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A Soft Multi-Axial Force Sensor to Assess Tissue Properties in Real-Time

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Abstract*— *Objective: **This work presents a method for the use of a soft multi-axis force sensor to determine tissue trauma in Minimally Invasive Surgery.**

Despite recent developments, there is a lack of effective haptic sensing technology employed in instruments for Minimally Invasive Surgery (MIS). There is thus a clear clinical need to increase the provision of haptic feedback and to perform real-time analysis of haptic data to inform the surgical operator.

This paper establishes a methodology for the capture of real-time data through use of an inexpensive prototype grasper. Fabricated using soft silicone and 3D printing, the sensor is able to precisely detect compressive and shear forces applied to the grasper face. The sensor is based upon a magnetic soft tactile sensor, using variations in the local magnetic field to determine force.

The performance of the sensing element is assessed and a linear response was observed, with a max hysteresis error of 4.1% of the maximum range of the sensor.

To assess the potential of the sensor for surgical sensing, a simulated grasping study was conducted using ex vivo porcine tissue. Two previously established metrics for prediction of tissue trauma were obtained and compared from recorded data. The normalized stress rate (kPa.mm^{-1}) of compression and the normalized stress relaxation (ΔσR) were analyzed across repeated grasps. The sensor was able to obtain measures in agreement with previous research, demonstrating future potential for this approach.

In summary this work demonstrates that inexpensive soft sensing systems can be used to instrument surgical tools and thus assess properties such as tissue health. This could help reduce surgical error and thus benefit patients.

I. Introduction

Minimally Invasive Surgery (MIS) is well known for bringing a number of advantages over traditional open surgery methods, including a lower degree of blood loss, improved cosmesis and faster recovery [1]. Despite these advantages, MIS still presents some significant issues. During the procedure, long and narrow instruments are used to enter the abdominal cavity through several small ports (also known as trocars) [2] (Fig.1a). Graspers are a key instrument used to manipulate and hold tissue in MIS procedures, in essence replacing the surgeon's fingertips. However, the mechanical nature of these tools significantly impedes the sensation of force feedback to the surgeon, thus increasing the likelihood of tissue over-compression and damage [3]. Up to 66% of damage in MIS can be attributed to grasping forceps [4], showing a significant clinical need for improvements in surgical tools to deliver force feedback and limit tissue trauma. Accordingly, this paper presents the development of a soft sensing system for real-time measurement and analysis of applied tissue pressure in MIS graspers, with an ultimate aim to minimize tissue trauma.

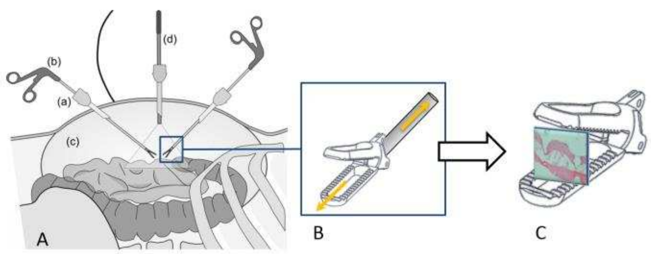

Figure 1. (a) Schematic of MIS, indication a – Trocars, b – Graspers, c – Expanded abdominal cavity, d – Laparoscope. (b) Indication of the forces in surgical retraction. (c) Indications of damages caused by the grasper teeth

Existing research seeking to reduce damage in MIS graspers has considered the geometry and characteristics of the grasper jaw. In general, sharp jaws are used to prevent cases of tissue slip [5], however sharp edges and corners on the grasping face increase the magnitude of point stresses applied to the tissue [6] (Fig. 1c). These stresses can be reduced by rounding of the geometries [7, 8] in an effort to reduce damage, although excessive rounding can increase the chances of tissue slip. Other groups have looked toward soft graspers as a method of stress reduction. Instead of a large variance of elastic modulus between the tool and tissue (found in grasper jaws made from stainless steels), a soft grasper would deform under higher forces, applying much lower stresses to the tissue. Such a grasper would deform to the geometry of the tissue providing a higher contact area to increase friction along the tissue-tool interface, reducing the chances of slip [9].

Complementing work on grasper jaw optimization, other research has explored the use of sensing systems to provide additional haptic feedback to the surgeon. The replication and enhancement of the tactile sensations experienced in open surgery could negate one of the main sources of surgical error in MIS. While many research groups have focused on simulated surgical environments for training [10], others have looked to embed sensors within surgical tools, to allow real-time feedback to be produced [11, 12]. The field is split into

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two main foci: 1) Indirect measurement of forces within the linkage mechanism, and 2) Direct measurement of the tissue-tool interface. The sizing constraints of the surgical graspers used in MIS [2] present a number of advantages and disadvantages for the two measurement methods above.

Indirect measurement [6, 12, 13] relaxes geometric constraints by situating force sensors at the distal end of the instrument and thus outside of the body. This also enables the grasper to use a conventional set of grasping jaws. Typically a standard tension load cell is used to detect the force applied to the grasper linkage, from which the force on the grasper faces can be calculated. While this method shows promise, it is best suited to determine the normal force exerted on the tissue by the grasper, further degrees of freedom are difficult to obtain since they would rely on modifications to the mechanical linkages within the grasper. It is also sensitive to mechanical losses in the system (i.e. friction along the actuation shaft).

The alternative approach is to measure interaction forces 'directly' with sensing elements situated at or within a modified pair of grasping jaws. In this instance, geometric and environmental constraints limit appropriate sensing modalities; for example the system must be biocompatible and robust to fluidic contamination from the body. The majority of direct measurement graspers employ prototype miniaturized force sensors [9, 14, 15] to measure the distribution of forces in direct contact with tissue. Other sensors have been introduced to give a multi-axial force reading, allowing axial forces as well as compressive to be read [16, 17]. This offers a benefit over the compressive only measurement as more complex actions, such as surgical retraction (i.e. pulling tissue) (Fig. 1b), apply shear forces onto the tissue in combination with the compressive forces used to obtain grip.

The tools used within MIS fall within two fields: disposable and reusable. A major economic problem with the use of disposable surgical tools is the increased cost to hospitals, with disposable tools costing approximately 10x as much as the reusable counterparts [18, 19]. Soft tactile sensors have become a specific area of research interest in recent years for their use in applications such as robotic manipulators. Their small size and compliance also make soft sensors an ideal candidate for integration into surgical tools [20]. Sensing principles for deformable/soft sensors vary, and are optimally suited to different applications. Microelectromechanical (MEMS) barometric sensors offer a precise single measurement of pressure applied, however are limited to sensing a single axis[21]. Soft capacitive and resistive sensors may be used to measure multi-dimensional force and 2D force maps, however the fabrication procedure is complex, increasing cost [22, 23]. Alternatively, recent work on magnetic field based sensors offer a simple fabrication method, with the ability to obtain multiaxial forces precisely [24, 25].

For a sensing system to help minimize tissue trauma intraoperatively it is crucial that it can measure and analyze appropriate data in real-time. Past research has focused on the recording of data from surgical tools for retrospective analysis of tissue-tool contact forces [6, 12, 13]. However, these applied forces could potentially be analyzed in real-time to assess tissue health. Previous work in our group has proposed a set of metrics with the potential to detect tissue damage in real time through the assessment of applied stress and strain rate [26]. A discrete differentiation of the changing pressure on a tissue sample was correlated with histological signs of damage. Such an analysis could allow a new generation of 'smart' surgical tools with the ability to detect approaching tissue damage and help prevent it from occurring by providing the controlling surgeon with appropriate feedback.

II. Methods

Our work focuses on the development of a prototype haptic sensor for an MIS grasper, leveraging our prior work in magnetic field (MF) based soft sensors which have been shown to be accurate, high bandwidth, and low cost [24]. This will be followed by an evaluation of performance and preliminary clinical feasibility through an *ex vivo* tissue study.

A. Sensor Requirements

To fully describe the interaction forces between grasper jaws and corresponding soft tissue, the sensor should be able to determine both normal and shear forces, as the latter may result in significant tissue trauma [12, 13].

Sensor performance in each degree of freedom must be sufficient to detect mechanical indicators of trauma and inform the user in real-time. Our prior work indicates that 500Hz sample rate is appropriate and that a load range of 300kPa coupled with high sensitivity (~6.25mN) is necessary [27].

Geometric constraints for typical MIS grasper instruments are 5mmx20mm [28]. Here we consider this target surface for a single sensor node, while noting that arrays of sensor nodes have been employed within these constraints previously [29].

B. Sensor Description

Based on the requirements detailed above, we chose to evaluate a soft MF-based tactile sensing technique, based upon work described by Wang et al. as a candidate for our instrumented surgical grasper. The technique exploits recent advances in MEMs Hall-Effect sensors to provide high performance, multi-axis force sensing at a low cost. Sensing characteristics can be readily 'tuned' through adaption of sensor design variables, principally the geometry and material properties of an elastomer core which deforms with applied load [24].

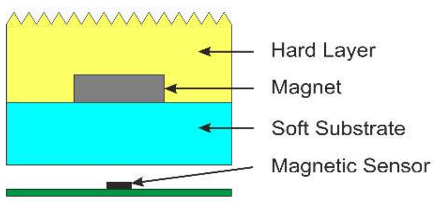
In this implementation, we integrated a sensing element into an idealized grasper jaw with cross-sectional area of 8x15mm (Fig. 2b) and adopting a typical 1mm 60° tooth pitch, as specified by Brown et al. [28]. This facilitates the preliminary laboratory-based evaluation described here and avoids manufacturer specific jaw profiles that often contain more complex geometries and fenestrations.

Our sensor consists of a 3D printed grasper face with an ABS-like elastic modulus (HTM140, Envisiontec). Embedded within the grasper face is a high-strength cylindrical neodymium permanent magnet (5mm radius x 2mm). The grasper face is layered above a compliant substrate of silicone sheet (DragonSkin 30 Shore 30A, Smooth-On), assembled with the magnet directly above a tri-axis Hall Effect sensor (ML90393, Melexis), as shown in Fig. 2a. Forces applied to the grasper face results in deformation of the compliant silicon substrate, associated movement of the permanent magnet relative to the Hall Effect sensor and accordingly variation in the measured local magnetic field. The Hall Effect sensor communicates with real-time control hardware (myRIO, National Instruments) using a dedicated I2C communication channel. This permits a sample frequency of up to 500Hz. Sensor measurements were then filtered with a second-order low-pass filter at 50Hz to attenuate electrical noise.

The system was calibrated using a Genetic Programming algorithm to define a discrete relationship between force and magnetic field [30, 31]. This enables fast and robust identification of the non-linear multi-dimensional relationships between force and magnetic field without the need to develop descriptive mathematical models of the system.

The calibration dataset was obtained by taking measures of applied load and resultant magnetic field across a parametric sweep of normal and shear loads, using the experimental equipment described in the next section.

A



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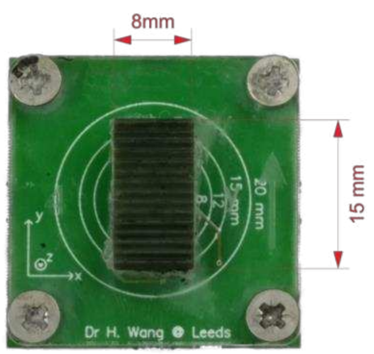


Figure 2. (a) Schematic of the sensing grasper face (b) A photo of the prototype grasper, indicating the dimensions and principal axes of compression *(z-axis)* and shear *(y-axis)*

C. Sensor Performance Characterization

After fabrication, the performance characteristics of the sensor were assessed using a bespoke test apparatus, based on the sensor evaluation equipment described in [24]. Two linear stages (T-LSR75B, Zaber Technologies Inc., Vancouver, Canada) were arranged in the z- and y-axes relative to the grasper face. A small manual stage was used to center the grasper jaws to be in line with interlocking teeth. A laser displacement sensor (LM10, Panasonic) was used to provide an independent measure of the separation of the grasper faces.

Firstly, due to the non-linear mechanical properties of the silicone substrate [32] the hysteresis of the sensor was examined. The hysteresis was analyzed against a commercial 6-axis load cell (Nano17, ATI). The outputs of both the MF sensor and load cell were recorded through 8 cycles of loading and unloading. From this, the maximum hysteresis error was taken.

Secondly, an experiment was conducted to determine the ability of the sensor to determine both compressive and shear forces. The sensor evaluation system (Fig. 3) was used to compress a sample of silicone elastomer (Sorbothane 00-50, Sorbothane) to a 35N hold pressure before tension was applied along the Y-axis of the sensor (Fig. 2b).

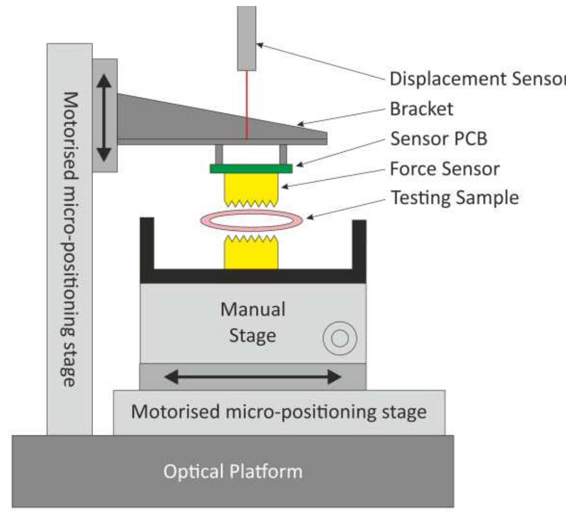


Figure 3. Schematic of the experimental setup used for assessing the sensor under controlled conditions.

D. Porcine Testing

To complement the basic characterization of sensor performance, a study was undertaken to represent the target application of MIS. In this work we focused on grasping of colon (large intestine) tissue. This is a common occurrence during abdominal MIS since the tissue must regularly be moved to expose different target regions during procedures. Porcine colon was selected as an analogue for human colon tissue, due to its similar size and morphology [33].

In order to apply a controlled compression to the tissue samples, a simulated grasping environment was produced. The sensor evaluation system was adapted to hold colonic tissue (at zero strain) such that grasping stresses of up to 300kPa could be applied to the tissue, replicating typical operating conditions found in a surgical environment [6]. The system has the ability to precisely control strain applied to the tissue and obtain measurements at high enough frequency (500Hz), appropriate for real-time analysis.

From the measured data we aimed to calculate two key metrics shown to be predictive of tissue trauma and thus evaluate the ability of the sensor system to reliably perform this function. Firstly, relaxation occurs when tissue is held under a constant compressive strain and the observed stresses will decay [34]. Since relaxation is dependent on material properties this may be indicative of tissue trauma. To normalize the metric for different grasping pressures, a percentage relaxation was measured, defined by equation (1).

$$\overline{\Delta\sigma} = (\Delta\sigma / \sigma_{max}) \quad (1)$$

Secondly, a metric termed the Normalized Stress Rate has been proposed as a potential metric for predicting tissue damage [26]. It is defined by equation (2) in which the applied stress rate is determined and normalized with respect to applied strain rate (i.e. speed of the grasper jaws). This metric was developed to be applied in real-time and allows tissue properties to be compared across a range of operating conditions within the surgical environment.

$$\bar{\dot{\sigma}} = \frac{(\sigma_i - \sigma_{i-n})/n\Delta t}{(x_i - x_{i-m})/m\Delta t} \quad (2)$$

Where: σ = Applied Stress (kPa)
x = Displacement (mm)
t = Time (s)

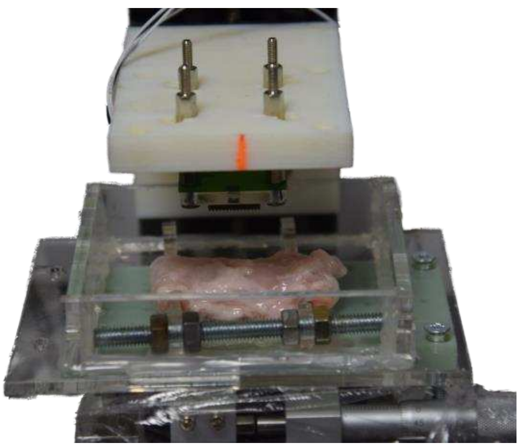
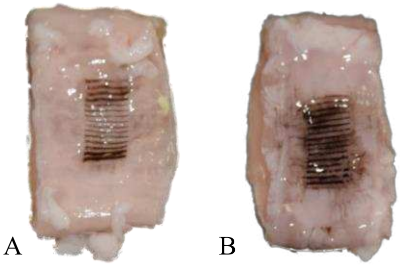


Figure 4. Photo of the tissue test setup, with porcine colon sample inserted.

E. Experimental Procedure

An experimental protocol was designed to parametrically investigate the effect of different grasper operating conditions on the above metrics. Applied strain and grasping speed (strain rate) were identified as key variables within surgical environments and accordingly varied to replicate different conditions within a surgery. Results were compared to previous work reported in the literature to assess the efficacy of the soft sensor for the real-time analysis of tissue health.

An experimental matrix of three strains (0.2, 0.4, 0.6) and three strain rates (2, 4, 6 $mm.s^{-1}$) was assessed, with five repeats for each condition. Samples of porcine colon, cut to obtain tubular sections of approximately 20mm in width, were stored in PBS prior to testing. Samples were then spread to flat with a mounting jig and mounted into the test system (Fig. 3). The grasper faces were oriented circumferentially across the whole section, so as to replicate the conditions of surgical grasping. Photographs of the samples were taken post-analysis where visible deformation was present on the surface, enhanced by the application of India ink. (Fig. 5).



A B

Figure 5. Damage caused by the grasper face to a colon tissue sample after a linear compression, highlighted with India ink, (a) low applied strain, lower visible deformation, and (b) high applied strain, higher visible deformation

III. RESULTS

A. Sensor Characterization

Results from hysteresis testing were obtained across 10 cycles. Fig. 6a shows the output of the sensor during a typical loading-unloading cycle, revealing a maximum hysteresis error in the Z- and Y-Axes of 4.5% (1.58N) and 4.3% (0.31N of the maximum applied forces (35N and 7N respectively).

The examination of the sensing performance is shown in Fig. 6b. A sample of silicone was inserted into the test setup (Fig. 4)

B. Tissue Application

The *ex vivo* study was completed successfully, using tissue samples from two porcine colons. A total of 360 grasps were recorded. Across the range of strains and speeds, the sensor can be seen to measure pressures up to around 120kPa at a rate of 500Hz. The output data was analyzed using a computational analysis package (MATLAB, Mathworks), and metrics were calculated for both Percentage Relaxation and Normalized Stress Rate of each sample.

For Tissue Relaxation, significant correlations were found between the relaxation percentage and both applied strain and repeated grasps (Pearson Correlation, $r = 0.548$ $p<0.001$ and $r = -0.610$ $p<0.001$, respectively) (Fig. 7a). The measurements also became more precise with multiple grasps indicated by a reduction in Standard Error from 1% to a plateau of around 0.4%.

The Normalized Stress rate metric indicated an opposite relationship, with the magnitude increasing with repeated grasps. The characteristics of this metric were investigated at a threshold of 40kPa in the high strain (60%) cases (Fig. 7b). This enables a direct comparison of a specific condition between the cases. A Pearson Correlation indicated a relationship between the Normalized Stress Rate and the number of repeated grasps ($r = 0.332$ $p<0.001$). No significant relationship was observed between Normalized Stress Rate and Speed.

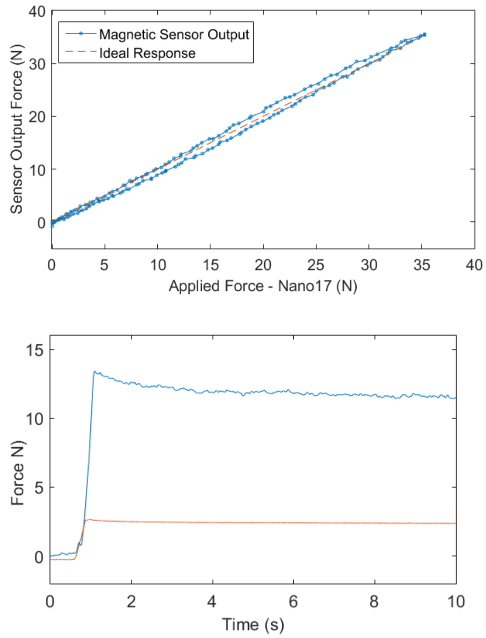


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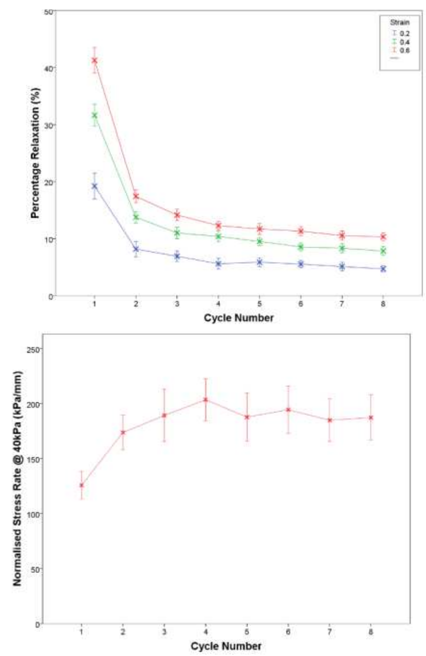
Figure 6. (a) Hysteresis error of the sensing grasper face (b) Data obtained in the grape crush, indicating little change in shear forces under compression *(Fy)* and a smooth compressive profile *(Fz)*. (c) The grape in the compression rig.



A

Figure 7. The mechanical response of tissue over multiple grasps, (a) normalized stress relaxation over 8 repeated cycles, of applied strains 0.2,0.4, and 0.6, (b) normalized stress rate over 8 repeated grasps for a 6mm/s loading rate and 0.6 applied strain.

IV. Discussion

This paper presents a new approach for real-time haptic sensing in surgical graspers and its application to prevent tissue trauma. The MF-based soft tactile sensor was based on a commercially available MEMS chip to give a low cost and simple to manufacture sensor. It consisted of a hard grasping face with embedded magnet, a soft silicone substrate, and a three-axis Hall Effect sensor. The sensor was designed to replicate a surgical grasper face, for potential incorporation to a laparoscopic device in the future. The simple design and low cost allow for the sensor to be disposable, giving benefits in a surgical environment.

The MF tactile sensor was assessed for its potential for data acquisition within surgery. The sensor successfully replicated both the compressive and shear forces a sensor would need to obtain in a surgical environment. The linearity and hysteresis were examined to assess the performance of the sensor under a loading sequence. The force response was linear across the full range of applied force. Although the mechanical response of the silicone is generally non-linear, the robustness of the calibration algorithm gives a low-noise linear response [31]. Hysteresis Error was relatively high in comparison to commercial load cells, as would be expected due to the non-linear viscoelastic response of the silicone substrate. This was deemed to be within acceptable limits for this application, complemented by the low cost of the MF tactile sensor. The sampling speed is also high, with I2C bandwidth sufficient to support frequencies of 500Hz or greater.

The ex-vivo tissue study allowed us to investigate the sensor's effectiveness for real-time surgical use. Through the evaluation of real-time tissue analysis metrics we were able to compare the application of the sensor to that of an external load cell. For both relaxation and Normalized Stress Rate, similar trends were observed to those obtained by Chandler [26]. However, there was a significant increase in measurement noise in comparison to that of the commercial load cell used in the previous study. This is likely to reduce the sensitivity of the metrics and may therefore limit clinical efficacy.

Future optimization of the sensing system may help to address some of these concerns. Wang proposes future adaptations to the MF sensing method through the use of multiple coupled Hall Effect sensors. This would enable sensor measurements to be based on a combination of measurement signals, thus reducing environmental noise and cross-coupling effects. A second consideration is a conversion of this single node sensor to a sensor array. The multiple sensors would allow a 2D force map of the surface of the grasper face to be obtained, similar to that shown by Hammond et al.[23].

Future work on this system will explore both sensor optimization and development and validation of tissue trauma metrics which encompass both normal and shear loads, building on the preliminary work reported here. An important step here will be to couple mechanical measures with clinically accepted measures of tissue trauma (e.g. histological analyses) to better assess the relationship between real-time variance in the material properties of tissue and subsequent damage.

V. Conclusion

In this paper we aimed to produce a methodology for the use of soft sensing for real-time tissue trauma assessment in MIS. Magnetic field based sensing was identified as a potential sensing method to be integrated into the face of a surgical grasper. The prototype sensor was assessed on its ability to collect data for the real-time assessment of tissue properties. The experiments demonstrated that the sensor had potential to fulfil these requirements, although future optimization is required to enhance functionality. The outcomes of this work have relevance in both haptic and surgical research fields, allowing surgeons and robotic systems to perform safer surgeries and increase patient benefit.

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