Achieving Biocompatibility in Soft Sensors for Surgical Robots

Andrew H. Gosline[1], Veaceslav Arabagi[1], Adil Kassam, Pierre E. Dupont[1]

[1]*Boston Children's Hospital, Harvard Medical School, Boston, MA, USA*

andrew.gosline@childrens.harvard.edu

INTRODUCTION

Minimally invasive surgical robotics has enabled clinicians to operate with reduced trauma due to small incisions and long, slender tools. Instrument guidance remains a limiting challenge, however, for procedures in which image quality is limited and haptic feedback is unavailable, e.g., beating-heart intracardiac procedures [1]. To address this need, our group has been investigating sensing technologies that can envelop a robot with a compliant sensing structure and provide clinicians with tissue contact and force information over its entire surface including its tip.

Soft sensors are a particularly promising candidate technology that, to date, have not been considered for medical applications. These sensors are composed of a compliant polymer structure containing channels filled with a conductive liquid. Under load, the channel cross sections deform providing pressure or strain measurements measured as changes in resistance [2].

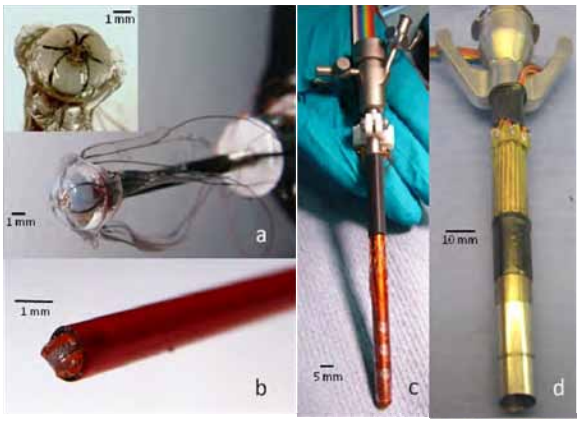

Fig. 1 Surgical soft sensors created in our lab. (a) Robotic catheter fingertip sensor measuring contact force and contact angle [3]. (b) Tissue contact sensor with diameter of 1.5mm. (c) Neuroendoscope pressure-sensing sleeve for monitoring contact pressure with parenchymal tissue. (d) Pressure sensing laparoscopic port for the Intuitive Surgical da Vinci robot.

Manufactured using soft lithography, these sensors can have a thickness of less than 1 mm, are extremely elastic and compliant, can be manufactured with high sensor densities and can exhibit high pressure sensitivities. Our group has created and evaluated several soft sensor morphologies as shown in Fig 1. These include tip force and contact angle sensors for intracardiac surgery [3], pressure sensing to prevent cortical damage for neuroendoscopy, and pressure sensing to monitor/prevent ischemia on the laparoscopic ports of the Intuitive Surgical da Vinci system. In our work, the soft sensors were made sufficiently thin, ~400µm thick, and sensitive to loads of order 5g.

The majority of prior work in soft sensors has used the conductive liquid metal Eutectic Gallium-Indium (eGaIn). The properties of eGaIn are conducive to soft sensing. Its low channel resistance facilitates simple DC circuitry; its low viscosity enables easy injection into micro-channel geometries and its high surface tension reduces bubble formation and facilitates channel filling and sealing during manufacture. Unfortunately, eGaIn is not biocompatible and the development of alternate conductive fluids is needed for medical applications.

In recent work aimed at green chemistry, the use of ionic salt solutions was proposed. The low viscosity of aqueous salt solutions, however, confounds the production of bubble-free channels and thus, thickening agents are added to facilitate manufacture [4]. Ionic solutions also require AC excitation to prevent ion polarization at the electrodes, and a low voltage must be used to avoid electrolysis of the fluid, which can cause bubble formation and corrosion [4]. This, however, proves advantageous since transforming the signal into the digital domain increases the signal to noise ratio while requiring a relatively simple oscillator circuit to monitor channel resistance via oscillation frequency [5].

While biocompatible, however, aqueous salt solutions do not produce reliable and robust sensors because the gas permeability of the encapsulating polymer is high, which allows the water from the solution to evaporate [4]. In fact, the permeability of siloxane based polymers is so high that they can be used to manufacture permeation pumps for micro-fluidic systems [6].

The contribution of this paper is to demonstrate that NaCl saturated glycerol provides all of the necessary properties for medical soft sensors. This solution is biocompatible [7], sufficiently conductive for sensing applications, adjustably viscous for ease of manufacture and, possessing a boiling point over 300^{o}C, it is extremely resistant to evaporation. An initial sensor design and experimental evaluation is presented below.

MATERIALS AND METHODS

To investigate the efficacy of NaCl saturated glycerol as a biocompatible conductive fluid, we constructed and tested several prototype sensors. NaCl saturated glycerol was made by mixing 99% pure glycerol (Sigma Aldrich G5516-500ml) with food grade NaCl until saturation. The solution was then carefully poured into a syringe through a filter to prevent any undisolved NaCl crystals from being captured. For ease of experimentation, the sensor channel geometry was manufactured using high purity silicone microtubes with a 305µm inner diameter, and 150µm wall thickness (McMaster-Carr #51845K65). These tubes were cast into a 1.1mm thick layer of PDMS, then injected with NaCl saturated glycerol with a 30g blunt tip needle. Electrodes composed of 280µm diameter nickel-titanium wire were used for corrosion resistance and biocompatibility. These electrodes were inserted through the glycerol filled micro-tubes, then tied off with 5-0 braided silk sutures to prevent leaks. Finally, the sensor was rinsed with isopropanol, then cast in another layer of PDMS to fully seal the liquid channels to the electrode junction. Fig 2 shows a semicircular sensor design, with a radius of curvature of 1mm, for highly localized pressure sensing. The oscillation circuit as proposed by [5] was used to generate an AC square wave. The conductivity of NaCl saturated glycerol is approximately 2 orders of magnitude lower than that of saturated saline, which necessitated a rail-to-rail voltage of +/-2.5V for stable operation.

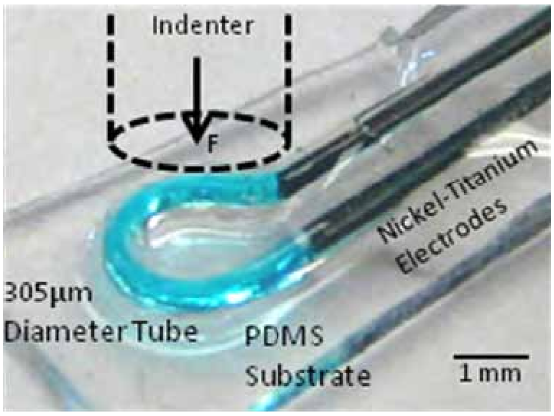

Fig. 2 Sensor prototype. The sensor is 1.1mm thick and uses glycerol dyed with food colouring for channel visualization.

Two experiments were performed using the sensor prototype shown in Fig. 2. The first experiment measured the change in oscillator period as a function of applied force using a 2.5mm diameter PDMS indenter. The second experiment evaluated the longer-term stability of the sensor in terms of robustness to evaporation, electrolysis and electrode corrosion. For this test, the sensor was put into continuous use for 14 hours with measurements of base resistance and sensor function made approximately every hour to ensure that no adverse electrochemical effects were present.

RESULTS

The applied force versus measured oscillator period is plotted in Fig. 3. The depicted curve is a two-term exponential function $L = 18.3\exp(0.59T) - 2785\exp(-12.4T)$, where L is the load in grams, and T is the period in milliseconds. Channel collapse was observed at ~43g load, yielding a loss of oscillator signal. Though the data was taken at increments of approximately 5g, it was possible to detect loads as low as ~2g based on the resolution of the oscilloscope's frequency analyzer.

In the second test, no bubble formation due to evaporation or change in base resistance was noted over the 14 hour test period.

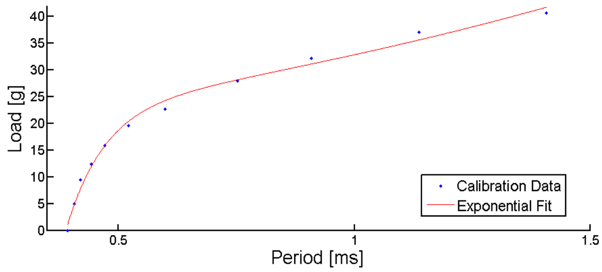

Fig. 3 Applied Force vs. Oscillator Period

DISCUSSION

NaCl saturated glycerol is a suitable biocompatible substitute for eGaIn in soft sensors for robotic surgical applications. The experiments discussed herein demonstrate that it is possible to fabricate a soft sensor using biocompatible materials exclusively. Future work will include: 1) sensitivity comparison between biocompatible ionic-based sensors and their eGaIn-based predecessors 2) optimization and refinement of the ionic solution to maximize conductivity and sensitivity 3) development of prototype support electronics with a focus on a low production cost, and 4) integration of the sensing system with a surgical robot platform, for bench-top and in-vivo testing.

Towards clinical adoption, we envision the use of biocompatible soft sensors with inexpensive electronics for amplification and sampling to be a disposable add-on to existing surgical robotic systems.

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