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A flexible tri-axis contact force sensor for tubular medical device applications

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Abstract

A capacitive, tube-shaped tri-axis contact force sensor is fabricated and characterized in this paper. The sensor is inserted into a tubular medical device just like a small section of tube in order to sense the applied contact force on the tip of this tubular device. It comprises two SU-8 plates, wherein the outer and inner rims of the plates with diameters of 2.3 and 1.1 mm, respectively, are assembled together. Three proof masses were hinged to the outer and inner rims of the first plate by membranes. The different signals received from the three electrodes provide collective information indicating the precision and accurate direction of contact forces exerted on the tip of the tubular medical device. In the test, applied force range from 10 mN to 1 N was detected, the performance of the sensor is stable from 25 to 65 °C, and the device was functional under a shock of 2.0 N for 10 s in a shock test. In addition, a two-wire readout approach was also developed to minimize the lead wires.

(Some figures in this article are in colour only in the electronic version)

Introduction

Less invasive therapy using tubular medical devices has grown rapidly in recent years. Currently it is difficult for operators to know the interactions between a tubular medical device and human tissues in these narrow and complex cavities in the human body, which sometimes leads to damage to tissues or causes critical accidents. Some studies have been performed to make ‘smart’ tubular medical devices for sensing contact conditions, including three-dimensional tactile sensors composed of a rubber layer and three strain gauges [1]. However, the size of the sensor is hard to reduce due to the assembly process. Three optical fibers were used to measure small deformations of the tube tip in a radio frequency ablation tubular medical device [2]. An endoscopic tactile sensor measured the tactile force by means of image processing of the IR cut pattern [3]. Optically reflective, corrugated Parylene diaphragms were fabricated for fiber-optic-linked pressure sensing in an ultralow pressure range [4]. However, the sensors in [2–4] are complex due to the optical sensing system. Therefore, these reported sensors are too large or complex to be embedded in many tubular medical devices, especially with smaller diameters, because tubular medical devices become multifunctional and all these functions need to be used in a limited space.

Normally the sensor is mounted between the tip and the remaining part of a tubular medical device. The sensor is a small section of a tube with similar sidewall thickness to that of the tubular medical device. The process is rather complex using silicon or glass substrates because a circular hole and a circular ring have to be drilled through the sensor. Another problem is that the sensor should be sealed, which makes it difficult to package the sensor if a silicon surface micromachining process, such as the sacrificial layer etching process, is used. SU-8, a kind of negative photoresist, was chosen because SU-8 can work as a permanent material for MEMS devices. A ring-shaped SU-8 structure can be patterned by the lithography process easily. With a lower elastic modulus, SU-8 also deflects more than silicon for a given contact force and diaphragm design, resulting in improved pressure sensitivity. In addition, SU-8 is biocompatible [5].

Multi-layer SU-8 processing was developed to prepare for three-dimensional MEMS architectures. For example, Conradie et al demonstrated that SU-8 was well suitable for a permanent material in mechanically active MEMS devices [6]. Agirregabiria et al reported yielding sealed cavities of an SU-8 process, and complex three-dimensional microfluidic channels without using a sacrificial layer [7]. An SU-8 MEMS Fabry–Perot pressure sensor for catheter application was presented
Figure 1. Cross section of the sensor assembled in a tubular medical device, schematic of the sensor structure, and optical microscopy picture of a fabricated device.

by Hill et al, and the sensor consists of a polymer cap with a reflective, pressure-sensing diaphragm mounted at the end of a fiber optic cable [8]; this report also demonstrates the biocompatibility of SU-8.

In this work, a tri-axis force sensor made of SU-8 photoresist is presented. Its capacitive readout approach is less expensive than its optical competitor, and only one additional lead is needed to read pressure signals from three pixels, which is especially attractive for tubular medical devices of small diameters.

Design

As shown in figure 1, the device comprises a top and a bottom SU-8 plate, and the outer and inner rims of the two plates are assembled together. Three proof masses were hinged to the outer and inner rims of the first plate by membranes. The tip of the tubular medical device only connects the top surface of the three proof mass blocks. The force applied on the tip will create a pressure on the three proof mass blocks. The outer and inner rings act as a solid frame, and the membrane around the proof mass deflects with applied pressure. The proof mass moves toward or away from the bottom plate if a force is applied on the tip, and changes the capacitance between the two plates, indicating the amplitude and direction of the applied force on the tip.

A ballpark estimate of detected force range was made to decide the initial structural parameters, and the thickness of the membrane is adjusted later in the process to fit the force range based on the experimental data in the first run.

In our study, the outer and inner diameters of the sensor are 2.3 and 1.1 mm, respectively, the area of each capacitor is calculated to be $1.3 \times 10^{-6}$ m², and the capacitance is about 1.5 pF for a gap of 10 μm. According to the required force range from 10 mN to 1.0 N, which was set by the system designer, the pressure on one of the capacitors is from 7.5 to 750 kPa, which is derived through dividing the force by the capacitance area.

The mechanical model of the proof mass and the membrane system is simplified as a rectangle-shaped plate, and the following formula was employed to get a rough estimate of the displacement of the proof mass [9]:

$$
\Delta x = \frac{F}{k} = \frac{F}{\frac{1}{2} A \Delta y}
$$
where $\Delta d$ is the maximum displacement of the proof mass, $L_x$, $L_y$ are the length and width of the rectangle-shaped plate, respectively, $\max(L_x, L_y) = L_x$ if $L_x > L_y$, and $\max(L_x, L_y) = L_y$ if $L_x < L_y$. $c_1$ is the coefficient decided by $L_x/L_y$, which is 1.57 in this case, and $c_1$ is decided to be 0.025, $p$ is the applied pressure, $E$ is Young's modulus of SU-8, which is 4 GPa in this calculation, and $h$ is the thickness of the membrane.

As a result of equation (1), the thickness of the membrane is 140 $\mu$m to make $\Delta d$ below 5 $\mu$m at the maximum pressure of 750 kPa. However, the linearity relationship of capacitance with the gap becomes poor if the gap was set as 5 $\mu$m. A gap of 12 $\mu$m was chosen in this design to balance the linearity and sensitivity of the capacitive force sensor.

Considering that the proof mass increases the stiffness of the membrane, a thinner membrane will reach a greater force range. In actual experiments, membranes with a series of thickness from 80 to 150 $\mu$m were tested and the sample with a membrane about 100 $\mu$m thick is decided to be able to meet the requirements of sensitivity, full range and linearity well.

**Fabrication process**

The top plate was fabricated by the three-layer SU-8 process, as illustrated in figure 2. First, a layer of LOR 3A (Microchem, USA), a kind of resist based on polydimethylglutarimide, was coated and prebaked at 150 $^\circ$C for 3 min, which acted as a sacrificial layer to release the SU-8 structure. As the first SU-8 layer, SU-8 3050 was spun coated at 1200 rpm, and prebaked at 65 $^\circ$C for 30 min and at 95 $^\circ$C for 30 min. Next, hard bake with the same procedure as that in the prebake step was performed after an exposure of 500 mJ cm$^{-2}$. The thickness of this layer was about 90–100 $\mu$m. The second SU-8 3050 layer was spin coated at 800 rpm, and prebaked at 65 $^\circ$C for 45 min and at 95 $^\circ$C for 45 min. This step was followed by an exposure at 800 mJ cm$^{-2}$ and a hard bake at 65 $^\circ$C for 45 min and 95 $^\circ$C for 45 min. The thickness of the second layer was about 120–130 $\mu$m. Another SU-8 2010 layer was coated at 3000 rpm, prebaked at 65 $^\circ$C for 3 min and at 95 $^\circ$C for 3 min, and hard baked at 95 $^\circ$C for 3 min after exposure at 150 mJ cm$^{-2}$. The thickness of the third layer was about 8–10 $\mu$m. In the developing process, the wafers were dipped in the SU-8 developer (Microchem, USA) for 45–50 min with shaking, followed by rinsing in isopropyl alcohol (IPA). These mentioned parameters for the fabrication process were extracted experimentally. The second layer being thicker was baked longer and exposed with more energy. The baking process of the second layer has little influence on the properties of the first layer, especially the residual stress of the exposed SU-8 layer. Finally, these wafers were dipped in an LOR photoresist developer (CD 26) for 30 s. The exposed LOR3A was removed by 30 s CD 26 dipping, and an undercut of LOR3A under the SU-8 layer was prepared. This undercut is necessary for the later peeling process. Otherwise, the LOR3A layer would be covered by sputtering metal and difficult to remove.

A double layer of chromium 20 nm thick and gold 200 nm thick was sputtered on the wafers. Next, these wafers were immersed in the CD 26 developer overnight and all the SU-8 chips were peeled up. These chips were collected and arranged on a dummy wafer with the uncoated surface upward, and were sputtered with chromium 20 nm thick and gold 200 nm thick, again. Now the whole surface of the first plate was metalized.

The bottom SU-8 plate was fabricated on one layer of SU-8. A layer of LOR 3A was coated and prebaked at 150 $^\circ$C for 3 min. SU-8 3050 was spin coated at 800 rpm for 60 s, and prebaked at 65 $^\circ$C for 45 min and at 95 $^\circ$C for 45 min. The same condition for hard bake was followed after exposure at 800 mJ cm$^{-2}$. The thickness of the bottom plate was about 130 $\mu$m. The wafers were immersed in the SU-8 developer for 30 min with shaking, and rinsed by IPA. The wafers were immersed in CD 26 for 30 s and rinsed by DI water. Next, the wafers were sputter coated with chromium 20 nm thick and gold 200 nm thick. The metal layer was patterned in a lithography process and a wet etching process to separate the metal into three individual electrodes. Here standard gold and chromium etchant were used successively. The wafers were immersed in CD 26 overnight and the SU-8 chips were peeled off and collected. A shield mask was prepared in advance and it was an SU-8 plate about 100 $\mu$m thick, too. The collected SU-8 chips were positioned on the shield mask under a microscope with the uncoated surface toward the mask surface, and were fixed by tape from the backside. The chips and masks were sputter coated with chromium 20 nm thick and gold 200 nm thick. Last, the chips were released from the tapes and rinsed in IPA. Optical microscopy pictures of the top and bottom plates are shown in figure 3.

An SU-8 mold plate was prepared for the assembly of the top and the bottom plates. Omnicoat (Microchem, USA) was spin coated at 3000 rpm and prebaked at 200 $^\circ$C for 5 min. SU-8 3050 was spin coated at 800 rpm for 60 s, and was patterned using the same recipe as that in the fabrication of the bottom plate. In this mold mask, some recessed cavities were arranged around the center of the wafer to contain the bottom plates. The bottom plates were placed into these recessed cavities, followed by SU-8 2010 spin coated at 5000 rpm. The top plates were aligned face to face with the bottom plates in the mold plate under the microscope. Three aligning bumps in the outer rims of both top and bottom plates were designed to make this alignment easier. Next, pressure was applied on the top surface of the pairs through a round bar. The pairs were removed from these indentations by tweezers, and prebaked at 65 $^\circ$C for 3 min and 95 $^\circ$C for 3 min, followed by exposure at 80 mJ cm$^{-2}$ with the transparent-edged bottom plate upward. The assembly process and an optical microscopy picture of the assembled force sensor are shown in figure 4.

**Force response in a bench test**

A plastic cap was bonded on the top surface of the sensor in a bench test. A plastic tip was driven to touch different locations of the cap, and the force was measured by a microbalance. Here a plastic tip was chosen to avoid the
parasitic capacitance from a metal tip. The three capacitances were measured by a multi-channel capacitance to digital convertor (EVAL AD7147EBZ). The input capacitance of AD7147 was converted to a dc voltage, and was further converted to a 16-bit digital output with the unit of codes. Hereby numbers from ‘0–65536’ indicate a certain value of capacitance. Figure 5 shows the output at a force pulse of 400 mN for about 10 s. Comparison of the three different digital outputs indicates the direction of the applied force. Here the applied force pointed at the boundary of electrodes A and B, and leant toward electrode A, because the output of electrodes A and B is larger than that of electrode C, and the output of electrode A is the maximum.

In a strategy of contact force control, the motion of the tubular medical device was operated by pushing and pulling three metal strings. The locations of the three electrodes were
Varied perpendicular forces were applied to the center of the plastic cap, as shown in figure 5. The digital output increment of one of the three electrodes was recorded with respect to different applied forces measured by a microbalance. In this calibration process, the sensing range from 10 mN to 1.0 N was demonstrated by the calibration curve, as shown in figure 6 (right). The digital output at 10 mN for 10 s is shown in figure 6 (left), the signal-to-noise ratio was larger than 2.0 to guarantee a resolution below 10 mN. The maximum detected force was more than 1.0 N. Theoretically, the digital output is proportional to the capacitance increment, $\Delta C$, while $\Delta C$ is proportional to $\Delta d/(d - \Delta d)$, where $d$ is the gap of a sensing capacitance in the device, and $\Delta d$ is the change of the gap with respect to the applied force. From equation (1), $\Delta d$ is proportional to the applied force for a given contact area in this experiment. Therefore, the force curve is almost linear in the small force range, where $\Delta d$ is much less than $d$. An initial gap of about 12 $\mu$m and a maximum $\Delta d$ below 5 $\mu$m were set in this design in order to extend the linear range. The output will be saturated in a shock if the force were further increased because the proof mass would contact the bottom plate. A thin layer of SU-8 was deposited on the electrode surface of the bottom plate in the assembly process to avoid shorting of the two electrodes of the capacitors.

No significant performance shift of the sensor was found at a temperature test varied from 25 to 65 °C. The sensor is required to keep stable performance in this temperature range because the tip of the tubular medical device will be heated by RF power for ablation. In the test, a thermal meter was placed near the sensor to indicate the temperature while the environment temperature was increased using an electric heater. As shown in figure 7 (left), the outputs of a sample with an applied force of 400 mN for 10 s were recorded at 25 and 65 °C. The difference between the outputs at 25 and 65 °C was negligible. Hereby the power of the heater was adjusted manually to keep a stable temperature for a few minutes.

Figure 3. Optical microscope photos of the fabricating parts. Top left: top view of the first plate, top right: bottom view of the first plate, bottom left: photograph of the second plate and bottom right: photograph of a wafer.

Figure 4. The assembly process of the sensor: the second plate was placed on a container wafer, and spin coated with SU-8; the first plate was aligned and assembled on the second plate under the microscope. Bottom right is the photograph of an assembled device after SU-8 curing.

Figure 5. Left: schematic illustration of the test apparatus; right: the outputs of three pixels at a force pulse of 400 mN for 10 s.
Figure 6. Left: the output at 10 mN for 10 s and right: the calibration curve of the force sensor.

Figure 7. Left: the output at 25 °C and 65 °C, 400 mN for 10 s and right: the output at 400 mN after 2.0 N shock.

The sensor was demonstrated to keep it functional after a shock of 2.0 N. Because of the possible attacks at the tip in the assembly process in factory, a desired maximum shock of 2.0 N was set by the system designer of this tubular medical device. As shown in figure 7 (right), a shock of 2.0 N for 15 s was applied on another sample, followed by a 10 s rest and a force of 400 mN for several seconds. As a result, the output at 400 mN was almost consistent with the output before shocking. Here the samples were from the same batch as shown in figure 7. The output linearity of the device is not good. However the linearity is not critical in our application because several threshold forces should be recorded. Furthermore, the output will be returned to zero for each measurement, and only the increment is used to indicate the force.

Two-lead readout approach

A two-wire readout approach was developed to minimize the number of lead wires. More and more sensors and actuators are integrated into tubular medical devices for more advanced functions, resulting in an inner cavity of crowded wires. Accordingly, it is very desirable to reduce the leads of the components in a tubular medical device. In the tri-axis force sensor, four wires are required to contact a common ground and three pixels if using a commercial capacitance to digital convertor, just like the setup in the bench test. However, there is no room left for the four wires in the inner cavity. Alternatively, three parallel LC circuits were utilized to deduce the capacitance through only two wires.

In principle, three parallel LC circuits give three pitches in the frequency spectrum, and the location of the pitches is decided by the product of the inductance and capacitance. Generally, the pitch of an LC circuit is expressed as

$$f = \frac{1}{2\pi \sqrt{LC}}$$

where $L$, $C$ and $f$ are the inductance, capacitance and frequency of the LC circuit, respectively. The changes in the three capacitances are detected by the shift of the corresponding pitches in the case of a force applied on the sensor. As shown in figure 8, three on-chip inductors (560 nH, 0402AF, Coilcraft Inc.) are conductively glued on the three electrodes of the sensor, and the other ends of the three inductors are electrically connected. An impedance meter (Agilent 4195A) was used to measure the impedance between the common ground of the sensor and the common electrode connecting the three inductors.

The relation between the change of capacitance, $\Delta C$, and the change of pitch, $\Delta f$, was deduced from equation (2), $C + \Delta C$ and $f - \Delta f$ were used to replace $C$ and $f$ in equation (2), respectively, and the following equation was derived:

$$\Delta C = \frac{2\Delta f}{2\pi L (f + \Delta f)^2}. \quad (3)$$

Considering equation (1), which shows the relation between the displacement $\Delta d$ and the applied pressure, and the relation between $\Delta f$ and $\Delta C$, a particular value of $\Delta f$ indicates a particular value of applied force. The three pitch shifts
with applied forces, $\Delta f_1$, $\Delta f_2$ and $\Delta f_3$, indicate the force applied on the three capacitors in the sensor, respectively. The direction of applied force can be derived by comparing the amplitudes of the three pitch shifts. Experimentally, the shift of the three pitches before and after an applied force at 400 mN is shown in figure 8(c).

Chen et al reported an implantable Parylene-based wireless passive intraocular pressure sensor using a similar microfabricated LC circuit to detect applied pressure, and a pressure resolution smaller than 1 mmHg was demonstrated [10]. In this study, the technology was extended to detect multi-pixels, and only two wires were employed. Considering the sensor can share the common ground with other electronics in a tubular medical device, only one additional lead is needed. Furthermore, this readout approach can be extended to more than three pixels, and is very attractive for a future tubular medical device with a small diameter and more functions.

Conclusions

A flexible, low-cost, tri-axis force sensor for application to tubular medical devices was demonstrated. In the fabrication process, SU-8 was used to construct and bond the sensor architecture. In the bench test, a force ranging from 10 mN to 1.0 N, operating temperature up to 65 °C and a tolerated shock of 2.0 N were verified. A readout approach was proposed to collect multi-channel signals by adding only one wire in the system. The new fabrication technique and readout approach are very promising for functional tubular medical devices with a small diameter and a limited inner space.

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