A MAGNETORHEOLOGICAL FLUID AS A HAPTIC DISPLAY TO REPLICATE PERCEIVED BIOLOGICAL TISSUES COMPLIANCE

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Abstract – **In this paper we present an innovative application of magnetorheological fluids. These materials are a suspension a micron-sized, magnetizable particles in a synthetic oil. Exposure to an external magnetic field induces in the fluid a modification in rheological behaviour changing it into a near-solid in few milliseconds. Just as quickly, the fluid can be returned to its liquid state with the removal of the field. MR fluids are already present on the market, used in devices such as valves, brakes, clutches, dampers. We report here about the possibility to use MR fluids to mimic the compressional compliance of biological tissues in order to realize a haptic display for surgical training in minimally invasive surgery applications.**

I. INTRODUCTION

Magnetorheological (MR) fluids are materials that respond to an applied magnetic field with a change in rheological behaviour. Typically, this change is manifested by the development of a yield stress that monotonically increases with applied field. Just as quickly, the fluid can be returned to its liquid state with the removal of the field. MR fluids are already present on the market but their application field is restricted to devices such as valves, brakes, clutches, dampers.

We report here about the possibility to use MR fluids to mimic the compressional compliance of biological tissues in order to realize a haptic display for surgical training in minimally invasive surgery applications [1][2]. This approach is justified by the observation that biological tissues viscoelastic properties could be mimicked by magnetically tuning the rhelogical properties of a MR fluid incorporated in the handle of a surgical end-effector.

Minimally invasive surgery is a technique developed to reduce the traumatic effect of some surgical operation, which underwent a dramatic development in recent years [3]. The reason of such a fast growth are know: 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 [4].

Nevertheless, the minimally invasive surgery is still afflicted with important limitations. 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 may manipulate patient's viscera only using long tools, observing actions and movements on a monitor visualizing abdominal environment. He can not either touch or see viscera directly and that restricts the application of this technique only to some specific fields. Diminished tactile sensibility causes a loss of surgeon palpation capability, in particular with regard to tissues compliance and viscosity.

Present work aim to solve this problem providing a haptic display able to mimic the compliance of biological tissues manipulated by surgical tools.

In the first phase of the work we experimentally characterized the MR fluid by performing indentation tests on a confined sample by means of a compressional indentors driven by an electromagnetic shaker.

We focused our attention on stress relaxation tests in order to formulate a suitable descriptive model of the MR fluid (MRF-132LD by Lord Corporation® , Cary, NC, Usa). The model parameters have been estimated through best fitting between the experimental data acquired by means of stress relaxation tests and the theoretical ones. Best fit was obtained using a generalized Kelvin model of the second order, which is composed by three springs and two dampers.

In order to justify our approach we also performed a set of psychophysical tests on a group of 20 voluntary human subjects. We selected six specimens of bovine biological tissues: brain, myocardium, spleen, liver, lower limb muscle and lung. Each specimen was cut to the same dimensions, and therefore same volume, of the MR confined sample. Then, volunteers were asked to simultaneously assess, by touch, the compressional compliance of a biological tissue sample, different each time, and the one of the MR fluid driven by a suitable intensity of the magnetic field. Volunteers were required to tune (through instructions given to an assistant) the value of the magnetic field till they were subjectively satisfied with the degree of resemblance of the perception originated from both specimens. The consequent value was recorded. For every biological tissue, the average values of intensity of the magnetic field established by the 20 subjects have been calculated. Then, we performed tests of stress relaxation on the specimens of biological tissues and we used these data to identify, through mathematical fitting, a Kelvin model of the second order describing the behaviour of biological tissues.

The final test we performed was the stress relaxation on MR fluid excited with the values of the magnetic field deriving from the psychophysical tests. From the comparison of these curves with those ones relative to biological tissues we can state that for liver, spleen and brain we have a good agreement. For myocardium, muscle and lung, instead, the agreement in not satisfactory. This is due to the fact that the magnetic field intensity needed to induce a compliance similar to these biological tissues, is beyond saturation of the MR fluid we used.

II. MR FLUIDS

Magnetorheological fluids are materials that undergo a change in rheological behaviour if an external magnetic field is applied. They consist of a suspension of a micron-sized, magnetiziable particles in a synthetic oil. In the absence of an applied field MR fluids exhibit a Newtonian-like behaviour, whereas with the application of magnetic field they develop a precisely controllable yield stress [5][6]. In order to better comprise the mechanism of functioning let us assume the fluid within a gap created between two ferrous plates of a device. In the absence of an applied magnetic field, the fluid flows freely through the gap, but upon application of a magnetic field the particles within the gap become magnetized and align themselves with the direction of the field. Particles chains restricting the movement of the fluid are created (see Fig. 1). By increasing or decreasing the intensity of the magnetic field we are able to alter the interparticle attraction and continuously control the rheological properties. Exposure to a magnetic field, however, can transform the fluid into a near-solid in milliseconds. Just as quickly, the fluid can be returned to its liquid state with the removal of the field. The degree of change in a MR fluid is proportional to the magnitude of the applied magnetic field. The behaviour of controllable fluids is often represented as a Bingham plastic having a variable yeld strength. In this model, the flow is governed by Bingham's equations:

$$\tau = \tau_0(H) + \mu \frac{dv}{dy} \qquad (1)$$

where τ_0 is yeld stress, μ is the viscosity and dv/dy is the fluid share rate.

III. EXPERIMENTAL APPARATUS

In order to apply a magnetic field on the MR fluid specimen we designed and built an electromagnet. Design process steps were to design a low reluctance steel flux conduit to guide and focus magnetic flux into region of MR fluid ; maximize the magnetic field energy in this region and minimizing the energy lost in the other regions. To do this we positioned MR fluid in the air-gap of an electromagnet with three coils in series composed by 2400 turns of copper wire with 0,8 mm in diameter, such that the maximum current flowing in it is 1,26 A sufficient to produce magnetic field of our interest.

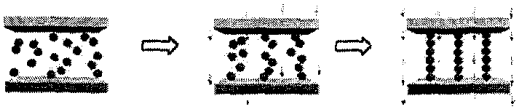

Fig. 1. Internal configuration of the MR fluid in the absence of the magnetic field (left figure), in intermediate configuration (middle figure) and with a magnetic field applied (right figure)

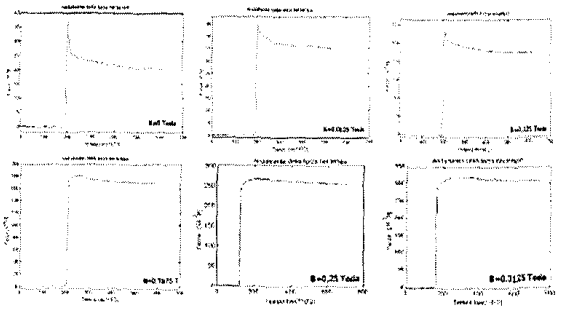

Fig. 2. MR fluid response to step function of deformation by magnitude 2,5 mm for increasing magnetic field up to MR phenomenon saturation.

IV. EXPERIMENTAL RESULTS

A. Model resolution and identification

Our goal was to identify a model able to describe MR fluid rheological behaviour. To do this several indentation tests on a MR specimen fluid are performed by using a compressional indentor. Since MR fluids exhibit viscoelastic features, our attention was focused on techniques suited to emphasize these properties.

In literature are referred two experimental tests called stress relaxation and creep in order to evaluate viscoelastic features. In the former the specimen is suddenly strained with the strain maintained constant afterward and the corresponding stress, decreasing with the time, is observed. In the latter the specimen is subjected to a force step maintained constant in a time lapse and the induced deformation is recorded.

We only focused our attention on stress relaxation tests because the instrumentation used better fits to this kind of experiments; nevertheless they are enough to elaborate a suitable descriptive model.

The first step was to apply on the MR specimen, submitted to increasing magnetic field, a strain step function with a given magnitude and to acquire the relative stress relaxation curve.

Tests have been repeated many times in order to confirm the repetitiveness of the phenomenon. The deformation imposed was 2,5 mm, while the application time (about 10 sec) has been chosen greater than the decay time of the phenomenon. In Fig. 2 are reported graphics relative to MR fluid response to a step function of deformation by magnitude 2,5 mm for increasing magnetic field up to MR phenomenon saturation.

Observing figures we can notice that the force value ranges from few thousandths of Newton in absence of

magnetic field to about 5,4 Newton when the MR phenomenon is already saturated (maximum intensity of the

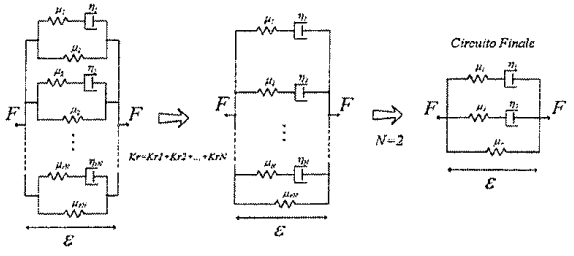

Fig. 3. Final generalized Kelvin model.

magnetic induction vector). Furthermore, for magnetic fields greater than 0,25 Tesla the MR fluid specimen does not show a significant stress relaxation.

For this reason graphs the magnetic fields range of our interest is restricted to 0-0,3 Tesla, where differences in the fluid behaviour are more pronounced. The magnetic field incremental step imposed on the specimen in order to evaluate the descriptive model was equal to dB=0,03125 Tesla, corresponding to an increment of the coil current of 0,05 A.

The next step was to elaborate a general viscoelastic model able to describe the experimental behaviour previously acquired.

Graphics reported in Fig. 2 look alike to stress relaxation curves relative to Kelvin model. In fact here are present an initial peak, a relaxation in the time and a final steady value as well as in the Kelvin model behaviour. A generalized Kelvin model of order n is comprised of n Kelvin bodies in parallel configuration. By comparing the experimental curves of stress relaxation with those of the theoretical model, we ascertained that an order equal to 2 was sufficient to describe experimental curves (see Fig.3).

This system is described by the differential equation of the II order:

$$F(t)+a_1\frac{dF(t)}{dt}+a_2\frac{d^2F(t)}{dt^2}=b_0\varepsilon(t)+b_1\frac{d\varepsilon(t)}{dt}+b_2\frac{d^2\varepsilon(t)}{dt^2} \qquad (2)$$

where F(t) is the input force to Kelvin model, e(t) is output deformation from model and coefficients a_1, a_2, b_0, b_1, b_2, c_0 are equal to:

$$a_1=\frac{K_1\eta_2+K_2\eta_1}{K_1K_2}$$

$$a_2=\frac{\eta_1\eta_2}{K_1K_2}$$

$$b_0=K_r$$

$$b_1=\frac{K_2\eta_2(K_r+K_2)+K_2\eta_1(K_r+K_1)}{K_1K_2}$$

$$b_2=\frac{(K_r+K_1+K_2)\eta_1\eta_2}{K_1K_2}$$

Applying the Laplace transform to differential equation we obtain:

$$\Sigma(s)=K(s)E(s) \qquad (3)$$

where $\Sigma(s)$ e $E(s)$ are the L-transform of F and ε, whereas $K(s)$ is equal to:

$$K(s)=K_r+\frac{K_1\eta_1 s}{K_1+\eta_1 s}+\frac{K_2\eta_2 s}{K_2+\eta_2 s} \qquad (4)$$

$K(s)$ can be explained as a stress-strain transfer function and expressed as a rational function with two zeros and two poles:

$$K(s)=A\frac{(s+s_{o1})(s+s_{o2})}{(s+s_{p1})(s+s_{p2})} \qquad (5)$$

where:

$$A=K_r+K_1+K_2$$

$$s_{p1}=\frac{K_1}{\eta_1}$$

$$s_{p2}=\frac{K_2}{\eta_2}$$

$$s_{o1}+s_{o2}=\frac{1}{K_1+K_2+K_3}\left[\frac{K_1}{\eta_1}(K_r+K_2)+\frac{K_2}{\eta_2}(K_r+K_1)\right]$$

$$s_{o1}\cdot s_{o2}=\frac{K_rK_1K_2}{K_1+K_2+K_3}\frac{1}{\eta_1\eta_2}$$

To identify the model means to determine numerical values of the lumped parameters. In our case we have three elastic parameters and two viscous parameters. Such values are determined by the comparison between the theoretical resolution of the model and the corresponding experimental data.

B. *Parameters determination*

Let us solve our differential equation. The applied deformation is a step function at the instant t=0+:

$$\varepsilon(t)=\varepsilon_o\cdot u(t) \xrightarrow{\ L\ } E(s)=\frac{\varepsilon_o}{s}$$

In the Laplace domain we have:

$$\Sigma(s)=K(s)\frac{\varepsilon_o}{s} \qquad (6)$$

By replacing $K(s)$ expression and expanding in partial fractions we obtain:

$$\Sigma(s)=\frac{K_r\varepsilon_o}{s}-(K_1+K_2)\varepsilon_o\left(\frac{c_1}{s+K_1/\eta_1}+\frac{c_2}{s+K_2/\eta_2}\right) \qquad (7)$$

where residuals c_1 and c_2 depend on the model parameters and their sum is equal to unity $c_1 + c_2 = 1$.

In the time domain the (7) can be written as:

$$F(t)=F_\infty+(F_o-F_\infty)\left[c_1\exp\left(-\frac{K_1}{\eta_1 t}\right)+c_2\exp\left(-\frac{K_2}{\eta_2 t}\right)\right] \qquad (8)$$

where:

$$F_o=(K_r+K_1+K_2)\varepsilon_0$$
$$F_\infty=K_r\varepsilon_0$$
$$c_1=\frac{K_1}{K_1+K_2}$$
$$c_2=\frac{K_2}{K_1+K_2}$$

The previous equation can be written in a form more appropriate to numerical fitting:

$$F(t)=F_\infty+F_1\exp\left(-\frac{t}{\tau_1}\right)+F_2\exp\left(-\frac{t}{\tau_2}\right) \qquad (9)$$

where:

$$F_\infty=K_r\varepsilon_o$$
$$F_1=(F_o-F_\infty)\frac{K_1}{K_1+K_2}$$
$$F_2=(F_o-F_\infty)\frac{K_2}{K_1+K_2}$$
$$\tau_1=\eta_1/K_1$$
$$\tau_2=\eta_2/K_2$$

Therefore the force is a non-linear function depending on the time and on five model parameters. The best fitting between the theoretical curve F and the experimental one F^* is performed using the least-squares criterion, that is the minimization of the following error function:

$$H=\sum_{i=1}^{N}\left[F(t_i)-F^*(t_i)\right]^2 \qquad (10)$$

where N is the number of experimental samples. In Fig. 4 experimental results, with a zoom view on the part of curve relative to the relaxation behaviour, are reported.

We can show that the model reproduces the real MR fluid behavior with a good accuracy.

C. *Tests on biological tissues*

Our goal was to verify the possibility of using the MR fluid to simulate the compliance of some biological tissues.

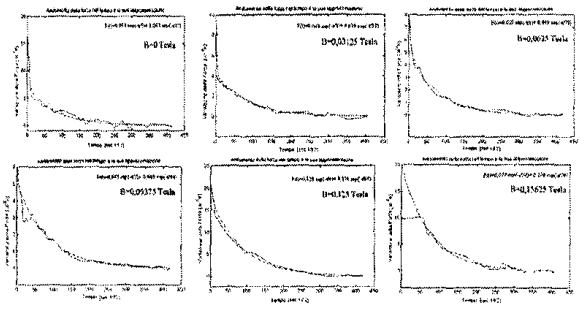

Fig. 4. Best fitting between experimental an theoretical curves with a zoom view on the part of the curve relative to the relaxation behabiour.

At first we proceeded with a psychophysical investigation. In this phase we selected six sample of bovine biological tissues: brain, myocardium, spleen, liver, lower limb muscle and lung. Each sample has been reduced to the same size of the MR fluid specimen in order to remove artefacts due to differences in test conditions. Volunteers were asked to manipulate at the same time using both hands the biological tissue sample and the MR fluid specimen duly excited with magnetic field. The magnetic field was changed on volunteers' suggestion till a good resemblance in compliance was attained.

When volunteers perceived the same tactile sensation from both biological tissue sample and MR fluid specimen, the corresponding magnetic field was recorded. Tests were repeated more times for each volunteer in order to calculate the average magnetic field necessary to excite the MR fluid specimen and to induce a compliance similar as much as possible to corresponding biological tissue sample.

In table I results are reported.

TABLE I
MAGNETIC FIELD TO BE APPLIED ON MR FLUID TO MIMIC CORRESPONDING BIOLOGICAL TISSUES

	|B| Tesla
Lung	0,34
Myocardium	0,62
Liver	0,132
Spleen	0,095
Muscle	0,44
Brain	0,033

The final step was to validate psychophysical results through the investigation about the biological tissues viscoelastic properties.

Using the same methodology, previously described, in order to identify a discrete descriptive model of the MR fluids, we performed stress relaxation tests on biological tissues samples. We ascertained that stress relaxation curves could be described by the same generalized Kelvin body of the second order whose parameters were evaluated through numerical fitting of the experimental data.

In this way we have a lumped model describing the six different biological tissues and a lumped model describing the MR fluid excited by different magnetic field.

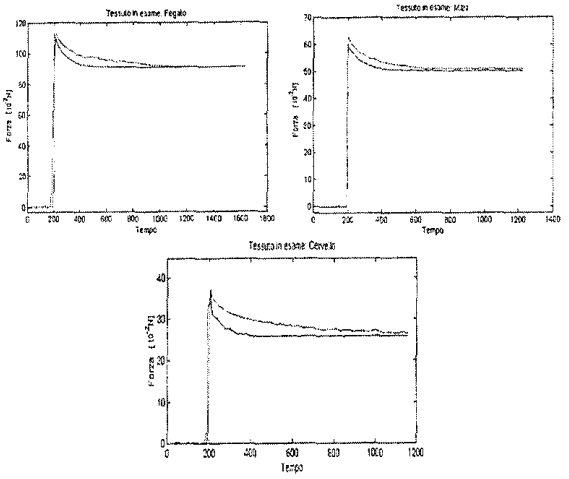

Fig. 5. Comparison between stress relaxation curves of biological tissues sample (liver, spleen and brain respectively) and MR fluid specimen.

The last assessment test was to calculate the MR fluid model parameters relative to the six magnetic field resulting by psychophysical tests and reported in table I and to evaluate the theoretical stress relaxation curves. We compared theoretical curves derived from calculated models of biological tissues and MR fluid specimen excited with the magnetic field estimated in psychophysical tests. Furthermore, to confirm this analogy in the behaviour we also compared the relative experimental curves. On the basis of these results we can affirm that with respect to liver, spleen and brain a good agreement between the behaviour of biological tissues and MR fluids was attained.

In each graphic of the Fig. 5 the higher curve is relative to stress relaxation acquired on biological tissue while the blue lower curve is relative to stress relaxation on MR fluid. The initial and final values are very near, while the relaxation times are different. In fact biological tissues exhibit relaxation times greater than those ones relative to MR fluid. Nevertheless the general behaviour is very similar and this is a confirmation of the results obtained with psychophysical tests. With respect to myocardium, lower limb muscle and lung the analogy was not satisfactory. This is due to the fact that the magnetic field intensity needed to induce a compliance similar to these biological tissues, is beyond saturation of the MR fluid we used.

IV. CONCLUSIONS

In this paper we performed several tests to investigate about MR fluid viscoelastic properties and evaluate the possibility to use MR fluid in order to mimic biological tissues compliance. At first we performed a detailed study in order to formulate a descriptive model of MR fluid through the traditional approach reported in literature to emphasize viscoelastic features. Then we evaluated a similar model for biological tissues and we performed a comparison between theoretical and experimental stress relaxation curves. Results are very encouraging and open new perspectives in realize a haptic display for surgical training in minimally invasive surgery applications. In particular biological tissues viscoelastic properties could be mimicked by magnetically tuning the rheological properties of a MR fluid incorporated in the handle of a surgical end-effector. Nevertheless some limitations are present due to the presence of a threshold on the biological tissues stiffness to be mimicked by MR fluids because the MR phenomenon exhibits saturation.

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