Sensors and Actuators A xxx (2016) xxx–xxx

Contents lists available at ScienceDirect

Sensors and Actuators A: Physical

journal homepage: www.elsevier.com/locate/sna

Piezoelectric tactile sensor for submucosal tumor detection in endoscopy

Cheng-Hsin Chuang[a,*], Tsan-Hsiu Li[a], I-Chinms Chou[b], Ying-Juimr Teng[b]

[a] Department of Mechanical Engineering, Southern Taiwan University of Science and Technology, Tainan 71005, Taiwan
[b] Kuang Tai Metal Industrial Co. Ltd., Kaohsiung, Taiwan

ARTICLE INFO

Article history:
Received 30 September 2015
Received in revised form 6 April 2016
Accepted 6 April 2016
Available online xxx

Keywords:
Tactile sensor
Piezoelectric
Endoscopy
Stiffness

ABSTRACT

In this study, we have fabricated a miniaturized piezoelectric tactile sensor (Ø = 1.4 mm) that is suitable for mounting on an endoscope to detect submucosal tumors. The sensing mechanism is based on a tandem spring model and consists of two components with varying stiffness, namely a hard inner structure embedded in a soft outer packaging. The voltage output of the PVDF sensing film is proportional to the localized normal stress exerted by each component and this differential output can be used to extract information about the elasticity of the test object/biological tissue. The sensor design has been tested by embedding different elastomers and artificial tumors in a pig's stomach for simulating the conditions of submucosal tumors in humans and shows good agreement with theoretical analysis and numerical simulations. The sensor response is proportional to the Young's modulus of test sample over a range of 1.01–3.51 MPa, making it suitable for detection of tumors present in soft tissues. The proposed miniaturized tactile sensor utilizes a passive sensing element, is wired to the external readout system through the metal catheter which can be inserted into the endoscopic channel, is safe for insertion into the stomach as it has a biocompatible packaging and can thus be utilized to provide tactile information during regular endoscopy.

1. Introduction

Cancer typically manifests as hard lumps (tumors) in the soft tissues of the stomach, breast, prostate, lungs and other body organs [1]. Many studies have also demonstrated that tumors are significantly stiffer than the surrounding tissue [2–4]. Submucosal tumors (SMT) in clinical terminology are protuberant lesions or bumps covered with intact mucosa and their etiology varies from non neo-plastic lesions to true neoplasia [5,6]. SMT of the stomach are difficult to diagnose at an early stage using luminal endoscopy or barium radiography as these lesions originate in the muscularis mucosa or submucosa instead of the surface of gastric mucosa and visual detection may not be sufficient. Usually, endoscopic ultrasonography (EUS) can easily show whether a submucosal lesion is a tumor and is considered the most accurate imaging technique for the diagnosis of gastrointestinal SMT [7,8]. However, the facility of EUS is not usually available in low resource settings due to its high set up cost. We propose an alternative way to detect SMT based on direct contact with a miniaturized tactile sensor that is compatible with traditional endoscopy for the detection of suspicious lesions in the upper gastrointestinal tract. Tactile sensors are devices that can measure certain properties of an object or contact event through physical contact between the sensor and the object [9] and can be categorized based on the transduction mechanism that they utilize such as capacitive sensing [10], piezoelectricity [11], piezo resistivity [12] and pneumatic-based methods [13] to name a few. During MIS, much of the tactile information that is normally available during open surgery is lost. Real time measurement of elasticity or stiffness of biological tissues would help the surgeon to locate blood vessels, determine the health and type of a tissue or identify cancerous tumors whose mechanical properties are known to differ substantially from healthy tissue.

Recently, many researchers have presented different medical tactile sensors with a variety of designs and principles to measure the tactile information during MIS. Sokhanva et al. [14] proposed a piezoelectric tactile sensor (size: $22 \times 4 \times 0.8$ mm) for tissue characterization by attaching three PVDF films to two stiff supporting beams and a flexible beam to acquire the voltage output as a mea-

* Corresponding author.
E-mail address: chchuang@mail.stust.edu.tw (C.-H. Chuang).

http://dx.doi.org/10.1016/j.sna.2016.04.020
0924-4247/

sure of compliance of the object in contact. Tanaka et al. [15] proposed a real time tactile sensor system (size: Ø = 4 mm) based on balloon expansion by injection of biocompatible water and can ensure sterilization of the fluid with safe contact while requiring no electrical power. The sensor output results can effectively study both the stiffness and surface condition with potential for intra-operative brain tumor diagnosis. Sangpradit et al. [16] proposed a force feedback sensor (size: Ø = 11.5 mm) for use in MIS. Its distinguishing characteristic as compared to other tactile and force sensors is the air cushion on which the main sensing element is located. Zhao et al. [17] used the two-spring model to fabricate a tactile sensor (size: 19 × 19 mm) utilizing two piezo resistive cantilevers of varying stiffness to detect the hardness of biological tissues. Our previous work employed a tactile sensor (size: 6.5 × 3.5 mm) that can be attached to a laparoscope for differentiating soft tissues in animals [18]. The mechanical characteristics of the contact object can be differentiated by extracting the voltage outputs from the piezoelectric film at the inner and outer electrodes. Åstrand et al. [19] developed a resonance based piezoelectric tactile sensor (size: 15 mm × 5 mm) that can detect stiffness variations located up to 4 mm in silicone and chicken muscle, mimicking the presence of tumors embedded in prostate tissue. Using this tactile sensor, the shift of the resonance frequency and the force at contact with tissue can be measured and combined into a tissue stiffness parameter. Kwon et al. [20] proposed a simple tactile sensor (size: 26 mm × 8 mm) that has multiple sensing points using a semiconductor micro strain gage for recognition of tissue stiffness. The sensor exhibits high linearity with comparable spatial resolution of a human finger and can be attached to the gripper of surgical instruments for in-vivo palpation during MIS. While the aforementioned concepts can aid in improving tactile sensing during MIS, they are limited by their size and design for practical application in conjugation with regular endoscopy for obtaining information about tissue elasticity and detection of submucosal tumors (endoscopic channel Ø = 2.3 mm).

An endoscope is a useful MIS tool as it not only provides an image for visual inspection but also enables biopsies and the retrieval of foreign objects from inner organs of the body that are difficult to access. However, a visual inspection for an early diagnosis of a tumor is still often insufficient and thus obtaining tactile feedback using an endoscope would be helpful for increasing the precision of examinations and thus improving the quality of treatments. In this study, we have developed a miniaturized tactile sensor that can be mounted on an endoscope and can enable the surgeon to locate and differentiate the SMT from normal tissues in the stomach. We have tested the efficacy of the proposed tactile sensor for elasticity detection of commercial elastomers and artificial tumors embedded in the submucosa of a pig's stomach. Furthermore, the theoretical analysis in ideal contact conditions based on the tandem spring model shows good agreement with observed simulation and experimental results and can operate in a dynamic measurement range suitable for identifying tumors in soft biological tissues.

2. Sensor design

2.1. Sensing mechanism

The miniaturized tactile sensor consists of two components of varying stiffness, namely a hard inner component (E_1) consisting of a copper ball and a soft outer packaging (E_2) made of PDMS. When the sensor contacts a test object under a uniform force, the two components undergo varying deformation depending on their stiffness in comparison to the stiffness of the test object. Since the Young's modulus of the inner structure is greater than that of the outer packaging material ($E_1 > E_2$), a non-uniform stress distribution will be experienced by the piezoelectric film, resulting in two varying output voltages obtained at the corresponding structural electrodes. Since the force transfer by the two sensor components of varying stiffness is different under a uniform applied load, the charge generated (voltage output) by the piezoelectric film under the two components will also be different. When the tactile sensor contacts soft objects, the deformation of the outer packaging will be significantly greater than when the sensor contacts hard objects. Thus the force transfer by the outer structure will be relatively larger when the sensor contacts soft tissues as compared to hard objects. Consequently, there is a lower stress distribution differentiation on the piezoelectric film corresponding to the two sensor components when contacting a soft object and the ratio of the output voltages (V_1/V_2) is relatively small as shown in Fig. 1(a). However, when contacting hard objects, the majority of the normal force will be transferred by the inner structure, resulting in a greater stress distribution differentiation and so the ratio of output voltages is greater, as shown in Fig. 1(b). Therefore, the ratio of output voltages from the inner structure and outer packaging material can be used to obtain information about the hardness (and corresponding stiffness or elasticity) of the test object.

2.2. Ideal contact analysis using the tandem spring model

The tandem spring model has been used to quantitatively analyze the sensing mechanism by simplifying the sensor/sample system as two sets of parallel one dimensional springs arranged serially as shown schematically in Fig. 2. In previous literature, Peng et al. [21] have used a simple spring model to analyze the deflections of the two springs as a measure of the stiffness of the test tissue while the sensor transduction mechanism is based on capacitive sensing. Based on a similar concept of applying two springs with considerably different stiffnesses to soft tissue for compliance detection, Fath El-Bab et al. [22] have developed a detailed design procedure with simulations that take into account the measurement ranges of soft tissue for optimizing the sensor design parameters to give high sensitivity and linearity of the sensor output while also considering the effect of crosstalk between the two springs due to tissue deformation. They have analyzed the soft tissue stiffness as a function of the different forces experienced by the two springs with varying stiffness and have designed a micro machined piezo resistive tactile sensor [23]. In our design, we have used a two-spring model to analyze the stress distribution on the piezoelectric film from the inner and outer sensor components. Since our sensor is relatively small (Ø = 1.4), we assumed that the sensor surface can keep flat as contacted with tissue and the difference between tissue deflection when contacted by the hard and low stiffness spring is negligible to avoid crosstalk due to tissue deformation.

The two springs, E_1 and E_2, with differing stiffness (Young's modulus values) represent the hard inner and soft outer components of the tactile sensor, respectively. The two identical springs E_{sub} represent the submucosal layer while E_o refers to the object under test. The thickness of the sensor, the submucosa and the test object are denoted as H_s, H_{sub} and H_o, respectively. The sensor indents a uniform vertical force on the test object, resulting in localized normal stresses (σ_1 and σ_2) which are proportional to the voltage output corresponding to its inner and outer components, respectively. In an ideal case, it is assumed that the sensor and object in contact are perfectly parallel to each other and the total vertical deformation, δ, of the two sets of parallel springs are identical:

$$\delta = \Delta_1 + \Delta_3 + \Delta_5 = \Delta_2 + \Delta_4 + \Delta_6 \quad (1)$$

where Δ_1 and Δ_2 correspond to the deformation of the inner (E_1) and outer (E_2) components of the sensor, respectively. Δ_3 and Δ_4

Please cite this article in press as: C.-H. Chuang, et al., Piezoelectric tactile sensor for submucosal tumor detection in endoscopy, Sens. Actuators A: Phys. (2016), http://dx.doi.org/10.1016/j.sna.2016.04.020