.9
13 / M	27	22.1	43.1±2.6	51.6±7.5	15.9±4.1	42±12.8
14 / F	32	21.5	44.5±3.7	42.4±3.7	21.9±7.5	39.1±3.8
15 / F	32	21.2	29.5±3.3	35.1±1.0	15.6±2.9	22.3±1.0
16 / F	24	22.3	32.4± 7.0	33.7± 11.0	11.5±2.9	43.4± 6.3
Mean	29.3±6.		E <sub>c</sub> =	E <sub>cb</sub> =		
± std	3	21.5±2.1	31.0±4.6	34.9±6.6	E <sub>f</sub> = 17.3±4.1	E <sub>l</sub> = 33.7±7.3

501 Table 2 Luboz

	Cheek $E_c$	Lower lip $E_l$	Cheekbone $E_{cb}$
Forehead $E_f$	0.000025	0.000007	0.000001
Cheek $E_c$		0.424	0.238
Lower lip $E_l$			0.733

502 Captions:

503 *Figure 1 – Cross section of LASTIC's two compartments. The lower part is the aspiration*  
504 *chamber with the mirror and the upper part contains the camera that images the deformation.*

505 *Figure 2 – The two different Finite Element models used to create the precomputed library*  
506 *of displacement heights / pressure curves, for a tissue thickness of (a) 5 mm (forehead), and*  
507 *(b) 15 mm or more (lip, cheek, cheekbone). The Von Mises strains are plotted on both sides,*  
508 *showing maximum deformations (of (a) 15 % and (b) 26 %) near the interface with LASTIC*  
509 *because of the contact. But the measured displacement height is at the top of the deformation*  
510 *dome.*

511 *Figure 3 – The measurements are done at four locations: the cheek, the cheekbone, the*  
512 *forehead, and the lower lip. The subjects are asked to position and hold LASTIC themselves*  
513 *under supervision of the experimenter. From the LASTIC measurements, the stiffness of the*  
514 *facial soft tissues at the different location can be estimated.*

515 *Figure 4 – Whisker box plot showing for each location: mean, minimal and maximal initial*  
516 *Young's modulus (in kPa). The result of the bilateral paired Welch's T-test between the*  
517 *forehead and the three other locations is also presented.*

518 *Figure 5 – Variation of the initial Young's modulus  $E_l$  (lower lip) for subjects #10 and #12*  
519 *(averaged for the five measurements), compared to the mean initial Young's modulus for all*  
520 *subjects. The mean stiffness computed by the inverse analysis using a Neo Hookean*  
521 *approximation is also plotted (continuous lines). Measurement points are also plotted.  $E_l(10)$*   
522  *$= E_l \times 0.83$  and  $E_l(12) = E_l \times 1.35$ .*

523        *Table 1 – Age, body mass index (BMI), sex (M for Male and F for Female), initial Young's*  
524        *modulus  $E$  (in kPa) and standard deviation (std) of each subject  $i$  for the facial soft tissues at*  
525        *four locations: cheek ( $E_c$ ), cheekbone ( $E_{cb}$ ), forehead ( $E_f$ ), and lower lip ( $E_l$ ).*

526        *Table 2 – P-value for each possible pair of locations where the stiffness is estimated.*

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