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TECHNICAL PAPER

3D force sensors for laparoscopic surgery tool

János Radó[1] · Csaba Dücső[1] · Péter Földesy[1] · Gábor Szebényi[2] · Zbigniew Nawrat[3] · Kamil Rohr[3] · Péter Fürjes[1]

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Abstract 3D force sensors were developed and integrated in a laparoscope grasping head of the Robin Heart surgery robot to provide additional tactile and force feedback to the surgeon. The Si sensors operate with piezoresistive transduction principle by measuring the stress induced signals of the symmetrically arranged four piezoresistors in a deforming membrane. The chip size was reduced to 1 mm^2 by applying deep reactive ion etching (DRIE) for membrane formation. DRIE opens the way to fabricate complex shaped membranes, thereby a monolith force transfer rod protruding over the chip surface could be integrated. This rod increases shear sensitivity of the structure and plays crucial role in tactile sensing. According to the medical and functional requirements the sensors were covered by biocompatible elastic polymer. The effect of elastic cover on the device performance was modelled by coupled finite element simulation to determine the appropriate geometric parameters. Sensors were covered with semi-sphere PDMS (polydimethylsiloxane) polymer and the effect of the elastic coating was studied in terms of sensitivity and response time. Preliminary test of the laparoscopic head integrated in the Robin Heart surgery robot was performed.

✉ János Radó
rado@mfa.kfki.hu

1 Centre for Energy Research, Institute of Technical Physics and Materials Science, Budapest University of Technology and Economics, Budapest, Hungary

2 Department of Polymer Engineering, Budapest University of Technology and Economics, Budapest, Hungary

3 Foundation of Cardiac Surgery Development, Biocybernetics Laboratory, FRK, Zabrze, Poland

1 Introduction

Minimally invasive surgery (MIS) offers several advantages for the patient and also for the society. The quicker recovery and the smaller trauma are obviously essential for the patient, whereas the reduced hospitalization and recovery time helps the society to utilize the medical costs more effectively (Nawrat 2012).

In order to improve the safety of the laparoscopic intervention the surgeon must get direct and immediate information about the physical and anatomic parameters during operation. The information needed is the feedback from the tools integrated on the head of the laparoscope, such as scalpel and grasper (Trejos et al. 2010).

The essential parameter required during safe cutting and knot tying is the measurement of force, whereas artificial palpation helps in identification of different organs and tissues (Richards et al. 2000; Puangmali et al. 2008; Trejos et al. 2010). These functions require 1D and 3D force sensors, respectively. Solutions have already been demonstrated operate with devices of various transduction principles, such as strain gauges (Menciassi et al. 2003), capacitive (Motamed and Yan 2005), piezoresistive (King et al. 2009), optical (Yip et al. 2010; Takashima et al. 2005) and piezoelectric (Sedaghati et al. 2005).

Our goal is to develop and test a novel laparoscope head with integrated sensors inside the grasper and also on the tip of the device to measure the grasping strength and to provide information about the hardness and surface roughness of the tissue the laparoscope touches (Fig. 1). Targeting a hysteresis- free 3D sensing structure, single crystalline Si with piezoresistive read-out was selected. System integration and development of protective, biocompatible coating technique is also in the focus of the present work.

Fig. 1 Elements of the surgery robots (FRK). Assembled robotic arm (*left*) and the laparoscopic head where the force sensors must be integrated (*right*)

2 Experimental

2.1 Design

2.1.1 Force sensor chip

Four half Wheatstone-bridges were formed on full Si membranes, each composed of two identical resistors and arranged such as to exhibit maximum out-of-balance voltages over mechanical deformation The two resistors form a simple voltage divider or a half Wheatstone-bridge, therefore, the readout is an analogue DC signal (Fig. 2).

In order to fit in the laparoscopic grasper the chip size was limited in 1 × 1 mm^2, whereas the diameter and the thickness of the membrane was selected according to the force range requirements. 1–20 N and 10–2000 mN force ranges were targeted for gripping force measurements and tactile sensing, respectively. The geometric parameters of the sensors are listed in Table 1.

The geometric design was aided by FEM simulation for the force ranges to be measured, e.g. Fig. 3 represents the results given by perpendicular load for the thinner membrane.

The Si wafer processing was complemented by a combined two step DRIE process for parallel formation of the monolithically integrated force transmitting rod on the circular membrane. The membrane thickness is controlled by using SOI wafers of appropriate device layers. An additional etching step was also introduced to form a recessed ring on the bulk Si to facilitate reproducible formation of a semi-sphere elastomer coating on the tactile interface.

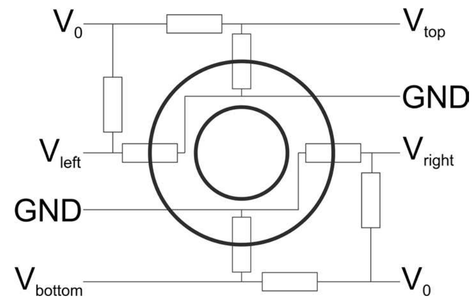

Fig. 2 Structure and read-out of the four half Wheatstone-bridges, each formed by an active piezoresistor and one identical reference element

Table 1 Geometric parameters of the sensor chips

Sensitivity range	Membrane diameter (μm)	Membrane thickness (device layer) (μm)	Force transferring rod diameter/length (μm)	Frame thickness (μm)
10–2000 mN	500	20	250/460	380
1–20 N	500	50	250/430	380

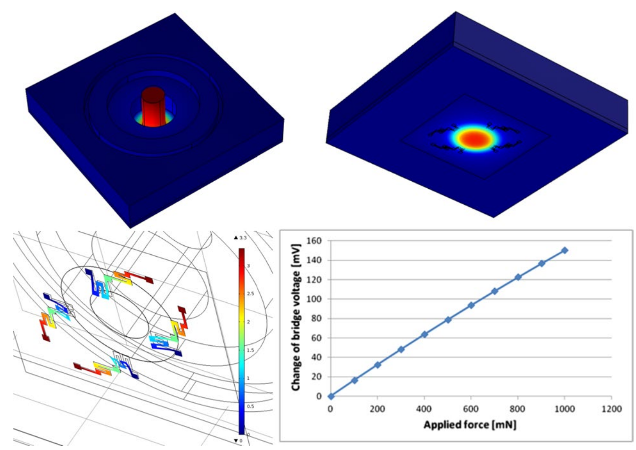

Fig. 3 Coloured representation of the coupled FEM simulation: deformation and mechanical stress distribution of the membrane (*top*) and the four piezoresistors (*bottom left*). The calculated sensor signal for perpendicular load is also shown (*bottom right*). Membrane thickness is 20 μm

The force sensor consists of four independent piezoresistor pairs (3 + 3 k Ohm) in order to resolve the vectorial components of the load. The linear relationship between the voltage changes and the vectorial components of the loading force in the centre of the sensor can be described by the following equations:

$$F_x = \frac{1}{V_0 \alpha_{ls} \pi_{44}} \left(\Delta V_{right} - \Delta V_{left} \right), \tag{1}$$

$$F_y = \frac{1}{V_0 \alpha_{ls} \pi_{44}} \left(\Delta V_{top} - \Delta V_{bottom} \right), \quad (2)$$

$$F_z = \frac{1}{V_0 \alpha_{ln} \pi_{44}} \left(\frac{\Delta V_{right} + \Delta V_{left} + \Delta V_{top} + \Delta V_{bottom}}{2} \right), \quad (3)$$

where F_x, F_y, F_z are the normal (Z) and tangential (X and Y) force components. V_0 an ΔV are the common and the measured voltages, π_{44} is the dominant piezoresistive coefficient, α_{ln} and α_{ls} are the linear normal and shear coefficients in the given geometric arrangement. Note, that an additional resistor was also formed in the bulk Si to measure the temperature and correct force read-out by considering the TCR of Si.

2.2 Processing

The processed wafer was anodic bonded to boron glass to provide enhanced mechanical stability, cavity underneath the membrane and wire contacts for assembly (Fig. 4).

The test chips were mounted on PCB header for preliminary tests. In order to investigate the effect of the elastic coating, identical chips were covered by PDMS layer and characterized. The characteristic responses of each sensing element were collected by a dedicated setup (Fig. 5). Fixed sensors were loaded by an ANDILOG force measuring system capable to measure in the 1 mN–10 N range. A round shape, 300 µm diameter glass pin was mounted on the head of the ANDILOG system to provide precise positioning during measurements. The three axis manual manipulator enables to load from any direction in the 1/4 of the semi-sphere. Due to the symmetric arrangement of the piezoresistors in the deforming membrane the sensor responses can be revealed for all loading directions in the full semi-sphere.

2.3 Electronic integration

Due to the small diameter of the 35 cm long laparoscope arm the volume of wiring must be minimized to leave enough space for the mechanical manipulation wires. Consequently, on-site digitalization of signals is essential to minimize the number of contact wiring and also to eliminate signal loss via the long wires. Therefore, the sensors and the electronic components were mounted on the head of a flexible PCB of appropriate length. The circuit of the pre-processing electronics is shown in Fig. 6. The applied reference voltage is divided by the sensors and digitalized by integrated AD converters mounted nearby. The analogue to digital conversion is performed by miniature high-end ICs (TI ADS1115). Beside the

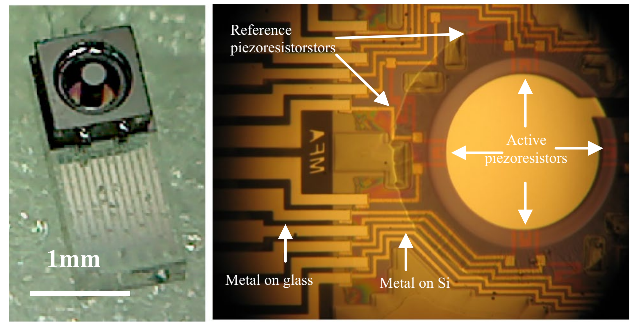

Fig. 4 Optical views of glass bonded chip (*left*) and the same chip as looking from the glass side (*right*). The recessed ring facilitates uniformly shaped semi-sphere elastomer coating

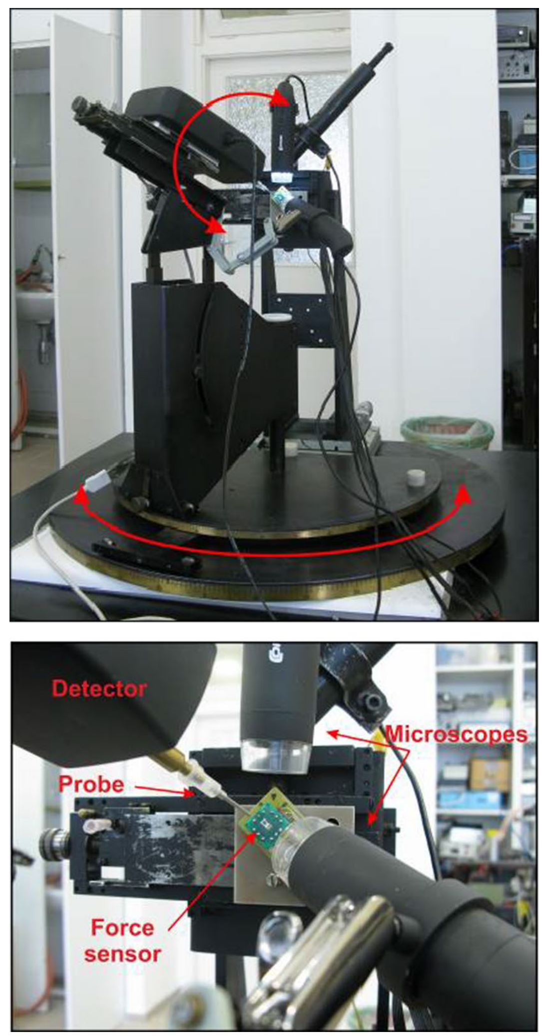

Fig. 5 Measuring setup (*top*) for characterization of the sensor chips exposed to any load direction from a ¼ of the semi-sphere. Close view of the probe head with a PCB mounted force sensor (*bottom*). Two cameras facilitate precise positioning

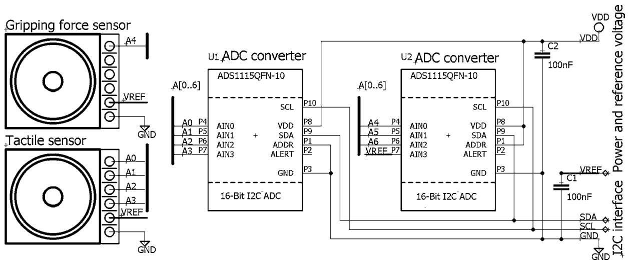

Fig. 6 Architecture of the PCB to be integrated into the laparoscope head. The hybrid circuit contains two AD converters and two force sensors

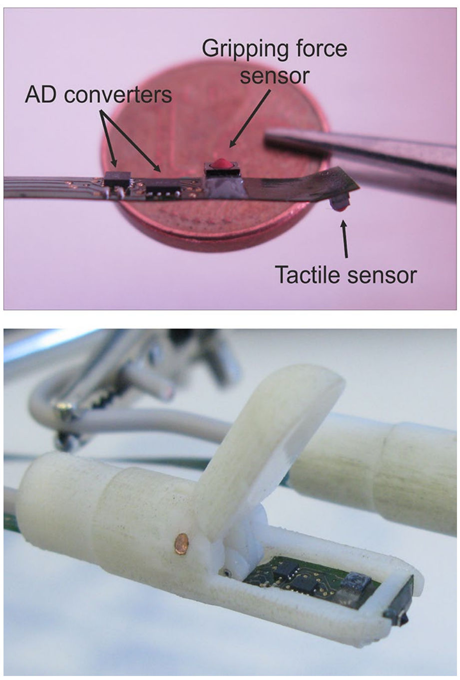

Fig. 7 Sensor chips and AD converters integrated on the head of the 35 cm long flexible PCB. This end of the PCB is inserted in the grasper. Note that the force sensor chips are individually covered with silicon semi-sphere bumpers (*top*). The flexible PCB inserted in the 3D printed test gripper (*bottom*) (color figure online)

force signals the AD converters also convert the signals of the integrated temperature sensor as well as the reference voltage. These digitalized data are accessible through the I^{2}C protocol bus.

Figure 7 shows the head of the flexible PCB with the mounted sensors and AD converter chips. Bending upwards the tip tactile sensor gets its final position in the grasper what is formed by 3D printing for preliminary tests.

2.4 Biocompatible coating

The function of the elastomer coating is twofold: prevents infection of the patient and protects the circuit from the environment. Moreover, it must be elastic to transmit the force and have to withstand sterilization. The low shrinkage of the elastomer during processing is also advised to minimize pre-stressing of the Si membrane and introduction large offset. Considering all the above requirements PDMS was selected for the encapsulation of the hybrid circuit fixed on the lower jaw of the grasper. The void-free filling of the complex shape gripper is also a challenge. Compared to casting technique, injection moulding is expected to meet better this requirement (https://www.researchgate.net/publication/225921553_Flexible_stretchable_and_implantable_PDMS_encapsulated_cable_for_implantable_medical_device) by providing no structure difference, orientation or anisotropy of the thermoset polymer during processing.

Having fixed the assembled circuit on the lower jaw of the grasper, the jaw was inserted in a 3D printed moulding form and PDMS of 4.1 Pas viscosity was injected manually by a medical syringe at room temperature. After 120 min at 80 °C curing the laparoscopic head could be de-moulded. The produced PDMS coverage exhibits 35 ShoreA hardness, 3 MPa tensile strength and more than 300% strain at break.

The moulding however, may introduce deformation of the force sensor. Due to the one side penetration of the elastomer the force transmitting rod may be tilted and introduce large offset. In order to eliminate this effect the sensor chips were individually covered by semi-sphere elastomers of 700 μm diameter before injection moulding (see red bumpers in Fig. 7.)

3 Force sensing

Three levels of testing were performed to characterize the system. In the first step the performance of individual, bare force sensor chips were investigated to determine sensitivity and the amplifying effect of the force transmitting rod on the shear components, followed by the second phase where identical sensors were coated with PDMS elastomer and tested for the same characteristic loads. Due to the elastic properties the PDMS coating is expected to significantly modify both the sensitivity and response dynamics. Data given by these tests are essential for the proper design of the robot's feedback system. Finally, the laparoscope was integrated in the surgery robot to elaborate the best methodologies facilitate the robotic operation.

3.1 Sensing characteristics

The standard test includes the measurements of the four out-of-balance voltage responses for the applied force range in perpendicular and in characteristic directions of load. Due to the symmetry of the system, investigation in 1/4 of the semi-sphere is enough for complete characterization. In this work we present the results given by the three characteristic directions.

3.1.1 *The effect of PDMS coating*

Responses of the same chips with and without PDMS coating were registered to reveal the effect of the elastomer coating of the given geometry and material properties (Figs. 8, 9). A set of eight chips were processed and characterized in terms of sensitivity and response characteristics of the 4 sensing elements. The coating process increased the deviation from ±3 to ±10% in terms of sensitivity, as measured in case of perpendicular load. The sensitivity change for the perpendicular component of the load is straightforward: drops to 1/20th of the bare reference. Nevertheless, the effect of the shear force components is more complex and can not be described as a simple sensitivity loss. The spreading deformation and the coupled hardening in the elastomer across the semi-sphere are manifested not only in sensitivity but also in transformation of response characteristics (see Figs. 8b, 9).

Consequently, detailed mapping and analysis are required to quantitatively describe the relationship between the load direction and responses. Precise determination the surface load by knowing only the strain of the resistors under the elastic cover requires the solution of the complex inverse problem considering the deformation dependent mechanical properties of the elastomer. This is out of the scope of our present work. Nevertheless, the data based recognition of load direction is enough in many applications, e.g. where touch-free steering is targeted.

3.2 Dynamics—the effect of the elastic coating

One may expect that the elastic layer between the rigid Si and the touched surface also introduces delay in response. Therefore, a setup was constructed to measure the difference between the two sensors. A pin was fixed on a piezoelectric crystal (PI Ceramic GmbH, P-153.10) and step-like driving voltage was applied. The resonant frequency was measured to calculate the dynamics of the excitation. According to equation

$$t_{response} = \frac{1}{3f_0}, \tag{4}$$

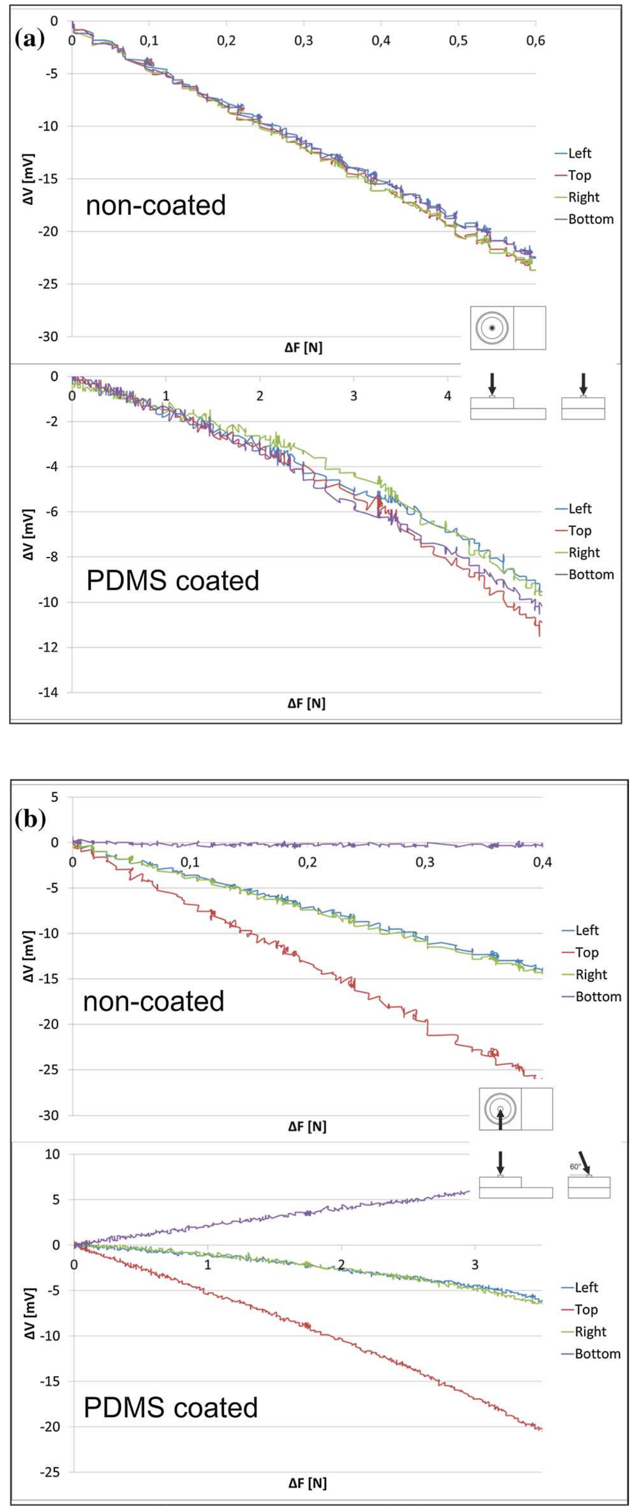

Fig. 8 Sensitivity and response characteristics of the force sensing elements for load directions indicated by the insets. Perpendicular load (**a**) and 60° in Y direction (**b**). The same chip was measured with and without elastic coating

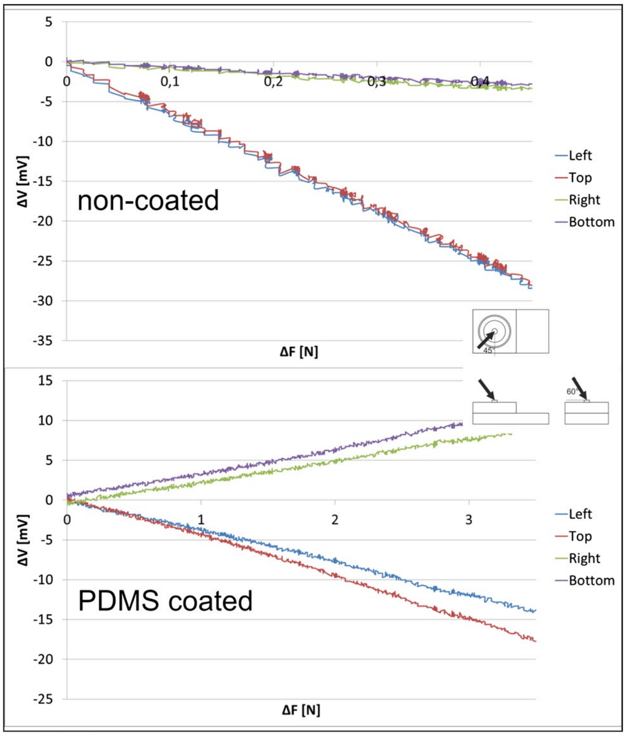

Fig. 9 Sensitivity and response characteristics of the force sensing elements for load direction indicated by the inset. The sensor is exposed to load of X = 60° and Y = 45° direction. The same chip was measured with and without elastic coating

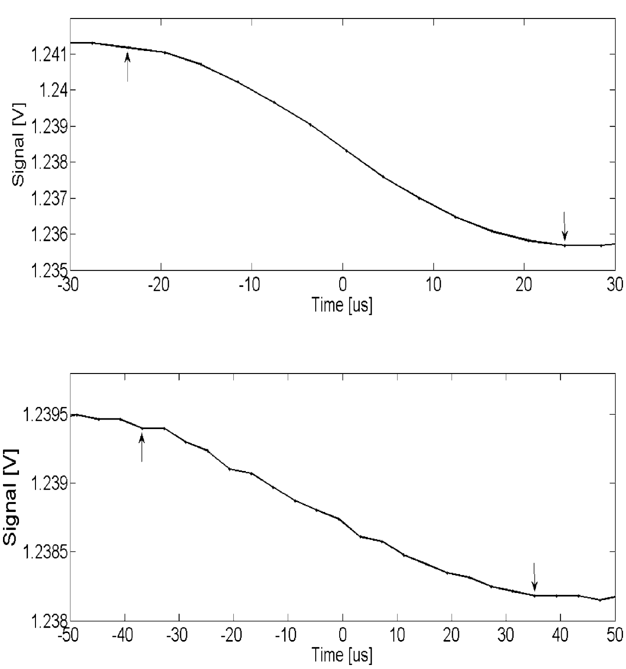

Fig. 10 Response time of the bare (*top*) and PDMS coated (*bottom*) force sensors for perpendicular load. The *arrows* indicate the excitation and the signal saturation. The measured value must be reduced by 11.8 µs delay of the actuator piezo crystal

where f_0 is the resonance frequency, the measured $f_0 = 28.3$ kHz corresponds to a delay time of 11.8 µs. The response time of the bare and the coated sensors are 36 and 60 µs for perpendicular loads, respectively (Fig. 10).

Note, that for reliable measurements the force sensors must have been pre-pressed by 5% of the respective maximum load. In the presented results 100 mN and 1 N were applied for the bare and coated sensors, respectively. Thereby, the measured delay time must be treated as the minimum value.

3.3 Integrated system

The laparoscope was fixed on the arm of the Robin Heart surgery robot and connected to the manipulator via an I^2C communication board (Fig. 11). In a simple preliminary test the signals of both sensors were utilized to control the movement of the arm and the jaw of the gripper. When the laparoscope touches a surface and the emerging force in the tip tactile sensor reaches a pre-set value, an actuation routine stops the forward movement, regardless the actions of the operator with the manipulator head. Also the appropriate strength of the grip can be indicated when the force reaches a pre-set value.

Although much more on-line information can be revealed from the signals of the sensors, the most effective feedback and actuation methods (e.g.: visualization, haptic sensing or automatic control) still have to be elaborated to provide optimal support for the surgeon. This complex task is far beyond the scope of the present paper. In an ongoing follow- up research the group of FRK will elaborate adequate methodologies for human-robot interactions.

4 Conclusions

Processing technology of 1 × 1 mm^2 force sensors was elaborated. The sensors operate with piezoresistive

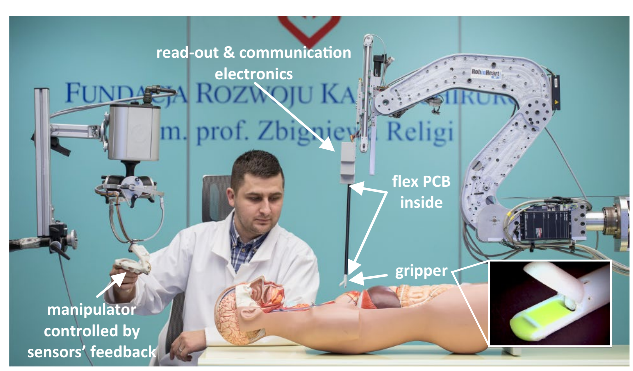

Fig. 11 The laparoscope integrated in the Robin Heart surgery tool. The inset shows the completed grasping head

transduction principle and are able to resolve the vectorial components of the load. The sensitivity of the elastomer coated 3D force sensor chips was measured and their characteristics were compared with their bare references to identify the effects of the elastomer. As the elaboration of reproducible elastomer coating process is crucial in device fabrication, a dedicated injection moulding technique was developed for the coating process. Two sensors and AD converters were integrated in a test version grasper head of a laparoscopic tool. The system can measure the grasping force inside the head and provides information about organs or tissues the head mounted tactile sensor touches. The laparoscope was integrated in the Robin Heart surgery robot and the feasibility of the system was demonstrated.

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