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A novel tactile softness display for minimally invasive surgery

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ABSTRACT

The use of Minimally Invasive Surgery (MIS) in various types of surgical procedures has increased significantly in recent years. However, its scope is limited due mainly to the fact that this procedure does not currently provide tactile feedback without which the surgeon can neither feel nor palpate tissue. In this paper, we describe a new tactile display that reproduces these missing constituent properties of the tissue directly to the surgeon. The softness of different objects is regenerated based on the mechanical properties of those objects and fingertip pulp. The force–displacement non-linear behavior of several different materials is simulated using force feedback and a Proportional-Integral-Derivative (PID) controller for the linear actuator. Experimental tests show that the proposed tactile display can closely regenerate the feeling of softness for different objects by rendering the force–displacement curve to the surgeon's hand.

1. Background

Minimally Invasive Surgery (MIS) is the practice of performing surgery through small incisions using specialized surgical instruments in order to gain access to internal organs. During conventional open surgery, large incisions are usually required and often cause more trauma than the procedure itself [1–3]. By contrast, MIS procedures cause less bleeding and discomfort with the additional benefits of reduced recovery time and, ergo, cost. However, notwithstanding these advantages, MIS procedures do have certain shortcomings and it is these that have been the subject of studies over the last decade. They include loss of tactile sensing and direct kinesthetic feedback, possible lengthier surgery time, greater difficulty in removing bulky organs and the need for increased technical expertise. Among these present disadvantages, however, it is the loss of tactile and kinesthetic feedbacks that has proven to be the most serious issue because it compromises the surgeon's abilities and limits the scope of MIS procedures. For this reason, it has become the focal point of intensive research.

Numerous research works have been undertaken on providing the surgeon with the means of applying force to sense softness, feel the pulse and detect abnormalities. The actuation technique is the key parameter of a physical tactile display which determines the whole system performance. For MIS and MIRS (minimally invasive robotic surgery) applications, the tactile display must conform to the medical device regulations imposed by the Food and Drug Administration (FDA) and the European Medicines Agency (EMA) while, at the same time, also meeting the physical requirements and other strictures that govern the safe and proper use of tactile displays in medical practice.

The minimum requirement of an ideal tactor (pin) type tactile display is that it must have a temporal resolution of 50 Hz, spatial resolution of $1\,\mathrm{mm}^{-2}$ and provide a minimum pressure of $50\,\mathrm{N\,cm}^{-2}$. Each actuator should be capable of indenting up to 4 mm with a height resolution of 10% (a power density of $10\,\mathrm{W\,cm}^{-2}$ with an actuator density of 1 per mm^2) [4–6]. The primary prerequisites for tactile displays for MIS and MIRS procedures, in order to make them suitable for integrating into current devices, are that they be small and quiet. In keeping with this, various actuation and stimulation techniques have been proposed. Pelrine et al. reported an electrostatic actuator composed of a polymeric elastic dielectric sandwiched between compliant electrodes [7]. In a similar design, Jungmann and Schlaak [8] used electrostatic actuators with elastic dielectrics as the tactile stimulator. By applying a voltage to the electrodes, the dielectric contracts in thickness and expands its area due to the attracting charges on the electrodes. When the voltage is reduced, the dielectric returns to its initial shape and produce forces due to its stored elastic energy. One notable disadvantage of this actuator, certainly in medical applications, is its relatively high operation voltages which increases the risk of electrical interference in other devices.

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Yamamoto worked on a tactile display in which electrostatic force and friction control was employed in order to recreate surface roughness [9]. This device consisted of stator electrodes and a thin film slider upon which an aluminum conductive layer was deposited. The user places his index finger on the slider and moves it horizontally to obtain a certain tactile sensation. By applying various voltage patterns to stator electrodes, various friction distributions are generated on the slider which, in turn, is transferred to the fingertip so as to generate a surface roughness sensation which, therefore, makes it applicable only for surface roughness perception. However, the main objective of a tactile display in medical applications is to recreate softness and contact force during surgery which assists the surgeon in finding out how hard he/she is pulling or pushing the tissue and to identify what hidden features may be present under the tissue.

Ottermo et al. worked on fabricating a shape display using micromotors on the handle of an endoscopic grasper. They employed a 35 mm $\times$ 10 mm array of 15 $\times$ 4 piezoelectric sensors, in conjunction with a tactile display consisting of thirty-two micro motors, having a total size of 27 mm $\times$ 20 mm $\times$ 18 mm [10,11]. This design, however, was limited by virtue of the fact that the actuator shaft was relatively large which decreased the tactor density in this display. Furthermore, the limited force capacity of the actuators also restricted the ability to regenerate the mechanical deformation behavior of hard biological tissues. Wellman et al. used shape memory alloy (SMA) as the actuating element for a tactile shape display [12]. A 3D shape display, using an array of bars actuated by the SMA, is presented in [13]. Despite its many advantages in medical applications, SMA based actuators suffer from non-linear behavior and hysteresis during loading and unloading cycles. Hayward et al. [14] worked on a tactile display using an array of piezoceramics which stimulates the skin in its lateral mode by vibrating two active piezoelectric layers, also known as a bimorph. In another research work, Yun et al. [15] described the development of a piezoelectric based planar-distributed tactile system that only displayed textures. This proposed tactile display comprised a 6 $\times$ 5 pin array, actuated by thirty piezoelectric bimorphs. Another tactile stimulation method was proposed by controlling the suction pressure [16–18]. This method is based on the tactile illusion that we feel when something like a stick pushes up into the skin surface when we pull skin through a hole by lowering the air pressure.

In [19], a tactile display is proposed utilizing thermo-pneumatic micro pumps and micro valves which works based on sealed cavities in which one side is flexible. Each cavity is filled with a low boiling point liquid, such as methyl chloride, with a resistive heater built inside. When the heater increases the temperature inside the cavity, the pressure increases because of the gas resulting from the liquid–gas phase transition and the flexible side of the cavity becomes swollen [20,21]. Though this design is simple and has many advantages, the response time in this actuation method is long. In [22,23] an electrocutaneous display is presented which directly stimulates sensory receptors within the skin with electrical current. An array of 5 $\times$ 6 electrodes is developed for generating pressure or vibration patterns without using any mechanical type of actuator.

Yamamoto et al. [24] developed a graphical overlay technique to display the location of hard objects hidden in soft tissue. While the phantom tissue is palpated using a surgical robot, the stiffness of a Hunt-Crossley model is estimated. At the same time, a visual overlay is created on a semi-transparent disc at the tissue surface by using hue-saturation-luminance. The hue corresponds to the stiffness at a palpated point and the saturation is calculated based on the distance from this point. Kalantari et al. [25] developed a 3D graphical tactile display for determining the presence of localized lumps. Their proposed algorithm reads the pressure distribution from 3 $\times$ 3 PVDF sensor arrays mounted on both jaws of a laparoscopic grasper and renders pressure distribution in three engineering views making the operator capable of detecting features that are hidden under the tissue.

Graphical tactile displays that provide the surgeon with pressure distribution of contacted tissue are easy to implement on current MIS and MIRS devices. However, because such displays are based on indirect feedback through vision, they are therefore less efficient than direct stimulation. In [26], a tactile display device is designed based on the soft-actuator-based wearable technology. This device can provide stimulation on the human skin without any mechanical transmission. An electroactive polymer is used for the construction of the tactile display device. Another work which has been developed is a pneumatically-driven balloon actuator array suitable for mounting on robotic surgical master controls [27,28]. The inflation of hemispherical balloons increases the force on the skin and causes skin deformation. The actuator consists of an array of balloons formed from a spin coated silicone film placed over a molded substrate. Because it is small and fast due to its pneumatic driven mechanism, this display is suitable for MIS and MIRS applications despite the less than ideal shape of its pins (tactors) and non-uniform contact area. In [29] Kim et al. developed a multi-fingered tactile display module in which each comprises a 4 $\times$ 4 piezoelectric ultrasonic actuator array. Various types of texture information can be generated using a static indentation or through the vibration of each pin. In [30], Kimura et al. presented a 2-DOF (Degrees of Freedom) controlled softness display, capable of recreating the asymmetric softness of the contact area. Their proposed display consists of a flexible sheet as the contact surface and two DC motors which independently control the height of both sides of the flexible sheet and, consequently, the contact area. Liu [31] developed a real-time softness display by controlling the deformable length of an elastic beam element. Burch and Pawluk [32] presented a 2-D multifinger tactile display employing piezoelectric actuators as both contact and stimulator elements on each fingertip to reproduce and represent texture images. Cameron et al. [33] presented an electroactive polymer-based tactile display. They employed an Electro-Active Polymer (EAP) in the form of a cantilever to produce light touch through bending force and vibratory sensations to the fingertips by varying the sinusoidal waveform. In [34], a compact and wearable 3-DOF tactile display was developed and affixed on the fingertip. It comprised two main parts, one fixed part on the backside of the finger which supported three DC motors, and the other active part on the fingertip which consisted of three wires whose lengths and strains were controlled by three DC motors. Watanabe et al. [35] proposed a tactile display using large displacement Micro-Electro-Mechanical Systems (MEMS) actuator arrays in which each was composed of a piezoelectric actuator and a hydraulic displacement amplification mechanism to achieve the minimum required stroke length of a tactile display element.

Small workspace on MIRS master controls and sensitive electrical instruments limit the number of tactile display technologies suitable for this application. Considering the current design of robotic surgical systems, only fast and small softness displays with a safe operational voltage and magnetic field would be suitable. Therefore, actuation techniques such as dielectric polymers, SMA (shape memory alloy), piezoelectric and electroactive polymers generally are not a good candidate for MIRS applications.

Most tactile displays that have been introduced which are driven by a servomotor, or other type of linear actuator, use the spring model to simulate the softness. From a simplistic point of view, most materials can be modeled as a spring which, for small strains, is good for characterizing the softness of the material. However, in large deformations, the spring model cannot determine the behavior of the material with any exactness. During MIS, a surgical