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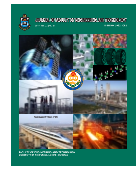

EVALUATION OF LC RESONANT PRESSURE SENSOR FOR SMART STENT APPLICATION

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Abstract

This paper aims to project the design and evaluation of inductor capacitor (LC) pressure sensor for smart stent application. The sensor was fabricated through ultraviolet (UV) patterning process using SU-8 polymer material because of its biocompatibility and suitability to fabricate small patterns. In the second phase of the research illustrates the measurement of pressure difference of blood and wireless monitoring inside the coronary artery having a stent implant equipped with LC resonant pressure sensor. This rectangular shaped pressure sensor (with and area of 4 x 4 mm^2), consists of an inductive coil and a capacitive plate, which exactly matches with the radio frequency-identification (RFID) design. The LC pressure sensor circuit is able to communicate with external antenna by matching the resonance frequency of both circuits. The fabricated sensor has 1.3 pF capacitance, 100 nH inductance with the resonance frequency about 440 MHz. The fabricated sensor is placed in a small vacuum chamber and the chamber pressure is controlled by using a syringe pump. The change of 100 MHz in the resonance frequency is observed in a pressure range of 0 to 16 kPa. Through a series of experimental analysis, it has been perceived that the discussed sensor can effectively measure pressure difference inside the vacuum and the change in resonance frequency can be further detected by using wireless equipment. The sensor can be widely used in the applications to effectively monitor the pressure difference of blood inside the human coronary artery.

Keywords : LC Sensor , Pressure Sensor, SU-8, Smart Stent, UV patterning

1. Introduction

Hardening of blood vessels known as “Atherosclerosis” is a type of disease which cause blockage in the arterial wall due to the deposition of plaque [1]. To treat partially blocked arteries a minimal invasive procedure of stent implant is followed in most of the cases which is known as percutaneous transluminal coronary angioplasty (PTCA) [2]. Commercial stents are mainly composed of tubular bodies fabricated by utilizing stainless steel or shape memory alloy material to reopen the blocked blood vessel. Every year more than 1.5 million stents are implanted in patients having partially blocked coronary arteries [3]. Even though PTCA procedure is an effective treatment of atherosclerosis, but about 30-40% patients

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suffer from a complication known as restenosis within 6 months [4]. During the stent deployment arterial wall may damage and immune system will begin to heal the damaged arterial wall. This healing process may deposit platelets and protein (fibrin) around the stented region leading to blood clot formation. After clotting, white blood cell starts to stick on the damaged surface leading re-narrowing of blood vessel around stented region [5].

To tackle restenosis, different kinds of stents are available in the market such as biodegradable stents [6] and drug eluting stents (DES) [7]. DES loaded with drugs such as sirolimus and paclitaxel provides a temporary solution to prevent restenosis but this technique has a risk of late stent thrombosis [8]. Restenosis is a serious threat to patient having PTCA treatment and an early detection of in-stent restenosis is very crucial to tackle the problem. Current diagnosing techniques such as electrocardiography, blood test, nuclear imaging, ultra sound testes and catheter based diagnoses are invasive, do not provide real time monitoring and lack accuracy [9]. Wireless inductor-capacitor LC resonant sensors operating under radio-frequency (RF) can be found in many biomedical devices, such as, wireless pressure sensor for continuous intraocular pressure (IOP) monitoring inside an eye for glaucoma patients [10], LC sensor for cardiac implants [11]. Most of these devices work by utilizing passive LC resonant circuits whose resonance frequency is changed in response to physical or biological variation.

We aim to design an LC resonant pressure sensor in a way that the stent can continuously monitor blood pressure inside the coronary artery having stent implant in order to improve the real time monitoring of stent and early detection of restenosis. The vessel containing stent if re-narrowed by restenosis will prevent the blood flow inside the artery. The reduction in the blood flow inside narrowed artery will cause a change in blood flow per unit area which will appear as a pressure drop [12]. This physical phenomenon is possible to measure as a change in pressure by using an LC pressure sensor. For this purpose, the pressure sensor should have the following features:

1. It should be smaller than the stent to be inserted within the artery.
2. It should have appropriate size and thickness so as not to interfere with the flow of blood.
3. Should be able to measure minute change in blood pressure inside the coronary artery.
4. Sensor should be fabricated by utilizing biocompatible material [11].

In the current study, a sensor of a size of several millimetres is designed and manufactured by using a biocompatible polymer named as SU-8. SU-8 is an acrylic-based type polymer which enables us to manufacture a sensor structure of the desired form [13]. It can also be used to produce a thin miniaturized pressure sensor with adjustable thickness by spin coating method, which has a possible advantage in the free form integration inside the stent. In addition, this LC resonant sensor can be integrated with the stent structure and can effectively measure the pressure changes in region having condition similar to coronary artery, this provide the first confirmed possibility for the development of an intelligent stent.

2. Sensor design and fabrication

2.1. Sensor Operating Principle

Figure 1 illustrates the schematic diagram of LC pressure sensor with top and cross section view. The sensor is designed to have a square form shape containing inductor coil and capacitor strips to build an LC resonant circuit. Polydimethylsiloxane (PDMS) is filled between the capacitor strips to keep the strips separated. Capacitance of the LC pressure sensor is altered as the distance between the capacitor plates is changed in response to the changing pressure. This change in capacitance as response of pressure change can be expressed as [14].

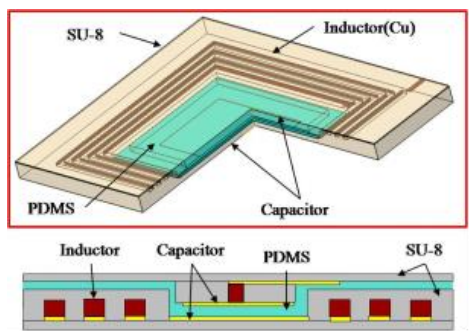


Figure 1. Schematics of L-C pressure sensor (Top), and view of cross section (Bottom)

$$C_s \approx C_0\left(1-\frac{\delta}{d}\right) \qquad (1a)$$

$$\delta = 0.02023p\left(\frac{a^4}{D}\right) \qquad (1b)$$

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

Where C_o gives the capacitance of a parallel plate capacitor at normal condition, d is separation distance between the capacitor plates, δ is the deflection of capacitor plate, p is the surrounding pressure, a is the half of the length of plate, D and h are flexural rigidity and thickness of the square capacitor plate, where *v* and E are the poison's ratio and young's modulus of the capacitor material respectively.

Inductor part of sensor is composed of coil form structure to incorporate with capacitor to form an LC resonant circuit. LC circuit has unique resonance frequency depending on the designed inductance and capacitance of the sensor, can be expressed as (2).

Inductors exhibit wireless measurable magnetic inductive coupling feature. An external antenna coil can be coupled with sensor by matching its resonance frequency; at this condition magnetic flux is generated inside the inductor. The generated electric current is passed to the capacitor which is electrically connected to the inductor and contains an electric charge. Inductance of device having n number of turns of coil is shown in equation (3); greater the inductance is, higher the amount of current that can flow inside the circuit [15].

$$f_s = \frac{1}{\sqrt{L_s C_s}} \qquad (2)$$

$$L_s = K_1 \pi_0 \frac{n^2 d_{avg}}{1 + K_2 \rho} \qquad (3)$$

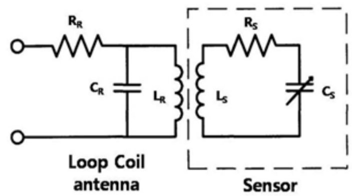


Figure 2. Block diagram of L-C sensor system. Resonance frequencies shifts following to the capacitance change of the sensor.

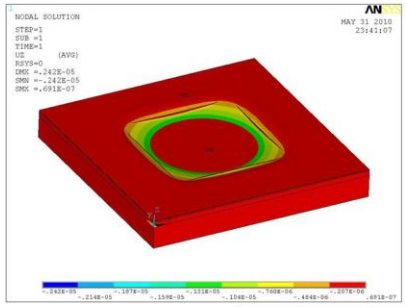
Inductor-capacitor LC sensor coupled to external circuit through electromagnetic force can be used to measure pressure changes by measuring the change in capacitance using network analyzer. Figure 2 shows block diagram of equivalent circuit of external antenna and sensor, where L_s is inductance, C_s is the capacitance, R_s indicates the total resistance of sensor. The characteristics of designed sensor are shown in Table 1.

Table 1. L-C pressure sensor data

Element	Value
Capacitor	2.26pF
Inductor	100nH
Cu-electroplating	Thickness 30mm
Resistivity	5 ohm.mm
Body size	4×4mm^2

2.2. FEM Analysis of Pressure Sensor

An FEM model of the LC pressure sensor is developed using ANSYS software to determine the maximum deformation of polymer based diaphragm. The response of capacitive part of device to the surrounding pressure of vessel is analyzed with respect to displacement of diaphragm, shown in Figure 3. Overall size of the sensor is 4 × 4 mm^2, size of the capacitor electrode is 1 × 1 mm^2 and thickness of the diaphragm is 2 µm. The separation between the capacitor plates is kept at 17.5 µm. Diaphragm behavior is observed by applying pressure in range of 0 ~ 16 kPa, which is average blood pressure inside a coronary artery of normal human. As a result, when pressure of 16 kPa is applied to the diaphragm a deflection of up to 2.4 µm is observed. Thin film manufactured in this study with center boss diaphragm structure, has advantage of excellent linearity at lower pressures over conventional diaphragm having rectangular thin film structure [16].



Material	Density	Young's modulus	Poisson's ratio
SU-8	1190Kg/m³	4.02GPa	0.22
PDMS	0.97Kg/m³	360~870KPa	0.5
Au	19300Kg/m³	18GPa	0.44

Figure 3. FEM Analysis results of the Diaphragm deformation. Maximum deformation is 2.4mm at a pressure of 16kPa.

Sensor characterization is done by measuring the change in its resonance frequency with respect to applied pressure on the sensor. The change in resonance frequency of sensor based on deflection of thin film is plotted on P-spice program. At normal condition, resonance frequency of the device is found to be 440 MHz.When 16 kPa pressure is applied on the diaphragm, this caused electrode spacing to decrease by one-half with 114 MHz reduction in the resonant frequency, results are plotted in Figure 4.

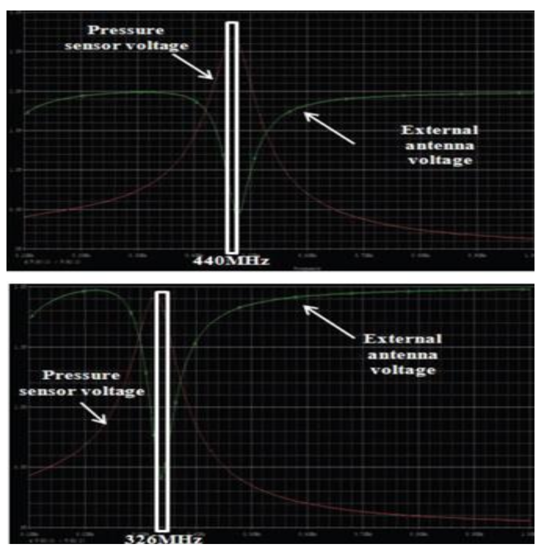


Figure 4. Analysis results of P-spice (Top) normal state of sensor without any applied pressure, (Bottom) Frequency shift about 114MHz due to 16kPa applied pressure

2.3. Device Fabrication

As described earlier, the device is designed to measure the pressure of blood inside coronary artery. The device should be flexible, safe, and biocompatible for this reason PDMS and SU-8, are used, which possess all three properties. The designed sensor is separated into two parts, bottom part consists of inductor, and top part consists of capacitor strips. Figure 5 shows the fabrication procedure, for inductor part firstly aluminum is deposited on silicon wafer to be used as a sacrificial layer. A 3 µm SU-8 layer is deposited above the sacrificial layer using AZ5214E followed by lift off process; afterwards layers of Cr / Au with 10 / 50 nm thickness are deposited by using E-beam evaporator. At last copper metal layer of 30 µm thickness is produced by using electroplating method. The inductor part is completed by SU-8 structure fabrication; total thickness of final product is about 40 µm.

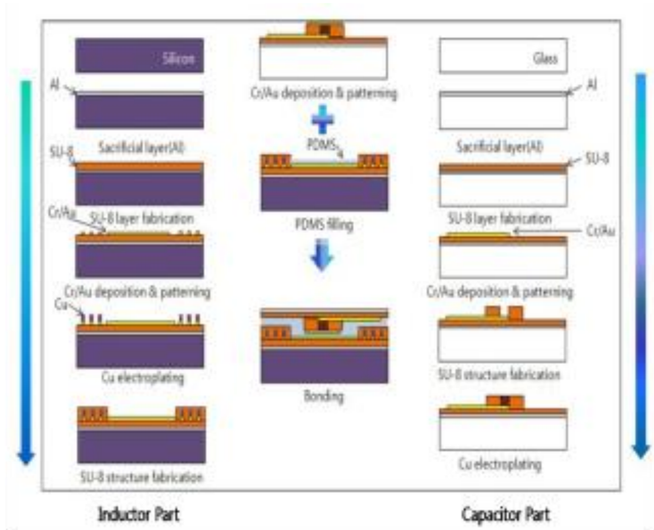


Figure 5. Process flow schematics of L-C pressure sensor. Left side shows inductor part and right side shows capacitor part. After fabrication, both parts are bonded using by PDMS layer for separation

In the next phase, capacitor part is produced by using a transparent glass substrate to align bonding which will take place later. First, an aluminum sacrificial layer is deposited on the glass, and then about 1 µm thick SU-8 layer is deposited above aluminum sacrifical layer using AZ5214E equipment by lift off process to form a pattern. To produce parallel plate of capacitor, layers of Cr / Au with 10 / 50 nm thickness are deposited using the E-beam evaporator. At the end, SU-8 structure of 30 µm thickness is formed followed by copper deposition by electroplating. Fabricated SU-8 structure is manufactured to have high linearity, as described in Section 2.2, so that center boss diaphragm structure should maintain the capacitor electrodes separtion to a distance of 17.5 µm. Upper panels and the lower panel of the fabricated sensor is completed by making the aligned bonding having a PDMS filling in between. Figure 5 shows the process flow schematic of LC pressure sensor and fabricated LC pressure sensor is shown in Figure 6

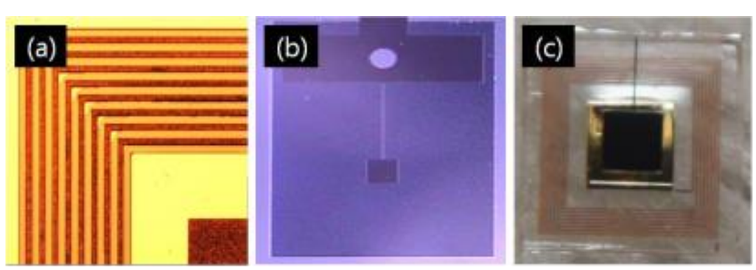


Figure 6. (a) Inductor area (b) capacitance area and (c) finished pressure sensor

3. LC Resonant Pressure Sensor Evaluation

Configuration of system for analyzing radio characteristics of the fabricated device is shown in Figure 7. A RFID reader loop antenna (AWG30, diameter: 0.25 mm, and 8 turns) is connected to the network analyzer (Agilent, 8510C, 400 MHz ~ 26.5 GHz). LC pressure sensor is placed inside the plastic chamber connected to the syringe pump through flow tube; pressure inside the chamber can be controlled through syringe pump. A magnetic field is formed along the external antenna as the test signal is fed to it through network analyzer. This cause current flow through the LC sensor due electromagnetic induction and frequency of the device can be detected.

The external antenna should be placed at a distance of about 10-20 mm from the sensor, to develop a magnetic connection between the external coil and the sensor. The electromagnetic field has a particular frequency band, when tuned to match the resonance frequency of sensor, will induce an electrical resonance current inside the sensor. Thus the generated electric current will be used as a driving power source of the sensor.

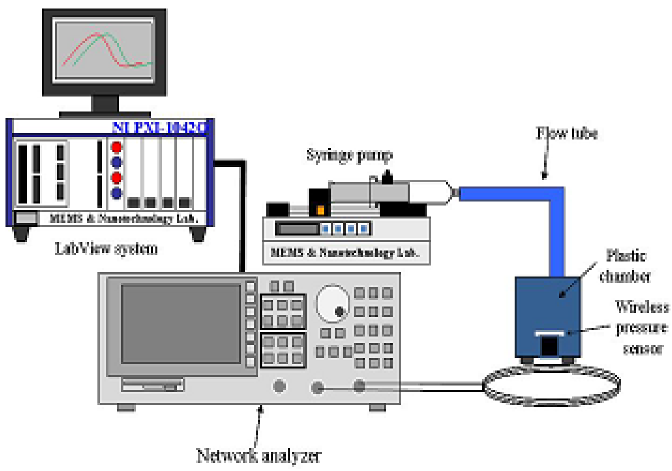


Figure 7. Sensor test system schematics

A syringe pump is used to alter resonance frequency of the sensor by controlling pressure inside the chamber. Change in the value of capacitance due to increase in pressure inside chamber changes the resonance frequency of the sensor; this change can be monitored on network analyzer using external antenna. To calculate the change in capacitance than the measured capacitance of 1.3 pF, pressure is altered inside the pressure chamber. During the practical demonstration wire connection is developed on the sensor to measure its resonance frequency using network analyzer. Due to consumption of conductive epoxy, thickness of PDMS or separation between the capacitor plates is slightly altered. As a result of this, the capacitance of sensor during practical demonstration is slightly varied from its calculated 1.3 pF value, which caused a change in resonance frequency of the sensor from 440 MHz to 510 MHz.

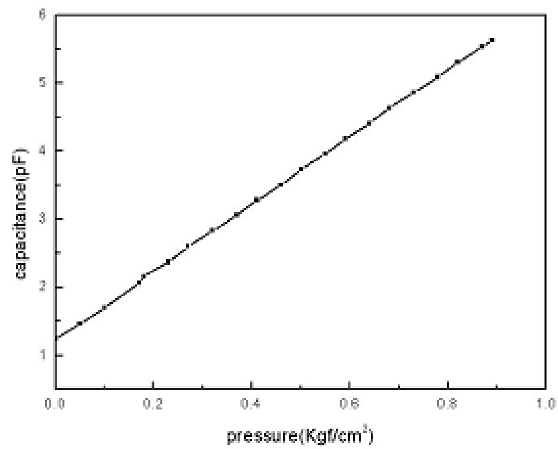
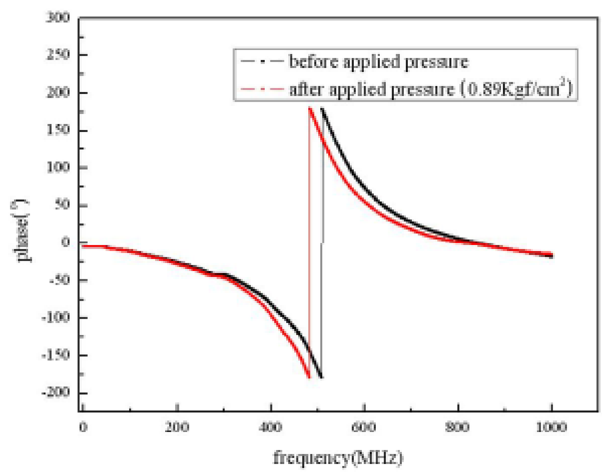


Figure 8. Test result of change in capacitance as a function of pressure

Results of change in resonance frequency obtained as a result of changing applied pressure are measured by using wireless equipment and plotted in Figure 8 and Figure 9. By varying applied pressure to the sensor between 0-16 KPa, capacitance measured to be between 1.3-5.6 pF shown in Figure 8. In absence of any applied pressure the resonance frequency of sensor is found to be 510 MHz and the resonance frequency is reduced to about 100 MHz when a pressure of 16 kPa is applied to the sensor shown in Figure 9.



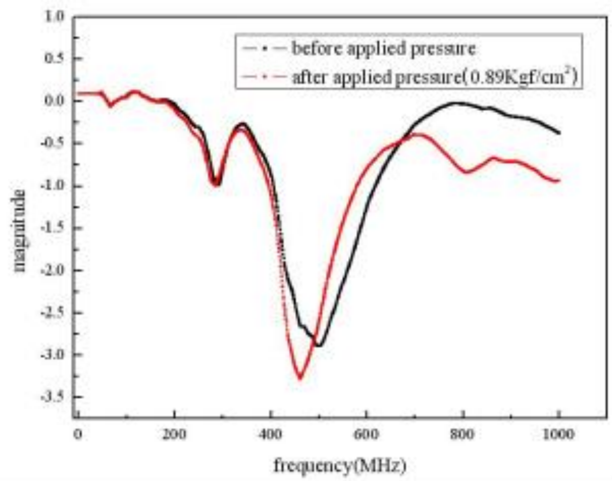


Figure 9. Test results of the L-C pressure sensor (Black line-atmospheric pressure, Red line- applied pressure). Phase change due to pressure rise (top) and magnitude change due to pressure rise (bottom)

5. Conclusions

A polymer-based RF micro pressure sensor to monitor blood pressure inside coronary artery is designed, manufactured and examined. The structure of sensor composed of polymer was designed by using the micromachining technology. Wireless sensor manufactured by utilizing simple manufacturing process with low manufacturing cost and biocompatible material SU-8 polymer does not require any separate packaging for bio-implants. As compared with silicon-based sensors, the present design has an advantage of being practically applicable. As the pressure change can be monitored from outside the body by the application of RFID technology. The experiment was performed to measure the change in resonance frequency due to change in capacitance of the sensor which is caused by change in applied pressure. The capacitance of fabricated device was measured to change from 1.3 pF to 5.6 pF when the pressure is altered from 0-16 KPa, and it is confirmed that the resonance frequency is reduced by an amount of approximately 120 MHz from initial value of 510 MHz. Although an exact match of the calculated values of sensor with the analysis result was not achieved perfectly. However, the change in frequency characteristics due to pressure difference indicates that this pressure sensor can be used in the applications for biomedical purpose. Additionally, restenosis inside stented arteries can also be detected by utilizing this sensor.

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