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On the feasibility of a liquid crystal polymer pressure sensor for intracranial pressure measurement

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Abstract: Intracranial pressure (ICP) monitoring is crucial in determining the appropriate treatment in traumatic brain injury. Minimally invasive approaches to monitor ICP are subject to ongoing research because they are expected to reduce infections and complications associated with conventional devices. This study aims to develop a wireless ICP monitoring device that is biocompatible, miniature and implantable. Liquid crystal polymer (LCP) was selected to be the main material for the device fabrication. This study considers the design, fabrication and testing of the sensing unit of the proposed wireless ICP monitoring device. A piezoresistive pressure sensor was designed to respond to 0-50 mm Hg applied pressure and fabricated on LCP by standard microelectromechanical systems (MEMS) procedures. The fabricated LCP pressure sensor was studied in a moist environment by means of a hydrostatic pressure test. The results showed a relative change in voltage and pressure from which the sensor's sensitivity was deduced. This was a proof-of-concept study and based on the results of this study, a number of recommendations for improving the considered sensor performance were made. The limitations are discussed, and future design modifications are proposed that should lead to a complete LCP package with an improved performance for wireless, minimally invasive ICP monitoring.

Keywords: hydrostatic experiment; intracranial pressure (ICP); liquid crystal polymer (LCP); MEMS; moist

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environment; piezoresistive pressure sensor; strain gauge; Wheatstone bridge.

Introduction

Traumatic brain injury (TBI) can lead to neurological disorders such as paralysis, loss of sensation and unconsciousness. TBI is mostly caused by car accidents in developing countries. An impact on the head can cause the brain to swell within the skull, which corresponds to brain shift and death. For brain injuries, the cerebral blood flow (CBF) is excessive and causes the rise of intracranial pressure (ICP) called intracranial hypertension (ICH). The normal range of ICP is 7-15 mm Hg in the supine position for adults; a higher pressure is regarded as harmful to patients [1]. The Glasgow Coma Scale (GSC) scoring method is used to classify the severity of TBIs in order to provide correct treatment. In all severe (GSC less than 9) and moderate (GSC from 9 to 12) head injury cases, the ICP must be monitored [2]. Some neurologic diseases such as brain tumors and meningitis can cause raised ICP. Typically, one of the four mechanisms: cerebral edema, vascular congestion, hydrocephalus and mass lesion causes raised ICP (see Ref. [3]). These patients need continuous ICP monitoring with a drainage system as part of their long-term treatment. This study aimed to develop an ICP monitoring technique for the acute treatment. In general, the ICP monitoring unit is connected to an embedded sensing device inside the head. Several conventional devices consist of a catheter with a pressure sensor connected to an external monitoring unit. The catheter is inserted by drilling through the skull and placed in a particular position. Each device is specific to the patient's condition and requirements as shown in Figure 1.

Conventional ICP measurement techniques and recent developments

A subarachnoid screw or bolt can be used for an instantaneous measurement of ICP; it is placed among the arachnoid membrane and cerebral cortex to penetrate the dura. The bolt lumen is filled with cerebral spinal fluid

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Figure 1: Conventional ICP monitoring devices.

(CSF) and is connected to an external transducer through a fluid-filled sterile tube. One limitation of this method is that the overall ICP can be underestimated due to the damping of pressure by the dura.

An epidural sensor can be placed in the epidural space beneath the skull without penetrating the dura; the catheter is inserted through the subarachnoid bolt and locked to prevent dislocation and to seal the device from the atmosphere. This method does not require the passage of fluids to external monitors and it shows the lowest infection rate among other methods [4]. The disadvantage of this method is that there is a measurement drift after a few days of continuous monitoring which requires a catheter replacement, and epidural monitoring is considered the least accurate method of those discussed here.

Two more reliable and accurate techniques are a ventricular catheter and an intraparenchymal system. The ventricular catheter is inserted into the lateral ventricle by performing ventriculostomy and connected to a transducer and a collection system through the fluidfilled tube. The device can act as both a monitoring unit and a CSF drainage system to reduce the raised ICP; it can also be used to administer medication intracranially. This technique is regarded as the "gold standard" and the most cost-effective method; however, it is the most invasive and requires multiple calibrations at every elevating head position. In addition, a raised ICP causes the brain to expand resulting in access difficulties, so this technique requires an expert neurosurgeon to carry out the procedure, which is not suitable for immediate monitoring.

The intraparenchymal system (fiber optic or wire type) consists of a strain gauge catheter tip and pneumatic devices. It can be inserted through the bolt and locked to place on the brain parenchyma in order to reduce the difficulty of the previous method. However, the fiber optic devices are fragile, expensive and cannot be recalibrated after insertion. Table 1 shows the advantages and disadvantages of these conventional techniques [5–7].

Table 1: The comparison among the conventional ICP measurement techniques (\checkmark = with and \times = without).

	Intraventricular catheter	Intraparenchymal system	Subarachnoid screw or bolt	Subdural and Epidural catheter
Accuracy	1st	2nd	3rd	4th
CSF drainage	✓	×	×	×
CSF sampling	1	×	1	×
Invasive to brain tissue	High	Low	×	×
Risk of infection	Highest	High	Low	Lowest
Placement Procedure	Difficult	Easy	Easy	Easy
Blocked by tissue or clotted blood	1	×	1	×
Risk of bleeding	1	1	1	1
Recalibration	1	1	1	×
Continuous fluid column	\checkmark	×	1	×

A comprehensive review of ICP monitoring was given by Kawoos et al. [8]. In their study, ICP monitoring was categorized as being direct or indirect. Indirect methods are non-invasive in nature and rely on evaluating ICP from some measurement other than ICP. The most common methods for ICP monitoring are invasive due to their accuracy and reliability. Kawoos et al. grouped invasive methods into three sub-categories: fluid-filled systems, transducertipped catheters and telemetric methods. It was shown that the trend for implantable and telemetric devices is increasing for ICP monitoring; these techniques can provide advantages in patient mobility and present a lower risk of infection. However, most of the research on implantable devices has been conducted for non-clinical applications.

Minimally invasive ICP measurement techniques have gained more attention recently as they offer the following advantages: implantation, wireless data/power transmission, biocompatibility and miniaturization. Campus Micro Technologies (Bremen, Germany) [9] developed short- and long-term ICP measurement devices based on an absolute capacitive pressure sensor. The implant does not require a battery but is powered via radio waves from an external unit. Millar Instruments (Houston, TX, USA) [10] remotely measured a rat's ventricular and arterial pressure by utilizing a solid-state pressure sensor. Kiefer et al. [11] developed a commercial telemetric ICP sensor; the device consists of a silicon-based microelectromechanical systems (MEMS) piezoresistive pressure sensor for absolute pressure measurement and a temperature sensor. This sensor is integrated with an application specific integrated circuit (ASIC) to digitize and process data, and data/power transmission is achieved through radio-frequency identification (RFID).

Integrated Sensing Systems (Ypsilanti, MI, USA) [12] has commercially developed variable silicon-based MEMS capacitive pressure sensors, which they applied in their wireless implantable pressure sensor. Signals can be transmitted via magnetic telemetry to a handheld device. Kawoos et al. [13–15] developed a wireless ICP sensor based on two commercial pressure sensors: piezoresistive (Merit Sensor Systems, South Jordan, UT, USA) and MEMS capacitive (Murata Electronics Oy, Vantaa, Finland) pressure sensors. Two analog devices integrated with fabricated planar inverted F antennas (PIFAs) were implanted on the epidural and subdural layers, respectively. These commercial sensors were shown to be influenced by a moisture-rich environment; as such, they require suitable encapsulation. Kawoos et al. concluded that the subdural device is more accurate than the epidural device and that wireless monitoring of ICP would be useful for long-term treatment. Presently, MEMS technology can be found in several biocompatible applications such as drug delivery systems,

surgical equipment, artificial organs and micro fluidic devices [16–23]. MEMS-based pressure sensors are attractive because they require less power consumption and low manufacturing costs. Furthermore, devices can be miniaturized and fabricated from biocompatible materials thus making MEMS devices well suited for medical applications.

Several researchers have focused on MEMS-based ICP sensors [23]. Ginggen et al. [24] proposed a longterm telemetric pressure sensor. This absolute pressure sensor consists of an ASIC to convert the capacitances to frequency-encoded signals due to the change of pressure. The device does not require external calibration, sensor protrusion or a burr hole in the skull. The capacitive pressure sensor was fabricated, integrated with ASIC and attached to a glass disk at the bottom of the device. Other electronic components were placed on the second layer, while all components were covered with a hermetical glass chamber and insulated with Parylene C. The device is 13 mm in diameter and 4.5 mm in height, and it measures pressure over a range of 600-1200 mbar (450-900 mm Hg). Ghannad-Rezaie et al. [25] applied the concept of optomechanical transduction: based on infrared communication, this fully implantable pressure sensor offers powerless and remote operation. A near-infrared fluorescent-based optomechanical or NiFO sensor was fabricated by using three layers of silicon wafer integrated with a twowavelength quantum dot (QD) micropillar, a mini-lens and a silicon nitride membrane. The changes in pressure deflect the backside membrane and the sensor converts pressure changes into the changes of fluorescent intensity ratio between two QD wavelengths (705 nm/800 nm ratio) when illuminated with a laser source on the readout platform. The operating pressure range is 0–40 mm Hg.

For implantable applications, polymer materials offer both insulation and encapsulation. The near-hermetic property of polymers is a useful mechanical property and aids the manufacturing process [26–28]. An example of successful polymer MEMS-based pressure sensor is from CardioMEMS (Atlanta, GA, USA), who developed capacitive-based sensors for the measurement of blood pressure. This implantable device used a passive inductor-capacitor resonator. The inductor structures were patterned on a liquid crystal polymer (LCP) material. The application of this device is to measure the pressure of abdominal aortic aneurysms (AAA) [29].

Advantages of using LCP

LCP is a thermoplastic polymer that contains links of rigid and flexible monomers. The structure is nearly isotropic such that rigid segments of molecules align in the shear flow direction in the liquid crystal state. The highly ordered structure occurs when it melts or is in a solid state; this characteristic separates LCP from other thermoplastic polymers. LCP has a low di-electric constant of ~3 in the range of 0 to 110 GHz and a low loss factor of 0.002-0.0045 at 110 GHz. It has a high dielectric strength that is suitable for high-voltage and high-power applications. LCP has low moisture absorption of 0.02–0.04% [30]. The density and Young's modulus are 1.4-1.6 g/cm³ and 11-24 GPa, respectively [31]. LCP is commonly used in high-density printed circuit boards (PCBs) and semiconductor packaging because of its various chemical resistances: it is resistant to acids, bases and solvents for long periods and at various temperatures. LCP is a good gas barrier and it can withstand humidity. LCP is an attractive material due to its easy fabrication thus leading to low manufacturing costs [30, 32, 33].

LCP has shown great promise in MEMS-based devices for biomedical applications. LCP provides biocompatibility, low moisture absorption, flexibility and durability [34–37]. LCP is chemically inert across a wide range of temperatures and is thus well suited for conventional microfabrication processes. LCP can be simply integrated with electrical components and other substrates to provide a variety of designs [38–41].

Design and fabrication

The novelty of the present study was to apply a Wheatstone bridge fabricated using LCP for a simple and compact wireless ICP monitoring device. The literature review shows that LCP has been used for the fabrication of capacitance-based pressure measurement, but to the authors' best knowledge an LCP Wheatstone bridge pressure sensor has not previously been developed for the purpose of ICP monitoring wirelessly. The proposed device satisfies the requirements of biocompatibility, miniaturization, implantation and affordability. The proposed sensor consists of three main parts: a pressure sensing unit, a power unit and a data telemetry unit. This study focused on the pressure-sensing unit (pressure sensor) regarding the design (Section "Mechanical and electrical characteristics of LCP MEMS pressure sensor"), fabrication (Section "Fabrication of LCP pressure sensor") and performance evaluation (Section "Performance evaluation of LCP pressure sensor").

The sensor is required to be biocompatible and flexible, thus an ULTRALAM[®] 3850 LCP sheet (Rogers Corporation, Chandler, AZ, USA) [42] is used for the fabrication of the main device. The size of the sensor including the contact pads is $8 \times 8 \text{ mm}^2$ and the thickness is 100 µm. The $2 \times 2 \text{ mm}^2$ sensing membrane is situated at the center of the structure. A 50-µm thick sensing membrane is designed for an applied pressure of 0–50 mm Hg. The sensing operation is based on the piezoresistive concept in response to internal strains. The built-in Wheatstone bridge circuit is designed to be placed on the top of the sensing membrane. The half-bridge configuration is applied by using two variable resistors and two fixed resistors. Two variable resistors are placed on the two edges of the sensing membrane. When the pressure applies on the sensor, the two variable resistors will be deformed which results in changes in resistance.

Mechanical and electrical characteristics of the LCP MEMS pressure sensor

To sustain a maximum pressure, the size of the sensing membrane is designed to provide the optimal deflection. A two-dimensional (2D) membrane is used to study the LCP membrane deflection. From the Kirchoff-Love plate theory [43], the square plate is considered with all edges fixed. A uniformly distributed pressure is applied on the top surface to observe the membrane deflection [15, 29, 44–46]. This LCP membrane is considered a linearly elastic material which is homogenous and isotropic, such that the deflection relies on Young's modulus (E) and Poisson's ratio (ν) of the material. Even if deflection occurs, there is no change at any point on the line that is perpendicular to the middle plane of the plate. Therefore, the membrane's deflection causes normal and shear stresses, which create moments and shear forces, respectively. An equilibrium state exists such that the summation of all forces and moments is given in Equation (1):

$$\frac{\partial^4 w}{\partial x^4} + 2 \frac{\partial^4 w}{\partial x^2 \partial y^2} + \frac{\partial^4 w}{\partial y^4} = \frac{p}{D}$$
(1)

where p is the uniform pressure, w is the normal displacement for a point of membrane at a location of (x, y) and D is the bending stiffness of the membrane as shown in Equation (2):

$$D = \frac{Eh^3}{12(1-\nu^2)}$$
(2)

Equation (1) is a complex partial differential equation and can be solved by Fourier series and the use of boundary conditions and results in the following equation:



Figure 2: 2D model of the LCP pressure sensor comprising the LCP outer membrane, sensing membrane and the gold piezoresistive material.

$$W = \frac{4pa^4}{\pi^5 D} \sum_{n=1,3,5}^{\infty} \frac{1}{n^5} \left(1 - \frac{\alpha_n \tanh \alpha_n + 2}{2\cosh \alpha_n} \cosh \frac{n\pi y}{a} + \frac{n\pi y}{2a\cosh \alpha_n} \sin h \frac{n\pi y}{a} \right) \sin \frac{n\pi x}{a}$$
(3)

Boresi et al. defined the maximum deflection (W_{max}) and provided an analytical solution:

$$W_{\rm max} = C(1 - \nu^2) \frac{Pb^4}{Eh^3}$$
 (4)

where *b* and *h* are the width and height of the membrane. The dimensionless constant (*C*) can be obtained from the assumption of all fixed edges, which is calculated as the ratio of the width and length of the membrane (α) as follows:

$$C = \frac{0.032}{1 + \alpha^4}$$
(5)

Equation (3) was used in the design of the sensing membrane to determine the membrane thickness with respect to the maximum pressure. The thickness of the sensing membrane must be less than the total thickness of the LCP outer membrane under maximum pressure. The thicknesses of the outer and the sensitive membranes are determined by the pressure range to be measured (0-50 mm Hg), the maximum strain that the sensing membrane can sustain, and the miniature size required. Hence, the pressure range that the sensor is designed to measure is between 0 and 50 mm Hg, in which case the maximum pressure at 50 mm Hg was used to optimize the size of the sensor. The thicknesses of the outer and sensitive membranes relate to the maximum strain induced by the maximum pressure. The sensing membrane situated at the middle of the sensor was designed to be able to sustain the maximum pressure and not exceed the outer membrane of the sensor. Applying this theory, the LCP pressure sensor is designed to consist of an 8 mm \times 8 mm \times 100 μ m outer membrane and a $2 \text{ mm} \times 2 \text{ mm} \times 50 \mu \text{m}$ sensing membrane above a bottom cavity, as shown in Figure 2. In this case, b = 2 mm, $h = 50 \mu$ m, v = 0.3, $\alpha = 1$ and E = 2255 MPa. Moreover, the maximum deflection under 50 mm Hg is approximately 5.509 μ m. Therefore, the center deflection to the pressure ratio is as follows:

$$\frac{W}{P} = 0.8265 \text{ nm}/\text{Pa.}$$
 (6)

A COMSOL (Burlington, MA, USA) based finite element method is used to simulate the membrane's deflection under different uniformly distributed pressures. This simulation can investigate the stress and strain in the LCP membrane to ensure no plastic deformation occurs during measurements and to assist with the placement of strain gauges. As LCP is a nearly isotropic material, the simulation can be applied with a linearly elastic stress-strain function. The total thickness of the outer membrane is limited to 100 µm. The 2-mm wide, 50-µm thick sensing membrane is designed to prevent deformations exceeding the outer membrane as shown in Figure 2. Figure 3 shows a three-dimensional (3D) model of the maximum membrane deflection that occurs at the center of the diaphragm. The maximum membrane stress (and strain) occurs at the middle of the edges of the sensing membrane



Figure 3: 3D model of the LCP membrane deflection with the applied pressure.

as shown in Figure 4. The simulations identify where the strain gauges should be placed for maximum deflection.

As the shear strain is omitted due to its infinitesimal translation to parallel resistors, the maximum strain variation is obtained from the strain tensor in the X component over a range of 0–50 mm Hg as shown in Figure 5. Figure 6A shows the deflection behavior of the LCP membrane at pressures of 0, 10, 20, 30, 40 and 50 mm Hg. There is no deflection at 0 mm Hg which corresponds to the atmospheric pressure in the bottom chamber. At the pressure of 50 mm Hg, the computed deflection is approximately 5.681 µm. The graphs present a linear deformation of the LCP membrane with the applied pressure. The results show that the designed thickness is appropriate to maintain the elasticity of the sensing membrane and to not exceed the total thickness. The numerical results of the membrane deflection are plotted according to the applied pressure. Using Equation (5), the maximum deflection can be calculated and plotted with respect to the pressure at 0,

10, 20, 30, 40 and 50 mm Hg. The numerical and analytical results have a 3.07% difference, which indicates a good agreement (see Figure 6B).

The built-in Wheatstone bridge circuit is designed to be placed on top of the sensing membrane and change under uniformly distributed pressure. When the pressure occurs, the strain of each resistor changes. The serpentine strain gauges are designed to increase the length of the resistor in order to provide effective resistance. The gold strain gauges are placed on the edges of the sensing membrane where the maximum stress occurs. According to the formula for resistance within a wire, the change of resistance is determined by the bulk resistivity (ρ), length and cross-sectional area of the wire. The resistivity constant of gold is $2.44 \times 10^{-8} \Omega \cdot m$, and the total resistance is approximately 180 Ω for each strain gauge. The maximum strain occurs at the center of each edge due to the membrane's deflection, and the maximum strain (ε_{max}) is obtained from ref [47].



Figure 4: 3D model of the membrane stress along the four edges of the LCP sensing membrane with the applied pressure.



Figure 5: The maximum strain variation with the applied pressure at the center of the edge.



Figure 6: Simulation results of the LCP membrane at pressures of 0, 10, 20, 30, 40 and 50 mm Hg (A) 2D simulation results showing the LCP membrane deflection with the applied pressure in the cross-section along the width; (B) the comparison of the maximum membrane deflection between the FEM and the analytical method with the applied pressure.

$$\varepsilon_{\rm max} = 0.308(1 - v^2) \frac{Pb^2}{Eh^2}.$$
 (7)

From Equation (7), the relative change in resistance for a resistor segment ($\Delta R/R$) deformed by being bonded to the top of the plate is as follows:

$$\frac{\Delta R}{R} \approx \frac{1}{1 - v_{\rm R}} \varepsilon_l + \frac{2v_{\rm R} - 1}{1 - v_{\rm R}} \varepsilon_{\rm w} \tag{8}$$

where ε_l and ε_w are the strains along the resistor length and width, respectively. ν_R is the isotropic Poisson's ratio of the gold resistor which is 0.42, respectively. By symmetry, the maximum strain can be estimated from Equation (7) and substituted in Equation (8) for the relative change in resistance in terms of the applied pressure (*P*) as follows:

$$\frac{\Delta R}{R} \approx (2.743 \times 10^{-7} \,\mathrm{Pa}^{-1})P \tag{9}$$

The theoretical strain can be found from the finite element method (FEM) in the range of $0-1.174 \times 10^{-3}$. Using the half-bridge configuration, the sensitivity is defined by the ratio of the output voltage (V_o) to the supply voltage (V_o) [44].

$$\frac{V_o}{V_s} \approx \frac{-\alpha_4 - \alpha_1 - \alpha_4 \alpha_1}{4} \tag{10}$$

where α_1 and α_4 are the relative changes in resistance for the resistors in terms of the surface strain. Hence, α_1 and α_2 are obtained from Equation (8) by using the theoretical strain from the FEM and substituted in Equation (10). So, it presents as follows:

$$\frac{V_o}{V_s} \approx (12.77 \times 10^{-8} \text{Pa}^{-1})P$$
(11)

Fabrication of the LCP pressure sensor

Conventional MEMS fabrication techniques can be used to make the proposed LCP pressure sensor [48]. As LCP is more delicate than silicon, the etching recipe is optimized to find the appropriate power to avoid overheating. The first step is to make the 50- μ m sensing membrane. The 100- μ m thick ULTRALAM 3850 LCP substrate is temporarily attached to a silicon wafer using an adhesive layer of photoresist (AZ9260) (Microchemicals GmbH, Ulm, Germany). The top copper cladding is then etched to expose the LCP layer, and the square pattern is transferred to the surface using lithography. The copper etching is processed to make a square copper mask for the LCP etching process. Deep reactive ion etching (DRIE) is periodically processed until the depth of the chamber reached 50 μ m as shown in Figure 7A.



Figure 7: Previous fabrication process for the LCP pressure sensor: (A) bottom view of a sensor with $2 \text{ mm} \times 2 \text{ mm} \times 0.05 \text{ mm}$ cavity at the center sensing membrane; (B) top view of a sensor with the transferring gold strain gauges and contact pads; (C) four serpentine strain gauges at the edges of the sensing membrane; (D) an array of $8 \text{ mm} \times 8 \text{ mm} \times 0.1 \text{ mm}$ LCP pressure sensors.

The second step is to make the strain gauges. The bare LCP side is flipped over and the etched side is attached to the silicon wafer by the aforementioned processes. To make the serpentine pattern on the LCP side, the serpentine pattern is transferred to the LCP substrate using the backside alignment method during the photolithography process as shown in Figure 7B. The recipes of exposure and developing time are optimized to achieve the complete strain gauge.

Due to the benefit in biocompatibility, Ti (20 nm)/Au (150 nm) is sputtered on the LCP substrate to make the piezoresistors. After the lift-off process, the complete strain gauges are obtained as shown in Figure 7C. The copper etching is processed to remove the last copper cladding. This LCP substrate is easily cut into 8 mm×8 mm LCP pressure sensors as shown in Figure 7D. The bottom cavity is sealed to prevent airflow in the pressure measurement experiment. For the primary testing, adhesive epoxy is used to seal the bottom cavity of the glass in ambient conditions. The conductive parts are coated with polydimethylsiloxane (PDMS) (Dow, Midland, MI, USA) in order to test in a hydrostatic pressure experiment.

Performance evaluation of the LCP pressure sensor

The LCP pressure sensor is designed to operate in moist environments in the pressure range of 0–50 mm Hg. The

Hydrostatic pressure measurement results

functional.

The resistance measurement of each strain gauge at the atmosphere is about 257.3–308.8 Ω . The sensor output

fabricated LCP pressure sensor is tested in a hydrostatic

pressure experiment to evaluate the feasibility and performance of the sensor in a moist environment. Theoreti-

cally, hydrostatic pressure is independent of the shape and

volume of the container [49]. The hydrostatic pressure at a

measuring point corresponds to the filling height of a con-

stant density liquid. In this case, the filling height at each

measuring point is calculated to be equal to the designed

pressure at 0, 10, 20, 30, 40 and 50 mm Hg, respectively.

The LCP pressure sensor is immersed at each filling height

and the pressure is measured via the induced change in

the resistivity. The experimental equipment consists of a

National Instruments (Austin, TX, USA) data acquisition

board (USB-6289), a power supply and a water tank. The Wheatstone bridge is supplied with 3 V; the output signal

is acquired in a LabVIEW Signal Express (NI, Texas, USA) at a sampling rate of 10 Hz and recorded for 60 s for each

pressure measurement. The experimental set-up is shown

in Figure 8. The relative change in voltage and pressure

presents the sensors sensitivity. The hydrostatic pressure

test was performed 6 times, the sensor was positioned at

different depths and the sensor output was recorded for 60 s. After the test was completed, the sensor was still



Figure 8: Schematic diagram of the experiment setup for testing the LCP pressure sensor.



Figure 9: Graph of the output voltage under 5, 10, 15, 20, 25 and 30 mm Hg pressure ranges recorded at each pressure for 60 s.

voltage is recorded at room temperature and is plotted against time for various pressures in Figure 9. The signal is passed through a low-pass filter with a cutoff frequency of 0.1 Hz. The output voltage decreases with increasing pressure; however, at pressures greater than 30 mm Hg, the voltage is constant and does not change with increasing pressure. Six repeated measurements were conducted with the same experimental conditions. The output voltage was averaged and plotted with respect to the pressure as shown in Figure 10. The repeatability of the sensor was obtained from the standard deviation of the repeated measurements and is presented as error bars in Figure 10. A linear relationship is seen between the output voltage and pressure. The output voltage at 0 mm Hg is 0.294 V. The output voltages at 0 mm Hg were measured at the atmospheric pressure before immersing in water. The average sensitivity was calculated as the gradient of the best-fit line fitted to the data points in Figure 10 and it was found to be 48.76 μ V mm Hg⁻¹.



Figure 10: Graph of the averaged output voltage under 5, 10, 15, 20, 25 and 30 mm Hg pressure ranges.

Discussion

The goal of the hydrostatic pressure experiment was to prove the device concept. The results show that the proposed LCP pressure sensor can operate in a moist environment, which replicates a biological environment. The simulation results showed the sensor is able to operate at pressures from 0 to 50 mm Hg, whilst the experiments showed the sensor is only able to operate at pressures from 0 to 30 mm Hg. The maximum pressure that the sensing membrane can respond to is 440 mm Hg, and this corresponds to the maximum possible displacement of the sensor membrane which is 49 µm. This upper limit is dictated by the geometry of the sensor. However, this estimation does not consider the strain gauges; for high pressures, the gold strain gauges may fail. The performance of the sensor is acceptable as the measurement is still within the range for measuring the ICP. This discrepancy may be caused by the primary packaging, which may have allowed fluid to fill the bottom chamber of the device.

The simulation results from the FEM give the maximum sensitivity of the strain gauge as 51.04 μ V mm Hg⁻¹. The FEM results correspond to the experimental results. The percentage difference between the simulation and experimental results is 4.47%. This discrepancy between the experiment and the FEM can be explained by several possible reasons. For example, the fabrication process may have caused a dissimilarity in the resistance of the strain gauges, and the sensing membrane may be thicker than the design. Additionally, it may be the assumption of ideal clamping at the edges of the membrane that may have influenced the simulation.

The pressure measurements presented in Figure 10 are clearly subject to large experimental errors. This is likely due to the fluctuating sensor output, which in turn is likely due to the interference in the Wheatstone bridge and the data acquisition rather than instabilities in the sensing membrane. Irrespective of these experimental errors, a clear trend is present, showing that voltage decreases in a linear fashion with increasing pressure. This encouraging result demonstrates the sensor works as intended. It should be noted that the only post-processing conducted was a first-order low-pass filter.

In future work, the developed simulation can be used to investigate the nature of the pressure distribution and the implications of a non-uniform pressure distribution. The sensor fabrication process will be implemented to improve the packaging using a heat bonding technique. The sensitivity of the sensor can be improved by decreasing the thickness of the sensing membrane. Furthermore, the size of the entire sensor can be further minimized to enhance its non-invasiveness. Further experiments could be considered to characterize the sensor in terms of response time, subtle sensor fluctuation and reliability.

Additional signal processing could be considered in future work to reduce the output signal noise and other error sources, and this additional signal processing should lead to improved repeatability and hence reduced measurement uncertainty. As mentioned in the design, the pressure sensing unit will be integrated with two other parts; the powering unit and the data telemetry unit. These two units will be assembled as a single electronic platform including the signal conditioning part that will prepare the measured signal to transfer to the data telemetry unit. The powering unit will supply the electrical power to the pressure sensing unit and other parts on the same platform. At the present stage, a 3.6-V, 120-mAh rechargeable lithium ion battery will be used as the power source in the feasibility studies of data transmission through a scalp phantom. The system will be operated in continuous mode and the power consumption will be considered in this future work.

Conclusion

The presented concept of a biocompatible LCP pressure sensor was designed, fabricated and tested *in vitro*. By using the piezoresistive concept, the sensor is composed of a built-in Wheatstone bridge formed by gold strain gauges; this configuration is easy to integrate with other circuits. The feasibility of the sensor was demonstrated in a moist environment in the pressure ranges of 0–30 mm Hg, which is a sufficient range for ICP measurement. This study shows that an LCP Wheatstone bridge-based pressure sensor is feasible and suitable for ICP measurement. The simple fabrication of the sensor can minimize manufacturing costs for both custom and mass production. Future work will focus on integrating to implantable telemetry communication, packaging, improving the performance of the device and testing *in vivo*.

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