FORCE FEEDBACK-BASED MICROINSTRUMENT FOR MEASURING TISSUE PROPERTIES AND PULSE IN MICROSCUREGY

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Abstract
Miniaturized and “smart” instruments capable of characterizing the mechanical properties of tiny biological tissues are needed for research in biology, physiology and biomechanics, and can find very important clinical applications for diagnostics and minimally invasive surgery (MIS). We are developing a set of robotic microinstruments designed to augment the performance of the surgeon during MIS. These microtools are intended to restore (or even enhance) the finger palpation capabilities that the surgeon exploits to characterize tissue hardness and to measure pulsating vessels in traditional surgery, but that are substantially reduced in MIS.

This paper describes the main features and the performance of a prototype miniature robotic instrument consisting of a microfabricated microgripper, instrumented with semiconductor strain-gauges as force sensors. For the (in vitro) experiments reported in this paper, the microgripper is mounted on a workstation and teleoperated. An haptic interface provides force-feedback to the operator. We have demonstrated that the system can discriminate tiny skin samples based on their different elastic properties, and feel microvessels based on pulsating fluid flowing through them.

Keywords: tissue characterization, palpation, micromanipulation, microgrippers, force-feedback.

1. Introduction
Measuring the mechanical properties of cells and tiny biological tissues is important for research in biology, physiology and biomechanics. Cells of the body are exposed to mechanical stresses and strains throughout life, and this is critical to the health and functions of various tissues and organs of the body [1-2]. It is clear that microfabrication and microrobotics will provide very powerful means for the development of a novel generation of research instruments. In fact, a variety of laboratory apparatuses and microfabricated instruments have been developed recently for investigating cells and tissues properties by mechanical stimulation [3-5]. Miyazaki et al. [6] designed a micro tensile test system for cells and fine biological fibers and determined tensile properties of collagen fibers isolated from the patellar tendon. Sato et al. [7] measured the viscoelastic properties of cultured porcine aortic endothelial cells and analyzed them using a standard linear viscoelastic model. Wang and Coceani [8] set up an in vitro preparation of fetal lambs pulmonary arteries and veins to examine local factors responsible for hemodynamic control.

In addition to basic research in biomechanics, miniaturized robotic instruments are fundamental tools for minimally invasive diagnostics and surgery. Perhaps the most critical factor in MIS is the severe reduction of sensory and dexterous manipulation capabilities of the surgeon. Restoring (or even augmenting) these capabilities by developing new “smart” surgical tools would have a major impact on the future of the whole field of MIS.

Palpation is a procedure that surgeons exploit regularly and “naturally” in traditional surgery in order to estimate tissue hardness and to locate blood vessels hidden beneath opaque tissues. This is very important because the accidental puncturing of blood vessels during MIS is a rather frequent life threatening complication.

Research on robotics palpation has received some attention recently. A tactile array system aimed at finding hidden arteries has been proposed by Howe et al. [9], and a commercial tool for MIS has been instrumented in order to enhance the surgeon’s haptic perception of the manipulated tissue by Bicchi et al. [10].

The authors are investigating a new class of robotic microinstruments whose ultimate goal is to augment the performance of the surgeon during MIS. In previous papers we have presented some preliminary work on the fabrication, characterization and control of microgrippers for micromanipulation tasks [11-13]. We have also described some initial results on the use of microgrippers for characterizing soft tissues [14].

This paper discusses the main features and the performance of a prototype miniature robotic instrument consisting of a microfabricated microgripper, instrumented with semiconductor strain-gauges as force...
sensors. *In vitro* experiments have been carried out on the microgripper mounted on a workstation and teleoperated. An haptic interface provides force-feedback to the operator. The system can discriminate tiny skin samples based on their different elastic properties, and feel microvessels based on pulsating fluid flowing through them.

2. The Modular Workstation

The apparatus we developed for testing the microrobotic system is modular and teleoperated. It comprises: a microfabricated instrumented probe which can exert controllable force-displacement cycles on soft tissues and measure force generated by pulsating flow in microvessels; a 3 d.o.f. motorized manipulator which moves the microgripper; a fiber optic microscope (50-200X) with monitor which allows the operator to visualize the sample and the microgripper position; a PC-based control unit; and a haptic interface (Phantom 1.0, SensAble Technologies Inc.) which provides force-feedback to the operator. A scheme of the apparatus is illustrated in Figure 1.

![Figure 1 Scheme of the apparatus](image)

**Figure 1** Scheme of the apparatus

2.1 The instrumented microprobe

The microprobe selected as the end-effector of the testing apparatus is a LIGA-fabricated microgripper made out of electroplated nickel coated with a thin gold layer [11]. The geometry of the microgripper (overall length 17 mm, overall width 7.5 mm, thickness 0.4 mm) is showed in Figure 2.

The microgripper exploits flexure joints in order to generate large displacement at the fingertips in a compact structure, relatively easy to fabricate and assemble by microfabrication technologies. The microgripper is actuated by a low voltage multi-layer PZT stack (TOKIN AE0203D16), whose maximum driving voltage is 150 V, and maximum displacement is about 17 µm. The voltage is supplied by a power amplifier which receives an input signal directly from a PC through an A/D interface (DAQ AT-A06-10, National Instruments Inc.). When the actuator pushes the rear part of the microgripper, the flexure joint-based deformable structure amplifies the displacement, thus producing much larger displacement at the tip.

![Figure 2. Microgripper design](image)

**Figure 2. Microgripper design**

The microprobe is instrumented with semiconductor strain gauge sensors (ESU-025-1500 Entran Devices Inc.) [12]. In order to obtain accurate and repeatable measurements of force, the symmetry of the microgripper structure should be matched by a symmetrical configuration of the strain gauge sensors. A symmetrical configuration is also useful for thermal compensation and better signal to noise ratio [15]. The optimal symmetrical configuration consists of a full Wheatstone bridge sensor comprising four active strain gauges. The sensor can be used to implement a PI closed loop force control, as described in [12-13]. The four strain gauges have been mounted in two pairs on the microgripper, each pair located at a flexure joint: one strain gauge of the pair measures compression and the other measures tension. Figure 3 illustrates the approximate location of the strain gauges in the microgripper structure.

![Figure 3. Scheme of the microgripper showing the location of the strain gauge sensors](image)

**Figure 3. Scheme of the microgripper showing the location of the strain gauge sensors**

As the fingertips grasp a tissue sample, the strain gauges measure the deformation of the microprobe structure. After proper calibration, the output signal of the strain gauge bridge can be read as a force signal. To this purpose, the strain gauge sensor system was calibrated by opening the microgripper fingertip against a load cell (Model GM2 3M, PTC Electronics Inc.; full scale 300 mN, accuracy 0.01 mN). Calibration tests showed the good linearity of the strain gauge sensors and indicated that the microstructural deformation, and hence the force exerted by the micro gripper, can be monitored rather accurately using the Wheatstone bridge configuration.
The output signal of the instrumented microgripper in the force range 0-6 mN is illustrated in Figure 4. The force signal measured by the instrumented microprobe is delivered to the Phantom haptic system, thus providing force feedback to the human operator.

Figure 4. Calibration of the strain gauge sensor mounted on the microgripper

We have also developed a new version of microgripper, fabricated in superelastic alloy (Ni50.8Ti49.2) by wire micro Electro Discharge Machining (µEDM). µEDM allows to fabricate high aspect-ratio structures made out of different conducting and semiconducting materials with good surface finishing and without any thermal alterations even in the smallest features. Moreover, even when the above characteristics are not strictly required (in our design the aspect ratio is ~5), µEDM is an elective choice to machine hard materials which could not be machined by other technologies.

Superelastic alloy is the material of choice because of its favorable mechanical properties. This alloy allows to obtain a more robust and stiffer microprobe and to fabricate flexure joints which reach a large displacement amplification factor still within the elastic range of the material. A photograph of the new microgripper is shown in Figure 5.

2.2 The haptic interface

The system control is implemented by means of a PC, which interfaces the microgripper actuating and sensing circuits with the Phantom haptic interface. A graphic interface continuously displays the state of the system to the operator. The system is activated through the graphic interface and various modes of functionality can be selected.

For calibration purposes, the Phantom can be by-passed by retrieving the closure commands via a mouse or via an automated sine wave generator. In normal operation, the operator drives the actuators by the haptic interface and, at the same time, "feels" the grasping forces measured by the instrumented microprobe. By using this apparatus, different micro-samples of soft tissue have been tested and pulse in microvessels has been "felt". However, in order to extract quantitative parameters from the tested samples, signals must be processed. To this aim a method has been devised for the identification of the probe-sample system, as illustrated in the following paragraph.

3. Testing Methods and Experimental Results

While for research in biology and physiology accurate measurements of tissue mechanical properties are required, but on-line monitoring is not strictly necessary, most microsurgery tasks (e.g. the identification of microvessels embedded in the organ or tissue to operate) require an instrument which is able to provide some information to the operator in real time, even if the information is less accurate.

In order to evaluate the performance of the proposed microrobotic system in different measurement tasks, we have selected two experiments: one directed to test the elastic properties of micro samples of biological tissue, and the second to detect pulsating flow in microvessels. These experiments are described in the next two sections.

3.1 Testing the elastic properties of microsamples of biological tissue

Although the main intended application of the apparatus is for in vivo experiments in physiology and for microsurgery, in this phase we elected not to make tests with animals for ethical reasons. Instead, we used tiny samples (width about 100 µm) of human skin freshly excised from areas around the fingernails of three volunteers. In order to obtain information on the elastic properties of the microsamples, the following procedure was used: the microgripper grasped the sample and, whilst grasping, a small signal step voltage was applied to the piezo-actuator resulting in a sudden closure of the gripper. This results in small oscillations of the gripper fingertips at frequencies dependent on the resonant frequencies of the mechanical system (“gripper plus sample”). The step response of the system in idling
conditions and for the three different skin samples is shown in Figure 6.

![Figure 6](image)

**Figure 6.** Step response of the system in idling conditions (a) and when grasping three different skin samples (b, c, d).

As clearly visible in Figure 6, the step response signals are not sufficient per se to discriminate the different skin samples. Additional processing is needed to provide the apparatus with the required discrimination capabilities. The method we elected to use for analyzing the experimental data presented in Figure 6 is based on system identification [16]. The microgripper system was identified by means of a small step impulse excitation, filtering and Fast Fourier Transform (FFT) of the output signals. The experimental data were frequency analyzed using MATLAB 5.3 (The MathWorks, Inc.). Results are illustrated in Figure 7.

![Figure 7](image)

**Figure 7.** Bode plots of the frequency response of the gripper microsystem in idling conditions (blue line) and when grasping three different skin samples.

In idling conditions the gripper exhibits two main resonant frequencies. When the gripper grasps a sample, the fundamental mode of oscillation is blocked, the second resonant frequency is shifted depending on the mechanical properties of the sample, and the damping increases due to internal friction. This behavior can be observed in the Bode diagrams of Figure 7 which refer to the idling condition and to the cases when three different skin samples are grasped. In the system identification procedure, the resonant frequency and the fading constant of the poles relative to the secondary mode of oscillation are calculated for each system “gripper plus sample” starting from the acquired oscillating signals. Figure 8 illustrates the Bode diagrams related to the identified systems for the cases of idling condition and when the three different skin samples are grasped.

![Figure 8](image)

**Figure 8.** Bode plots of the identified systems. The signals corresponding to the gripper grasping three different tissue samples are much easier to discriminate.

In idling conditions, the following experimental transfer function was obtained:

\[
G(s) = K \cdot \frac{s^2}{s^2 + 2\zeta \rho_0 s + \rho_0} + \frac{s^2}{s^2 + 2\zeta \rho_1 s + \rho_1}
\]

with:

\[
K = 10^{-13}
\]

\[
\rho_0 = 195Hz, \quad \zeta = 0.00005
\]

\[
\rho_1 = 900Hz, \quad \zeta = 0.025
\]

When grasping skin samples, the following resonant frequencies were derived from the diagrams in Figure 8:

Skin sample 1: \( f = 690 \text{ Hz}, \quad \zeta = 0.1, \quad K = 0.2 \)
Skin sample 2: \( f = 800 \text{ Hz}, \quad \zeta = 0.25, \quad K = 0.2 \)
Skin sample 3: \( f = 820 \text{ Hz}, \quad \zeta = 0.04, \quad K = 0.2 \)
In order to extract information on the mechanical properties of each sample, the resonant frequency and the fading constant of the poles relative to the secondary mode of oscillation should be interpreted using models for the cases of the gripper alone (idling condition) and of the complete system “gripper plus sample”. The system “gripper and grasped sample” has been modeled as a 2 d.o.f. system and this simplified model has been used to calculate the first two resonating frequencies of the system [14]. The characteristic equations of this system have been obtained and the relative frequencies of the modes of oscillation have been calculated. Finally, the poles of the system have been calculated by resolving the following second order polynomial equation:

\[ P(\omega) = m_1 m_2 \omega^4 + \left[ m_2 (k_1 + k_2) + m_1 (k_2 + k_c) \right] \omega^2 + \left[ (k_1 + k_2)(k_2 + k_c) - k_c^2 \right] \]

where \( \omega \) is the resonant frequency of the system; \( k_1 \) and \( k_2 \) are the elastic constants of the microgripper (according to our 2 d.o.f. model); \( m_1 \) and \( m_2 \) are the characteristic masses of the system; \( k_c \) is the elastic constant of the sample.

The resonant frequencies \( \omega_i \) of the system “microgripper and grasped sample” obtained experimentally (see Figure 6) corresponds to the roots of the equation above. By knowing the value of \( k_1, k_2, m_1 \) and \( m_2 \) from mechanical simulations of the microgripper, the experimental resonant frequencies can be related to the elastic constant \( k_c \) of each sample.

### 3.2 Testing pulse in microvessels

The same system (connected to the haptic interface) was used to to measure and “feel” pulse in microvessels. For ethical reasons, microvessels were substituted by polymeric microtubes which simulate microvessels (for example those in the coronary tree) with good approximation. Pulsating blood flow was simulated by a simple microfluidic circuit comprising a Micro Annular Gear Pump (mzr-2903, Mikrosysteme GmbH) and polymeric microtubes (Zeus Scientific Inc), with an external diameter of 800 \( \mu \)m and an internal diameter of 500 \( \mu \)m. The micro pump is PC-controlled by a software interface that allows dosing and continuous flow of fluid with an accuracy of more than 99.5% on delivered volume. Experimental parameters such as acceleration time, maximum speed and time interval among dosing pulses are variable in a wide range. Physiological blood flow in different sites and conditions can be simulated quite accurately in the microfluidic circuit, and therefore the performance of the proposed microrobotic system can be evaluated in realistic conditions. Water with a flow rate of 40 \( \mu l \) and a pulsating frequency of 8.3 Hz (reproducing physiological parameters in coronary system of mice) was supplied through the microtube grasped by the instrumented microgripper, as illustrated in Figure 9.

![Figure 9. Microgripper grasping a microtube under microscope](image)

The pulse signal was measured by the strain gauge sensor and also “felt” distinctly by the operator through the Phantom interface. Figure 10 shows the recorded strain gauge signal and the operator hand who “feels” the pulse signal through the haptic interface.

![Figure 10. Pulsating microtube signal and haptic interface sensing](image)

### 4. Conclusion and Future Work

In this paper we have presented a technique suitable for measuring \textit{in vivo} the mechanical properties of different tissues, although in a rather qualitative way. The proposed technique can be adopted for research and clinical purposes, as it does not require that the biological sample is excised and prepared by drying and fixation, like in traditional (and more accurate) techniques. The main features and components of the apparatus developed in order to implement the proposed technique have been described. The key component of the apparatus is a LIGA microfabricated and instrumented microgripper which is used to grasp and measure the mechanical properties of the tissue sample. A method has been presented for the identification of the microgripper-sample system. Different samples of human skin have been tested \textit{ex vivo} and successfully distinguished based on their mechanical characteristics. Furthermore the...
system has been used successfully to sense pulse in a microvessel. Further work is in progress along the two different directions described in this paper. The first direction aims to improve the performance of the microfabricated probe. In fact, a limitation of the LIGA microgripper used for the experiments presented in this paper is its low stiffness, comparable to that of some tissue micro-samples. When the probe is much stiffer than the tested samples, the discrimination accuracy of the apparatus improves considerably [14]. The new probe fabricated in superelastic alloy by using micro Electro Discharge Machining has these characteristics and thus it will be extensively used in the near future.

A second direction aims to further exploit the force feedback capabilities of the system. In fact, the haptic interface provides the most intuitive approach to tissue and microvessel palpation and provides real time, high bandwidth feedback signal. This feature may make the Phantom interface very useful to provide a "feeling" of tissue "hardness" to the operator, and even an "alarm signal" during delicate microsurgical operations. We intend to exploit these characteristics in order to develop a new a class of robotic microinstruments designed to augment the performance of the surgeon during MIS.

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