

MiniTac: An Ultra-Compact 8 mm Vision-Based Tactile Sensor for Enhanced Palpation in Robot-Assisted Minimally Invasive Surgery

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Abstract—Robot-assisted minimally invasive surgery (RAMIS) provides substantial benefits over traditional open and laparoscopic methods. However, a significant limitation of RAMIS is the surgeon's inability to palpate tissues, a crucial technique for examining tissue properties and detecting abnormalities, restricting the widespread adoption of RAMIS. To overcome this obstacle, we introduce MiniTac, a novel vision-based tactile sensor with an ultra-compact cross-sectional diameter of 8 mm, designed for seamless integration into mainstream RAMIS devices, particularly the Da Vinci surgical systems. MiniTac features a novel mechanoresponsive photonic elastomer membrane that changes color distribution under varying contact pressures. This color change is captured by an embedded miniature camera, allowing MiniTac to detect tumors both on the tissue surface and in deeper layers typically obscured from endoscopic view. MiniTac's efficacy has been rigorously tested on both phantoms and ex-vivo tissues. By leveraging advanced mechanoresponsive photonic materials, MiniTac represents a significant advancement in integrating tactile sensing into RAMIS, potentially expanding its applicability to a wider array of clinical scenarios that currently rely on traditional surgical approaches.

Index Terms—Robot-assisted minimally invasive surgery, miniature tactile sensor, tumor detection

I. INTRODUCTION

ROBOT-ASSISTED minimally invasive surgery (RAMIS) represents a pivotal advancement in surgical methodologies, providing a plethora of benefits. It not only ensures smaller incisions, accelerated patient recovery, diminished postoperative pain, and a reduced risk of infection compared to traditional open surgeries but also surpasses laparoscopic surgeries in terms of enhanced 3D visualization and superior precision and control over surgical instruments [1, 2].

Despite these advantages, a critical limitation of RAMIS is the absence of tactile feedback, which restricts the surgeon's

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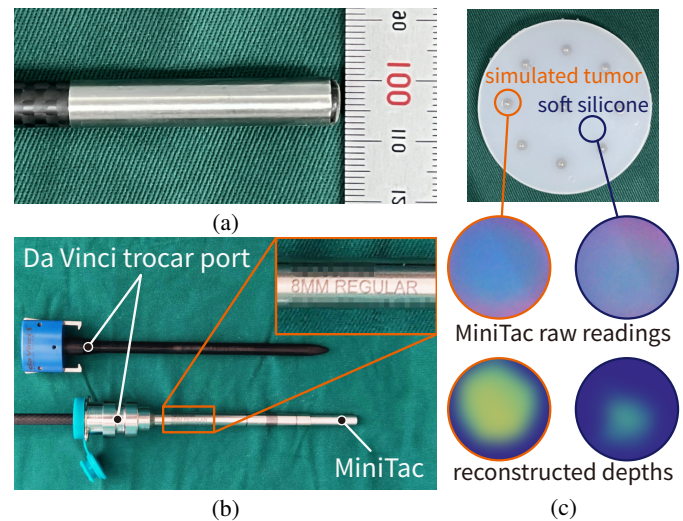


Fig. 1: MiniTac is compatible with Da Vinci robotic surgical systems for palpation. (a) MiniTac features an ultra-compact design with a cross-sectional diameter of 8 mm, suitable for standard trocar ports. (b) Its design ensures seamless integration with Da Vinci systems, commonly used in RAMIS. (c) MiniTac provides high spatial resolution and sensitivity, which facilitates detailed deformation information at the contact site, enabling precise detection of hard tumors within soft, simulated normal tissue.

ability to perform palpation [1–3]. In both open and laparoscopic surgeries, surgeons palpate the patient's soft tissues using their fingers or surgical probes to identify underlying anatomical structures invisible to the naked eye or endoscopic cameras [3]. The tactile feedback from this kind of manipulation is crucial for accurately localizing tumors, as solid tumors can be distinguished from normal tissue by their increased firmness [4, 5], which is vital for ensuring minimal resection margins [1]. The lack of this capability in RAMIS significantly curtails its application across a broader range of surgical scenarios [6].

In addressing this limitation, there has been significant interest in integrating tactile sensors into RAMIS systems [7, 8]. However, the trocar port size—typically an 8 mm diameter circle, as is standard in the widely-used Da Vinci robotic surgical systems [9]—presents a unique challenge. This challenge involves developing a sensor that not only provides high-density tactile sensing but is also compact enough for insertion through the port. Despite extensive research, current tactile sensing solutions continue to struggle with achieving miniaturization while maintaining a sufficient number of taxels (sensing units) within the limited space (reviewed in detail in Sec. II). This challenge has hindered the integration of tactile sensors into RAMIS systems and limited their overall impact. The lack of a commercially available solution further

highlights the complexity of this challenge.

In this paper, we introduce MiniTac, a novel vision-based tactile sensor specifically designed for RAMIS systems. The ultra-compact sensor has a diameter of 8 mm (Fig. 1(a)), ideally suited to fit through standard trocar ports of Da Vinci robotic surgical systems (Fig. 1(b)). MiniTac features a mechanoresponsive photonic elastomer membrane that alters its color distribution in response to different levels of contact [10]. This altered color distribution is captured by a miniature camera at a high spatial resolution of 10 μ m, endowing the device with sensitivity akin to having 300,000 taxels across its surface. This enables MiniTac to accurately map detailed deformations at the contact site and reliably identify simulated embedded tumors within the contact region, as depicted in Fig. 1(c).

The two principal contributions of this work are as follows:

- We present MiniTac, an innovative design for miniature tactile sensors with high-density taxels. This design offers RAMIS systems nuanced tactile sensing capabilities. As a result, it becomes feasible to palpate tissue surface for tumor detection during RAMIS procedures.
- We demonstrate the functionality of MiniTac using both phantoms and ex-vivo tissue samples. Our tests show clear distinctions in sensor readings between normal and pathological tissue areas, confirming the MiniTac's effectiveness.

The remainder of the paper is organized as follows: Sec. II reviews relevant literature. Sec. III introduces MiniTac, detailing its design objectives, fabrication processes, calibration methods, and sensitivity. Sec. IV validates the capabilities of MiniTac in detecting hard embedded tumors in both phantom models and ex-vivo tissue. In Sec. V, we discuss the unique features and advantages of MiniTac, emphasizing its potential to enhance palpation-equivalent capabilities in RAMIS as compared to other tactile sensors. We conclude with key findings and implications in Sec. VI.

II. RELATED WORK

Research into enhancing RAMIS systems with palpation capabilities continues amid substantial general progress within the field of RAMIS. Traditionally, surgeons depend on advanced stereo vision feedback as offered by RAMIS systems such as the Da Vinci robotic surgical system [1]. These systems enable surgeons to infer tissue characteristics by observing the deformations resulting from contact. However, this method demands extensive training, and the accuracy of the outcomes can vary significantly depending on the surgeon's experience level. As a result, there is a growing desire among surgeons for the integration of augmented haptic feedback.

Tactile sensors designed to measure contact properties have shown promise in various robotic applications. However, their integration into RAMIS systems poses unique challenges, principally due to the confined operational space. The sensing technologies currently used in RAMIS are typically piezoresistive [11], piezoelectric [12], capacitive [13], or optical-fiber-based [14]. These technologies generally provide measurements at a single or a few points, limiting spatial resolution and effectiveness for tasks requiring the examination of large

areas. Although rolling mechanisms have been developed to extend coverage [3], they increase the complexity and duration of surgical procedures.

Emerging technologies in electronic skins (E-skins) leverage advancements in materials science, nanotechnology, and electronics to offer flexible, high-resolution, and multi-modal tactile sensing [15]. Despite their potential, E-skins are known to have durability issues and their production is subject to complex manufacturing processes and high costs.

Vision-based tactile sensors utilize an embedded camera to capture optical changes of a front soft membrane, decoding its deformation and providing tactile feedback. These sensors offer extremely high spatial resolution at the pixel level, straightforward manufacturing processes, and low costs. Based on their working principles, they can be broadly categorized into photometric stereo types [16–20], darkness mapping types [21], binocular stereo types [22], and compound-eye stereo types [23]. Extensive research has demonstrated that these sensors can precisely detect object texture, hardness, and surface roughness, closely mirroring the efficacy of manual palpation [24–26]. Indeed, their effectiveness in identifying simulated embedded tumors has been demonstrated, highlighting their potential in medical applications [27–29]. However, these sensors tend to be bulky within the context of RAMIS, posing significant challenges for their integration into RAMIS systems. The characteristics of these sensors are summarized in Tab. I.

Structurally colored materials present a promising approach for developing miniature vision-based tactile sensors due to their minimal illumination requirements [10, 15, 30, 31]. Our research capitalizes on one type of structurally colored materials called mechanoresponsive photonic materials [10] and introduces a concise and effective illumination strategy, facilitating the miniaturization of vision-based tactile sensors to millimeter scales while ensuring extensive coverage of the effective measurement area. Our device, termed MiniTac, replaces traditional surround-illumination systems with a simplified white lighting setup positioned directly above the sensor's outermost layer. This innovative design enables the compact construction of vision-based sensors without sacrificing the scope of sensing regions. The performance of MiniTac in detecting tumors has been rigorously confirmed through tests on both phantoms and ex-vivo tissues.

TABLE I: Comparison of compact high-resolution tactile sensors: P.S. (Photometric Stereo), D.M. (Darkness Mapping), B.S. (Binocular Stereo), C.S. (Compound-Eye Stereo), and M.P.M. (Mechanoresponsive Photonic Material).

| Sensor | Working Principle | Illumination (Channel) | Dimension (mm ²) ↓ | Sensing Region (mm ²) | SR/D [†] |
|-----------------------------|-------------------|------------------------|-----------------------------------|-----------------------------------|-------------------|
| GelSight [16] | P.S. | 3–RGB | 35 × 35 | 18 × 14 | 0.21 |
| GelSight-Mini [17] | P.S. | 3–RGB | 32 × 28 | 19 × 15 | 0.32 |
| DIGIT [18] | P.S. | 3–RGB | 20 × 27 | 19 × 16 | 0.56 |
| GelSlim3.0 [19] | P.S. | 3–RGB | 80 × 37 | 675 | 0.23 |
| L ³ F-TOUCH [20] | P.S. | 3–RGB | 40 × 25 | 20 × 14 | 0.28 |
| 9DTact [21] | D.M. | 1–White | 32.5 × 25.5 | 24 × 18 | 0.52 |
| GelStereo2.0 [22] | B.S. | 1–White | 30 × 30 | 23 × 23 | 0.59 |
| CompdVision [23] | C.S. | 1–White | 22 × 14 | 170 | 0.55 |
| MiniTac | M.P.M. | 1–White | $\pi \times 4^2$ | $\pi \times 3.5^2$ | 0.77 |

III. THE MINITAC SENSOR

This section outlines the developmental process for MiniTac, starting with design specifications that would ensure compatibility with RAMIS, as detailed in Sec. III-A. We then describe the design and fabrication processes developed to meet these specifications—see Sec. III-B. This leads to a discussion about the calibration method employed to correlate raw sensor readings with deformation data at the contact site, as outlined in Sec. III-C. Finally, we test MiniTac's sensitivity, as detailed in Sec. III-D, and its repeatability and hysteresis in Sec. III-E.

A. Design Objectives and Specifications

The primary objective is to facilitate effective robotic palpation during RAMIS for tumor detection. A critical requirement is that the sensor has to fit through trocar ports typically used in robotic surgical systems. Its cross-sectional diameter therefore has to be within the dimensions of the trocar's diameter, which, using the Da Vinci system as a guideline, means it can be no more than 8 mm.

Apart from size constraints, the ideal sensor must be highly sensitive to variations in deformation, as tumors are generally harder than the normal surrounding tissue [24]. It should also provide high spatial resolution to enhance the efficacy of palpation. A further consideration is that any parts of the tactile sensor that come into contact with human tissue must be constructed from soft materials to minimize the risk of injury during palpation.

B. Design and Fabrication Details

We carefully design and manufacture the MiniTac sensor to meet the given objectives and specifications, as illustrated in the exploded view and schematic diagram—see Fig. 2(a) and Fig. 2(b). Consideration is given to the design and selection of MiniTac's components, as each element will have an impact on the overarching goal.

Mechanoresponsive photonic membrane and its supporting layer: Taking into account the stringent requirements in terms of physical dimensions and sensitivity to deformation, we opt for a mechanoresponsive photonic membrane, as illustrated in Miller *et al.* [10]. Unlike conventional color-changeable membranes [16], which necessitate complex and bulky lighting systems, this membrane boasts a remarkable feature: on deformation, it is susceptible to changes in reflection wavelength, leading to color change even under basic illumination. Images of the undeformed and deformed membrane are depicted in Fig. 3(a) and (b), respectively. We refer the readers to the original paper [10] for details of the principles that underlie the workings of this mechanoresponsive photonic membrane.

To enhance visualization of the color change, we introduce the concept of tactile imprint, representing the difference between images of deformed and undeformed membranes, denoted as \mathcal{I}_w and \mathcal{I}_n respectively. This imprint is further amplified by a constant factor α . The function f that yields the augmented tactile imprint is defined as follows:

$$f(\mathcal{I}_n, \mathcal{I}_w) = g(\alpha(\mathcal{I}_w - \mathcal{I}_n) + \beta), \quad (1)$$

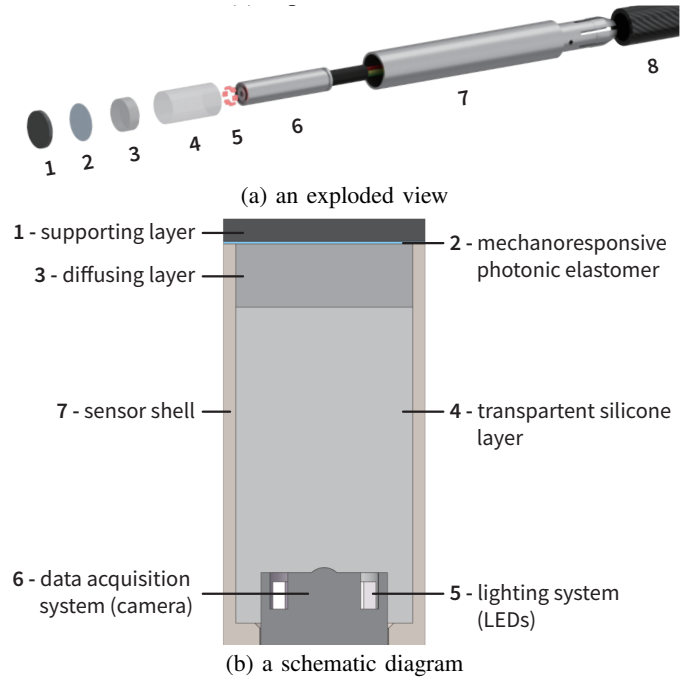


Fig. 2: **Mechanical design of MiniTac.** (a) An exploded view of MiniTac, highlighting its components. Notably, component 8 is a carbon cube acting as an extended handle to ensure compatibility with RAMIS. (b) A schematic diagram of the fully assembled MiniTac is presented, showing how the components fit together.

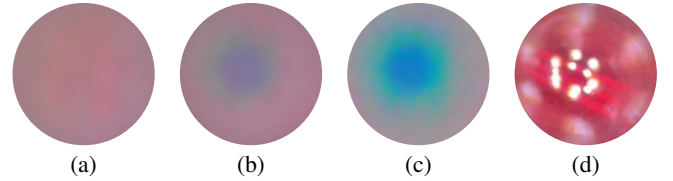


Fig. 3: **Properties of the mechanoresponsive photonic membrane.** (a) The membrane appears red when exposed to a red laser but prior to any contact. (b) Upon contact, the membrane undergoes a color change due to deformation. (c) For enhanced visualization, an augmented tactile imprint is created by applying the function f defined in Eq. (1), using a specific setting of $\alpha = 5$. (d) The membrane's speckled surface results from reflective glare when under illumination, necessitating the use of diffused lighting to reduce the effect.

where β is an offset, typically set to $\frac{255}{2}$, to convert negative differences to positive values, and g is a clipping function ensuring the value falls within the image range $[0, 255]$, so as to be compatible with the RGB value range for visualization. The augmented tactile imprint of the contact shown in Fig. 3(b) can be seen, more clearly, in Fig. 3(c).

To fabricate the membrane, we use a commercially available holographic photopolymer, Litholo C-RT20, and employ the Lippman photographic technique. Initially, we vertically expose the holographic photopolymer membrane, which has been peeled off from its initial support and layered onto a stainless-steel mirror sheet, in a darkroom under a 5 mW red laser at a distance of 40 cm for 5 min. Following this exposure, the film exhibits a uniform red pattern generated from standing waves, resulting from the reflection of the red laser on the mirror sheet. These waves capture structural color patterns as

periodic refractive index variations (distributed Bragg reflectors) within the material. The material then undergoes plasma surface treatment for 1 min to enhance its surface adhesion. Recognizing the limited strength of the thin membrane, we utilize black silicone DOWSIL 700 as the soft backing layer to improve durability. The backing layer is laminated onto the treated side of the membrane, and the mixture is then cured in an oven at 70 °C for 2 hours.

Data acquisition system: The ability to monitor subtle color changes in the mechanoresponsive photonic membrane at high spatial resolution is critical. For this purpose, we utilize an off-the-shelf miniaturized camera module equipped with an OV9734 CMOS image sensor. This sensor offers a resolution of 1280×720 pixels, which is ideal for capturing the detailed images necessary for accurate analysis. Taking into account the ultra-compact size requirements of our sensor, the camera is paired with a lens that offers a 76° field of view (FOV), chosen to minimize optical distortion while fitting within the limited cross-sectional diameter of the sensor.

To maximize the fidelity of the color capture, the camera operates at full resolution using the uncompressed YUV format, which ensures that the images are detailed and free from compression artifacts. This setup not only enhances sensitivity to subtle color changes, but also contributes to the production of clearer, less noisy images. Additionally, the incorporation of USB connectivity in the data acquisition system allows for easy integration with standard computing devices, ensuring versatility and ease of use in a variety of applications.

Lighting system and diffusing layer: The mechanoresponsive photonic membrane, essential for our sensor's functionality, shifts its reflection wavelength upon contact. To capture this change accurately, we use six white surface-mounted LEDs arranged in a circular pattern around the camera lens, ensuring comprehensive coverage across all wavelengths.

Because the membrane's reflective surface can cause significant glare, (see illustration in Fig. 3(d)), we incorporate a diffusing layer made of translucent silicone, specifically Posilicone DRSGJ02, positioned directly behind the membrane. The chosen material not only surpasses traditional methods such as diffuser paper in terms of light distribution uniformity but also adheres naturally to the photonic membrane. This adhesive property is beneficial as it mitigates the use of additional bonding agents whose constituent chemicals could interact with the membrane causing color alterations or other unwanted issues. The translucent quality of the silicone ensures even light distribution, enhancing the sensor's accuracy in detecting and responding to color changes.

Transparent silicone layer: To ensure robust support and optimal functionality within the sensor's assembly, a transparent silicone layer, made from Smooth-on Solaris, is positioned between the camera and the diffusing layer. This material is prepared by mixing equal quantities of Part A and Part B and is then degassed using a vacuum pump and poured directly above the camera within the sensor's metal shell to form a stable base.

Once cured at room temperature after 24 hours, a second layer of Posilicone DRSGJ02 silicone is mixed in the same 1:1 ratio, degassed using a vacuum pump, and poured over

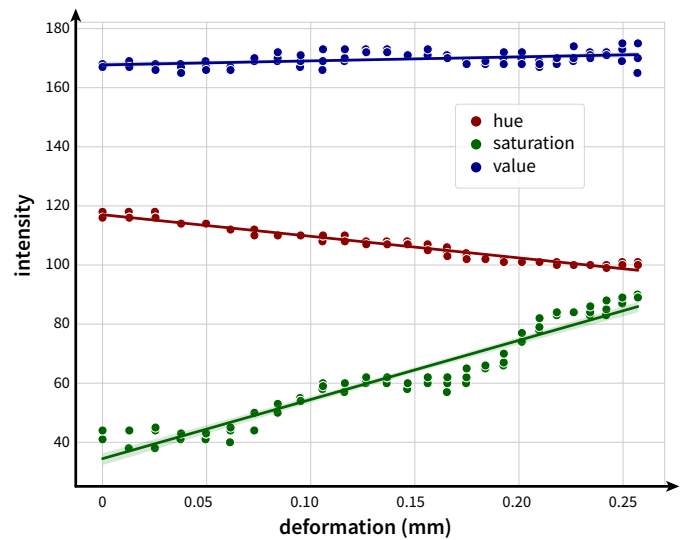


Fig. 4: **Relationship between membrane deformation and its color.** The mechanoresponsive photonic membrane exhibits color changes that correlate directly with its deformation, providing a visual indication of mechanical stress.

the base to create the diffusing layer, which helps to ensure uniform light distribution. The mechanoresponsive photonic membrane and its support structure are then placed on top of this layer, and the entire assembly is left to cure for 12 hours at room temperature. This step-by-step process ensures that all components are chemically bonded, enhancing the sensor's structural integrity and functional performance with clear optical paths and effective light diffusion.

Sensor shell and extended handle: To enhance the sensor's robustness and durability, we designed a metal shell with an outer diameter of 8 mm and a minimal wall thickness of 0.5 mm, targeting a large sensing region during RAMIS. This shell securely houses the camera module as shown in Fig. 2. The lower end of the shell is attached to a carbon tube via a custom fixture, which acts as an extended handle. The design strengthens the sensor's structure while enhancing its usability and ergonomic handling, ultimately facilitating its seamless integration into RAMIS systems.

C. Calibration

This section details the calibration process for MiniTac, which is crucial to ensure that it can extract accurate deformation data from sensor readings. We focus on converting pixel color changes into surface deformation data, utilizing the HSV color space as recommended by Miller *et al.* [10]. This color space has been shown to correlate strongly with the deformation of mechanoresponsive photonic elastomers.

We analyze and illustrate these correlations to understand the relationship between membrane deformation and the HSV channels, as shown in Fig. 4. Ideally, a uniform membrane color would enable straightforward mapping from color variation ($\Delta H, \Delta S, \Delta V$) to surface deformation D using a single function. However, real-world variations in membrane color, as observed in Fig. 3, necessitate a more complex approach. To address this, we introduce positional factors into our calibration model. We define a function m that maps

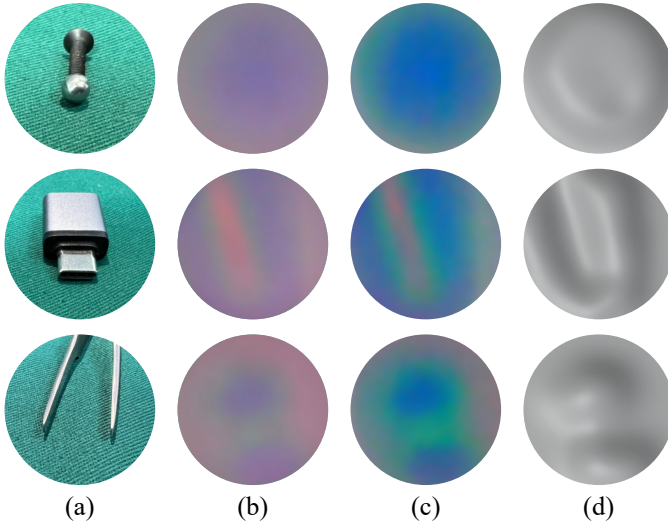


Fig. 5: **Deformation reconstruction at the contact site obtained by MiniTac.** MiniTac analyzes color shifts within captured images to accurately reconstruct surface deformation at the contact site, offering high spatial resolution: (a) test objects (b) raw readings from MiniTac (c) augmented tactile imprints (d) final reconstructed deformations.

color changes ($\Delta H(u, v)$, $\Delta S(u, v)$, $\Delta V(u, v)$) at each pixel position (u, v) to the corresponding deformation depth. The function is mathematically formulated as follows:

$$D(u, v) = m(\Delta H(u, v), \Delta S(u, v), \Delta V(u, v), u, v). \quad (2)$$

For the practical implementation of this function, we use an multilayer perceptron (MLP) to approximate the function m . Following the recommendation by Wang *et al.* [32], we adopt a three-layer architecture (5-32-32-32-1), which has proven effective. This setup utilizes the hyperbolic tangent (tanh) activation function, the Adam optimizer with a learning rate of 0.001, and mean squared error (MSE) as the loss function. To train the network, we generate a dataset by applying a known-sized metal sphere to the sensor surface and recording the resulting color changes at the contact site. A dataset comprising 30 captures is sufficient to train a single sensor.

Fig. 5 presents qualitative results, illustrating the reconstructed sensor surface deformation from various kinds of contact, demonstrating our calibration method's efficacy.

D. Sensor Sensitivity

Once calibrated, MiniTac accurately measures surface deformations at the contact site. Its sensitivity to external force is critical to its ability to distinguish between soft and hard objects. Indeed the key capabilities required for robotic applications can be summarized as follows:

- 1) The sensor must be able to detect deformations caused by minimal force in order to discern slight differences in hardness.
- 2) It should identify incremental deformation differences caused by minor increases in force, enabling differentiation between similar hardness levels.
- 3) The sensor should not reach saturation unless subjected to a relatively large force, so it retains the ability to discern substantial variations in hardness.

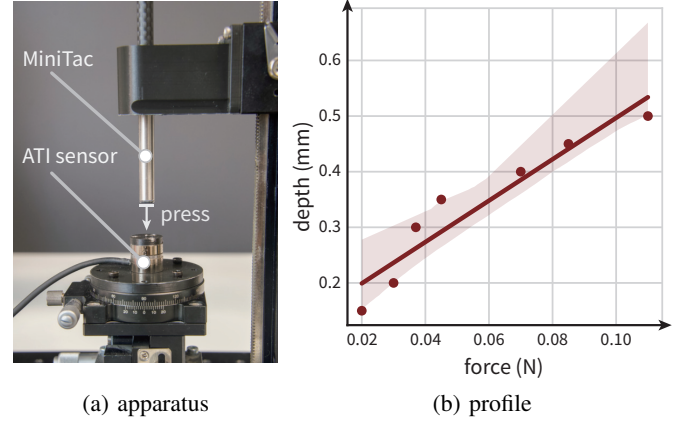


Fig. 6: **Sensitivity of MiniTac.** (a) The apparatus, equipped with an indenter, is used to measure deformation under varying external forces recorded by an ATI force sensor. (b) MiniTac detects deformation starting from 0.02 N, resolves force increments at approximately 0.6 mN, and remains unsaturated under force of up to 0.11 N.

To evaluate the sensitivity of MiniTac, we use an ATI Nano17 force sensor to apply controlled pressure to the sensor's surface and record its readings (Fig. 6(a)). As shown in Fig. 6(b), the sensor demonstrates high sensitivity, is capable of detecting deformations at 0.02 N, can resolve force increments at approximately 0.6 mN, and does not saturate under forces up to 0.11 N.

E. Sensor Repeatability and Hysteresis

With the same setting in Fig. 6(a), the repeatability of MiniTac r is calculated as:

$$r = \frac{\Delta \hat{d}_{tM}}{d_M} \times 100\% = \frac{0.11}{0.5} = 22\%, \quad (3)$$

where $d_M = 0.5$ mm is the maximum ground truth depth applied in this study and $\Delta \hat{d}_{tM} = 0.11$ mm is the maximum measurement difference at the same step across all trials. The standard deviation of the difference image captured before loading and after unloading is 0.7.

The red curve in Fig. 7 is the smoothed average of the five trials measured above. Based on this curve, the maximum measurement difference between two processes $\Delta \hat{d}_{pM}$ is observed as 0.19 mm, and the hysteresis h is calculated as

$$h = \frac{\Delta \hat{d}_{pM}}{d_M} \times 100\% = \frac{0.19}{0.5} = 38\%. \quad (4)$$

Given that the loading process yields output values higher than those of the unloading process, it is likely that this discrepancy is due to the viscoelastic properties of the elastomer.

IV. EXPERIMENTS

Initially, we evaluate the efficacy of MiniTac in detecting hard embedded tumors using phantoms (Sec. IV-A), establishing a decision boundary so as to distinguish tumor tissue from normal tissue. Further validation is then performed ex-vivo to verify the efficacy of MiniTac as well as the established decision boundary (Sec. IV-B).

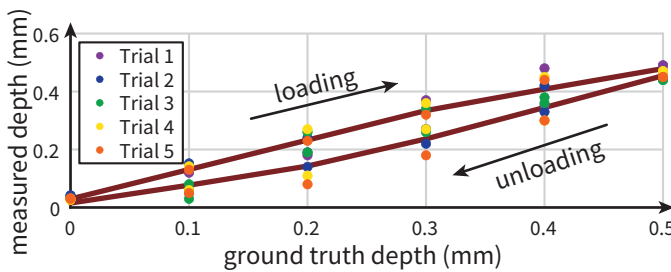


Fig. 7: **Repeatability and hysteresis properties of MiniTac.**

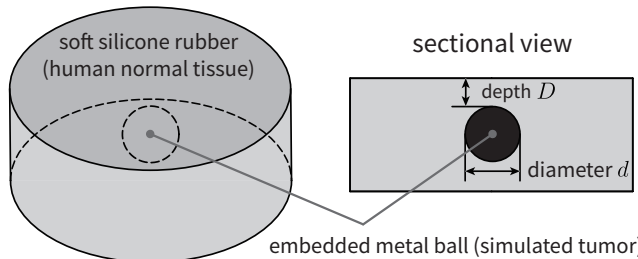
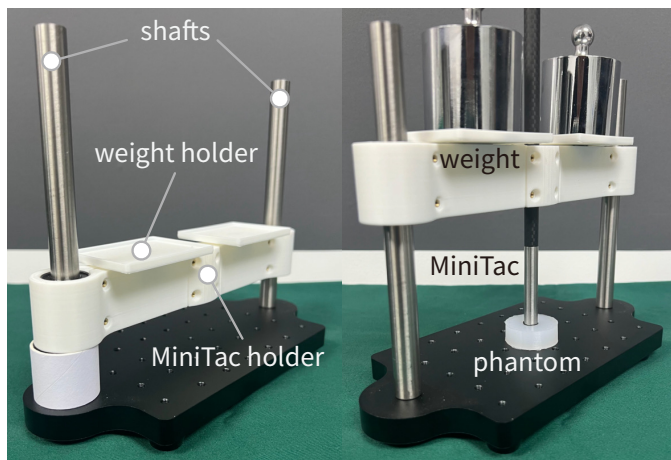


Fig. 8: **Illustration of phantoms.** Phantoms made of soft silicone rubber mimic human tissue—in this case, a metal ball of diameter d is embedded at depth D to represent a hard tumor.



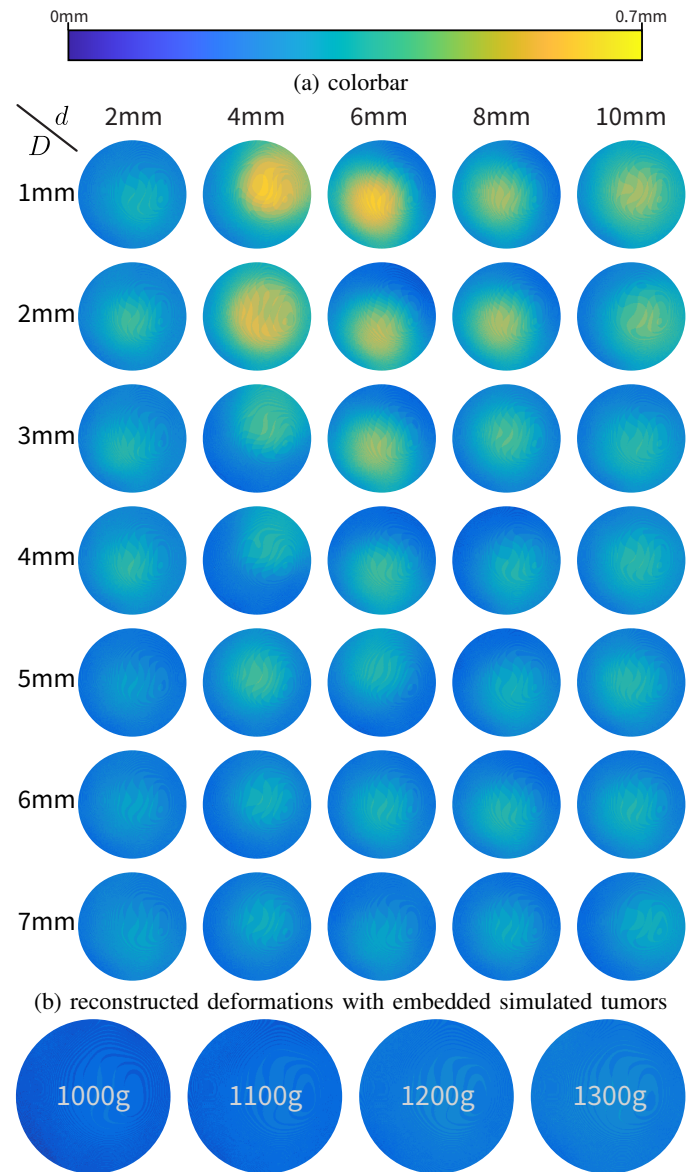
(a) apparatus

(b) experiment with MiniTac



(c) various phantoms

Fig. 9: **Experimental setup for MiniTac to detect embedded simulated tumors in phantoms.** (a) Side view of the device used to regulate the applied force. (b) MiniTac applying a consistent force of 1000 g to the phantom. (c) Various phantoms are used to establish the discrimination boundary between those with and without embedded simulated tumors.



(c) reconstructed deformations without embedded simulated tumors

Fig. 10: **Reconstructed deformations of MiniTac when in contact with phantoms.** (a) Colorbar indicating the mapping from color to deformation value. (b) Reconstructed deformations obtained by applying a consistent force of 1000 g to a selection of phantoms with embedded simulated tumors. (c) Reconstructed deformations obtained by applying varying forces, from 1000 g to 1300 g, to phantoms without embedded simulated tumors.

A. Experiments on Phantoms

We begin by evaluating the efficacy of MiniTac in detecting embedded hard tumors using controlled phantoms, aiming to establish a decision boundary between healthy and pathological (tumor-laden) tissue.

To create phantoms that closely resemble human tissue, we use Smooth-On Ecoflex 00-30 silicone rubber, which has a Shore 00 hardness of 30, similar to that of human tissue. The methodology follows the guidelines described in [27]. To assess the performance of MiniTac in a variety of tumor scenarios, we embed metal balls of assorted diameters d across different depths D , simulating a range of embedded tumor

situations. The phantom setup is illustrated in Fig. 8.

To establish a robust decision boundary between phantoms with and without simulated tumors, we collect samples under two distinct conditions:

- **Positive samples:** Phantoms are embedded with metal balls ranging in diameter from 2 mm to 10 mm and placed at depths from 1 mm to 7 mm at 1 mm intervals. We apply a consistent force of 1000 g on these samples, repeating the process four times for each. This relatively small force is chosen to generate fewer tactile features, thereby increasing the classification challenge to fashion a robust classifier. This force level is within the range that humans apply during manual palpation without causing discomfort.
- **Negative samples:** Phantoms without embedded metal balls are subjected to variable forces ranging from 1000 g to 1300 g at 100 g intervals. These increased forces, as compared to those used for positive samples, are intended to generate more tactile features for data augmentation, further complicating the classification task and further ensuring the robustness of the classifier. To balance the sample distribution, we apply each force to the empty phantom 35 times to equalize the number of positive and negative samples.

To precisely control the force applied during data collection, we construct a specialized device as depicted in Fig. 9. It features a platform that securely holds the MiniTac in place and also allows for the placement of weights to regulate the applied force. The movement of the platform is constrained by two smooth vertical shafts, ensuring that the force is directed strictly downwards.

One example for each scenario is showcased in Fig. 10(b) for positive samples and Fig. 10(c) for negative samples, respectively. The data from positive samples (Fig. 10(b)) distinctly indicate the presence of embedded tumors, with larger diameters and shallower depths resulting in more pronounced deformation signatures. In contrast, Fig. 10(c) demonstrates that deformations in phantoms without tumors tend to have flatter features due to the uniformity of the phantom material.

Based on the collected data, we initially divided it into training and test sets in a 4 : 1 ratio. We then employ a support vector machine (SVM) classifier with a linear kernel to distinguish between positive and negative samples, using the mean (μ) and standard deviation (σ) of the deformation data as features. We apply standardization to normalize these features to zero mean and unit variance. The classifier achieves 100% accuracy in both training and test sets. The decision boundary defined by the SVM is given by:

$$0.33z(\mu) + 4.80z(\sigma) = -4.53. \quad (5)$$

B. Experiments on Ex-Vivo Tissues

To further validate the performance of MiniTac and the robustness of the established decision boundary, we conduct tests on ex-vivo tissue. This setup introduces a more realistic setting that better simulates clinical conditions.

Fig. 11 illustrates our ex-vivo testing process. In Fig. 11(a), we show an ex-vivo tissue sample with areas containing and lacking an embedded tumor, as identified by experienced

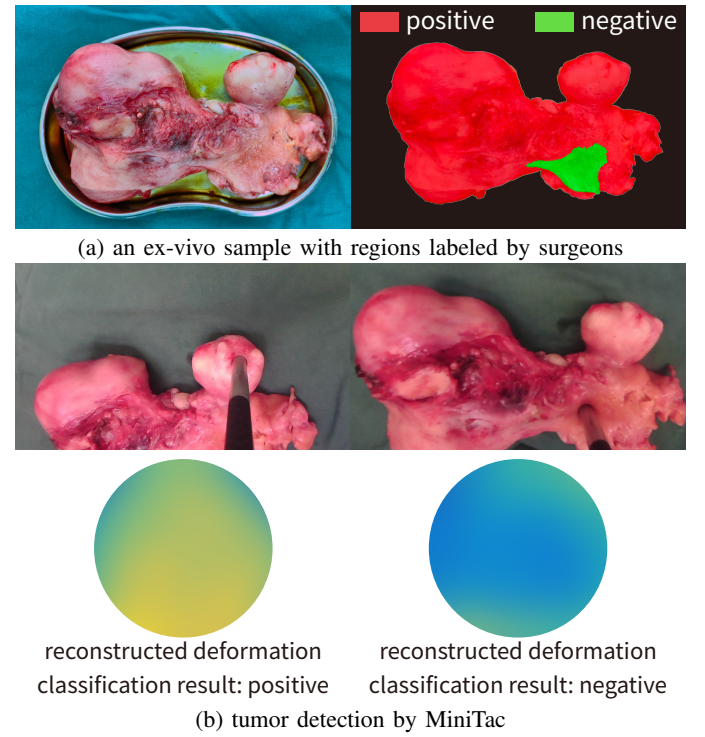


Fig. 11: **Effectiveness of MiniTac in detecting embedded tumors in ex-vivo human tissue.** (a) An ex-vivo sample with regions labeled by surgeons, indicating the presence (positive) or absence (negative) of tumors. (b) The tumor detection results from MiniTac align with those of the surgeons, demonstrating the effectiveness of MiniTac.

surgeons through manual palpation. Using MiniTac, surgeons then performed 50 random presses across this tissue. Our established decision boundary (Eq. (5)) achieved 100% accuracy in these trials, perfectly aligning with the surgeons' ground truth assessments. Fig. 11(b) provides two representative samples of reconstructed deformation and classification results. This strong correlation between MiniTac's outputs and expert surgical evaluations demonstrates the sensor's effectiveness in accurately detecting tumors within human tissue.

V. DISCUSSION

This section highlights the unique benefits of MiniTac in enhancing palpation capabilities within RAMIS systems. Unlike traditional tactile sensors such as piezoresistive [11], piezoelectric [12], capacitive [13], and optical-fiber-based sensors [14], MiniTac integrates 300,000 taxels within a 8 mm-diameter area. This configuration allows for detailed surface mapping, effectively capturing widespread deformations.

The ability to accurately map deformations and calculate mean and variance has proven effective in the detection of hard objects such as tumors, as demonstrated in controlled phantom experiments (Sec. IV-A) and ex-vivo tissue applications (Sec. IV-B). Harder objects induce more pronounced deformations, enhancing detectability and increasing the deformation data's mean and variance. The decision boundary analysis confirms the importance of standard deviation, which remains significant even when larger forces are applied (Eq. (5)). Although similar results might be achieved with high-resolution vision-based tactile sensors such as GelSight

[27] and DIGIT [29], MiniTac's simple illumination system allows for a significantly more compact design, facilitating integration into RAMIS systems (Fig. 1).

VI. CONCLUSION

In this paper, we introduce MiniTac, an ultra-compact vision-based tactile sensor designed to enhance RAMIS by providing high-resolution tissue palpation capabilities within the confined spaces of surgical environments. With its slender 8 mm diameter, MiniTac integrates seamlessly into mainstream RAMIS systems, such as the Da Vinci robotic surgical system, effectively overcoming the traditional limitations of tactile feedback. The sensor's ability to differentiate between normal and pathological tissue (tumors) has been validated through extensive testing on both phantoms and ex-vivo samples, demonstrating its sensitivity and efficacy.

While MiniTac's ultra-compact design shows promise for RAMIS integration, further clinical trials are needed to assess its performance in live surgical settings and compatibility with established procedures. Nevertheless, MiniTac has the potential to improve surgical outcomes and advance medical robotics.

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REFERENCES

- [1] K.-W. Kwok, H. Wurdemann, A. Arezzo, A. Menciassi, and K. Althoefer, "Soft robot-assisted minimally invasive surgery and interventions: Advances and outlook," *Proceedings of the IEEE*, vol. 110, no. 7, pp. 871–892, 2022.
- [2] J. Vilsan, S. A. Maddineni, N. Ahsan, M. Mathew, N. Chilakuri, N. Yadav, E. J. Munoz, M. A. Nadeem, K. Abbas, W. Razaq *et al.*, "Open, laparoscopic, and robotic approaches to treat colorectal cancer: a comprehensive review of literature," *Cureus*, vol. 15, no. 5, 2023.
- [3] H. Liu, J. Li, X. Song, L. D. Seneviratne, and K. Althoefer, "Rolling indentation probe for tissue abnormality identification during minimally invasive surgery," *Transactions on Robotics (T-RO)*, vol. 27, no. 3, pp. 450–460, 2011.
- [4] S. Vaccarella, S. Franceschi, F. Bray, C. P. Wild, M. Plummer, and L. Dal Maso, "Worldwide thyroid-cancer epidemic? the increasing impact of overdiagnosis," *New England Journal of Medicine*, vol. 375, no. 7, pp. 614–617, 2016.
- [5] N. Riggi, M. L. Suvà, and I. Stamenkovic, "Ewing's sarcoma," *New England Journal of Medicine*, vol. 384, no. 2, pp. 154–164, 2021.
- [6] A. Gerald and S. Russo, "Soft sensing and haptics for medical procedures," *Nature Reviews Materials*, pp. 1–3, 2024.
- [7] J. Konstantinova, A. Jiang, K. Althoefer, P. Dasgupta, and T. Nanayakkara, "Implementation of tactile sensing for palpation in robot-assisted minimally invasive surgery: A review," *IEEE Sensors Journal*, vol. 14, no. 8, pp. 2490–2501, 2014.
- [8] W. Othman, Z.-H. A. Lai, C. Abril, J. S. Barajas-Gamboa, R. Corcelles, M. Kroh, and M. A. Qasaimeh, "Tactile sensing for minimally invasive surgery: Conventional methods and potential emerging tactile technologies," *Frontiers in Robotics and AI*, vol. 8, p. 705662, 2022.
- [9] I. Surgical, "Da vinci robotic surgical systems," accessed: 2024. [Online]. Available: <https://www.intuitive.com>
- [10] B. H. Miller, H. Liu, and M. Kolle, "Scalable optical manufacture of dynamic structural colour in stretchable materials," *Nature Materials*, vol. 21, no. 9, pp. 1014–1018, 2022.
- [11] M. Kalantari, M. Ramezanifard, R. Ahmadi, J. Dargahi, and J. Kovacs, "Design, fabrication, and testing of a piezoresistive hardness sensor in minimally invasive surgery," in *IEEE Haptics Symposium*. IEEE, 2010, pp. 431–437.
- [12] J. Lee, W. Choi, Y. K. Yoo, K. S. Hwang, S.-M. Lee, S. Kang, J. Kim, and J. H. Lee, "A micro-fabricated force sensor using an all thin film piezoelectric active sensor," *Sensors*, vol. 14, no. 12, pp. 22 199–22 207, 2014.
- [13] U. Kim, Y. B. Kim, J. So, D.-Y. Seok, and H. R. Choi, "Sensorized surgical forceps for robotic-assisted minimally invasive surgery," *IEEE Transactions on Industrial Electronics*, vol. 65, no. 12, pp. 9604–9613, 2018.
- [14] H. Xie, H. Liu, L. D. Seneviratne, and K. Althoefer, "An optical tactile array probe head for tissue palpation during minimally invasive surgery," *IEEE Sensors Journal*, vol. 14, no. 9, pp. 3283–3291, 2014.
- [15] Y. Luo, M. R. Abidian, J.-H. Ahn, D. Akinwande, A. M. Andrews, M. Antonietti, Z. Bao, M. Berggren, C. A. Berkey, C. J. Bettinger *et al.*, "Technology roadmap for flexible sensors," *ACS Nano*, vol. 17, no. 6, pp. 5211–5295, 2023.
- [16] W. Yuan, S. Dong, and E. H. Adelson, "Gelsight: High-resolution robot tactile sensors for estimating geometry and force," *Sensors*, vol. 17, no. 12, p. 2762, 2017.
- [17] Gelsight, "Gelsight-mini," accessed: 2024. [Online]. Available: <https://www.gelsight.com/gelsightmini/>
- [18] M. Lambeta, P.-W. Chou, S. Tian, B. Yang, B. Maloon, V. R. Most, D. Stroud, R. Santos, A. Byagowi, G. Kammerer *et al.*, "Digit: A novel design for a low-cost compact high-resolution tactile sensor with application to in-hand manipulation," *IEEE Robotics and Automation Letters (RA-L)*, vol. 5, no. 3, pp. 3838–3845, 2020.
- [19] I. H. Taylor, S. Dong, and A. Rodriguez, "Gelslim 3.0: High-resolution measurement of shape, force and slip in a compact tactile-sensing finger," in *International Conference on Robotics and Automation (ICRA)*. IEEE, 2022, pp. 10 781–10 787.
- [20] W. Li, M. Wang, J. Li, Y. Su, D. K. Jha, X. Qian, K. Althoefer, and H. Liu, "L³ F-TOUCH: A wireless gelsight with decoupled tactile and three-axis force sensing," *IEEE Robotics and Automation Letters (RA-L)*, 2023.
- [21] C. Lin, H. Zhang, J. Xu, L. Wu, and H. Xu, "9dtact: A compact vision-based tactile sensor for accurate 3d shape reconstruction and generalizable 6d force estimation," *IEEE Robotics and Automation Letters (RA-L)*, 2023.
- [22] C. Zhang, S. Cui, S. Wang, J. Hu, Y. Cai, R. Wang, and Y. Wang, "Gel-stereo 2.0: An improved gelsight sensor with multimedium refractive stereo calibration," *IEEE Transactions on Industrial Electronics*, 2023.
- [23] L. Luo, B. Zhang, Z. Peng, Y. K. Cheung, G. Zhang, Z. Li, M. Y. Wang, and H. Yu, "Compdivision: Combining near-field 3d visual and tactile sensing using a compact compound-eye imaging system," *arXiv preprint arXiv:2312.07146*, 2023.
- [24] W. Yuan, M. A. Srinivasan, and E. H. Adelson, "Estimating object hardness with a gelsight touch sensor," in *International Conference on Intelligent Robots and Systems (IROS)*. IEEE, 2016, pp. 208–215.
- [25] Y. Yan, Z. Hu, Y. Shen, and J. Pan, "Surface texture recognition by deep learning-enhanced tactile sensing," *Advanced Intelligent Systems*, vol. 4, no. 1, p. 2100076, 2022.
- [26] Z. Zhao, Y. Li, W. Li, Z. Qi, L. Ruan, Y. Zhu, and K. Althoefer, "Tac-Man: Tactile-informed prior-free manipulation of articulated objects," *arXiv preprint arXiv:2403.01694*, 2024.
- [27] X. Jia, R. Li, M. A. Srinivasan, and E. H. Adelson, "Lump detection with a gelsight sensor," in *World Haptics Conference (WHC)*. IEEE, 2013, pp. 175–179.
- [28] S. Cui, Y. Tian, J. Hu, R. Wang, Y. Wang, and S. Wang, "Prototype of a novel visuotactile endoscope probe," in *IEEE 13th International Conference on CYBER Technology in Automation, Control, and Intelligent Systems (CYBER)*. IEEE, 2023, pp. 1368–1373.
- [29] J. Di, Z. Dugonjic, W. Fu, T. Wu, R. Mercado, K. Sawyer, V. R. Most, G. Kammerer, S. Speidel, R. E. Fan *et al.*, "Using fiber optic bundles to miniaturize vision-based tactile sensors," *arXiv preprint arXiv:2403.05500*, 2024.
- [30] Y. Yue and J. P. Gong, "Tunable one-dimensional photonic crystals from soft materials," *Journal of Photochemistry and Photobiology C: Photochemistry Reviews*, vol. 23, pp. 45–67, 2015.
- [31] Y. Cho, S. Y. Lee, L. Ellerthorpe, G. Feng, G. Lin, G. Wu, J. Yin, and S. Yang, "Elastoplastic inverse opals as power-free mechanochromic sensors for force recording," *Advanced Functional Materials*, vol. 25, no. 38, pp. 6041–6049, 2015.
- [32] S. Wang, Y. She, B. Romero, and E. Adelson, "Gelsight wedge: Measuring high-resolution 3d contact geometry with a compact robot finger," in *2021 IEEE International Conference on Robotics and Automation (ICRA)*. IEEE, 2021, pp. 6468–6475.