Measurements of Mechanical Interactions between Ovaries and a Rigid Instrument.

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Abstract

In order to program a training simulator for laparoscopic surgery and to reproduce realistic haptic feelings, we performed measurements on isolated ovaries of sows. Two types of 1-D mechanical tests were performed: loading-unloading (strain/stress) and relaxation tests. The first results show that the ovary behaves as a viscoelastoplastic body. The loading-displacement curves illustrate a non linear elasticity. We identified the parameters of a Maxwell model simulating the ovary. The presented works will be pursued to take into account the dynamic behaviour, investigate the heterogeneity of the ovarian structure and carry out in vivo measurements under true surgical conditions.

Keywords : biological organ, mechanical model, viscoelasticity, deformable bodies

Introduction

The medico-surgical training simulators are of major interest in the initial or continuing education of the surgeons. By dissociating the formation from the operative practice, they associate a lack of risk for the patient with the possibility given to the resident to cope with a greater diversity of pathological situations. The Scientific Interest Group " Simulators of medico-surgical instruments ", created in Lille in 1995, undertook to design and develop prototypes of instrumental simulators. It is working in close cooperation with the clinicians of the concerned disciplines ; the participation of the physicians starts from the initial definition

of the objectives to the final validation of the carried out devices. Various fields of minimally invasive surgery were thus already approached, among which ophthalmology - in laser photocoagulation -, gynaecologic laparoscopy, shoulder's arthroscopy.
One of the main specificity of medico-surgical simulation consists in that the operative field includes alive biological bodies, which can be warped under the action of the operator. For example in gynaecological surgery, the lower part of the abdominal cavity is considered, with the accounting organs, mainly the uterus, the fallopian tubes, the ovaries, the colon and the bladder. The operator carries out gestures involving either the touch (tactile sense) or an operative instrument (haptic sense). The realism of the training process demands a faithful restitution of the visual (geometric deformations of the bodies) and haptic (perception of forces and couples) interactions coming with handling of the organs. This requires to be capable to establish models suited to describe in real-time the mechanical behaviour of these very deformable bodies. The importance of this haptic skill was highlighted, as well from the qualitative [1] as quantitative [2,3] point of view.
In the Virtual Reality domain, various works were undertaken aiming at suggesting deformable geometrical models [4,5,6]. Most of them are based on physical laws but they are generally empirical. There is in this field two major kinds of models: discrete and continuous models. On the other hand, more recent studies referred to the mechanical characterization of in-vitro [7,8] or in-vivo [2,9,10,11] tissues.

Problematics

The theory of elasticity relates to the small continuous and reversible deformations of elastic media [12,13]. In particular, it bases relationships between the deformation or strain (ε) and the applied loading or stress (σ). The Hooke's law relates to the linear relation existing between these two variables for relatively weak deformations. The Young's modulus - or elastic modulus - (E) is one of the parameters characterising a given material. In a simplified way (1-D), it is considered that:

$$E = \sigma / \varepsilon \qquad (1)$$

An usual way for the experimental determination of elasticity consists in carrying out the loading diagram, obtained by drawing the variations of σ according to ε. It is a widely accepted fact [14,15] that biological tissues have a viscoelastic behaviour, i.e. a time-dependent elasticity. Moreover, former works [16] had highlighted the non-linear viscoelasticity of some particular tissues.
Our goal is to carry out an experimental mechanical study of the whole interaction between alive biological organs and operative instruments. The experimental study of the forces and deformations undergone by a biological body can thus allow to get a more precise quantitative characterization by using these parameters in a analytical simulation model.

Material and methods

The submitted work is a preliminary study, including some assumptions and constraints:

- the diversity of the living bodies prohibits to intend a generic and exhaustive study. We restricted ourselves here to the study of the ovary, the reason being that one of our application lies in gynaecological surgery. The ovary constitutes one of the main target of the laparoscopic operations in gynaecology.

- obvious ethical reasons and greater convenience of experimentation led us to investigate isolated organs from animals. One can however underline the involved limits, because there are noticeable differences between in vivo and in vitro organs: in particular those relative to the absence of blood perfusion and to the conditions of temperature. We choose sow as animal model, the same as during the surgeons' trainings. The major difference in its anatomical characteristics with the human ones lies in the heterogeneity introduced by the presence of multiple folliculi.
- the available testing bench having no control on the motion's speed, all measurements were made in a static way (quasi steady-states).
- lastly, we considered that the operative instrument interacting with the deformable body is a rigid object. That makes it possible to consider that the force felt on the operator's hand (force which has to be simulated) is identical to the force (really measured) in contact with the body. Anatomical knowledge on the ovary, as well as the observation of peroperative records make us regard this organ as an ovoid body, with a large axis of approximately 4 cm, a small axis of approximately 2 cm, and weighing approximately 6 to 8 g. It is in relation to the neighbouring structures via two main ligamenta (cf fig 1).

1 : round ligament

2 : own ligament

5 : ovary

8 : suspensory ligament

9 : fallopian tube

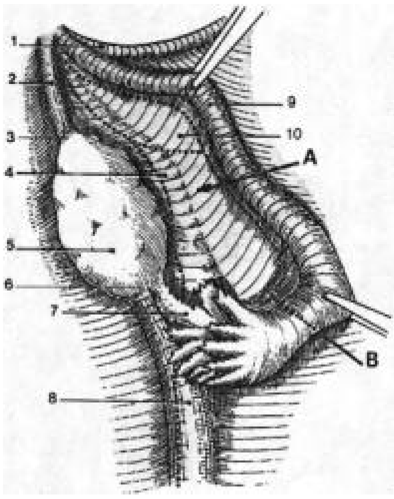

Figure 1: Diagram of the human ovary and its anatomical insertions [17]

The operative instrument consists of a grip located at the end of a 30 cm long stem, threaded itself in a trocar to cross the abdominal wall. The analysis of the movements of this instrument during a surgical operation allows to completely describe them in reference to the cutaneous plan (fig. 2) by means of 5 degrees of freedom.

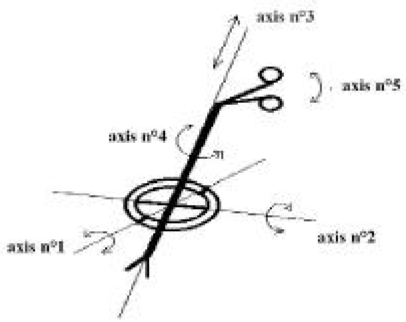

Figure 2: Degrees of freedom of the grip.

Thus maintained by its ligamenta and handled by such an instrument, the ovary is capable of moving (translations, torsions, rocking motions) and of becoming warped. By use of the superposition's principle, we assumed that the total behaviour of this body can, during this kind of interaction, be fully described by the addition of the behaviour of a rigid and mobile object and of the behaviour of a deformable and fixed object. In this work, we were only dealed with the deformation process.
We thus studied the uniaxial pushing in of an indenter vertically moved over the ovary lying on a fixed plate. The ovary was put on gauze to avoid any side slip and to keep constant the axis of pushing in. To prevent the indenter sticking to the ovary during the moving up, this indenter was always covered with chalk.
One of the main difficulties was to measure low forces (of about the hundred mN) and displacements (of about the mm). We used a strain gauge force sensor (model GM3, Scaime™; its effective range is from 0 to 560 mN and its resolution of 1mN). An endless potentiometric position encoder (Bourns™) was coupled with the transmission screw of a micrometric XY stage (effective range = 25 mm, resolution of 60 µm) holding the force sensor. The position and force values were coded on 12 bits and recorded simultaneously on PC by means of an analog/digital acquisition card (RTI 815, Analog Device™).
In agreement with the literature, we used a mechanical model a priori viscoelastic (generalised Maxwell model), made up of (fig. 3) several cells of different viscoelasticity (springs k_i, shock absorbers c_i) laid out in parallel. This model includes a purely elastic branch (spring k_{el}), for which can be written the relationship:

$$F = k_{el} X, \qquad (2)$$

F being the loading and X the displacement

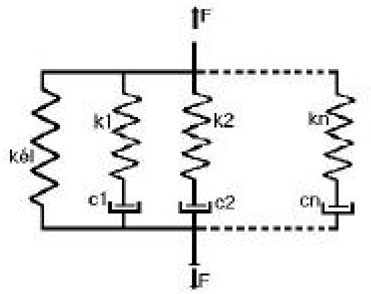

Figure 3: diagram of the general mechanical model.
The general differential equation of the model can be written:

$$F = k_{el} X + \sum_{i=1}^{n} \left[ci \frac{dX}{dt} - ci \frac{d(\frac{Fi}{ki})}{dt} \right] \qquad (3)$$

The quantification of the parameters of this model will be obtained by use of the two modes of experiments classically performed:

- loading/ unloading experiments, by increasing then decreasing pushing in: they characterize the elastic properties by relationships loading = function (displacement).
- relaxation experiments, obtained with a constant pushing in: by relationships loading = function (time), they make it possible to determine all its viscoelastic properties.

Results

1.1 Loading/ unloading tests

Usually, the tests are carried out at sufficiently low speeds of pushing in so that viscous damping can be neglected. in which the pushing in of the indenter was carried out by the manual rotation of the micrometric sliding gauge. These experimental conditions make difficult a precise control of the speed. However, under this assumed low speed condition, one can then consider that the total elastic behaviour (k) results from the contribution of all *n* springs of the model, that is to say:

$$k = \sum_{i=1}^{n} k_i \tag{4}$$

A hundred tests were carried out. The obtained curves show several results. First, the loading/displacement relationship is strongly non-linear. Then, exhibited elasticity varies a lot with speed. Lastly, every loading/unloading curve shows an hysteresis phenomenon illustrated by a shift between its increasing and decreasing parts. The presence of a final offset expresses the fact that the body can remain deformed (plasticity).

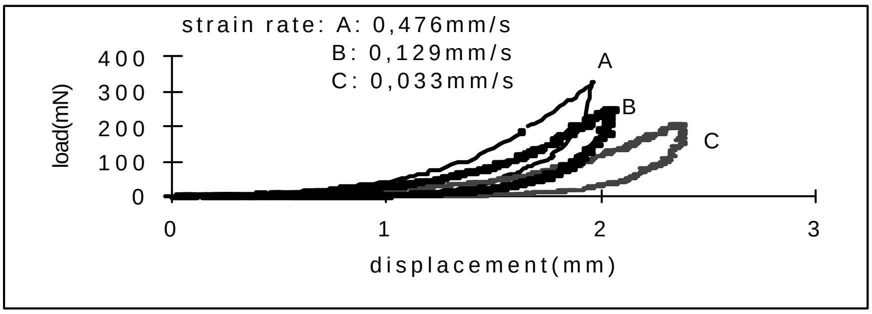

Figure 4 : Loading/unloading curves. These tests were performed at the same location of an ovary at different constant strain rate ; the maximum deformations corresponding to curves A, B, C are respectively 13.73%, 16.07% and 21.56%.

The curves can be fitted either by an exponential function,

$$F = k_e(e^{aX} - 1)$$

or a power function

$$F = k_p X^n$$

The identification of the parameters of each two models was done while taking into account:

- the quality of the adjustment of the model: measured by a distance (least squares) to the experimental values, it appears to be comparable for the two models.
- the dispersion of the identified parameters is illustrated by table n° 1. The performances of the two models look also greatly comparable here.

Table 1: Identified coefficients of the loading and unloading models

Loading process						Unloading process					
Power function			Exponential function			Power function			Exponential function		
	Mean	Std dev		Mean	Std dev		Mean	Std dev		Mean	Std dev
k_p	11.4	5.6	k_e	4.8	2.45	k_p	69.2	44.3	k_e	1.68	0.74
n	2.6	0.44	a	1.6	1.23	n	3.43	0.48	a	4.36	4.18

Having arbitrarily retained the power function model, the expression of pure elasticity becomes the following:

$$F = [k_p X^{n-1}]\, X \tag{5}$$

that is to say:

$$\sum_{i=1}^{n} k_i = k_p X^{n-1} \tag{6}$$

1.2 Relaxation tests

These tests consist in observing the temporal decrease of the force when the body is put through a constant deformation (figure 5). This decrease takes a multiexponential form which can be explained by the Maxwell model that we considered. We restricted the observation of the curves to a duration of about 3 mn, because the true gestures indeed never remain in contact with the organs during a so long time.

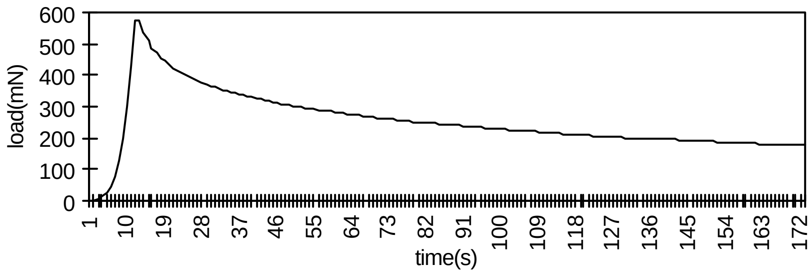

Figure 5 : An example of relaxation curve (deformation was set at 22.5%)

Under these conditions, it is possible to reduce to two the number of adequate cells: a purely elastic branch (k_{el}) and a viscoelastic branch (k_1, c_1). the differential equation (3) of the model can be simplified:

$$F = k_{él} X + c_1 \frac{dX}{dt} - c_1 \frac{d\frac{F_1}{k_1}}{dt} \tag{7}$$

associated with the other equation coming from the loading/unloading tests:

$$k_{él}(X)+k_1(X)=k_pX^{n-1} \tag{8}$$

The corresponding reduced model can then be drawned according the figure 6:

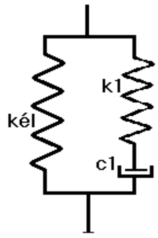

Figure 6: Reduced Maxwell for the isolated ovary

We started the identification of the parameters by imposing various values of pushing in at a same location of a given ovary, by keeping the speed as constant as possible. The curves were plotted for 8 values of pushing in, ranging from 1,25mm to 3mm. An average curve was calculated and compared point by point with each of the 8 series of measurements. Table 2 gives the values of the averages of the differences (in %) between each series of points and the average.

Table 2: Measurements' variability

Pushing in (in mm)	Mean variation (in %)
1.25	8.1
1.5	6.2
1.75	4
2	5.4
2.25	2.5
2.5	7.6
2.75	1.8
3	3.3

From this constancy of the temporal decrease, we deduced constancy of the viscosity coefficient (c_1) and thus of the associated elasticity (k_1). This brings a simplified expression of the equation (8):

$$kél(X)=k_pX^{n-1}-k_1 \tag{9}$$

By applying the deformation at 5 different locations of the ovary, we could check a good reproducibility (coefficient of variation lower than 10%) for 7 different ovaries. However, the variability inter ovaries is more significant and the dispersion of the average values obtained on the 7 different ovaries reaches 20%.

The values measured at late time (1000 s) give an estimation of the purely elastic component (k_{el}), thanks to the relationship:

$$F(t_\infty)=k_{él}(X_0)X_0 \tag{10}$$

, where X_0 is the imposed pushing in.

The identification of the parameters gives:

$k_1 = 61.4 \pm 28.8$ Nm^{-1} $c_1 = 652 \pm 478$ Nm^{-1} s $k_p = 11.4 \pm 5.6$

Discussion

The full description of the mechanical behaviour of biological organs is very complex. For our part, our major interest lies in the analytical description of the interactions occurring between the (rigid) surgical tool and the (very deformable) organ. That is the reason why we only kept the initial (fast-dropping) part of the relaxation curves : surgical gestures rarely extend beyond few seconds. This allows us to keep a simplified model (Maxwell model) with only one elasticity and one viscoelasticity.
Our investigations exhibited the influence of the speed of the interaction and of the heterogeneity of the ovarian structure. This first study outlined a great dispersion of some numerical values, often about 50% and sometimes near to 100%. This dispersion can be partly imputed to the natural biological dispersion. Consequently it could be introduced in an adjustable model.
Our further works will investigate the dynamic behaviour of isolated organs, with the help of a more sophisticated testing bench including the control of the indenter‘s speed and more precise measurements of lower forces and of higher speeds. A test machine will be designed to make multiaxial studies in compression and stretching. One particular problem will be to check the existence of the observed phenomenon of plasticy.
This work will also be carried on with dynamical studies on living organs. That is the reason why we are about to achieve a specific measurement tool : built from a true surgical grip fitted out with position and force sensors, this instrument will afford to perform spatio-temporal analysis of usual gestures under real operative conditions.

Aknowledgement

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