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Smart Laparoscopic Grasper Utilizing Force and Angle Sensors for Stiffness Assessment in Minimally Invasive Surgery*

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***Abstract*— As an alternative to open surgery, minimally invasive surgery (MIS) utilizes small skin incisions as ports to insert an endoscope and surgical tools. MIS offers significant advantages, including reduced pain, shorter recovery times, and better cosmetic outcomes than classical surgeries. However, MIS procedures come at the cost of losing the "sense of touch," which surgeons rely on to examine the tissues under operation, palpate organs, and assessing their conditions. This has encouraged researchers to develop smart MIS tools that provide artificial tactile sensation, mostly using electrical- or optical-based tactile sensors. In this work, we introduce a prototype of a smart laparoscopic grasper integrated with force and angle sensing capabilities via off-the-shelf sensors. The specification and design of the smart grasper are presented, as well as a demonstration on stiffness assessment of elastomeric samples and chicken meat. Overall, our prototype exhibits great potential for MIS applications, with room for future improvements.**

***Clinical Relevance*— The development of a smart laparoscopic grasper for MIS applications helps in restoring the tactile sensation to surgeons and enables safe grasping and manipulation of human organs.**

I. Introduction

Minimally invasive surgery (MIS) has recently become a gold standard for many frequent surgical procedures, e.g., cholecystectomy and appendectomy [1]. During MIS, surgeons grasp and manipulate internal organs using specialized instruments designed with long shafts. However, missing vital information about applied forces during these actions remains a challenge [2]. The demand for MIS tactile sensing has also been highlighted with the recent advent of robot-assisted MIS [3]. As a response, many researchers attempted to develop artificial MIS tactile sensations by integrating force sensors at different locations of MIS instruments, i.e., at the end effector [4], the shaft [5], or the base [6].

The development of artificial tactile sensation in MIS is an ongoing research trend. From one side, the tactile sensation is essential for safely maneuvering organs, tissues, and sutures, as well as getting reliable determination of the consistency of the tissues. With force feedback, surgeons can avoid applying excessive pressure or accidentally damaging healthy body parts [7]. Besides, force feedback helps in reducing slippage occasions by guiding the operator towards secure grasps. From the other side, the tactile sensation can be developed one step further to assess the stiffness of organs. As palpation of organs remains the most popular method used for tumor detection, MIS surgeons would potentially benefit from the artificial tactile sensation in determining the presence of harder, stiffer tumor tissues and in detecting potential hidden lumps [8]. One MIS-related study demonstrated the calculation of tissues' modulus of elasticity using a force sensor integrated on the grasping tip of an endoscopic tool, with a good agreement between the theoretical and experimental results [9].

Among the various tactile sensing techniques attempted for MIS, electrical-based force sensing methods are the most commonly used and widely spread due to the ease of fabrication, simple circuitry, and low cost [10]. In this context, the recent developments in microelectromechanical systems (MEMS) have revolutionized the tactile sensing technologies. Through microfabrication approaches, e.g., photolithography, silicon-based sensors and actuators can be miniaturized down to the micron-level and manufactured in batches with excellent signal-to-noise ratio and low hysteresis [11]. Consequently, several MEMS-based force sensing devices were oriented towards MIS applications. For example, Qasaimeh et al. introduced a fully micromachined polyvinylidene fluoride (PVDF)-based jaw sensor aiming to create a sensorized endoscopic tool [12]. The design of the piezoelectric jaw sensor incorporated a patterned PVDF film sandwiched between a micromachined silicon layer with tooth-shaped protrusions and a Plexiglas layer. After characterization, the proposed sensor was proven capable of measuring the full range of forces associated with MIS and detecting small, hidden irregularities in objects. Nevertheless, there are quite a few issues concerning such a piezoelectric-based sensor since it requires charge amplifiers and complicated electronic setups and is limited to only measuring dynamics forces [13]. Similarly, capacitive and piezoresistive MEMS-based force sensors were considered for integration with MIS tools [14]. All these efforts come as a response to the lack of sense of touch in MIS and robotic surgeries.

Force sensitive resistors (FSRs), type of piezoresistors, are thin, printed, and flexible electrical devices that measure applied forces with decent precision. Based on their structural design, common FSRs are classified into either "thru mode" or "shunt mode." In shunt mode sensors, two interlaced traces, serving as the sensing electrodes, lie on top of a resistive ink-coated membrane [15]. The increased contact between the electrodes and the resistive layer when forces are applied causes a decrease in the electrical resistance. On the other hand, FSRs employing thru mode technology utilize two metal circles sandwiching a pressure-sensitive layer [16]. Under loading, the electrical resistivity of the piezoresistive layer decreases, and the output signal is used to estimate the

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magnitude of applied forces. Thru mode has better linearity, sensor drift, and dynamic measurement accuracy than the alternative shunt mode [17]. Both FSR modes are available as off-the-shelf sensors.

This work aims to develop a smart laparoscopic grasper equipped with two off-the-shelf sensors: an FSR and an angle sensor, to measure applied forces and assess the stiffness of organs and tissues in MIS. Compared to Silicon-based MEMS force sensors, off-the-shelf FSRs are cheaply available, thin, and can be used as plug-and-play with MIS tools. In addition to being disposable, these commercially available sensors are flexible and can accommodate different sizes and shapes of grasper jaws. Also, they have a large working range and can measure from milli- to tens of newtons. By having libraries of the force sensors available in the market, specific sensors can be matched and integrated with specific laparoscopic tools on-demand, based on the surgery, patient, and surgeon—a LEGO-like approach. Our work also aims to assess the functionality of adding an angle sensor off-the-graspers to see if organ stiffness can be accurately estimated without interacting with it, eliminating the need for sterilizations and biocompatibility. Once successful, these new integrative LEGO-like concepts can be applied to other fields, such as robotic manipulators.

This paper is organized as follows: Section 2 presents the design and specification of the assembled prototype and the two designated sensors. In section 3, the experimental setup and sample preparation are described in detail. The results are presented and discussed in Section 4. Towards the end, potential improvements for next-generation prototypes are highlighted to overcome the current limitations.

II. Design and Integration

MIS surgeons rely on long surgical instruments to reach internal organs through small incisions in the human skin (~10 mm). On the market, there are many available laparoscopic tools designed to do specific tasks. Our developed prototype is a modified version of ratcheted-type laparoscopic grasping forceps (Inovus Medical, UK). In this grasper, the two jaws at the end effector are 5 mm wide, and the angle of opening is manually controlled via the tool's handle. Two commercially available sensors were installed onto the grasper to make it smart (Figure 1). Specifically, a force sensor was attached to one grasping jaw, and an angle sensor was fixed at the handle's pivot (center of rotation).

A. Force Sensor

First, an FSR, serving as a force sensor, was attached to one of the grasping jaws of our prototype. The FSR of choice is a FlexiForce A201 (Tekscan, USA), with a sensing area of 9.7 mm and 0.1 – 111 N working range. FlexiForce sensors are thin, flexible, cost-effective, customizable, and can be easily wired to a data acquisition board through a voltage divider circuit with minimal power requirements. A decrease in the electrical resistance R_{sensor} due to external forces causes a voltage increase at the divider output node [16]. Since the sensor's surface area was larger than that of the jaw, a 3D-printed part was needed to support the hanging part of the sensor. The force sensor was glued to the jaw and the 3D object, achieving accurate measurements of grasping forces.

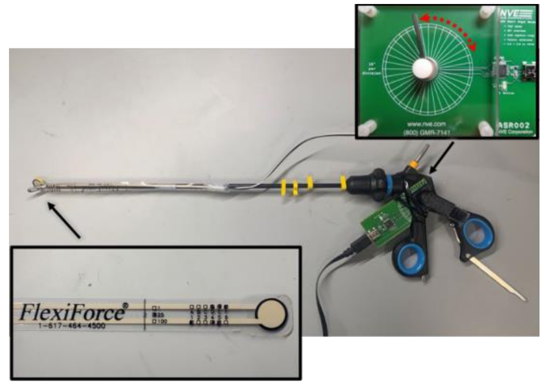

Figure 1. Image of the developed smart laparoscopic grasper prototype: ratcheted type laparoscopic grasping forceps integrated with an FSR and angle sensors shown in the two insets.

B. Angle Sensor

Information about the opening angle of the jaws can be indirectly obtained at the handle of the tool. Therefore, an ASR002 Smart Angle Sensor (NVE Corporation, USA) was installed at the handle's pivot of our smart grasper to measure the jaw's opening angle. This sensor has a resolution of 0.1°, and the jaws' rotation is two folds of the handle. With information from both sensors, the stiffness of the grasped object can be estimated based on the ratio between the force and angel value of the grasping. In other words, soft tissues will deform easily without reacting with large forces on the sensor installed at the grasper jaw, while stiffer tissues will resist deformation and react with larger forces on the sensor.

III. Experimental Setup

A. Force Sensor Calibration

Before testing the prototype, the FlexiForce sensor was calibrated using Instron 5540 Series electromechanical testing system (Instron Inc., USA). The Instron load frame was equipped with a 50 N load cell capable of ±0.5% reading accuracy down to 1/250 of the cell capacity (200 mN). The Instron load frame applied normal force to the top surface of the sensor by compressing it at a slow rate. Upon reaching a compressive load of 25 N, the Instron load frame retracted to its initial level. Based on repetitive loading processes, a correlation was obtained between the sensor output and the force applied. This force sensor serves the purpose of proof-of-concept. Yet, FSRs are known to be nonlinear, which might impact the repeatability of the method. Testing the repeatability and reliability of these sensors within this technique is beyond the scope of this study, and will be tested and characterized in our future work.

B. PDMS Samples

Polydimethylsiloxane (PDMS) is a silicone-based organic polymer that is widely used in microfluidics and biomedical applications due to the ease of fabrication and molding. Usually, PDMS substances were produced by mixing a pre-polymer (base–A) and cross-linker (curing agent–B), where the ratio of the two substances holds control over the mechanical properties. Here, four PDMS samples were prepared with a weight of 25 grams and A:B mixing ratios of 10:1, 20:1, 27:1, and 40:1, respectively. Then, each mixture was poured into an individual Petri dish (5 cm in diameter), filling 1 cm of it. Next, the four circular molds were placed inside a 60 °C oven for 2 hours. After being fully cured, the samples were removed from the dishes, creating: hardest, hard, soft, and softest PDMS samples.

C. Chicken Meat Samples

In addition to elastomeric samples, we aimed at testing our prototype against biological samples represented by chicken meat. To do that, a frozen chicken breast was bought from a grocery store and left for few hours at room temperature to defrost. Then, a total of three samples: raw, cooked for 10 mins, and raw embedded with a metal bead, were prepared with a thickness of 1 cm. Then, those samples were enclosed with a thin transparent cover to prevent the contamination of the grasper.

IV. Results and Discussion

After preparing the elastomeric and chicken meat samples, the prototype was manually operated to do repetitive grasp-hold-release events as an equivalent to palpation. The same samples were tested several times, and the average of the readings was calculated.

A. PDMS Samples

The first set of experiments was performed using the four PDMS samples. The signals from the integrated sensors during repetitive grasp-hold-release events are plotted on the same graph as shown in Figure 2(A). At each grasp, the angle sensor reflects the change in the opening angle of the grasping jaws (where 0° represents the initial state of fully open jaws), and the force sensor measures the amount of force being applied. At the same closing angle, softer samples show lower force magnitudes than harder ones. We demonstrated stiffness assessment by dividing the maximum force (N) by the maximum angle (θ) of each grasping occasion, as shown in Figure 2(B).

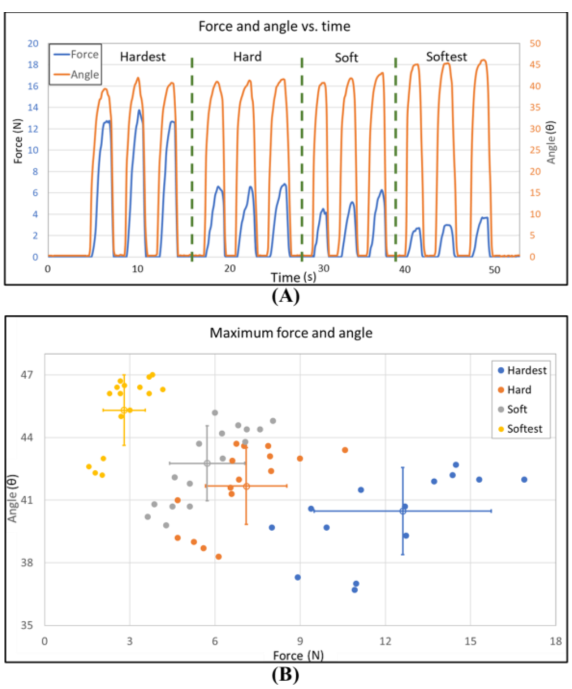

Figure 2. Data from the FSR and angle sensor integrated with the laparoscopic tool during grasp-hold-release events performed using four PDMS samples. (A) Plot of the value of force and angle versus time. (B) Maximum force versus maximum angle.

From the average of the calculated value of each sample, the samples can be remarkably classified based on their estimated stiffness, as shown in Figure 3.

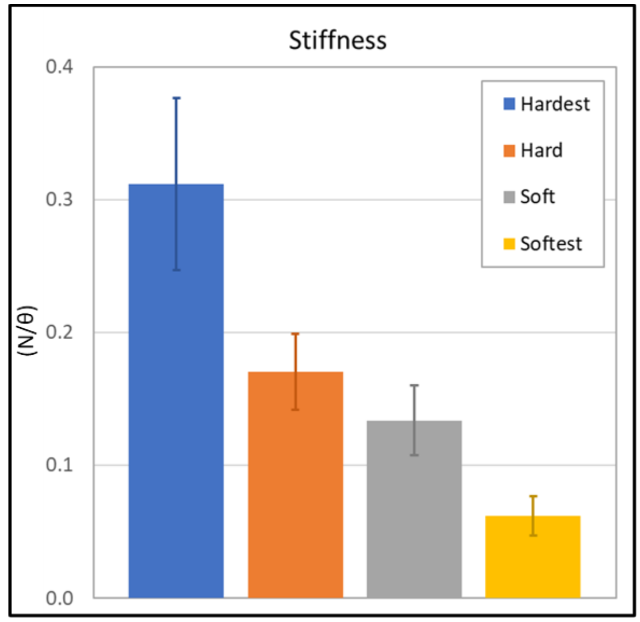

Figure 3. Stiffness estimations based on the average of the ratio of maximum values of force (N) and angle (θ) readings from the integrated sensors.

B. Chicken Meat

After testing the prototype with the elastomeric PDMS samples, the prepared chicken meat samples were similarly tested. Subsequently, the data from our smart grasper indicated that the raw chicken meat is softer than the cooked one (Figure 4). From the maximum values of force and angle, the stiffness of the meat can be estimated, similar to the way shown earlier for PDMS samples.

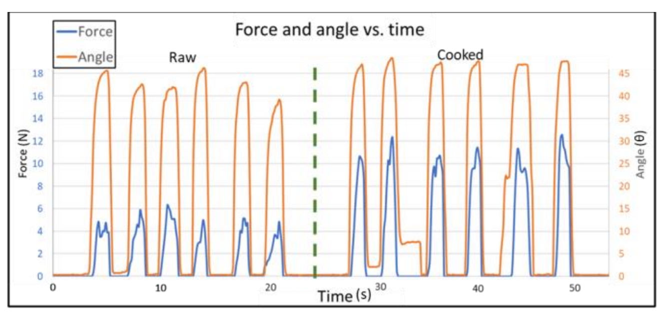

Figure 4. Data from the FSR and angle sensor during grasp-hold-release events performed on chicken meat samples. Combined readings from both sensors during palpations can clearly evaluate the stiffness of raw (soft) and cooked (hard) meat samples.

C. Lump Detection

The last aim of our smart grasper is to detect deep stiffer lumps inside grasped organs and tissues, which was conducted using a chicken meat sample embedded with a hidden metal bead. The test started with the grasping jaws being at position-1, 2.5 cm away from the location of the metal bead. For the following grasping location, the grasping tip was moved 0.5 cm closer to the bead each time until reaching the bead at position-5. From the force and angle signals, the location of the metallic bead can be identified with a higher magnitude of force at the same closing angle (Figure 5).

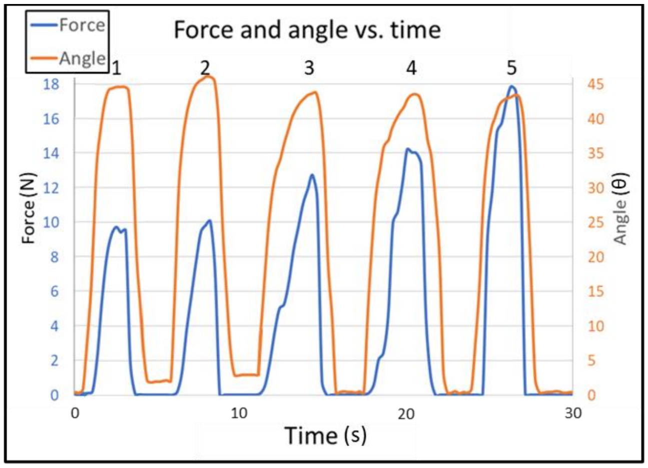

Figure 5. Data from the FSR and angle sensor during grasp-hold-release (palpation) events performed at five positions on the chicken meat sample embedded with a metallic bead (localized at position 5).

Overall, these samples are just models to show the applicability of the technique with real flesh taken from a biological source, i.e., chicken. Hence, they do not necessarily reflect human tissues involved in actual surgical treatments. Nevertheless, the technique was sensitive enough to show differences between cooked and raw chicken meat. Also, the metal bead was used as a model for representing embedded lumps. In the future, more realistic models of flesh embedded with lumps will be used and characterized in our work.

V. Conclusion

In this work, we presented our working prototype of a smart laparoscopic grasper and demonstrated its ability to distinguish the stiffness of grasped samples/objects based on the signals from two off-the-shelf integrated sensors, i.e., an FSR and an angle sensor. Additionally, our prototype's capability of obtaining perceptions of hidden lumps stands promising for tumor detection scenarios. Following this proof-of-concept study, more characterizations and repeatability studies will be performed in our future work. Also, we will consider using a more robust MIS grasper for a longer lifetime of the prototype, as the current one showed some signs of damage. Additionally, having one piece of code that processes the data from both sensors will be more convenient than the current off-line analysis process. Lastly, other properties of grasped tissues should be explored as well, e.g., thickness.

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