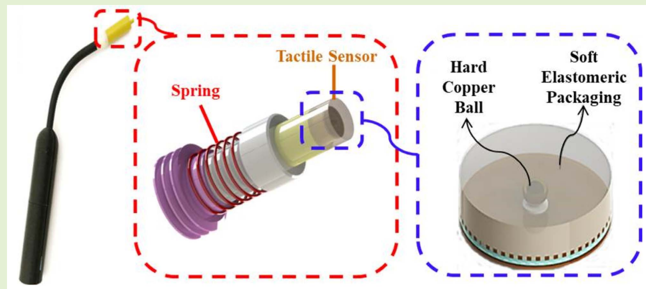
Portable Pen-Like Device With Miniaturized Tactile Sensor for Quantitative Tissue Palpation in Oral Cancer Screening

Muhammad Omar Shaikh, Chi-Mao Lin, Da-Huei Lee, Wei-Fan Chiang, *Member, IEEE*, I-Hung Chen, and Cheng-Hsin Chuang, *Member, IEEE*

***Abstract*—Oral cancer has consistently been ranked among the top ten cancers in terms of mortality with five year survival rates being among the lowest for all major cancers. Consequently, great efforts are being made to develop new diagnostic tools that can enable early detection in a clinical setting. In this study, we report on the design and development of a novel pen-like handheld device, with a miniaturized tactile sensor mounted on the front-end and integrated with a portable back-end readout module. The device can be used for tissue palpation in the oral cavity and obtain quantitative information regarding the elasticity of oral lesions and abnormalities, thus eliminating the need for manual palpation by the clinician. The sensing mechanism of the proposed tactile sensor is based on the two-spring model and employs two components with varying stiffness values, namely a hard copper ball embedded in a soft elastomeric packaging. The differential voltage output of the piezoelectric sensing film corresponding to the two components can provide information regarding the elasticity of the tissue under contact. To simulate feasibility of the tactile sensor in differentiating tissue elasticity, we have contacted the sensor with different elastomeric silicone materials under the same applied force. The experiments confirmed that the sensor response could be used to differentiate five silicones with varying Young's moduli ranging from 0.2 to 3.1 MPa and a corresponding Shore hardness of 2 to 56 Shore A. A two-stage linear trend was observed with a sensitivity of 0.356/MPa for softer silicones (2 ∼ 9 Shore A) and 0.059/MPa for harder silicones (24 ∼ 56 Shore A). Furthermore, the portable readout module that we have developed can analyze the data from the tactile sensor and transmit it wirelessly via Bluetooth for cloud computing. The proposed device is easy-to-use in general clinical settings and could enable early detection and improved prognosis of oral cancer.**



***Index Terms*—Tactile sensor, pen-like handheld device, piezoelectric, tissue elasticity.**

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I. Introduction

ORAL cancer is a deadly disease which can often cause disfigurement, impaired speech and difficulty in swallowing [1]. It is the most aggressive and malignant cancer of the head and neck region and has a poor response to chemotherapy and anticancer therapies [2]. In Taiwan, oral cancer was ranked fifth among the top ten cancers based on the mortality rate [3]. The relatively high prevalence of oral cancer in Taiwan is mainly attributed to presence of about 2.5 million high risk individuals who exhibit habits of betel nut chewing as well as cigarette smoking [4]–[6]. Unfortunately, over 50 % of new cases diagnosed already have stage 3 or stage 4 cancer lesions, resulting in a low 5-year survival rate. Diagnosis at an early stage, especially for high risk individuals, can significantly improve survival rates and enable improved personalized treatment strategies.

Since a majority of all head and neck cancers originate in the oral cavity [7], [8], conventional screening methods primarily consist of visual examination with white light and manual palpation of the oral cavity. In case of a suspicious lesion, a biopsy is taken for further diagnosis. However, early detection is still very challenging since the submucosal tumours are often invisible to the naked eye and benign oral lesions can be misdiagnosed as premalignant lesions [9]. Consequently, a range of adjunct technologies are being developed to supplement conventional methods and enable a more effective screening of the oral cavity [10]. Currently, most available technologies like ViziLite and VELscope utilize optical detection techniques such as autoflorescence, chemiluminscence and reflectance confocal microscopy [11]–[16]. However, approaches that are based solely on visual discrimination of soft tissue are susceptible to missing potential dysplastic regions and providing false positive results. Consequently, there is still a need for technologies that can provide quantitative tactile feedback which mimics manual palpation for distinguishing between healthy and cancerous tissue based on their elasticity. Furthermore, manual palpation is often subjective in nature depending on the clinician's experience, and is hindered in patients that experience difficulty opening their mouth completely, like those suffering from oral submucosal fibrosis (OSF) or those that have had facial reconstruction surgery. The ability to quantify tactile feedback in a minimally invasive manner, in addition to visual examinations, has the potential to dramatically improve reliability and accuracy of oral cancer detection and prognosis.

In the past two decades, a range of tactile sensor technologies have been proposed that are particularly aimed at improving haptic perception of the surgeon during minimally invasive surgeries [MIS]. In general, tactile sensors have been used to measure contact force/pressure [17]–[20] and detect hardness/stiffness [21]–[26] or elastic modulus [27]–[31] of soft tissues in MIS. Electrical tactile sensors utilizing transduction modes like capacitance, piezoelectricity and piezoresistivity have been mostly proposed in literature due to their simplicity, direct force measurement capability and scalability to different sizes. Electrical measurements for tissue hardness detection can be categorised as being either time-domain or frequency-domain measurements. For the first case, the sensor contact force and tissue hardness determine the degree of deformation of the tissue [32]. For the second case, the natural frequency of the tactile sensor shifts as it contacts tissues of different hardness and these sensors generally employ piezoelectric actuators for driving components like axial rods and cantilevers [33], [34]. In this study, we use time-domain measurements and propose a miniaturized tactile sensor which can be attached to the front end of a pen-like handheld device and used for quantitative palpation to measure elasticity and hardness of oral abnormalities like premalignant lesions and early-stage cancerous tumours. Furthermore, we have also developed a portable readout module that can analyse the data and transmit it wirelessly for cloud computing, thus enabling real-time tactile feedback in a clinical setting.

II. Experimental Methods

A. Sensing Mechanism

The sensing mechanism of the miniaturized tactile sensor is based on the concept of applying two-springs of varying stiffness, represented by the hard inner component (copper ball, E_1) and soft outer component (elastomeric silicone packaging, E_2) as shown in Fig. 1. As the sensor comes in contact with the test tissue under an applied normal force, a non-uniform stress distribution develops on the piezoelectric sensing film since the inner copper ball is considerably stiffer than the soft outer packaging. This results in two varying output voltages being generated at the structural electrodes corresponding to each component. When the sensor contacts soft tissues, the deformation of the outer component is significantly larger as compared to when contacting hard tissues as shown in Fig. 1b. Consequently the stress exerted on the piezoelectric film by the outer component is relatively larger when contacting soft tissues, resulting in a higher output voltage ratio (V_2/V_1). However, when the sensor contacts hard tissues, most of the applied force is transferred by the inner component, thus resulting in a lower output voltage ratio (V_2/V_1). Consequently, the ratio of the output voltages corresponding to the soft outer and hard inner components can be utilized to quantify the elasticity and hardness of the test tissue.

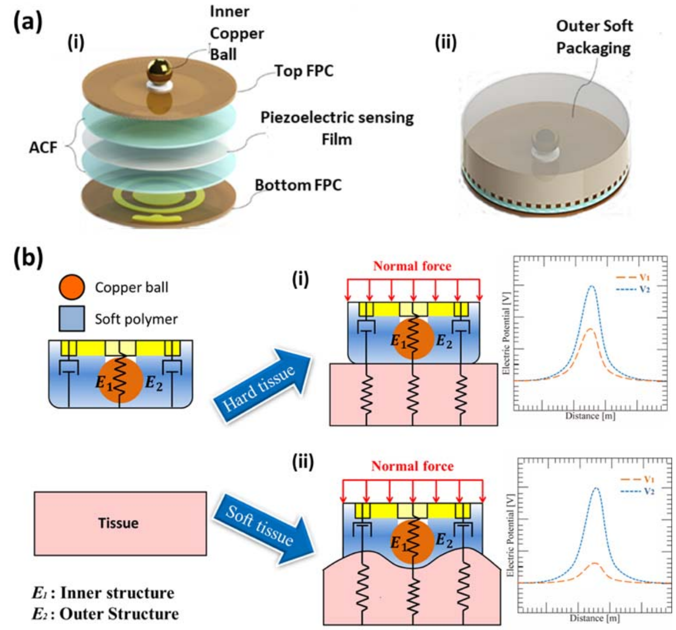

Fig. 1. (a) Schematic of (i) proposed tactile sensor with hard inner copper ball and (ii) after packaging with soft outer elastomeric polymer. (b) Sensing mechanism of the tactile sensor when contacting (i) hard and (ii) soft tissues under an applied normal force.

There have been previous reports in literature that have utilized a similar two-spring model to fabricate tactile sensors for detecting stiffness/compliance of tissues. Dargahi *et al.* designed a tactile sensor for detecting the stiffness of biological tissues by utilizing two concentric cylinders with varying moduli of elasticity [35]. Peng *et al.* performed finite element analysis to prove that an increase in sensor sensitivity can be achieved by utilizing a spring material that is harder than the test object [36]. They also developed a MEMS based capacitive tactile sensor using two sensing elements with varying stiffness for tissue elasticity and force measurements [37]. Fath El Bab *et al.* also utilized the two-spring model to fabricate a micro machined piezoresisitive tactile sensor for detecting the compliance of soft tissues [38]. While these studies show the feasibility of the concept, there

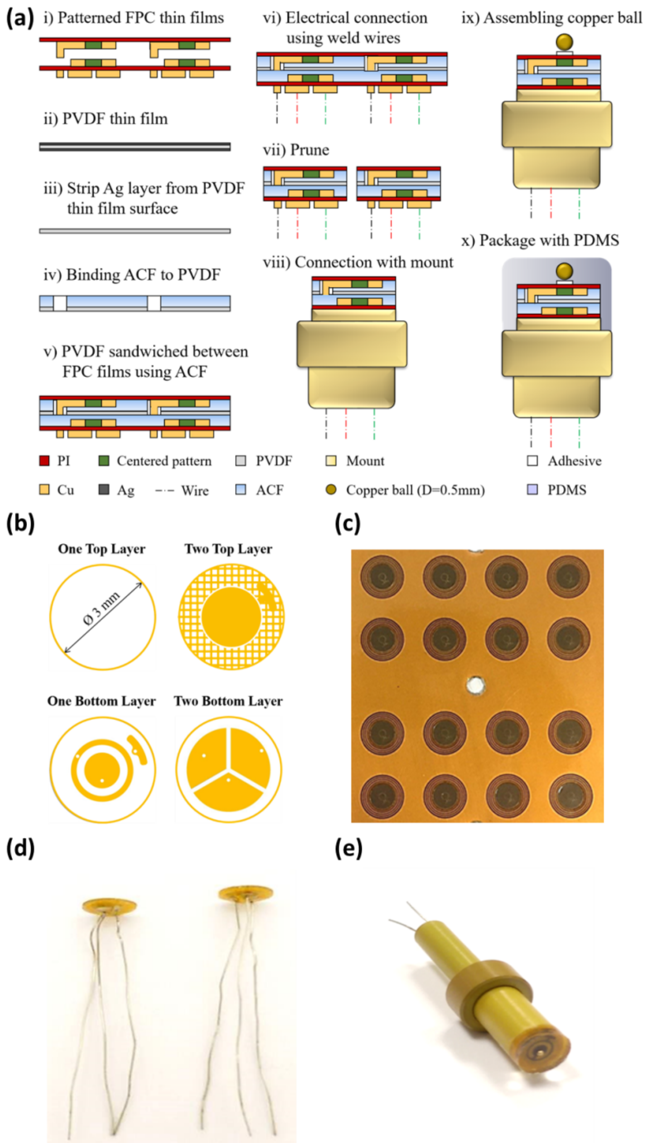

Fig. 2. Schematic of (a) systematic fabrication protocol of the tactile sensor and (b) the FPC based top and bottom structural electrodes. Photographic images of (c) Patterned FPC for fabricating 16 sensors (4 x 4 array) simultaneously, (d) completed individual tactile sensors and (e) after packaging with the plastic mount.

have been issues relating to reliability when testing tissues with larger surface irregularities. This causes an inclination angle to form between the sensor and the test tissue where one part of the sensor is in contact with the test tissue while the other part is not. We have tried to circumvent this issue by using an embedded copper ball with a diameter of only 0.5 mm (high-stiffness spring) that is surrounded on all sides by the elastomeric packaging (low-stiffness spring). Our proposed sensor design and small size ($\Theta = 3$mm) ensure improved contact with the tissue surface, thus enabling reliable detection of tissue elasticity and hardness.

B. Sensor Fabrication

The systematic fabrication protocol of the miniaturized tactile sensor is schematically illustrated in Fig. 2 and size

TABLE I
DIAMETER AND THICKNESS OF MATERIALS USED FOR THE FABRICATION OF THE TACTILE SENSOR

Materials	Diameter (mm)	Thickness (mm)
Outer Component (PDMS)	4.3	1
Inner Component (Copper Ball)	0.5	
PVDF Film	3	28×10^{-3}
ACF	3	50×10^{-3}
FPC	3	1×10^{-1}

dimensions of the materials used in sensor fabrication are listed in Table I. A standard photolithographic protocol was used to pattern the flexible printed circuit board (FPC) that consists of a polyamide (PI) film coated on both sides with copper to obtain the top and bottom electrodes for sixteen sensors simultaneously. A commercial polyvinylidene fluoride (PVDF) film with a thickness of 28 μm was utilized as the sensing layer owing to its passive sensing capability, mechanical flexibility and chemical stability. After removal of the double-sided silver coating from the PVDF film by acetone, the exposed film was sandwiched between the top and bottom patterned FPCs by utilizing an anisotropic conductive film (ACF). Conducting wires were welded to the bottom electrodes to enable electrical connection between the sensor and the external readout module. The upper surface of the top electrode was used as the reference point to manually attach the copper ball using an adhesive followed by packaging with soft medical grade Polydimethylsiloxane (PDMS) using a moulding technique. We have utilized medical grade PDMS as the packaging material since it is biocompatible and can be easily sterilized, thus increasing suitability for use in-vivo. The completed array of sixteen sensors were cut into individual pieces by using a hole-puncher and finally packaged in a plastic case for attachment to the front-end of the pen-like handheld device.

C. Sensor Fabrication

The systematic fabrication protocol of the miniaturized tactile sensor is schematically illustrated in Fig. 2 and the dimensions of the materials used in sensor fabrication are listed in Table I. A standard photolithographic protocol was used to pattern the flexible printed circuit (FPC) to obtain the top and bottom structural electrodes for sixteen individual sensors simultaneously. A commercial polyvinylidene fluoride (PVDF) film with a thickness of 28 μm was utilized as the sensing layer owing to its passive sensing capability, mechanical flexibility and chemical stability. After removal of the double-sided silver coating from the PVDF film using acetone, the exposed film was sandwiched between the top and bottom patterned FPCs by utilizing an anisotropic conductive film (ACF). Conducting wires were welded to the bottom electrodes to enable electrical connection between the sensor

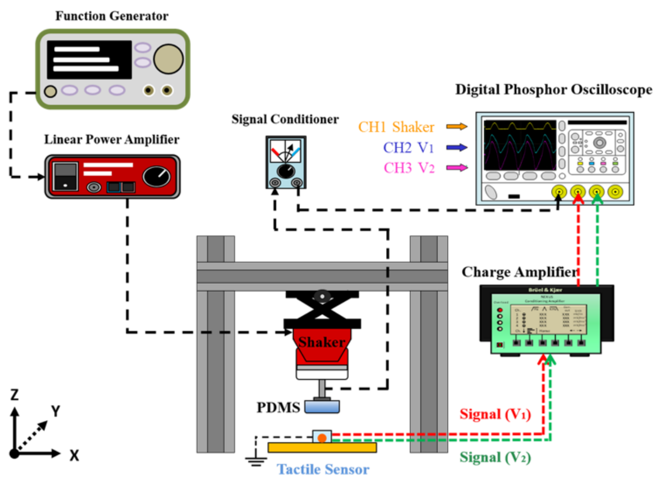

Fig. 3. Schematic of dynamic force measurement platform for analyzing the output signal from tactile sensor.

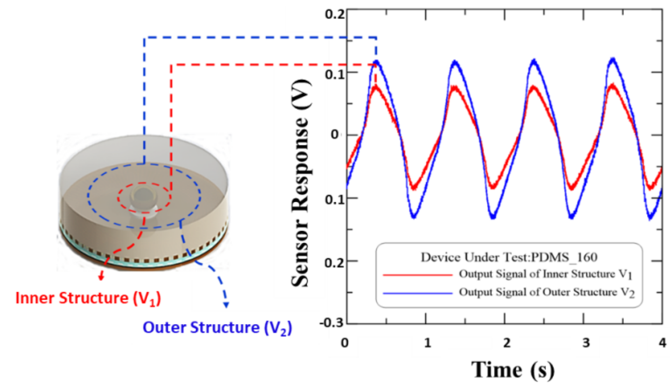

Fig. 4. The voltage waveforms corresponding to the inner (V_1) and outer (V_2) components as the PDMS 160 periodically contacts the tactile sensor under a normal applied force.

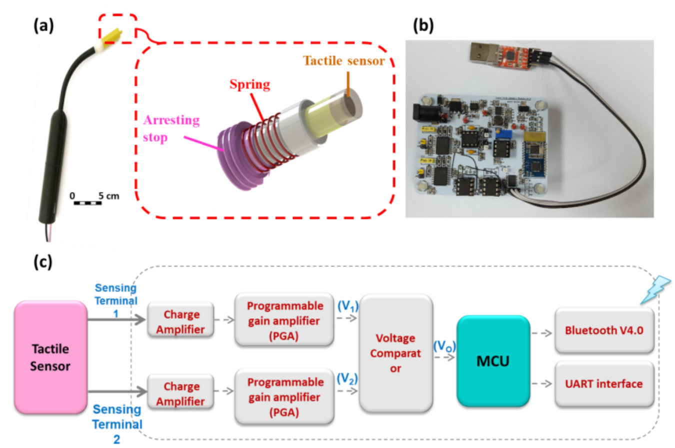

Fig. 5. (a) Schematic of the pen-like handheld device with attached miniaturized tactile sensor. A spring mechanism was used to ensure constant force between the sensor and test object. (b) An image and (c) block diagram of the prototype portable readout module.

and the external readout module. A hole-puncher was utilized to obtain the individual sensors which were then packaged in a plastic mount for attachment to the front-end of the pen-like handheld device. The upper surface of the top electrode was used as the reference point to manually attach the copper ball using an adhesive. This is followed by packaging with soft medical grade Polydimethylsiloxane (PDMS) using a moulding technique. We have utilized medical grade PDMS as the packaging material since it is biocompatible and can be easily sterilized, thus increasing suitability for use in-vivo.

D. Dynamic Force Measurement Platform

The preliminary feasibility of the fabricated tactile sensors was tested using a dynamic force sensing platform as schematically illustrated in Fig. 3. A signal generator (AFG3022, Tektronix Inc., USA) was used to drive the shaker at a frequency of 1 Hz to periodically contact the tactile sensor. The force sensor (PCB Piezotronics Inc., USA) was attached to the front-end of the shaker to control the applied force between the attached PDMS film and the tactile sensor in the range of 0.3 to 5N. The force sensor output signal was passed through a signal conditioner to the digital phosphor oscilloscope (AFG3022, Tektronix Inc., USA) via channe1 1. Additionally, the two voltage output signals from the tactile sensor (V_1 and V_2) were amplified by a charge amplifier (B&K NEXUS 2690-A) and sent to the oscilloscope via channels 2 and 3, respectively. For e.g., two voltage output waveforms were obtained when the tactile sensor was contacted periodically with PDMS 160 under a uniform applied force as shown in Fig.4.

E. Integration With Pen-Like Device and Readout Module

To create a dynamic sensing system during oral tissue palpation, the miniaturized tactile sensor was mounted on the front-end of a pen-like device and wired out to a prototype readout module that we have developed, as shown in Fig. 5. An underlying spring based mechanism, as shown in Fig. 5a, was used to apply a dynamic force on the sensor surface as it contacts with the test tissue. We have used this mechanism to ensure that the magnitude of the applied force is constant each time the sensor contacts the test tissue, thus reducing measurement errors during palpation. The magnitude of the applied force can be tuned depending on the stiffness of the spring utilized. Herein, we have compared the sensor response for three different springs with varying stiffness corresponding to an applied force of 0.3, 0.5 and 1 N.

Furthermore, we have also designed a portable prototype readout module that can analyze the voltage output from the two sensing electrodes as shown in Fig. 5b. The readout module consists of two charge amplifiers, two programmable gain amplifiers, one voltage comparator, one microcontroller unit (MCU) and one Bluetooth device as shown in the block diagram in Fig. 5c. The weak charges received from the two sensing terminals will be converted to varying voltage signals by the charge amplifiers. In order to set the voltage signal to a reasonable level, programmable gain amplifiers are used to enlarge the output voltages from the charge amplifiers, which are indicated as V_1 and V_2. The voltage comparator produces an output voltage, V_O, where $V_O = V_2/V_1$. Finally, the output

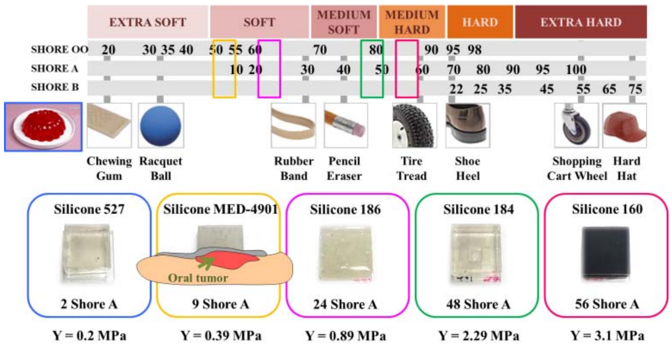

Fig. 6. Images of the five different silicone based elastomeric materials used for elasticity measurements and their corresponding Shore hardness and Young's modulus (Y) values.

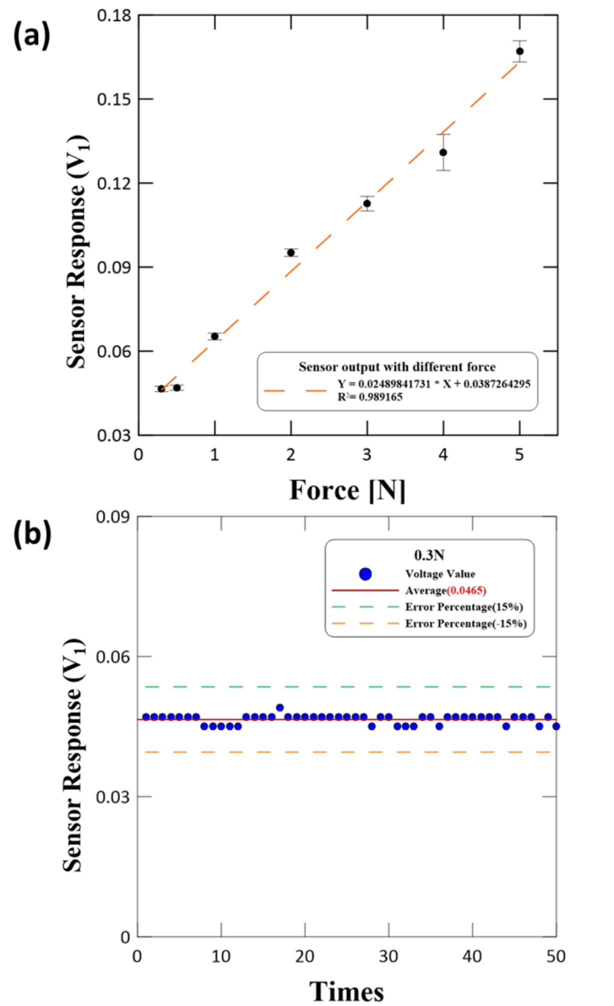

Fig. 7. (a) The sensor response observed for applied dynamic forces ranging from 0.3 to 5 N. (b) Repeatability analysis where the sensor response at each applied force (0.3 N in this case) was tested 50 times.

voltage is fed into the MCU, and converted to a digital format by the built-in analog-to-digital converter. The digital data can then be transmitted to a computer through the Universal Asynchronous Receiver/Transmitter (UART) interface or wirelessly to a smart phone by Bluetooth.

In this study, we have used the proposed pen-like device with integrated back-end readout module to test different elastomeric materials. Five silicone-based elastomers, namely Sylgard 527, Nu-Sil MED-4901, Sylgard 186, Sylgard 184 and Sylgard 160, with varying Shore hardness and Young's modulus were tested. Each silicone was prepared according to the manufacturer's suggested ratio of base and curing agents and moulded into cuboids with dimensions of $15 \times 15 \times 10$ mm^3. An image of the five silicones that have been prepared and their corresponding Shore hardness and Young's modulus values as obtained from the manufacturer are shown in Fig. 6.

III. Results and Discussion

A. Force Sensing

The force sensing capability of the proposed tactile sensor was tested using the dynamic force measurement platform.

Normal dynamic forces of 0.3, 0.5, 1, 2, 3, 4 and 5 N were applied to the sensor surface and each force was tested using 10 different sensors to ensure reproducibility. Furthermore, the sensor response at each applied force was tested 50 times to ensure repeatability. For force sensing, we only utilize the voltage output corresponding to the inner component (V_1) as the sensor response. Since the copper ball is significantly harder than the soft outer elastomeric packaging, it acts as a force concentrator, especially at higher applied forces, and can effectively transfer the force to the PVDF sensing film.

Consequently, a higher linearity in sensor response can be obtained over a wider force range simply by using the voltage output corresponding to the inner component. As seen in Fig. 7a, the sensor response shows a linear dependence on the applied force with an observed sensitivity of about 25 mV/N and a reproducible response as demonstrated by the short error bars. Furthermore, the sensor response shows high repeatability with an error value within $\pm 15\%$, as shown in Fig. 7b, for an applied force of 0.3 N for 50 cycles. Consequently, these results clearly demonstrate the feasibility of the proposed tactile sensor for reliably differentiating applied dynamic forces.

B. Testing of Elastomers

The developed pen-like handheld device was used to contact five different silicone elastomers under a constant applied force of 0.3, 0.5 and 1 N, owing to the in-built spring mechanism, and the resulting sensor response (V_2/V_1) was calculated by the prototype readout module. To ensure reproducibility, the average sensor response observed for each elastomer was obtained using 10 different sensors as shown in Fig. 8. The sensor can successfully differentiate between the five elastomers which have a Young's modulus ranging from 0.2 to 3.1 MPa and a corresponding Shore hardness ranging from 2 to 56 Shore A. The sensor response displays a two-stage linear response to the Young's modulus with a higher sensitivity observed for softer silicones as compared to harder silicones. While the sensor response follows a similar trend for each contact force, the highest reproducibility was observed for a contact force of 1 N as demonstrated by the shorter error bars in Fig 8c.

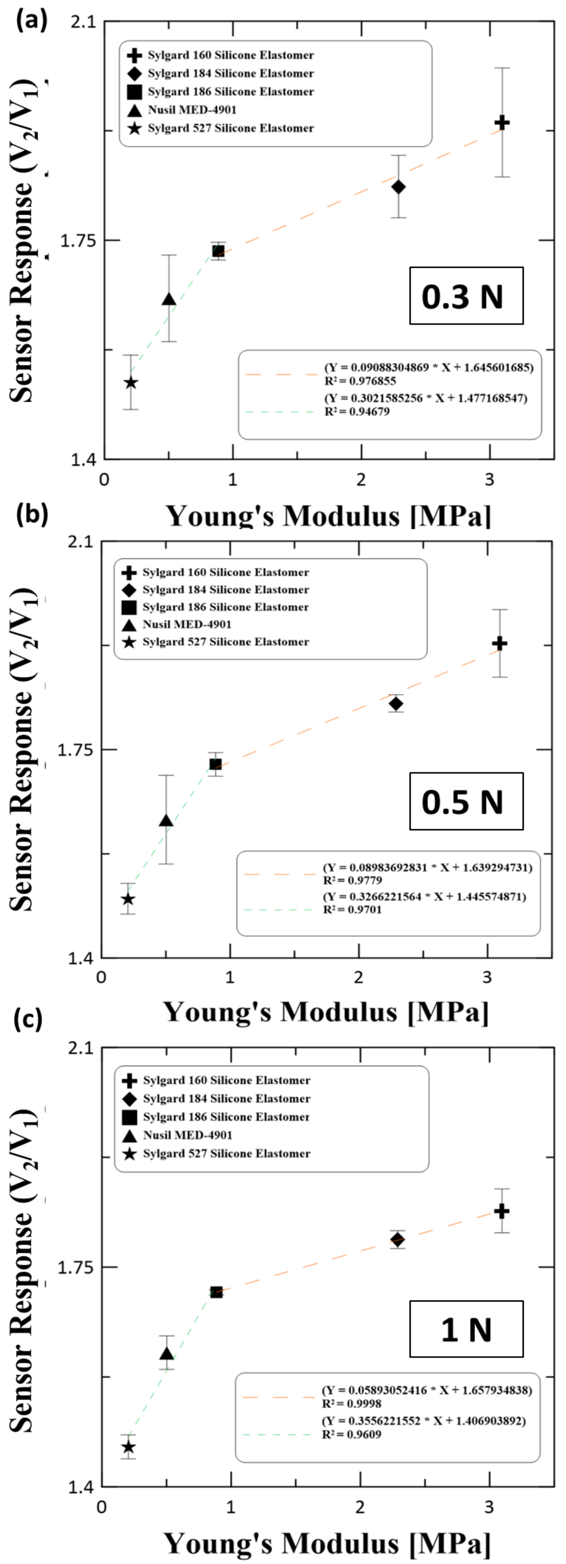

Fig. 8. Sensor response when contacted with different silicones under a constant applied force of (a) 0.3 N, (b) 0.5 N and (c) 1 N.

The elasticity of human soft tissues, including the oral mucosa, is typically in the kilopascal range (KPa) [39]. However, squamous cell carcinomas or oral cancer tumours are known to be statistically harder than the normal mucosa [40]. While our sensor shows higher sensitivity for detecting softer and compliant elastomers (Young's modulus of 0.2-0.89 MPa), it can also reproducibly detect harder and stiffer elastomers (Young's modulus up to about 3 MPa), thus covering the complete clinical range for detecting cancerous tumors. Consequently, the proposed pen-like device with integrated sensor shows practical feasibility for oral cancer screening far more precisely and quantitatively than with the human finger alone.

IV. Conclusions

In summary, we have developed a pen-like handheld device with integrated miniaturized tactile sensor and backend readout module for quantitative tissue palpation in oral cancer screening. The differential voltage output of the piezoelectric sensing film, corresponding to the two components with varying stiffness, was used to determine elasticity of the test sample. The sensor response (V_1) demonstrated a linear dependence on the applied dynamic force in the range of 0.3 to 5 N with high repeatability. Furthermore, the sensor response (V_2/V_1) was used to successfully differentiate between elastomers with varying Young's moduli. Our future work involves clinical testing to ensure reliable detection of oral abnormalities and submucosal tumors in human patients. Based on the preliminary results, the proposed device shows good practical applicability for quantitative palpation of the oral cavity in a clinical setting and can aid in early diagnosis of oral cancer.

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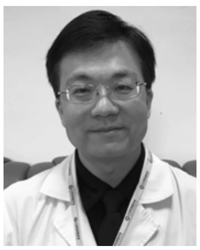

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