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A multi-purpose optical microsystem for static and dynamic tactile sensing

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ABSTRACT

The demand for miniaturized tactile sensing devices have greatly increased ever since the introduction of minimally invasive surgical (MIS) procedures. In this work, a tactile sensor is presented that provides essential artificial fingertip perception during such procedures. The proposed sensor can measure tactile information under various static/dynamic loading conditions by precisely measuring the quantity and position of a concentrated force, the local variation in force distribution, relative hardness and local discontinuities in the hardness of soft objects. The sensor, which is fabricated using fiber-optics and microsystems technology, can be used for most MIS tasks since it meets the necessary requirements of being electrically passive and magnetic resonance compatible.

1. Introduction

The use of minimally invasive surgical (MIS) procedures offers significant advantages for both surgeons and patients. In recent years, MIS has rapidly expanded from conventional endoscopic devices to sophisticated computer-controlled robotic-assisted surgical platforms [1]. The da Vinci® surgical system developed by Intuitive Surgical Inc. in the USA and its Canadian counterpart, Amadeus®, developed by the Titan Medical Inc. [2], are among the most popular examples of such surgical platforms. Although they are equipped with features such as enhanced vision systems, tele-surgical applications and force-feed back systems, they do not provide tactile information of tissues nor detect the lumps and arteries. It is essential to have a tactile sensor that can be relied upon to provide this missing sensory information in order that the surgeon is able to characterize and diagnose abnormal tissue, tumorous lumps, blood vessels and ureters. Therefore, our proposed tactile sensor has numerous applications in both minimally invasive robotic surgery (MIRS) and catheter-based cardiovascular techniques (CBT).

Although a number of tactile sensing techniques have been documented [3–5], they are not capable of precisely mimicking the perception of the surgeon's fingertips, which is crucial when making a diagnosis or performing MIS. The most important aspects, when developing a tactile sensor that is intended for use in advanced surgical systems, are that it is miniaturized, ultra-sensitive, electrically passive and compatible with magnetic resonance imaging (MRI) environments [6]. Most existing tactile sensing innovations are neither MRI compatible nor electrically passive because they are based on technologies such as strain gauges [7], piezoelectric [8–11] or capacitive [12]. Hence, tactile sensing based on optical technology is becoming the more attractive option [13–15]. Optical tactile sensors are also successfully employed in industrial robotic systems for detection analysis of small notches [16] and pressure measurement [17]. Most optical tactile sensors, as described in current literature, appear capable of only measuring the force acting on the tissue and not the actual hardness of the tissue itself. Furthermore, they are unable to locate an artery, lump or any other abnormalities in the tissue with any degree of accuracy [18]. In a recent work [19], we have reported a force distribution sensor using a combination of three optical fibers integrated into a single sensor with splice couplings. This, in conjunction with an algorithm we developed, compared the signals in three optical fibers to predict the quantity and location of the force in order that the force distribution could be precisely measured.

In this work, we propose an optical tactile sensor having the following seven attributes contained in one single small-size package: (i) it measures the amplitude as well as the position of a concentrated force; (ii) it measures the local variations in force distribution along its length; (iii) it measures the relative hardness of soft objects

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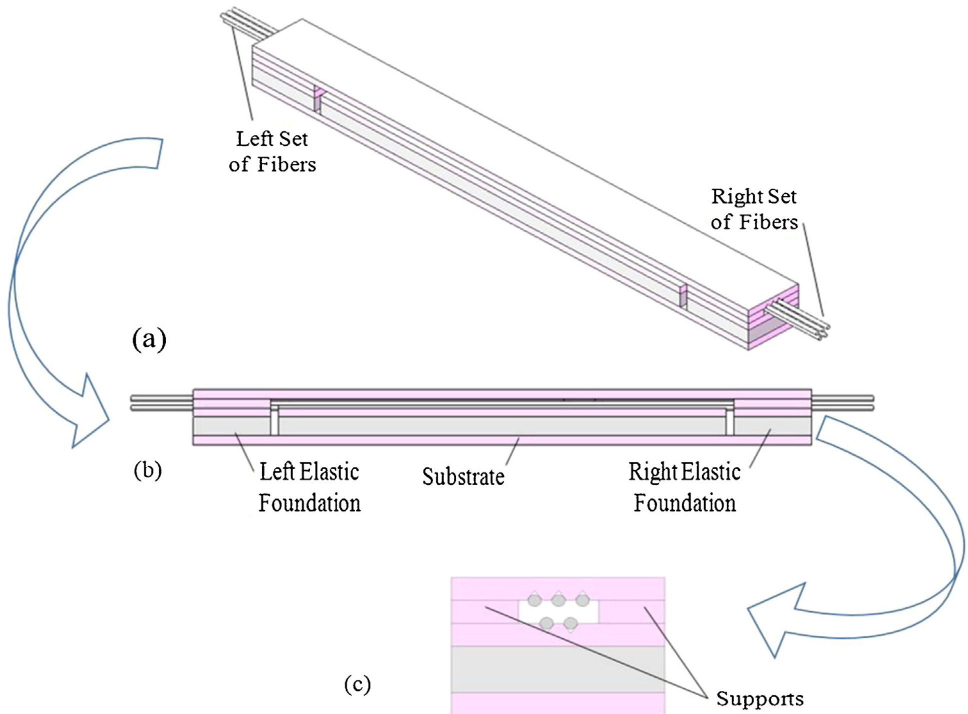

Fig. 1. Sensor schematic. The 3D view (a) front view (b) and side view (c).

such as tissues; (iv) it measures local discontinuities in the hardness of soft objects along the contact area; (v) it is magnetic resonance compatible; (vi) it is electrically passive; and (vii) it performs under both static and dynamic loading conditions. Furthermore, because micro electro mechanical systems (MEMS) batch fabrication techniques was used in its fabrication, it was possible to miniaturize the sensor such that it can fit into the limited narrow space on the end-effectors of surgical tools and was, also, inexpensive to produce.

The present study discusses the design, modeling, simulation, fabrication, and testing of the proposed tactile sensor. After the explanation of its design, the theoretical model of the sensing principle of the sensor, as well as the finite element model of the sensor–tissue interaction, is discussed. The prototype version of the sensor was fabricated and tested using micro-systems technology. The experimental test results prove that the sensor measures the relative hardness of artificial tissues as well as the local discontinuities in the force and hardness distribution along the contact area between the sensor and tissues.

1.1. Sensor design

In the design of the sensor, the objectives as well as the constraints were considered. The objective is to measure the amplitude/position of force, the variations in force distribution, the relative hardness, and the discontinuities in the hardness. Consequently, the sensor must comprise various sensing elements to address these required objectives. In the structure of the sensor, multiple sensing elements were reserved for sensing the force distribution and force position. They alone, however, were unable to satisfy the hardness measurement requirements because, for the hardness measurement of soft objects, the contact force as well as the resulting deformation must be measured simultaneously. As a result, another sensing element was required to measure the resulting deformation. In the structural design of the proposed sensor, two sensing layers were incorporated. One sensing layer measures the contact force whereas the other layer performs relative measurements of the resulting deformation. Moreover, the multiple sensing elements integrated into the sensor structure measure the variations in force distribution and the discontinuities in hardness.

More pointedly, the design constraints should also be taken into account. Such constraints consist of the following: (a) the constrained size of the available space at the tips of surgical end-effectors; (b) the interference emanating from magnetic resonance imaging (MRI) and ultra-sound devices common in surgical rooms; (c) the electrical disruption of the organs of the body such as the heart; and (d) the performance under dynamic and static loading conditions. In fact, the MEMS-compatible design of the sensor addresses the first constraint. Choosing the optical fiber sensing principle addresses the second, third and fourth constraints. In the design phase, each component and the whole assembled structure of the sensor are dealt with.

As shown in Fig. 1 the sensor comprises five narrow layers and five pairs of optical fibers. The first, the third, the fourth, and the fifth layers, counting from the bottom, are made of silicon whereas the second layer is made of polydimethylsiloxane (PDMS) material. The first layer, which is a single part, is the substrate of the sensor. The second layer, which consists of three separate parts with the same thickness, is the elastic foundation of the sensor. The third layer consists of three separate parts cut from a micromachined silicon wafer. On the top surfaces of these three parts, two rows of v-grooves are micromachined. The fourth layer includes four silicon chips cut from a silicon wafer. Each two of these four chips is located one on the left side and one on the right side of the sensor. The fifth layer, which is a single part, is a beam cut from a micromachined silicon wafer. On the bottom surface of the beam, three parallel rows of v-grooves are micromachined. All of the optical fibers of the sensor are single-mode. They are integrated into the v-grooves of the third layer and the fifth layer. Whereas two pairs of fibers are integrated into the third layer, three pairs of them are integrated into the fifth layer.

Fig. 2(a) shows the fibers that measure the force integrated into the third layer of the sensor. Fig. 2(b) shows the magnified view of (a) and Fig. 2(c) shows the fibers integrated into the fifth layer,