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LCP MEMS Implantable Pressure Sensor for Intracranial Pressure Measurement

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Abstract—**Numerous devices for the measurement of Intracranial Pressure (ICP) have been proposed; these include both wired and wireless systems, each of which having their own shortcomings. In this work, the following requirements for minimally invasive measurement of ICP are considered: low infection, biocompatibility, small size, accuracy and ease of use. With these requirements in mind, we propose an implantable ICP sensor based on Microelectromechanical Systems (MEMS). Liquid Crystal Polymer (LCP) is chosen as the biocompatible and flexible structure. This paper describes the significant characteristics of LCP that make it well suited for biomedical applications, and comparisons are made to other polymers in terms of biocompatibility and certification. The proposed LCP pressure sensor has been successfully fabricated, with a membrane area of 2mm x 2mm x 50µm.**

Keywords—Liquid crystal polymer (LCP); Piezoresistive pressure sensor; Implantable device; Intracranial Pressure (ICP); Microelectromechanical systems (MEMS)

I. BACKGROUND

In the case of severe neurological diseases and injuries the brain is able to swell which leads to an increase in pressure called intracranial pressure (ICP). It is necessary to measure, monitor, and maintain ICP, a normal level is 0-20 mmHg, above 20 mmHg the pressure regarded as harmful to patients. Traditional methods for measuring ICP require drilling the skull and inserting the catheter into a specific position depending on the type of device. In general there are two effective methods in terms of reliability. First the ventricular catheter; this 'gold standard' technique using a fluid-coupled concept, relies on inserting the catheter to the lateral ventricle. The system not only monitors ICP but is also useful for Cerebral Spinal Fluid (CSF) drainage. However, it has the highest risk of causing infection, and is particularly inconvenient to use since it requires recalibration if the patient moves their head; furthermore it requires skillful neurosurgeon for insertion. Intraparenchymal systems are a second method of measuring ICP, these systems may rely on fiber-optics, strain gauge catheter tips, or pneumatic devices. These systems are designed based on the concept of a catheter tip transducer, which is placed on the brain parenchyma; such systems are all siutable for bedside monitoring. The advantages of intraparenchymal systems include: low chance of infection, small drift, and one time calibration; the disadvantages include: the inability to drain CSF, fragility and expense [1,2]. It is clear more accurate techniques for measuring ICP are needed, alongside an a need to reduce the risk of infection through open drilled holes. Patients are also required to change the device every two weeks, resulting in high costs. The success of a minimally invasive device that is able to accurately measure ICP, and transmit wirelessly is dependent on its size, material and the longevity of the primary sensing element. ICP sensors based on MEMS technology are biocompatible, low-cost and miniature, all of these attributes are in accordance with the regulations specified by the Association for Advancements in Medical Instruments (AAMI). Micromachining of medical pressure sensors is common in MEMS technology due to high sensitivity and regularity; commercially this technology is based on conventional silicon fabrication. Kawoos et *al.* [6] used commercial MEMS capacitive pressure sensors in combination with a telemetric system; their sensor was covered with cylindrical case and positioned in the sub-dural space to make contact with the CSF. Kawoos et *al.* found errors in their pressure measurements in humid conditions; this was a result of in *vivo* testing [6,7]. Beyond the usage of the commercial sensors, Mostafa et *al.* [16] proposed near infrared communication with an ICP sensor based on silicon wafer structures integrated with a flexible membrane and micropillar on the sensing area. It has been commenced in MEMS fabrication and still developed in biocompatible materials. The concept of biocompatibility has been overcome with polymer MEMS pressure sensor as seen by a research on flexible ICP sensor made of polyvinylidene fluoride trifluoroethylene (PVDF-TrFE) which was able to performed both resonant and capacitive function though the properties of high dielectric constant material and piezoelectric diaphragm structure[8]. This work has been being attracted in Liquid Crystal Polymer (LCP) which was originally used in high-density printed circuit board (PCB) fabrication and electronics packaging. With the beneficial features in low water absorption, flexible substrate and mechanical robustness, this material has been developed in various microdevice applications [3,4,5]. There have been few biomedical applications shown the uses of LCP in MEMS based fabrication. For the neurological fields, LCP has been a role in microfabrication processes because of flexibility, durability and chemical inert across wide ranges of temperature. Wang *et al.* used LCP in fabrication process to make a lamination layer with sputtered iridium oxide film to construct flexible nerve electrodes [9]. S. W. Lee *et al.* also proposed a flexible neural depth probes by using LCP substrate. The fabrication was constructed with two different

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types of LCP film, high temperature film as the cover and the substrate layers and low temperature film as the adhesion layer. It has been used to study the neurological disease and examine the brain function in vivo, and developed as a low cost custom-made stimulating system based on LCP electrode array [10,11]. The fabrication and development of pressure sensors in biomedical applications have been designed with LCP material as sensing diaphragm based on capacitive or piezoresistive structures. By conventional fabrication processes and developing techniques, the deflectable membrane is made of LCP sheets in different types and thickness then assembled with the individual designed systems to vary with applied pressure [6,7,8].

II. Our Approach

This paper is described in the use of LCP as polymer based implantable pressure sensor to measure the elevated intracranial pressure through piezoresistive approach. To verify the feasibility of sensing system and the biocompatibility with organic environment, the structural fabrication is based on LCP substrate and deposited metal strain gauges. This pressure sensor is designed based on the concept of biocompatibility and flexibility, so LCP sheet is mainly used in the sensing structure of device. The size of sensor included contact pads is $8x8mm^2$ and 100μm thick at the entire structure, the dimension is able to be reduced to $5x5mm^2$ by minimizing the size of contact pads. The sensing membrane is $2x2mm^2$ square shape situated at the center of structure, the thickness of this area is designed approximately 50 µm to respond with the applied pressure as shown in figure 1. Based on piezoresistive concept, wheatstone bridges are configured within the sensor resulted in irrespective of external circuits. Four strain gauges sit on the measuring point to operate when the pressure is occurred, inducing the changes in resistances. This aspect will be implemented in future design of processing data through wireless system for external monitoring of ICP.

A. *LCP Characteristics and Biocompatibility*

Liquid Crystal Polymer(LCP) is constructed with rigid and flexible monomers linking together. Through it is classified as thermoplastic material, it has specific properties in term of uniformity. With the advantages of low moisture absorption (~ 0.02%) and permeability, LCPs are widely used in semiconductor packaging. Others are chemical resistance including acid, base and solvents in long duration and wide temperature range. These LCP materials are commercially supplied in 25μm to 3mm. thickness with 18μm thick copper cladding on one or both sides of each sheet. And each of product is slightly different in properties [5]. From FDA approval, some kinds of LCPs are applicable in medical field. In our work we used ULTRALAM® 3000 series liquid crystalline polymer circuit materials with double clad laminates from Roger Cooperation. The thickness of LCP was 100 µm with copper cladding both sides. The water absorption is 0.04% and chemical resistance is 98.7%. The melting temperature is 315°c which is able to withstand the heating in fabrication. However the scientific revelation of biocompatibility has been infrequently found in vivo testing for biomedical equipment. According to these characteristics, the potential of LCP and other polymers are shown in Table I to clarify the advantage and application in biological system [3].

B. *Characteristics and Comparison among LCP and other polymers*

TABLE I. List of Properties, Medical Class and Conclusion of LCP and Other Polymers

	Characteristics		
	Properties	***Med. Class***	***Remarks***
LCP	- High mechanical strength at high temperature - High chemical resistance - Low moisture absorption and Permeability - High biocompatibility - Good barrier for gases	USP class VI	- Specially designed types for medical application - Newly developed material in neurological field - Lower processing cost and easier bonding than Kapton
Polyimide (Kapton)	- Thermoxidative stability - High modulus - High insulator - High chemical resistance	*NC	- Still in study of Biocompatibility, Cytotoxicity, Hemolytic capability - No photo definable type for in vivo
PDMS	- Physiological indifference - Biodegradable resistance - High Biocompatibility - High gases and vapors permeability	USP class VI	- Most widely used polymer in biomedical application - No photo definable type for implantable grade
SU-8	- High optical transparency - Chemical and mechanical stability - Biocompatibility - Low cost processing	*NC	- Still in study of Biocompatibility, Cytotoxicity, Hemolytic capability - Low Processing cost than silicon micromachining - Unable to integrate with microelectronic circuitry
Parylene C	- Good biocompatibility - Chemical and biological inertness - Good barrier - Lubricious surface - Electrical insulator	*NC	- Deposited by using chemical vapor deposition (CVD) process under room temperature - Suitable material for chronic implants - More breakable in thin sheet

*Not Certified

C. Design of Structure and Sensitivity

As design of built-in wheatstone bridge circuit on the top of sensing diaphragm is changed under uniform pressure applies on strain of each resistors, the deflection of diaphragm is considerable to make the proper size of diaphragm under maximum loading condition. Considering the all edges fixed square plate, the maximum deflection $w_{\max}$ appears at the center of the plate and can be approximated by [13],

$$W_{max} = c(1-\nu^2)\frac{Pb^2}{Eh^3} \quad (1)$$

where

$$c = \frac{0.032}{1+\alpha^4} \quad (2)$$

α is the ratio of width and length of diaphragm. From above equation, p is the uniform applied pressure, E is the Young's modulus and v is the Poisson ratio. Moreover, b is the length and h is the thickness of diaphragm. This equation utilizes the design of diaphragm which concerns about the deflection to be smaller than the thickness of diaphragm under maximum pressure. According to the design of 0 - 50 mmHg pressure range, the maximum thickness of LCP membrane is defined to be 50μm thick with the cavity size of 2000μm x 2000μm x 50μm at the bottom, which does not exceed the maximum thickness of LCP when it is applied to the highest pressure. One the top of diaphragm, the gold strain gauges with 180Ω resistance value are spluttered on the four sides of diaphragm. Two of them are on edges of diaphragm and the others are placed close to the other sides that are operated as half bridge configuration.

At the edges of diaphragm, the maximum strain happens at the center of each edges due to the deflection of diaphragm, the maximum strain, $\varepsilon_{\max,}$ of the diaphragm is [14,15]

$$\varepsilon_{max} = 0.308(1-\nu^2)\frac{Pb^2}{Eh^2} \quad (3)$$

According to the relative change in resistance for a resistor segment deformed by being bond to the top of a plate is

$$\frac{\Delta R}{R} \approx \frac{1}{1-\nu}\varepsilon_l + \frac{2\nu-1}{1-\nu}\varepsilon_w \quad (4)$$

where ε_l and ε_w are the strains along resistors length and width. As $\varepsilon_l = \varepsilon_w$ can be defined from the stress at the center of an edge in four fixed edges plate[12], and substituted with v =0.3, b = 2000μm, h = 50μm and E = 2.1GPa (LCP). The relative change in resistance in term of applied pressure P is

$$\frac{\Delta R}{R} = (1.831\times10^{-7}Pa^{-1})P \quad (5)$$

D. Fabrication Processes

The fabrication of piezoresistive pressure sensor is shown in figure 2, 100μm thick LCP3850 double copper cladding is temporarily attached to a silicon support wafer by using adhesive layer of Photoresist (AZ9260) by spin coating. The wafer is heated up at 80°c for 20 minutes to make bonding securely. The Copper Etching is processed on the top copper cladded side as shown in figure 2(a) to expose LCP substrate. Then photoresist is removed to attach that bare LCP substrate on the silicon wafer. The conductive adhesion is applied on the edges before attaching LCP on thick silicon wafer. Two layers of 9μm thick photoresist are spin-coated on the surface of LCP and hard-baked at 80°c for 7 minutes then cooled down to a certain extent. After that the first pattern is transferred by lithography process, copper etching is performed again to make a hole as a copper mask for LCP etching process in figure 2(b). Figure 2(c) shows the capability of deep and high-aspect ratio feature, deep reactive ion etching (DRIE) is selected for this process, even there have been some research groups working on LCP etching but the etching recipes is still changeable depended on the condition of machine. With regard to optimization, the etching recipe is analyzed for this purpose. The etching is processed for 180

Fig. 1. SEM of cross sectional view of 50μm thick liquid crystal polymer (LCP) substrate after the optimization process by using DRIE: (a) the thickness of copper cladded layer and LCP substrate and (b) thick 40μm etched LCP

minutes with 800watts power until reached 50µm thickness as shown in figure 1(a) and (b). This process is delicate and needed to be under controlled. High power would have an affect on PR and copper mask to reflow. Next LCP is released from silicon wafer by soaking in acetone and flipped over to another side, copper cladding side is placed on support silicon wafer again by the same previous process. Subsequently, 9µm thick PR is spin coated on LCP for patterning and lift-off process and this wafer is baked at 60°c for 45 minutes. In term of biocompatibility, Ti(20nm)/Au(150nm) is selected to make piezoresistors , They are sputter-deposited and passed to lift-off process as shown in figure 2(d) and (e).

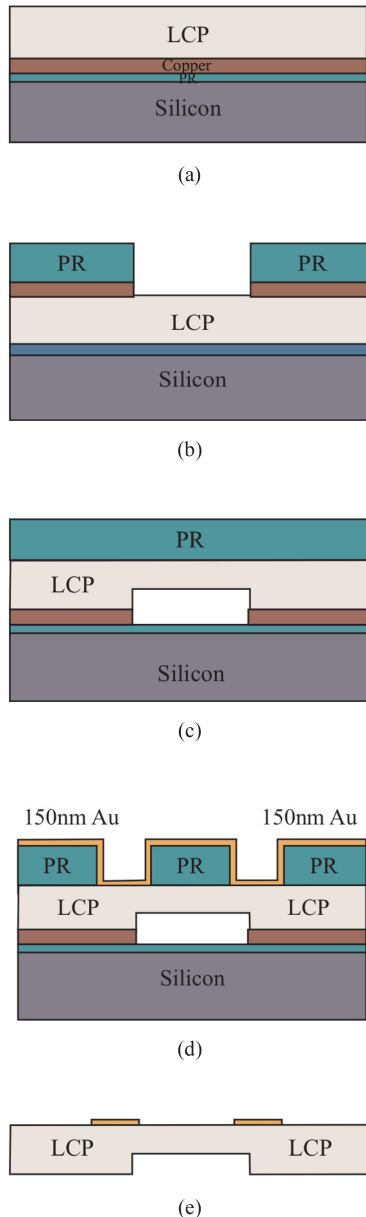

Fig. 2. A schematic of the fabrication processes: (a) one-side etched LCP with copper cladding attached on silicon substrate, (b) Photoresist patterning, (c) DRIE etching of LCP, (d) Ti(20nm)/Au(150nm) sputter-deposition and (e) the lift-off process to make piezoresistors

III. RESULT AND DISCUSSION

As shown in figure 3, the design of 0 - 50 mmHg pressure measurement is constructed as flexible pressure sensor and the maximum thickness of LCP deflecting membrane is defined to be 50µm thick with the cavity size of 2000µm x 2000µm x 50µm at the bottom of structure, which does not exceed the maximum thickness of LCP when it is applied to the highest pressure. Thick 150nm Au resistors are placed on the four sides of sensing diaphragm with the designed resistance 180 Ω as shown in figure 4 and 5. Due to theoretical result, the relative change in resistance for a resistor segment deformed by being bond to the top of a plate is (1.831×10^{-7} Pa^{-1}) P. In practical, this sensor is shown the changes of resistivity respect to applied pressure. The sensitivity and characteristics of the sensor will be performed in uniform pressure demonstrate system to evaluate the operation of the sensor. The experiment will be conducted in air and water to find the drift of the sensor. In the next step, this sensor will be integrated with ASIC (Application Specific Integrated Circuit) which is consisted of sensor interface, voltage controlled oscillator, RF powering system and wireless transmission to communicate with the external device through RF telemetric system .

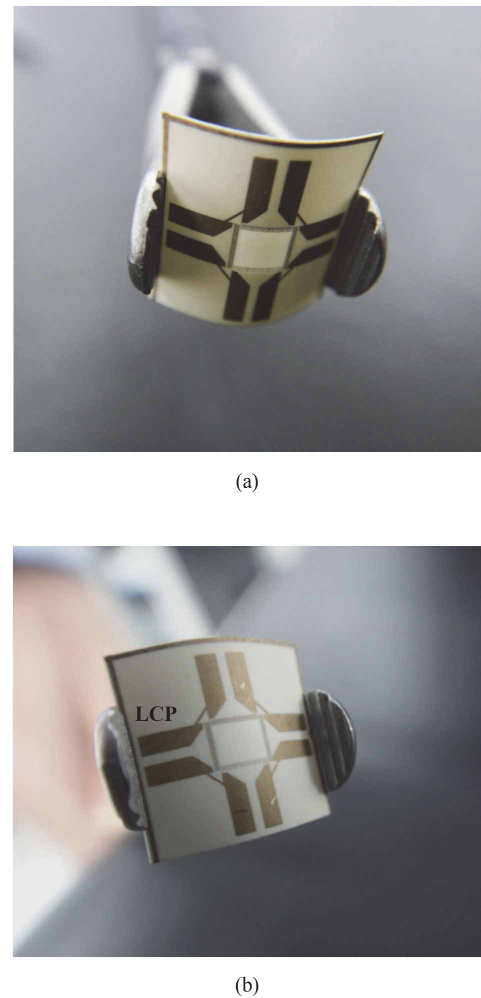

Fig. 3. Photographs of LCP pressure sensor: a) flexible structure with 50µm thick square membrane at the center and b)

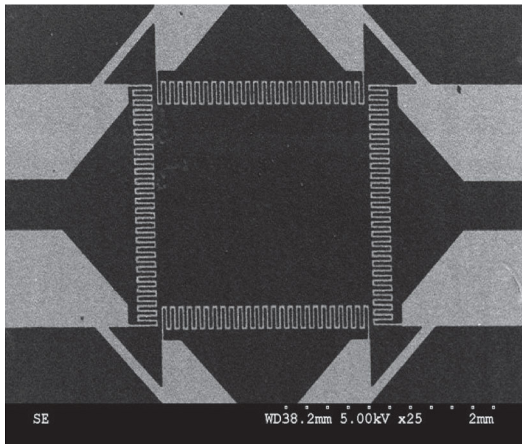

Fig. 4. SEM Photographs of LCP pressure sensor

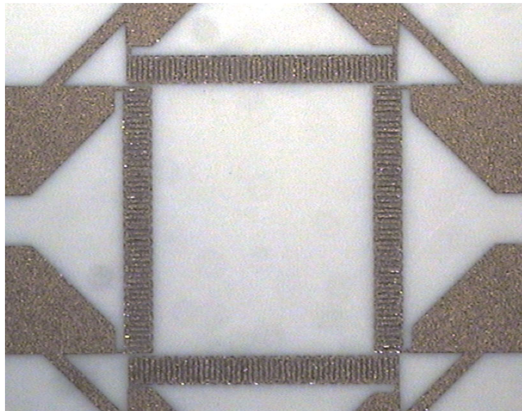

Fig. 5. Optical microscopic image of LCP pressure sensor and half bridge configuration with strain gauges

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