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A Facile one-step injection novel composite sensor for robot tactile assistance

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Abstract— **Tactile information is the research hotspot of wearable flexible sensors due to its importance and complexity. With the innovation of wearable technology and robotics in healthcare, researchers are increasingly integrating wearable flexible sensors on the front end of robots to reproduce the hand tactile manipulation of human tissues. Therefore, it is hoped to develop a thin-film sensor that can be deployed in a small area to assist robots in surgery and data collection of human tissues. Here we use a one-step injection method to fabricate a novel composite sensor based on liquid metal. By laminating multiple PDMS microfluidic layers, the two parameters of pressure and deformation are measured simultaneously in a decoupled manner. The sensor is small and thin, making it easy to integrate into fingers/robot fingers for assistance. The finger/robot finger exerts pressure on the sensor and the sensor deforms with the material to identify the hardness of the material being touched. Separate performance tests of the two sensors show that the strain and pressure functions are decoupled from each other, and their ratios can identify and classify the hardness of different touched materials (glass, PDMS and silicone). This novel composite sensor we proposed can assist robots in manipulating human tissues during medical surgeries. At the same time, its function in tactile information feedback also has broad applications in medical treatment, rehabilitation and services.**

I. INTRODUCTION

The development of miniaturized robots has brought a disruptive paradigm shift to the surgical field [1]. Robot-assisted surgery can minimize the risks of traditional invasive surgeries [2]. Integrating robots with wearable sensors for surgical procedures enhances the accuracy of surgery, and the collection of information during surgery also provides guidance for postoperative recovery [3]. The robot's front end integrates tactile sensors that replicate the force and tactile sensations experienced by the surgeon's hand during surgery [4]. However, there are still many unknown aspects of bones, skin, organs, and other tissues in surgical operations [5] [6], including details of their tactile characteristics such as hardness, surface tension, contact force, and response to mechanical loads [7] [8]. The mechanical properties of a surgical subject are crucial for diagnosis and treatment, including the evaluation of their current status and recovery success rate. Therefore, the development of hardness sensors integrated on the front end of robot fingers is an important aspect of tactile sensing for medical robots [9], which can help medical robots complete surgeries and collect data more accurately and efficiently.

Current methods for measuring surface hardness and physical durability of biological soft tissues include nanoindentation [10], aspiration [11], torsion [12], and elastography [13]. These techniques can provide important insights into the mechanical properties of human soft tissue. However, human tissue remains highly hydrated and sensitive during surgery. Additionally, exsiting high-precision equipment can be bulky and require hand-holding for measurement [14] [15]. Therefore, the related devices are not suitable for integrating wearable devices on the front of robots or for non-invasive hardness measurement of sensitive tissues (such as internal organs). In recent years, researchers started using tactile sensors to detect soft materials [16]. Bao's group used the principle of metal strain sensors and fine micro-pyramid structure designs to detect samples with different degrees of softness [17] [18]. Other complete system integrates actuators and sensors to continuously detect the Young's modulus of human skin [19].

In this work, we designed a novel composite sensor that couples a pressure sensor with a strain sensor to identify the hardness of the touced materials. The sensor can both capture the applied pressure by human/robot fingers and the surface deformation of the contact object. The manufacturing process of laminate bonding and one-step injection of liquid metal (Galinstan) is described, and materials of varying hardness are tested using the sensors. When external force is applied (such as the force of the operating instrument contacting the human tissue), the sensor deforms differently based on the material it touches, allowing for the measurement of mechanical characteristics of different touched material. The sensor has great potential in robotic surgical operations due to its convenient and rapid manufacturing process, small and compact structural design, and integration of external circuits and displays. Additionally, the tactile characteristics of the contact surface softness or hardness are crucial in medical treatment, rehabilitation, and service.

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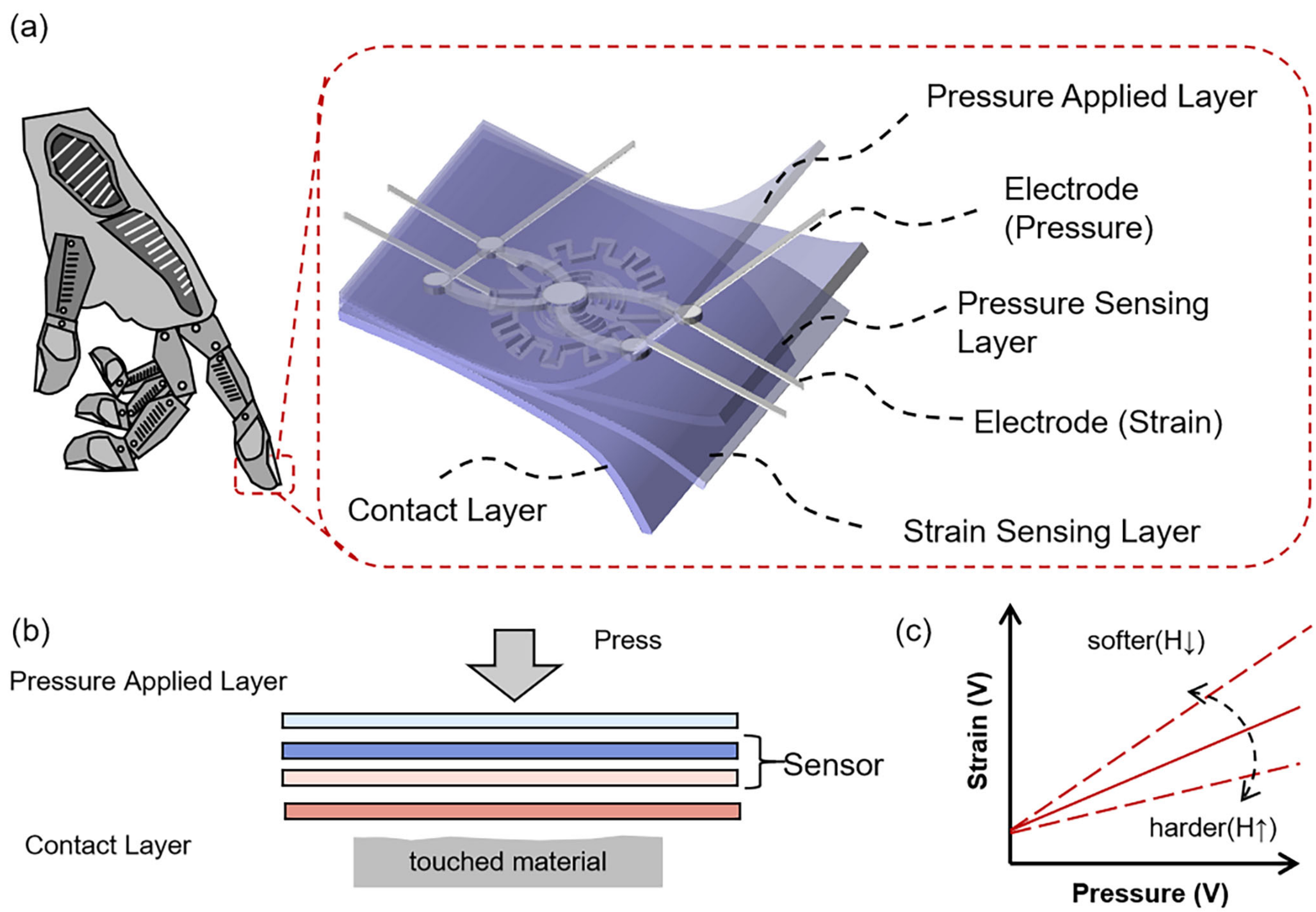

Fig. 1. Overview of the novel composite sensor (a) Exploded view of the arrayed configuration showing laminated layers (b) Schematic diagram of sensor application (c) An example output plot of the proposed sensor. The strain sensor provides y-axis data as either resistance change, while the pressure sensor provides x-axis pressure data as resistance change. The hardness of the object is indicated by the slope.

II. Design and Fabrication

A. Hardness Measurement Principle

The process of grasping or touching an will subject its surface to normal pressure. According to the bar structure model [20], normal pressure (represented by F) can be expressed as:

$$F = kx = \frac{EAx}{L} \tag{1}$$

Where k is the structural stiffness of the touched material, x is the deformation, E is Young's modulus, A is contact area, and L is length of the bar along the pressure direction.

The hardness of a touched material, which is the most intuitive feeling in tactile information (Fig 1b), depends on two parameters of the contact surface: the positive pressure exerted on the object and the deformation of the surface in response to the pressure. To obtain the hardness value H of the touched material (Fig 1c), calculate the ratio of deformation and pressure. The larger the value of H, the softer the surface of the touched material.

B. Strain Sensing and Pressure Sensing

Our novel composite sensor comprises a pressure sensor and a strain sensor arranged consecutively. The strain sensor is positioned on the lower layer and is in close proximity to the surface, enabling it to detect material deformation more accurately. Additionally, the pressure sensor is located on the upper layer and is directly affected by external forces. It also acts as an isolation layer, preventing external forces from affecting the strain sensor below.

A liquid metal flow channel was designed with a micropump structure to measure surface pressure. The pressure sensing area, which is a semicircle with a radius of 2 cm, sinks under external force, and the liquid flows to small reservoirs at the ends, causing a continuous increase in overall sensor resistance. When the load is removed, the liquid metal is pushed back into the large reservoir in the center. The sensor contains five circular liquid reservoirs that act as resistors, while the microchannels function as wires (Fig 3a). When the microchannel filled with liquid metal deforms, the resistance of the liquid metal in the channel increases due to the increase in channel length and decrease in cross-sectional area, or both [21]. To detect the deformation, we designed an equivalent Wheatstone bridge consisting of four connected sensing networks (Fig 3a). When the sensor is stretched, the resistance of the four gate networks increases. The Wheatstone bridge structure enhances the sensor's sensitivity while minimizing external interference.

C. Laminated Forming and One-step Injection

The novel composite sensor was fabricated using soft lithography and laminated bonding, as shown in Fig. 2. The process can be divided into five main steps: mask soft lithography, mold reversing, layer bonding, liquid metal injection, and wire connection. Polydimethylsiloxane (PDMS) was used as the base material due to its high flexibility and ease of casting microstructures.

The first step is to design and soft lithography the mask to make the mold (Fig 2a). The minimum width of the flow channel is 100 μm, and the thickness is 100 μm. PDMS was then prepared in a 10:1 ratio of elastomer to hardener and carefully mixed before being poured onto the molds. Each layer is dried for two hours and then demolded (Fig 2b). In the third step (Fig 2c), plasma bonding is used to laminate the layers together to form a complete sensor. Appropriate heating can make the bond between the layers tighter. Two syringes are used to inject the liquid metal (Galinstan) in the fourth step (Fig 2d). A syringe inserted at the end of the cavity can help expel air and make the injection process smoother. The final step (Fig 2e) is to connect the wires and encapsulate the entire sensor. The thickness of each layer of the sensor is approximately 7mm and the total thickness of the sensor is 25mm.

III. Result

A. Sensor Output and Characteristics

The designed novel composite sensor can be easily deployed on an object's surface, and external pressure (using a finger) can obtain the hardness of the contact object. The measurement process does not require any structural modifications to the detection surface structure or complex equipment for applying external force. After analyzing the performance of the two sensors separately, it is evident that the pressure sensor responds solely to pressure, while the strain sensor is insensitive to positive pressure and only responds to deformation.

To assess the sensor's performance under pressure, we applied a 100g weight vertically to the force-bearing area of the sensor for 4 seconds, removed it for 1 second, and repeated the process 5 times. Fig 3b illustrates the resistance change of the pressure sensor under static pressure. The resistance gradually increases during the placement process and quickly returns to its original value after the pressure is removed. By using the same method to measure the pressure sensing performance of the strain sensor and increasing the weight, it is clear from Fig 3c that the resistance of the strain sensor fluctuates narrowly under continuously increasing pressure, indicating its insensitivity to pressure. The strain performance of the sensor was ultimately characterized using

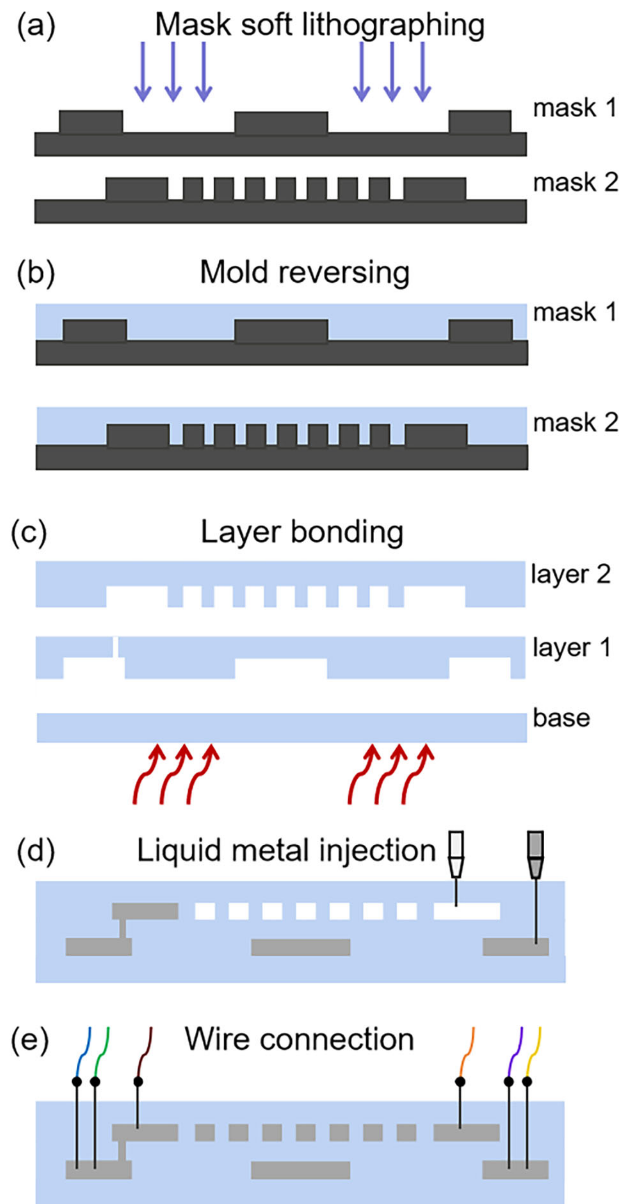

Fig. 2. Sensor manufacturing process (a) Mask soft lithographing (b) Mold reversing (c)Layer bonding (d) Liquid metal injection (e) Connect wire by inserting electrodes

a self-made in-situ stretching device (Fig 3d), where the resistance of the sensor increased linearly with the amount of stretching.

Finally, the novel composite hardness sensor was fabricated and tested. Vertical platform and force gauge used to apply force to the sensor (Fig. 4b). The contact materials chosen were glass (red), PDMS (orange) and silicone (green). The results indicate that under a certain pressure, soft materials such as silica gel deform more significantly, while hard glass hardly deforms.

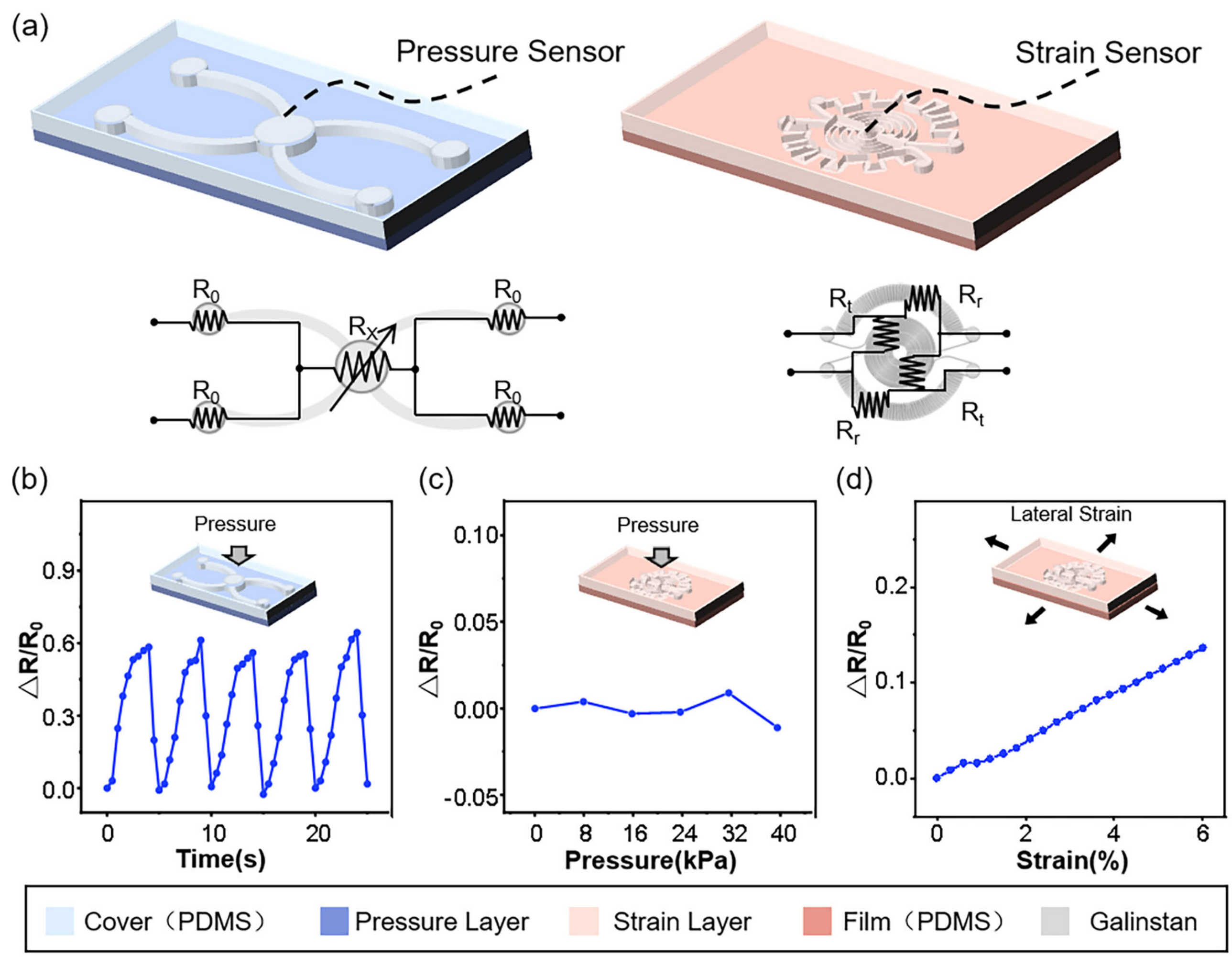

Fig. 3. Pressure sensor and strain sensor characterization (a) Illustration of pressure sensor and strain sensor. The plan shows the equivalent circuit of the sensor. (b) Pressure sensor response to 100g weights. (c) Resistance remian unchange under continuously rising pressure, which showing the insensitivity of strain sensor to pressure. (d) Resistance change of strain sensor as a functional of lateral strain.

B. System Integrated

To enable real-time monitoring in practical applications, a printed circuit board (PCB) is designed to aid in sensor detection. The test system comprises sensors, wires, and a PCB. As depicted in Fig 5b, the analog-to-digital converter in the STM32 microcontroller unit is utilized to control the OLED and display the measurement results. The schematic for reading signals from two sensor layers is also shown by the OLED. Both sensors utilize a constant current source to generate a voltage drop across the sensor. The circuit for the strain sensor utilizes a constant voltage of 2.5V to power the Wheatstone bridge. The signal is then amplified and the resistance change is measured through the analog-to-digital conversion port of the STM32.

C. Sensor on a Robot Hand

We have designed a novel composite sensor to assist medical robots and significantly improve their tactile sensing performance. The film sensor is lightweight, compact, and highly adherent, allowing us to place the thin-film sensor on the robot's finger without altering its structure. This enables the robot to obtain the hardness of the touched material by simply touching it (Fig 5a). To assess the application's feasibility, we created a 25mm thick sensor the size of a fingertip (a circle with a force area of 2cm radius). We integrated the sensor on one side of a human/robot finger and read the feedback through an external circuit. Different materials (glass, PDMS and silica) were used for detection. Fig 5c displays the sensor readings when pressing three different materials with the index finger. The experiment involved pressing the pressure sensor to three different resistance change rates (0.2, 0.4, and 0.6) and recording the corresponding strain results for three different materials.The results demonstrate the effectiveness of our novel composite sensor in detecting the hardness of touched materials.

IV. Discussion

Here we use an one-step injection method to manufacture a novel composite hardness sensor based on liquid metal

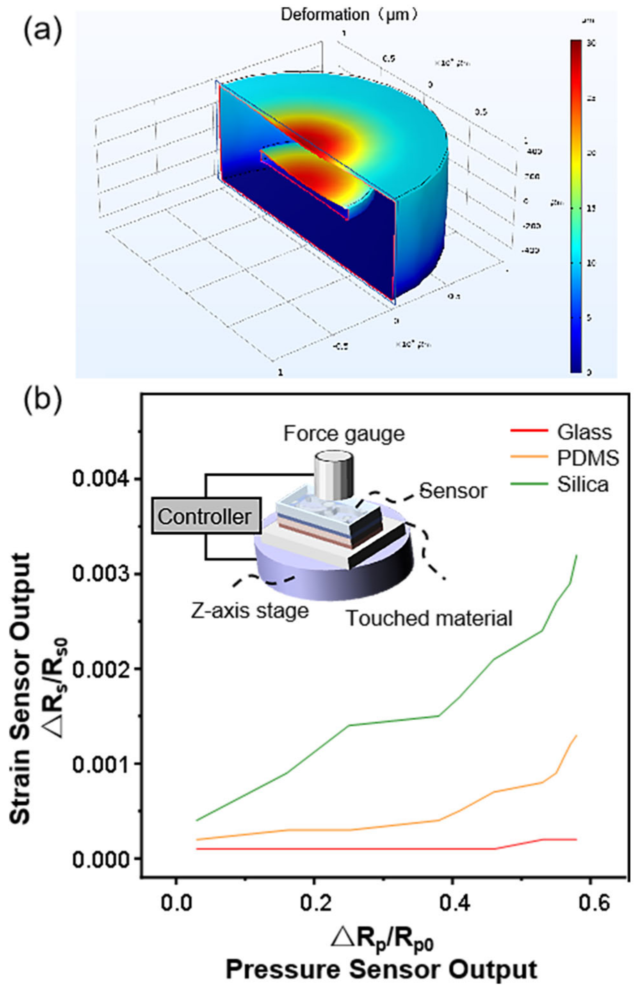

Fig. 4. Sensor application. (a) Compression deformation simulation. (b)The resistance change of the pressure sensor is represented on the x-axis, while the resistance change of the strain sensor is represented on the y-axis. These values indicate the hardness characteristics of three different materials.

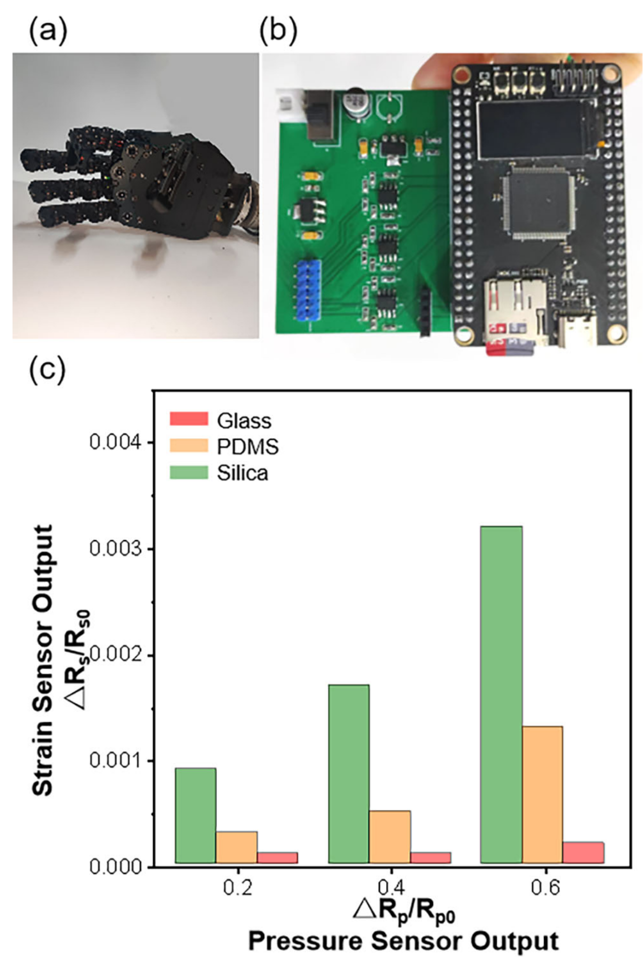

Fig. 5. Demonstration of sensor use (a) A robotic hand with novol composite sensor. (b) Physical diagrams of the test circuit. (c) Three materials were identified through the use of sensors. Use finger to press the glass, PDMS and silicone respectively, and press the pressure sensor to a resistance change rate of 0.2, 0.4, 0.6.

(Galinstan). The sensor measures both pressure and deformation parameters in a decoupled manner by laminating multiple PDMS microfluidic layers.

Previous studies have utilized various materials and sensing mechanisms to achieve two sensing functions of pressure and strain respectively. Lu's group used AgCP composites to detect strain and Ag Flake/CNT/PDMS (AFCP) water-in-oil emulsion to detect pressure [16]. Although this work achieved better sensitivity (a gauge factor of 1.45 and a force sensitivity of $0.46N^{-1}$), the manufacturing steps were too cumbersome and the problem of displacement delamination of the two sensors occurred [22]. We have constructed two different sensors using a one-step injection method. This approach allows for better discharge of air bubbles in the multi-layer flow channel, resulting in an improved success rate of sensor manufacturing. The one-step injection method has the advantage of creating two sensors as a single unit, which prevents delamination and mechanical deformation under various distortion changes, ensuring high stability. Additionally, the sensor is constructed with PDMS and has a thickness of only 25mm. It is lightweight, compact, and highly compatible. Its film-like shape allows for wearable hardness testing of robots or prosthetics without requiring any structural modifications to the robot's fingers prior to testing.

Due to the characteristics of liquid metal, such as low viscosity, high surface tension, and high conductivity, embedding it into elastomer microfluidics to form solutions has been widely studied in the field of flexible sensors in recent years. This approach typically produces pressure sensors based on multilayer capacitive [23] or resistive [24]. Here, we designed a liquid metal micropump structure to detect pressure and multiple grid-mounted structural mechanisms to detect strain. Both of them are resistive sensors, but due to their different structures and positions, they are only sensitive to pressure or stress, respectively, thereby achieving different functions based on the same conductor. A better decoupling effect can be obtained by adjusting the thickness of the PDMS by reasonably setting the layout of the pressure and strain sensors. The simulation in Figure 4a shows that the deformation of the material during the pressing process leads to changes in the liquid metal channl

We verified the sensor's ability to detect pressure and strain respectively by optimizing and characterizing the individual performance of the two sensors. A laminated structural design was used to position the strain sensor closer to the surface being touched and away from the applied pressure. This design better decouples the pressure and strain values. Next, the multi-layer PDMS microfluidic layer was laminated through plasma bonding and combined with the pressure and strain sensors to detect the hardness of the touched materials. Three materials with different hardnesses (glass, PDMS, silicone) were identified by applying finger pressure. Therefore, our proposed novel composite sensor is highly feasible and attractive for medical robot applications due to its laminated structural design and one-step injection manufacturing method.

V. Conclusion

In summary, we have proposed a novel composite sensor. The sensor is laminated with multiple layers of PDMS microfluidic channels, including a resistive pressure sensor and strain sensor. Resistance is formed by continuously injecting liquid metal into the microfluidic channels in a one-step process. To detect pressure, we have designed a liquid metal micropump structure, while multiple grid-mounted structural mechanisms are used to detect strain. Two sensors are laminated sequentially using plasma bonding, and the liquid metal is injected in a one-step continuous injection. This manufacturing method makes two sensors more closely coupled at the physical level, resulting in more stable performance. In summary, the sensor can simultaneously capture the external force and surface deformation of the touched material, finally identify the hardness of the material. Its thin-film and compact shape make it versatile, allowing for integration on the front end of a robot finger to identify material hardness through touch or pressure. To test the sensor's ability to classify materials, glass, PDMS, and silicone were used. These materials correspond to different levels of hardness found in human tissue, such as bone, skin, muscle and internal organs. Therefore, the rapid and convenient fabrication of our proposed novel composite sensor makes it highly appealing for tactile sensing applications, particularly in surgical operations and information collection of human tissues.

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