AN INKJET PRINTED PIEZORESISTIVE BACK-TO-BACK GRAPHENE TACTILE SENSOR FOR ENDOSURGICAL PALPATION APPLICATIONS

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ABSTRACT

The paper presents a tactile sensor design mimicking human finger touch to differentiate tissue hardness for endosurgical palpation applications. The sensor comprises two inkjet-printed piezoresistive graphene-based sensing elements linked back-to-back for force and displacement detection, respectively. Experimental results indicate the sensor registers 2.1 and 5.3 mN force feedback from the fat and muscle tissues of pig, respectively, when pressed to the tissues with the same 100 μm displacement. This difference of ~2.5 times in force feedback provides a compelling method by which doctors can more intuitively perceive hardness and tissue differences during endosurgery in comparison with the prior arts.

INTRODUCTION

Nowadays, with the advent of an aging society, the elderly population and the prevalence of chronic disease increase year by year. Minimally invasive surgery (MIS) has been a widely accepted treatment in clinical medicine due to its limited damage to tissues and shorter recovery time in comparison with traditional open surgery. The MIS is operated by directly inserting surgical instruments into the patient's body via one or two tiny incisions for medical operation where an endoscope is used to help the surgeon perform visual inspection. Therefore, palpation, a common practice using a doctor's hands to assess the texture of a patient's tissue, cannot be performed in MIS. The development of endosurgical instruments with force feedback function for surgeons to access surgical tool status and organ information such as tissue structure and hardness, etc. in real-time during MIS has become a critical research subject. Similar to palpation, the force feedback from surgical tools could help doctors to identify lesions, such as a cancerous lump within healthy tissues, due to its contrasting hardness [1].

Many researchers have developed and implemented different MEMS-type tactile sensors to detect hardness differences based on a tactile feedback stress or strain ratio in MIS. For instance, Sokhanvar et al. [2] presented a piezoelectric tactile sensor comprising a suspended beam attached with two sets of PVDF films to sense the beam flexure and total load, respectively. The hardness of the contact object was then derived from the voltage output of the films associated with the corresponding geometry and mechanical properties of the beam. Zhao et al. [3] proposed a two-spring model from which a tactile sensor comprising two piezoresistive cantilevers with different stiffness is developed for characterizing the hardness differences of

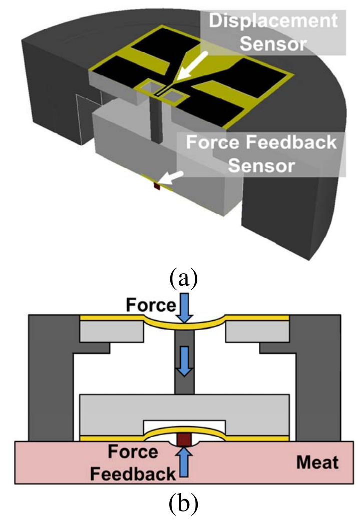
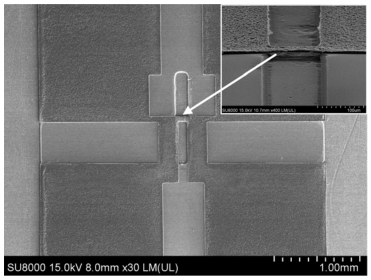


Figure 1: (a) The ideal Scheme of the back-to-back tactile sensor and (b) cross sectional view of the tactile sensor operation.

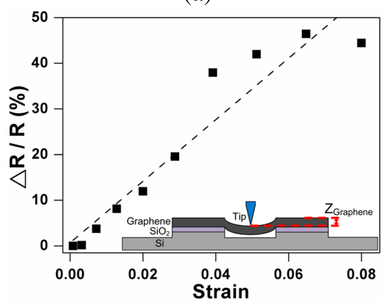
biological tissues. Chuang et al. [4] further employed the two-spring model to design a piezoelectric tactile sensor with a hard copper ball embedded in a soft material (PDMS), which can differentiate artificial tumors from normal tissues in a pig stomach. Nevertheless, none of these sensors can truly reflect the real force feedback of tissues for providing deep sensing capability on organs like real palpation practice. In this paper, we propose a back-to-back two-spring linked tactile sensor design to mimic human fingers for MIS applications as shown in Fig. 1. The ideal scheme of two sensing elements linked by a freely standing rod are accommodated on the top and bottom sides of a package (Fig. 1(a)). The top element will result in a deformation directly transferred to press the bottom element into the tissue by the rod. With the same deformation inside the tissues, the bottom element can sense the hardness difference via the force feedback from the tissues (Fig. 1(b)).

EXPERIMENTAL METHODS

A 3-D printed plastic structure with a central truss to freely link the sensing elements imitating the aforementioned scenario is employed in this work to validate the sensor concept. For the palpation application where the Young's modulus of soft tissues is only tens of kPa, the tactile sensor structure made of polymer materials with comparable Young's modulus can have detectable deformation and force



(a)



(b)

Figure 2: (a) SEM micrographs showing the top and cross sectional views of the free-standing inkjet-printed graphene beam using the CPLOP process. (b) The piezoresistive characteristic of graphene using a micro tip to press the center of the free-standing graphene beam (Displacement up to 20 μm, each point 2 μm).

feedback. As a result, both sensing elements in this work are designed with a cross SU-8 beam structure where graphene resistors are inkjet printed as piezoresistive sensors. In recent years, graphene has drawn considerable research attention for the fabrication of graphene-based microstructures for various applications due to its superior electrical and mechanical properties, such as high carrier mobility, thermal conductivity, and Young's modulus [5]. For example, Li et al. [6] demonstrated graphene cantilever beams as nanoswitches with excellent linearity of actuation force vs. beam displacement suitable for many sensor and actuator applications. Secor et al. [7] demonstrated the first inkjet-printed graphene electrodes with high electrical conductivity claimed for flexible electronics. Moreover, the gauge factor of graphene has been characterized with values ranging from 1 up to ~1.8×10^4 [8-11]. Due to the low temperature and patternable process characteristics of inkjet printing, printed graphene resistors on the aforementioned cross SU-8 beams are used in this work for strain sensing.

For the piezoresistivity characterization of the graphene, a free-standing inkjet-printed graphene beam (beam length = 100 μm, width = 30 μm) with a four-point measurement structure is fabricated using the combined process of lift-off and printing (CPLOP) technique [12]. Because the printed graphene flakes must be sintered above 200 °C for good adhesion and better property on a silicon substrate with silicon dioxide, Cu instead of SU-8 mold is used for the structure fabrication. Figure 2 shows SEM micrographs of the top and cross sectional views of the as-fabricated four-

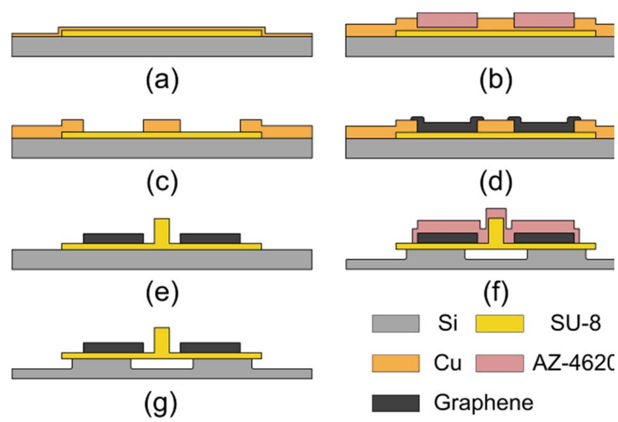


Figure 3: Process flow for the fabrication of the bottom sensing element of the back-to-back graphene tactile sensor.

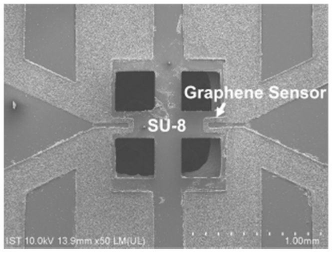
point measurement structure with the corresponding piezoresistive measurement of a 6.5 μm thick free-standing beam whose bottom part has been fully etched by XeF_2 (Fig. 2(a)). The beam is deformed at the center by a microprobe manipulated by a positioner with a resolution of 0.1 μm. As shown in Fig. 2(b), the measured resistance change of the graphene beam is 44.4 % with a center deflection of 20 μm, equivalent to a maximum strain of 0.08 in the beam structure based on the equation as follows [8]:

$$\varepsilon = \frac{2h^2}{l^2} \tag{1}$$

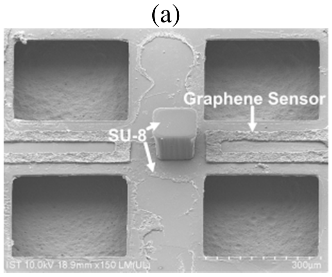
where h and l are 20 and 100 μm, respectively. From the measurement, the average gauge factor $\frac{\frac{\Delta R}{R}}{\varepsilon}$ is derived as ~6.2, which is comparable with the prior findings [8-10] and similar to the gauge factor of graphite [13]. This result can be attributed to the thick graphene structure since the resistor is patterned and printed via a Cu mold.

The fabrication of the bottom sensing element of the back-to-back graphene tactile sensor is depicted in Fig. 3: (a) coating and patterning SU-8 on a Si substrate followed by a Ti/Cu seed layer deposition, (b) photo-patterning AZ-4620 on the substrate for electroplating a Cu mold, (c) stripping the photoresist followed by the removal of the seed layer using Cu etchant and BOE, (d) inkjet printing graphene flakes [7] onto the mold followed by thermally sintering the microstructures at 260 °C for 30 min., (e) etching away the mold followed by photo-patterning SU-8 tip for contacting the tissue, (f) photo-patterning AZ-4620 photoresist to protect the graphene pattern followed by XeF_2 Si etching for the structure release, (g) stripping the AZ-4620 photoresist. The top sensing element is fabricated using a similar process except that the silicon under the sensing beams is DRIE first before undergoing the same process flow. It is noted that a SU-8 tip devised on the bottom sensing element is designed to ensure that the feedback force is directly transferred from the contact object to the sensing beam.

Fig. 4(a) and (b) show SEM micrographs of as-fabricated top and bottom piezoresistive sensing elements, respectively.



(a)



(b)

Figure 4: The SEM micrographs of as-fabricated (a) top and (b) bottom sensing elements of the back-to-back graphene tactile sensor.

Both sensing elements are designed with a cross SU-8 beam structure where graphene resistors are inkjet printed as piezoresistive sensors. Fig. 5 shows the experimental setup to validate the concept of back-to-back two-spring linked tactile sensor design to mimic human fingers. The tested samples, fat and muscle tissues of pig, are placed under the platform and in contact with the bottom sensing element. The same microprobe for the prior piezoresistivity characterization is utilized to provide controlled displacement to the upper sensor.

RESULTS AND DISCUSSION

Figures 6 and 7 show the measurement results. The piezoresistor response of the top sensing element indicates the same deformation is applied to the fat and muscle tissues of pig as shown in Fig. 6(a). Meanwhile, distinct piezoresistor responses from the bottom sensing element illustrate different force feedback from the fat and muscle tissues, with relative changes of resistance of 3.7 % and 10.6 %, respectively (Fig. 6(b)). To further elucidate the force feedback response in terms of force unit, the piezoresistive behavior of the bottom sensing element is analyzed based on the experimental measurement of the bottom element. Similar to the prior gauge factor measurement, the SU-8 tip in the bottom sensing element sitting on a load cell is deformed by the microprobe. The measured resistance changes corresponding to the force obtained from the load cell are depicted in Fig. 7(a). By linear interpolation, the force feedback from the tissue measurement, i.e. Fig. 6(b), can be replotted as shown in Fig. 7(b) with force unit instead. It is found that the muscle tissue indeed exhibits a larger rigidity than the fat one (Fig. 7(b)). For a load of 100 µm displacement, the detected force feedback from the muscle and fat tissues are 5.3 and 2.1 mN, respectively.

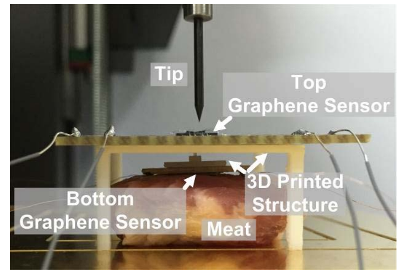
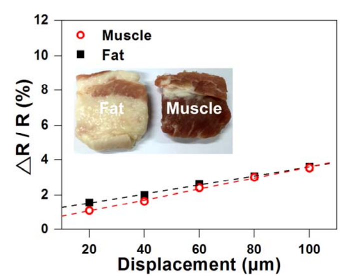


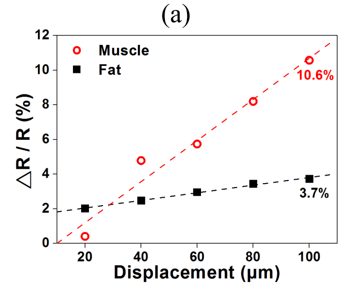
Figure 5: The experimental setup for the characterization of the back-to-back graphene tactile sensor.

A COMSOL simulation has been performed to analyze the force feedback response and indicates a small discrepancy between the simulation and measurement results in Fig. 7(a). The simulated force is based on the input of the 4.02 GPa of SU-8 Young's modulus and the maximum strain at the center of the cross beam derived from the piezoresistivity characterization with the corresponding resistance change. The discrepancy can be attributed to the improper material property and incidence of feedback force, which is off-center to the sensing beam due to the misalignment during the SU-8 tip fabrication as shown in Fig. 4(b). Further investigation to understand and mitigate this effect is ongoing.

In the back-to-back tactile sensor design, the applied displacment to the contact object is controlled by the top sensing element. The longer the cross SU-8 beam is, the larger the displacement that can be achieved. This may affect the spatial resolution once the sensor design is applied to form a 2-D array for mapping the hardness distribution of the tissue to be detected. Further optimization of the materials and sensor design will therefore benefit the continued development of this system.



(a)



(b)

Figure 6: Sensor measurement results for fat and muscle. The piezoresistive characteristics of (a) top graphene sensor and (b) bottom graphene sensor for the different tissues.

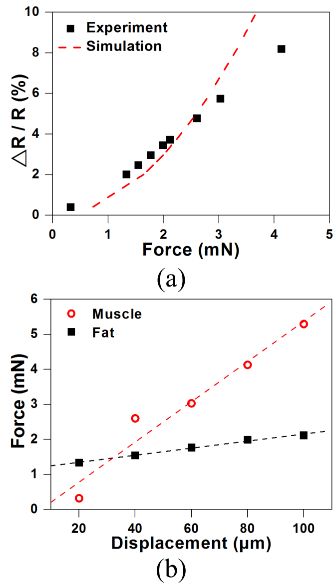


Figure 7: (a) Relative change of resistance vs. force between the measurement and simulation using tip to press the bottom graphene sensor and (b) relative displacement vs. force feedback from the tissues.

CONCLUSION

In summary, an inkjet-printed back-to-back linked graphene tactile sensor concept has been demonstrated. This fabricated sensor is applied to both fat and muscle tissue, revealing a difference in force feedback of ~2.5 times. This demonstration validates the sensor design, confirming a capability to differentiate tissue hardness and also revealing great potential for practical clinical application regarding endosurgical palpation.

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