Design a Wireless Pressure Sensor With an Ellipse and a Circular Shape to Monitor the Pressure Within the Coronary Artery

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ABSTRACT This paper put forward the design and modeling pertaining to a wireless pressure sensor that can monitor blood pressure within the coronary artery with the help of a capacitive pressure sensor. To adjust the resonance frequency (26.78–27.09 MHz) with regards to this sensor’s applied pressure (0–30 KPa), sensor’s dimension as well as human blood pressure (0–220 mmHg), a variable capacitor in the pressure sensor has been put forward. The shape of the capacitive pressure sensor was designed to look like an Ellipse and a circle, which were assessed for their performance with regards to capacitive sensitivity and diaphragm deflection. Also, the diaphragm thickness pertaining to the capacitive pressure sensor with Ellipse and circular shapes was altered (0.1 µm to 0.5 µm) while the cross-section area was kept constant at 1 mm². As per the results, an invasive sensor with circular and Ellipse capacitive pressure shape that had smaller size (1) mm² displayed high sensitivity. The sensitivity readings for Ellipse and circular shape capacitive pressure sensor were 7.73 ff mmHg⁻¹ and 9.94 ff mmHg⁻¹, respectively. Diaphragm deflection was simulated based on the COMSOL Multiphysics software, while MATLAB was employed for simulation with regards to changes in resonance frequency, capacitance, and capacitive sensitivity.

INDEX TERMS Capacitive pressure sensor, in-stent restenosis, wireless pressure sensor, stent.

I. INTRODUCTION Normal atherosclerotic cardiovascular disease is characterized with the formation of plaque as well as contraction in the inner walls of blood arteries, which could result in stroke, myocardial infarction, and transient cerebral ischemia episodes [1]. To deal with these difficulties, we employed a stent or a balloon catheter to enlarge the blood artery via physical means. By inflating a balloon, the inserted metal stents would widen the diameter of blood artery. On the other hand, the mechanical expansion pertaining to the metal stent could lead to many harmful impacts for the patient. In most of these cases, formation of neointima forms occurs due to thrombus in the peripheral section of the blood vessel, which could lead to a new constriction of the inner diameter pertaining to the blood vessel [2], [3]. In addition, diabetic patients do not feel chest pain when the narrowing in the arteries returns, and this may lead to a sudden and dangerous deterioration of the health condition. X-ray-based inspection devices can be employed to continuously monitor the metal stents but is costly. Thus, it is important to identify an alternative approach for monitoring these metal stents as well as blood pressure in real time [4]. MEMS pressure sensors are better than traditional pressure monitoring systems due to their...
small size, better efficiency, and cost-efficient production [5]. There are various techniques employed to distinguish pressure sensors: piezoelectric sensing, capacitive sensing, and piezoresistive sensing, among others. The capacitive pressure sensors offer several benefits versus piezoresistive and piezoelectric, including high sensitivity, low power consumption, small temperature drift and high resistance to-wards packaging stress [6]. The capacitive pressure sensors have different designs, including circular, square and Ellipse shapes. In the square shape, the sharp angles result in greater chances of fibrosis inside the artery. Thus, in most cases, elliptical and circular shapes are favoured [2]. Various MEMS-based pressure sensors have been developed to monitor intracranial pressure [7], [8], [9]. An experimental prototype system was employed to show the put forward design that employed a steel stent as an RF antenna to offer space saving as well as reduce the number of antenna components used in the wireless pressure sensor and unified it with the sensor [10]. Another researcher designed a 20-mm long antenna from a stainless-steel stent of 3.5 mm-diameter and planar foil with 50 µm-thick sheets. The coating material was selected to provide a thin, homogeneous, and conformal covering that can be deemed as biocompatible, chemically inert, and non-conductive to reach the sensitivity requirement of 273 ppm ton−1 with spacing <1 cm [4]. Biodegradable polymers poly-caprolactone, poly-L-lactide and conductor material were employed as structural and dielectric materials. They were employed to design the wireless pressure sensors with a 1 cm² area, with the help of an inductor and MEMS. This was done by maintaining 50 MHz as the frequency of opera-tion [11]. The MEMS pressure sensor was designed by employing a photosensitive SU-8 polymer and then inductance integration when the area size would be 3.13 × 3.16 mm × 150 µm. The stent was designed under an operating frequency of 200 MHz by employing 3D-printer biodegradable polymer. The pressure sensor was in the range of 0–230 mmHg, 0.043 MHz mmHg⁻¹ sensitivity and 10 mm reader distance. A micro-pressure sensor was build based on polymer and integrated with a 3D-printed polymer stent. The SU-8 bonding process helped to get uniform sensitivity pertaining to the pressure sensor value of 160 KHz mmHg⁻¹ as well as an area size of 4 × 4 mm². This also helped to measure the pressure in the range of 0–220 mmHg [12]. Later, the designed sensor was connected to a biocompatible polycaprolactone stent possessing 4 × 4 × 0.15 mm³ sensor area. The sensitivity was found to be at 160 KHz mmHg⁻¹ while the operation frequency at 179 MHz [8]. The researcher continued with the design of a 3D-printed biocompatible as well as biodegradable polymer stent, which were combined with pressure sensor based on poly(D-lactide) (PDLA) along with a pressure range of 0-220 mmH and a sensor surface area of 4 mm × 4 mm × 0.15 mm. At 148 MHz, the pressure sensor’s sensitivity was found to be 60 kHz mmHg⁻¹ [13]. Medical-grade stainless steel was designed by employing a smart stent with in-built MEMS pressure sensor. The helix-like pattern design of the stent allowed it to function as an inductive antenna and, at the same time, made it compliant with the generally employed balloon catheter stenting technique. Electroplating of the stent was done with a gold layer to minimize its series resistance. The length and diameter of the stent were 20 mm and 5 mm, respectively. In free space, its sensitivity was found to be 302–335 ppm mmHg⁻¹ accommodating a pressure of up to 250 mmHg, while the in vitro sensitivity was found to be 146 ppm mmHg⁻¹ when frequency was maintained in the range of 30–80 MHz and the size of the sensor was (1.5 × 1.5 mm² × 200 µm) [14]. The design of the smart stent included an integrated capacitive MEMS pressure sensor based on stainless steel, which was electroplated with a layer of gold to minimize stent’s series resistance. Parylene C layer was employed to passivze the stent and to make its sur-face biocompatible and electrically isolated. At frequency of 10 MHz in free space and in vivo, the chip sensor size was (1.5 × 1.5 mm² × 200 µm) and stent length was 30 mm [15], [16]. Despite the efforts made by the researcher, the pressure sensor is still faced with issues such as large size and low sensitivity.

This paper puts forward an approach of employing COMSOL Multiphysics to develop a biocompatible pressure sensor as well as smart stent. The pressure sensor employed on the smart stent platform can track pressure of the blood vessel in real time. The unique design of the wireless pressure sensor helps decrease the number of routine medical tests. The preliminary investigations employing simulation rate showed that the proposed design can also be integrated and used for health monitoring applications.

II. MATERIALS AND METHODS

The wireless pressure sensor has been designed based on the inductor-capacitor resonant circuit technique. The schematic design pertaining to the put forward wire-less pressure sensor system is shown in Figure 1. The wireless pressure sensor system is made of two parts, internal and external components. The external component is placed outside the body, but it touches the chest; while the internal component is placed within the coronary artery inside the human body. The internal part has length of the inductor coils (stent) is 30 mm and has been designed to be helical shape when expanded by a balloon placed within the artery along with a diameter of 5 mm, and a zigzag helical shape with a diameter of 2 mm prior to expansion and when inserted via a catheter. The basic principle behind this system is to reach a resonance frequency as well as sufficient transfer power from the outer circuit towards the inner circuit that is implanted within the coronary artery to enable it to work without a battery. It can measure the pressure by altering the resonance frequency because of change in the capacitor value, which happens due to the deflection of diaphragm sensor caused by the applied pressure. This, in turn, results in altering of the distance between the two plates, thereby leading to variation in the resonance frequency as well as capacitance value.

For the external component, the efficient design of the external circuit allowed power transfer at 20 mm, thereby
giving power 82% transfer efficiency at a resonance frequency of 27 MHz [17]. The external circuit can also allow measuring the resonant frequency pertaining to the sensor, which denotes the difference in capacitance value. Based on an LC circuit, Equation (1) allows calculating the resonance frequency pertaining to an ideal sensor, wherein $L_r$ and $C_r$ represent the sensor’s inductance and capacitance, respectively, while $\omega_0$ signifies the resonance frequency.

$$\omega_0 = 2\pi f = \frac{1}{\sqrt{L_r C_r}}$$

(1)

Since the coronary artery’s diameter does not exceed 5 mm, it includes certain restrictions pertaining to the use of spiral coil that are rectangular or circular in shape, which impacts the flow of the blood within the artery. Thus, the helical coil is regarded to be the optimum solution that can be placed within the coronary artery. Mathematically, equation (2) can be employed to compute the helical coil and self-inductance [18]

$$L_{helical \ coil} = \frac{\mu \pi r^2 N_{stent}^2 T}{l_{stent}}$$

(2)

where $\mu$ is the free-space permeability, $r$=$d_{stent}/2$, and $T$ is Nagaoka’s coefficient, estimated as

$$T = Y_z \left( \ln \left( 1 + \frac{1}{Y_z} \right) \right)$$

(3)

$$U = \frac{1}{X_0 + X_1 \left( \frac{d_{sent \ wire}}{d_{stent}} \right) + X_2 \left( \frac{d_{sent \ wire}}{d_{stent}} \right)^2 + \frac{a_1}{a_2^2 \pi \frac{d_{sent \ wire}}{d_{stent}}}}$$

(4)

$$Y_z = \frac{d_{sent \ wire}}{\pi r}$$

(5)

where $d_{sent \ wire}$ is the diameter of the wire used in the coil, $d_{stent}$ is diameter of helical coil, $X_0= 2.30038$, $X_1= 3.437$, $X_2= 1.76356, a_1 = -0.47$, $a_2= 0.755, v= 1.44$, $l_{stent}$ is length of wire used in helical coil.

Stent inserts were carried out via PCI operation with a size of 2 mm, which is expanded by the balloon to reach 5 mm. Gold is used as the material since it is biocompatible and possesses a good electric conductivity to allow wireless transfer of power. Based on Equations (6-9), calculation of the stent parameter is done prior to expansion and post expansion as demonstrated in Figure 2.

$$l_{one \ revolution} = \sqrt{H^2 + C^2}$$

(6)

where $H=2\pi r$ and present the rise of helical in one revolution (pitch), where $r$ is the radius of helical coil

$$l_{one \ revolution} = N_{strut} \left[ 2l_{strut} + 2l_{arc} \right]$$

(7)

$$l_{strut} = \frac{\pi}{2} - 2\left[ \frac{s}{2} + th \right]$$

(8)

$$l_{arc} = \left[ \frac{s}{2} + th \right] \pi$$

(9)

where $N_{strut}$ is number of struts in one revolution $l_{strut}$ Presents the length of strut $l_{arc}$ presents the length of arc connecting between the struts $th$ presents the thickness of strut $s$ presents the space between the struts.
the basic capacitance value is generally greater. Monitoring of the separation gap parameter is done cautiously to make sure that a suitable capacitance value gets read as output. Figure 3 shows the various layers of capacitive pressure sensor pertaining to two different geometries (Ellipse and circular), i.e., dielectric, substrate, diaphragm, and a bottom electrode.

A dielectric layer (Si3N4) that has a thickness of 0.8828 µm in circular shape and 0.8808 µm in an elliptical shape separates out the diaphragm and electrode at the bottom. When a certain pressure is applied, the diaphragm tends to lower and bend the plate separation gap, thereby leading to an increase in the capacitance value between the plates. There is an incremental increase in applied pressure by 5 kPa. By keeping constant the diaphragm thickness for a selected design model, there is a rise in the capacitance value with decrease in separation gap due to the applied pressure. Due to this, there is a linear variation in capacitance with pressure. When pressure is not applied on the diaphragm, there exists a specific amount of capacitance, also called as base capacitance, between the plates, which can be calculated by employing equation (10).

\[ C = \frac{\varepsilon_0 \varepsilon_r A}{d} \]  

where \( \varepsilon_0 \) is the air permittivity, \( \varepsilon_r \) the dielectric permittivity in the gap is the distance between the diaphragm and bottom electrode, and A is the sensor surface cross-section area. A circular = \( \pi r^2 \), where r is the radius of the circular diaphragm, A ellipse = \( \pi ab \), where a is the elliptical diaphragm’s semi-major axis, and b is its semi-minor axis.

The diaphragm dimensions are determined based on the selected shape. With regards to an applied uniform pressure of value P, the maximum deflection equation for computing a circular diaphragm’s displacement is given by equation (11), and equation (12) provides the maximum deflection of Ellipse diaphragm.

\[ \Delta_d = \frac{P}{64D} r^4 \]  

\[ \Delta_d = \frac{P}{8D} \left( \frac{a^4 b^4}{3(a^4 + b^4) + 2a^2 b^2} \right) \]  

where P denotes the applied pressure and D represents the flexural stiffness, as defined by:

\[ D = \frac{E d^3}{12(1 - v^2)} \]

where v is Poisson’s Ratio of Diaphragm, and E is Young’s Modulus.

Calculation of the capacitive pressure sensor’s sensitivity is carried out mechanically. Mechanical sensitivity can be defined as the difference in capacitance values to the applied pressure values. If the device possesses higher sensitivity, it demonstrates more accurate reading of the output. For any device, it is always favourable to have a high sensitivity value. Calculation of the sensitivity can be done theoretically based on the following equation (14) [19]

\[ S = \frac{\Delta_d}{P} = \frac{C_{max} - C_{min}}{P_{max} - P_{min}} \]  

For a sensor with ellipse shape, the substrate material is polyamide that possesses dimensions of 0.750 × 0.424 × 200 µm³. The dimensions of the diaphragm would be 0.750 × 0.424 × 0.1 µm³, and gold is selected as the material due to its biocompatibility and highest capacitance value. The bottom electrode includes identical material that is used in the diaphragm, with dimensions of 0.550 × 0.224 × 0.1 µm³. The same material is employed by the circular capacitive pressure sensor as those used in the ellipse capacitive pressure sensor, but it differs in size. The substrate dimensions are 0.564 × 0.564 × 200 µm³, the bottom electrode with 0.364 × 0.364 × 0.1 µm³, and the diaphragm with 0.564 × 0.564 × 0.1 µm³, as shown in Table 2.

**TABLE 2. Shows the dimensions and materials utilized in the two suggested designs.**

| Parameter | Ellipse | Circular |
|-----------|---------|----------|
| Diaphragm | \( \mu m \times 424\mu m \times 0.1\mu m \) | \( \mu m \times 564\mu m \times 0.1\mu m \) | Gold |
| Dielectric | \( 550\mu m \times 224\mu m \times 880\mu m \) | \( 364\mu m \times 364\mu m \times 882\mu m \) | Silicon Nitride |
| Bottom electrode | \( 550\mu m \times 224\mu m \times 0.1\mu m \) | \( 364\mu m \times 364\mu m \times 0.1\mu m \) | Gold |
| Substrate | \( 750\mu m \times 424\mu m \times 200\mu m \) | \( \mu m \times 564\mu m \times 200\mu m \) | Polyamide |

III. RESULT

Figure 4 indicates the circular and elliptical diaphragms when deflected with various pressures at the maximum. For these
structures, the uniqueness of the diaphragm and bottom electrode arrangement is evident. Designed for operating in a pressure range of 0-30 KPa, the sensors are suitable to measure blood pressure within the coronary artery. The COMSOL Multiphysics software was utilized to demonstrate the results of the simulation for capacitance change and electromechanical coupling for capacitive sensitivity and diaphragm deflection. The pressure sensor with an electromechanics operating system was designed using a 3D model builder. A stationary study analysis is conducted using the Multiphysics software. The diaphragm of the electromechanics interface is flexible, while the bottom electrode should be fixed. During the conduct of the electromechanical analysis, voltage is received by the top electrode since it deforms under uniform pressure and is flexible. On the other hand, since the ground is fixed, it is supplied to the bottom electrode. Boundary load pressure is consistently applied to the diaphragm electrode.

Table 3 illustrates the results of the simulated diaphragm deflection. Five feature designs are shown with the following diaphragm thicknesses: 0.1 µm, 0.2 µm, 0.3 µm, 0.4 µm, and 0.5 µm for both Ellipse and circular geometries.

**TABLE 3.** Shows the values for diaphragm deflection for different thicknesses and shapes of diaphragms.

| Pressure (KPa) | 5  | 10 | 15 | 20 | 25 | 30 |
|---------------|----|----|----|----|----|----|
| d=0.1 µm      | 0.003 | 0.007 | 0.010 | 0.01 | 0.017 | 0.02 |
| d=0.2 µm      | 0.003 | 0.007 | 0.010 | 0.01 | 0.016 | 0.01 |
| d=0.3 µm      | 0.003 | 0.006 | 0.009 | 0.01 | 0.016 | 0.01 |
| d=0.4 µm      | 0.003 | 0.006 | 0.009 | 0.01 | 0.015 | 0.01 |
| d=0.5 µm      | 0.003 | 0.006 | 0.009 | 0.01 | 0.015 | 0.01 |
| µm            | 7 | 8 | 9 | 30 | 1 | 92 |
| d=0.1 µm      | 0.003 | 0.005 | 0.008 | 0.01 | 0.013 | 0.01 |
| d=0.2 µm      | 0.002 | 0.005 | 0.007 | 0.01 | 0.012 | 0.01 |
| d=0.3 µm      | 0.002 | 0.005 | 0.007 | 0.00 | 0.012 | 0.01 |
| d=0.4 µm      | 0.002 | 0.005 | 0.007 | 0.00 | 0.012 | 0.01 |
| d=0.5 µm      | 0.002 | 0.004 | 0.007 | 0.00 | 0.011 | 0.01 |
| µm            | 72 | 99 | 26 | 95 | 79 | 40 |

**FIGURE 5.** (A) Shows the relationship between diaphragm thickness and pressure for circular diaphragms. (B) Shows the relationship between diaphragm thickness and pressure for elliptical diaphragms.

FIGURE 4. Shows the maximum deformation (A) in a circle and (B) in an ellipse.

Compared to elliptical diaphragms, deflection is more significant in circular diaphragms. The most noteworthy deflection can be seen in the design with the thinnest diaphragm. Figure 5 demonstrates how thickness increases as pressure sensitivity decreases. For a particular pressure range, the optimum diaphragm deflection can be reached by striking a balance between diaphragm thickness, plate separation gap, and maximum diaphragm deflection.

If no pressure is applied, the value of the base capacitance is 97.41pf. While this value does not factor in the thickness of the diaphragm, it is dependent on the size of the diaphragm’s area and the separation gap between the plates.
TABLE 4. Shows capacitance values for circular and elliptical geometries with various diaphragm thicknesses.

| Pressure (KPa) | 5   | 10  | 15  | 20  | 25  | 30  |
|---------------|-----|-----|-----|-----|-----|-----|
|               | d=0.1 µm | 97.77 | 98.14 | 98.51 | 98.89 | 99.26 | 99.65 |
|               | m    | 6   | 5   | 7   | 2   | 9   | 0   |
|               | d=0.2 µm | 97.76 | 98.11 | 98.47 | 98.83 | 99.20 | 99.56 |
|               | m    | 1   | 7   | 6   | 7   | 2   | 9   |
|               | d=0.3 µm | 97.74 | 98.09 | 98.44 | 98.79 | 99.14 | 99.50 |
|               | m    | 8   | 4   | 2   | 2   | 5   | 1   |
|               | d=0.4 µm | 97.73 | 98.07 | 98.41 | 98.75 | 99.09 | 99.44 |
|               | m    | 7   | 4   | 3   | 4   | 7   | 3   |
|               | d=0.5 µm | 97.72 | 98.05 | 98.38 | 98.72 | 99.05 | 99.39 |
|               | m    | 8   | 7   | 7   | 0   | 6   | 3   |

TABLE 5. Pressure sensor sensitivity for various diaphragm shapes.

| Pressure (KPa) | 30 |
|---------------|----|
|               |    |
| Circular      |    |
| d=0.1 µm      | 9.94 |
| d=0.2 µm      | 9.46 |
| d=0.3 µm      | 9.2  |
| d=0.4 µm      | 9.02 |
| d=0.5 µm      | 8.8  |
| d=0.1 µm      | 7.73 |
| d=0.2 µm      | 7.46 |
| d=0.3 µm      | 7.33 |
| d=0.4 µm      | 7.06 |
| d=0.5 µm      | 6.93 |

Setting the pressure to 30 KPa makes the circular model design’s diaphragm thickness 0.1 µm; it also pushes the capacitance value to the maximum at 99.65 PF. Conversely, when the circular design model is set with a diaphragm thickness of 0.5 µm, it generates a 99.39 PF maximum capacitance value. Moreover, testing in the Ellipse diaphragm form, defining the pressure at 30 KPa results in the model design having a 0.1 µm diaphragm thickness and reaching a 99.16 PF maximum capacitance value. Table 4 shows the 98.99PF maximum capacitance produced for the design model, with a 0.5 µm diaphragm thickness.

Figure 6 illustrates how the maximum capacitance between the bottom electrode and the diaphragm is determined by the distance of the plate separation. This is consistent no matter how thick the diaphragm is.

The variance in the ratings for the different design types’ capacitive sensitivity shows that sensitivity decreases as the diaphragm thickness grows. For a circular diaphragm 0.1 µm thick, the capacitive sensitivity is determined to be 9.94 fl/mmHg. Compared to other diaphragm thicknesses, this is the greatest value. As for an Ellipse diaphragm with a similar thickness (0.1 µm), its capacitive sensitivity is 7.73 fl/mmHg; this diaphragm thickness is also the greatest among the said values as exhibited in Table 5.

The dissimilarities in capacitance values generated between the plates for Ellipse and circular diaphragms with different thicknesses at varying pressure levels are being demonstrated. The design model that shows the best potential output is the circular diaphragm shape: since it demonstrates more sensitivity to pressure, it is the most recommended method for coronary artery blood pressure monitoring.
TABLE 6. Presents the results of validating the proposed study with the work of other researchers.

| References | Coil Shape | Inductance | Material of Sensor | Sensor shape | Dimension of Sensor | Frequency | Sensitivity | Pressure Range |
|------------|------------|------------|-------------------|--------------|---------------------|-----------|------------|--------------|
| [11]       | Stent Length 4 mm | 0.02 µH | Au, glass, Silicon | Square | (1.2 mm×1.4 mm×0.5mm) | 201 MHz | 5.0 fF/mmHg | - |
| [13]       | Stent Length 35 mm | - | - | Square | 1 mm×2×300 µm | 2.4 GHz | - | - |
| [14]       | Stent Length 30 mm | - | - | Square | (5 x 5) mm2 | 2.4 GHz | - | - |
| [20]       | Stent       | 3.37 nH  | Gold-tin          | Square | 3 mm×6 mm×300 µm | 2.4 GHz | 6.64 fF/mmHg | 0-50 mmHg |
| [15]       | Stent Length 20 mm | 0.530 µH | Stainless steel, parylene C | Square | (1.5 mm×1.5 mm×200 µm) | 50 MHz | 146 fF/mmHg | 0-250 mmHg |
| [17]       | Stent Length 30 mm | 268 nH | Au, Stainless steel 316 L, titanium, SiO2 | Square | (1.5 mm×1.5 mm×200 µm) | 10 MHz | - | - |

Proposed design Stent length 30mm 0.350 µH Gold, Polyamide, Silicon Nitride Ellipse 750 µm×424µm×20 µm 0.1 MHz (26.78 – 27.09) 7.73 fF/mmHg 0-240 mmHg

Proposed design Stent length 30mm 0.350 µH Gold, Polyamide, Silicon Nitride Circular 564 µm×564µm×200 µm 0.1 MHz (26.78 – 27.09) 9.94 fF/mmHg 0-240 mmHg

FIGURE 7. Depicts the relationship between frequency and pressure.

In addition, the coronary artery pressure is measured utilizing a wireless pressure sensor that has different frequencies which are different from the resonance frequency of the heart that have a low frequency of 0.1Hz (6 breaths per minute); this method alters the capacitance value upon application of pressure. In Figure 7, the shifts in frequencies for the wireless pressure sensor are shown with the application of varying pressures that ranges from 0 to 240 mmHg. As for the stent, its inductance value before expansion is 0.350 µH, while after expansion, the value decreased to 0.320 µH.

In view of the difficulty of conducting the experimental research and implanting sensors inside the human body, and there are no animal arteries that are identical to the human arteries so we evaluated our work by comparing it with other related works. Table 6 compares the previous research and the current design proposal for a capacitive pressure sensor using Ellipse and circular structures. The table also illustrates the size features of both designs: lowest in size, having no sharp corners or edges to be safer in increasing sensitivity and minimizing restenosis.

IV. CONCLUSION
This research provides a description of how a capacitive pressure sensor (MEMS) in circular and Ellipse geometries is designed, analyzed, and modelled. This MEMS, which may be utilized for measuring blood pressure, operates efficiently within the coronary artery in the 0-30 KPa range. The unique stent design can also result in good mutual coupling and staying within fracture limits in times of balloon expansions. Capacitive pressure sensors are employed since they are more sensitive to pressure and dependable in measuring low-pressure levels. COMSOL Multiphysics is employed during the design and simulation for the sensor and its electromagnetic analysis. With its 0.1 m diaphragm thickness, the circular capacitive type is identified to have a 9.94 fF/mmHg maximum sensitivity and a 97.776 PF capacitance value. Consequently, the proposed design is found to be more suitable for measuring coronary artery blood pressure.

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