Sensors and devices to enhance the performances of a minimally invasive surgery tool for replicating surgeon's haptic perception of the manipulated tissues

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Abstract

Minimally invasive surgery is afflicted with important limits related to tactile uncoupling between the surgeon's hand and the tissues manipulated by surgical tool. Our work suggests a solution for this problem. We designed a sensorization system in order to acquire information about the force exerted on the tissues and the deformation induced. We elaborated these information through an identification algorithm to extract typical materials parameters. We applied a control law using evaluated parameters for replicating rheological dynamic behaviour of surgical tissues by means of a tactile display. In this paper we describe the functioning of the sensor and devices utilized, and the experimental results.

1 Introduction

Laparoscopic surgery is a technique, alternative to open surgery, where the surgeon operates through small openings (between 3 and 12 mm) in the abdominal wall of the patient using a set of elongated slim rigid tools. In recent years this technique underwent a strong development because it offers a lot of advantages: reduction of risks, disfigurement, and patient pain, shorter immobilization (about 24 hours), shorter hospitalization (about 2-24 hours) and an earlier return at work (about 7 days). These advantages may be translated into a total health care cost reduction for commercial and governmental institutions as well as for the patient [2]. Nevertheless, there are still many problems related to the use of this technique. One of the most important is the surgeon loss of both tactile and kinesthetic sensibility due to the trasmission mechanism of the elongated tools. The surgeon can only avail himself of the aid of images acquired by an optical fiber videocamera inserted in the abdominal cavity in order to execute the operation without his touch sense. This limitation restricts the application of this technique only to some specific fields. Our work aims to solve this problem by planning a suitable sensorization system able to take signals from the force exerted on the tissues and the related deformation induced. There are two ways of using these signals: the first is to visualize during the manipulation, the stress-strain graphic and the typical rheological parameters of the surgical tissues on a computer monitor, the second is to elaborate the signals through an identification algorithm, to submit the result to the tactile display, and drive it trying to replicate rheological behaviour of the tissues. The first possibility is not completely satisfactory, but is the first step for dealing with this problem.

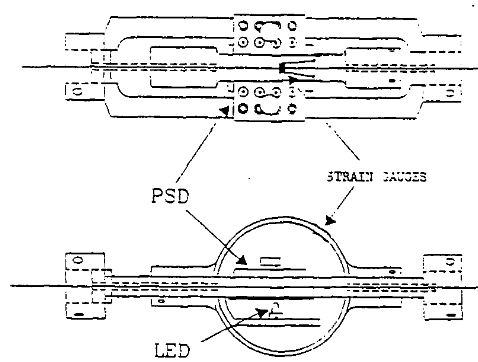

Figure 1: Sensorization system applied on the surgical tool.

2 Sensors applied to the surgical tool

The laparoscopic forceps has a very simple mechanical structure: a rigid beam is actuated by the handle. Its forward-backward movement closes and opens the jaws. The goal of our work was to measure the force applied on the manipulated tissues and the deformation exerted. The force sensor (see Fig. 1) is realized applying two strain gauges to opposite side of an alluminum ring fixed to the rigid beam. The force applied on the jaws of the la-

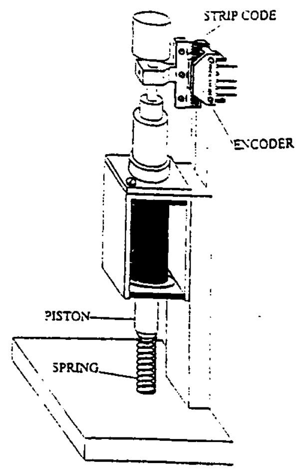

Figure 2: Structure of the haptic display.

paroscopic tool causes the ring's deformation that causes gauges resistance variation. The tissues deformation can be approximated by the change in the position of the jaws and it was measured by using an optical device on which a light emitting diode (LED) was placed. The LED is integral with the rigid beam and its forward-backward movement is followed by the change in the position of the light spot on the optical device. A linear relationship exist between the jaws angular variation and the optical device output. An 8 bit microcontroller (TMS370C756) with 8 A/D inputs, an asynchronous and a synchronous serial port, three 8 bit input-output digital ports was used to acquire the sensors signals. The microcontroller was programmed to acquire the sensors signals, and to process and send them to a PC by means of the asynchronous serial port. Rheological data were shown directly on the laparoscopic monitor or on a separate display.

3 Actuator for replicating rheological behaviour of manipulated tissues

The visual display gives a great aid to discriminate the anatomical nature of the manipulated tissues, but is not completely satisfactory, because the surgeon needs tactile perception to evaluate without hesitation. Motivated by this, we have realized an haptic display (see Fig. 2) designed to returning the surgeon with the haptic sensation of the compliance of manipulated tissues. The display is comprised of a linear actuator, a spring, and a linear position encoder. The actuator is realized by a piston fitting into a solenoid, being its vertical stroke limited by the spring. The presence of the spring is motivated by electrical reasons, because the solenoid needs a offset current to keep its equilibrium position and it causes overheating problems. The inputs of the haptic display are the force exerted on the piston and the current flowing into the solenoid. The outputs is the variation of the piston position with respect to the equilibrium position. The operator force is opposed by the spring's force and by the piston inertia, while the magnetic force generated by the current flowing into the solenoid concurs to modulate the virtual stiffness of the piston to make it equal to that of the tissues we want to simulate. The first step is to characterize the solenoid.

The magnetic energy stored in the system is given by

$$U = \frac{1}{2} L I \tag{1}$$

and the magnetic force is (taking downward vertical orientation for the x axis)

$$F_m = \frac{dU}{dx} \tag{2}$$

Using an inductance meter we measured experimentally the inductance of the solenoid obtaining a linear relationship between the piston position and the inductance (infact the inductance mainly depends on the quantity of ferromagnetic material inside the solenoid). By this, we can write

$$L = l_0(\mu_b b x + \mu_a a(l - x)), \tag{3}$$

where l is the length of the solenoid, a is the section of ferromagnetic part of the piston (with permeability μ_a), b is the area of non-ferromagnetic part (with permeability μb) (see Fig. 3).

Therefore, deriving

$$\frac{dL}{dx} = l_0(\mu_b b - \mu_a a) = constant = -2\alpha < 0. \tag{4}$$

The term $\frac{dL}{dx}$ is constant and negative (since $\mu_a >> \mu_b$ and $a > b$), hence F_m is always directed upwards, and its intensity is

$$F_m = -\alpha I^2. \tag{5}$$

In these assumptions the magnetic force does not depend on the piston position, and it changes with the square of the current. This type of behaviour was validated experimentally by inductance measurements.

The dynamics of the display's piston are easily derived as

$$M_p \ddot{x} + K_p x + \alpha I^2 = F_{extp} \tag{6}$$

In our prototype construction, we have used a transconductive amplifier of unitary gain such that the solenoid current I can be replaced by the amplifier input voltage V in eq. 6. In order to obtain numerical data for the constants appearing in eq. 6 we weighted the piston mass with an electronic gauge ($M_p = 83.35gr$). To calculate

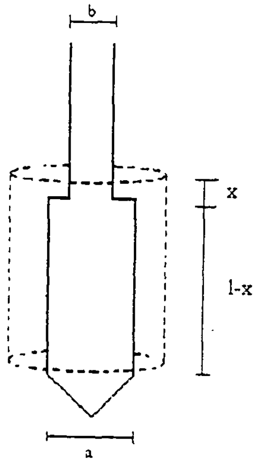

Figure 3: The size of the piston fitting into the solenoid.

the other two parameters (K_p, α) we have realized experimental tests by varying the piston position with a micrometric screw from 0 to 15 mm, and the solenoid input voltage from 0 to 1,5 V.

Average parameters, after several trials, were obtained as

$$K_p = 27(grp/mm)$$
$$\alpha = 12e - 5(grp/(mV)^2)$$

4 Replication of the dynamic behaviour of the material and experimental results

To identify rheological parameters of the tissues we applied a simple pseudoinverse algorithm. We used a nonlinear model of the material in the form

$$K f_0(x(t)) + S f_1(\dot{x}(t)) + M\ddot{x}(t) = F_{extm}(t) \qquad (7)$$

where K, S, M are three typical parameters of the material to be determined, and f_0 and f_1, are functions of the displacement and velocity of the material specimen, respectively. f_0 and f_1 are assumed to be nonlinear functions of known type, common in the rheology of materials, e.g. $f_0(x) = x$ or $f_0(x) = x^3$ for stiffness, and $f_1(\dot{x}) = \dot{x}$ or $f_1(\dot{x}) = \dot{x}^2$ for material damping. We made an acquisition using the sensorized surgical tool at n+1 consecutive sampling instants, building a system of n+1 equations. The vector of the three parameters was obtained evaluating pseudoinverse matrix:

$$\mathbf{p}_i = \mathbf{A}_i^+ \mathbf{F}_{extm}.$$

We repeated this operation m times obtaining m parameters vectors and we chose the best fitting vector based on comparison of residuals:

$$\mathbf{p} = \begin{bmatrix} K \\ S \\ M \end{bmatrix}.$$

These parameters were inserted in an algorithm using inverse model technique. Given the display's model in eq. 6, where the solenoid current I is replaced by the amplifier input voltage V, and considering for simplicity a material with linear model

$$M_m \ddot{x}_m + S_m \dot{x}_m + K_m x_m = F_{extm},$$

we considered the displacement error between the tissue specimen and the display as

$$e = x_p - x_m.$$

Our goal was to impose a control law on the solenoid voltage V such that, being the same force applied to the specimen and the display, also the deformation of the display should track that of the specimen, i.e. the error should converge to zero.

Imposing

$$F_{extp} = F_{extm} = F_{ext},$$

we obtained

$$M_p \ddot{x}_p - M_m \ddot{x}_m - S_m \dot{x}_m + K_p x_p - K_m x_m = -\alpha V^2.$$

By some algebraic manipulations, we had

$$M_m \ddot{e} + S_m \dot{e} + K_m e = -\alpha V^2 - (M_p - M_m)\ddot{x}_p + S_m \dot{x}_p - (K_p - K_m)x_p \qquad (8)$$

By choosing the control law

$$V = \sqrt{\frac{S_m \dot{x}_p + (K_m - K_p)x_p + C_v \dot{e} + C_p e}{\alpha}} \qquad (9)$$

from eq. 8 we obtained

$$M_m \ddot{e} + (S_m + Cv)\dot{e} + (K_m + C_p)e = (M_m - M_p)\ddot{x}_p.$$

By choosing suitable positive constants C_p and C_v, the model tracking error e can converge asymptotically with desired dynamics to within a ball of radius proportional to the maximum deformation acceleration $||\ddot{x}_p||$. As in most cases the acceleration is very small, converge of the tracking error to zero can be practically assured. In order to evaluate the quality of the system we manipulated a specimen material and we plotted the Figure 4, where the displacement imposed by the user on the piston is compared with replication through the identification of the parameters of the material.

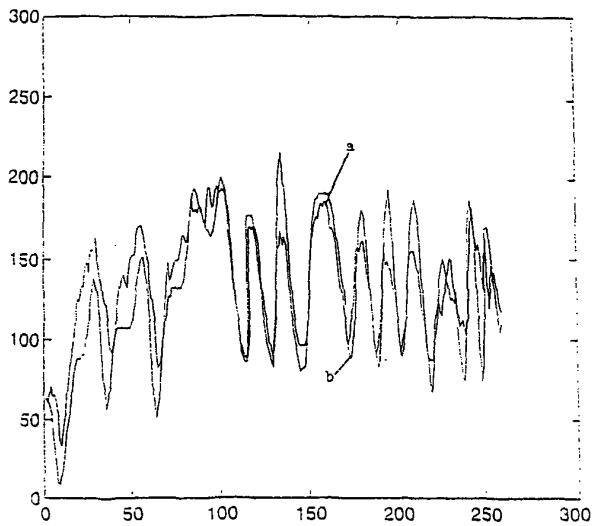

Figure 4: Comparison between the displacement exerted by the user on the piston (a) and that evaluated by identification algorithm of the material (b).

5 Conclusions

The results are encouraging, but there are still some limits due to the off-line nature of acquisitions and to the physical size of the haptic display, which needs to be miniaturized in order to be viable as an aid to the surgeon in the operating room. In the future we aim to extend the range of the materials to be simulated to biological tissues wich have a strongly nonlinear viscoelastic behaviour.

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