IJISET - International Journal of Innovative Science, Engineering & Technology, Vol. 8 Issue 8, August 2021
ISSN (Online) 2348 – 7968 | Impact Factor (2020) – 6.72
www.ijiset.com

MEMS PZR Sensor for Healthcare Applications

Sajith K[1], T Shanmuganantham[2], D Sindhahaiselvi[3]

[1]Assistant Professor, Department of ECE, GEC Wayanad, Wayanad, Kerala- 670644, India
[1]sajithrajan999@gmail.com

[2]Associate Professor, Dept. of Electro. Engg. Pondicherry University, Puducherry- 605014, India.
[2]shanmugananthamster@gmail.com

[3]Assistant Professor, Dept. of ElE., Pondicherry Engineering College, Puducherry- 605014, India.
[3]sindhanaiselvi@pec.edu

Abstract

A novel square diaphragm-based piezoresistive MEMS silicon sensor has been designed for intraocular pressure monitoring applications. The four piezoresistive elements are strategically placed on the square diaphragm for improving the sensitivity of the sensor. The dimension of MEMS sensors is the range of micrometer scale, 125x125x11μm^3. The designed MEMS Piezoresistive sensor has been simulated by the Finite Element Method (FEM). The analytical and simulation results show the designed sensor has Full-Scale Output (FSO) is 28.5mV/V for its linear operating range of pressure from 1KPa to 8KPa. This work mainly focused on the surface displacement, surface stress tensor and strain tensor of the diaphragm, and sensitivity. The simulated results are well suitable for the glaucoma intraocular pressure monitoring applications

Keywords: *MEMS, PZR sensors, SMT fabrication, Sensitivity, Ocular implant.*

1. Introduction

Recently, Micro-electromechanical systems (MEMS) technologies have been used to develop micro-sized sensors and actuators. The MEMS devices have smaller dimensions, low volume, higher reliability, easier mass-production, and low cost. K.V Menna et al [1] disused MEMS-based piezoresistive biomedical sensors with catheters has used to detect the blood pressure and temperature of the human body. The main advantages of this device are the highest sensitivity and linearity. The integrated catheter with PZR sensor classification, fabrication process, material selection and shape, and packing technique are well described [1]. PZR is a material asset where the bulk resistivity has prejudiced by the mechanical stresses applied to the material [2]. The monocrystalline silicon has a high elasticity value and excellent mechanical properties. It is particularly suited for the conversion of mechanical deformation to an electrical signal. The other advantages of use silicon for PZR sensors include reduced mass and dimensions, batch fabrication, and easy interfacing with electronic circuits and integrated circuits [3],[4].

The PZR sensors are classified as diaphragm-based and beam structure-based sensors. Depends upon the input pressure, PZR pressure sensors are classifying as absolute pressure sensor, differential pressure sensor, and gauge pressure sensor [9]. The main differences between these two configurations are the fabrication methods. Most cantilever-based sensors use the surface micromachining method (SMT) [6]. The diaphragm or membrane or thin or thick plate sensors have to use both the bulk micromachining technique BMT, and SMT fabrication methods depend upon the applications.

Glaucoma is a disease that leads the irreversible blindness. According to WHO affect about 60.5million people have glaucoma in 2010, of which 4.5 million will develop bilateral blindness [2]. The typical IOP pressure is in the range of 10-21mmHg. The main aim of our design is to monitor IOP pressure continually for twenty-four hours. The capacitive sensing mechanism is not linear, but the PZR sensing is linear that has been widely using in IOP monitoring applications. E Y Chow et.al [2] introduced the MEMS-based piezoresistive pressure sensor and ASIC for monitoring IOP pressure, as shown in Fig.1.1. The problem with this works the ASIC and RF devices are not that much compatible. Another issue is the battery life in an additional wireless charging device required for its proper operations. Fig.1.2 indicated the internal cross-sectional view of the ocular implant.

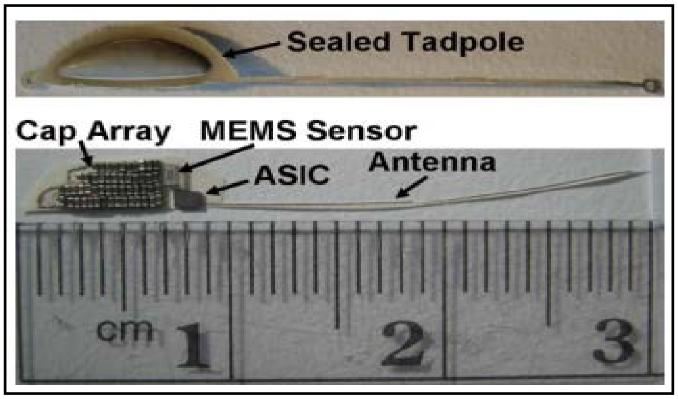

Fig.1.1. MEMS PZR for glaucoma monitor [2]

The first part of the paper discussed the literature review and introduction of piezoresistive MEMS sensors in biomedical applications. The second part consists of design equations and modeling. The result and discussion part are mentions in part four. The conclusion of the work is mention in part five.

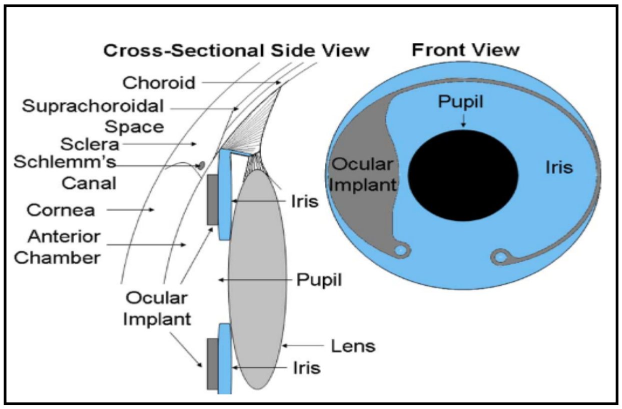

Fig.1.2. Cross sectional view of ocular implant [2]

2. DESIGN OF PZR SENSOR

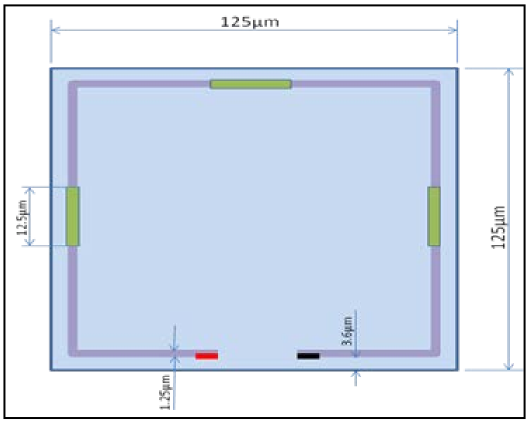

Fig.2.1.Geometry of proposed Model

The structure consists of three parts, first PZR elements which are placed on the top of the substrate. The material chosen for PZR is a p-type semiconductor (silicon doped with boron) which provides a high value of the PZR coefficient. The dimensions of PZR are 12.5x1.25x0.1 μm^3. These ultra miniaturized elements can be available to detect IOP pressure range from 1KPa to 8KPa [8]. These tiny PZR elements are capable of producing linear output. The second part is the silicon-based square diaphragm. The mono-crystalline silicon material is used for the construction of the diaphragm. The dimensions of the square diaphragm are 125x125x1 μm^3. The gap between the PZR elements and substrate is 3.6 µm. The doping density of PZR elements is 1.82x10 15cm^{-3} [7].

The silicon is an anisotropic material so the stress acting on the PZR elements is non-uniform for the everywhere. The flicker noise is more dominant concerning thermal noise in the pressure above 8KPa. The rectangular-shaped PZR is placed at the edges concerning the substrate because the highest value of stress value appeared at the edge of the diaphragm. The PZR is the change in resistance to initial resistance; this is due to a change in the stress of the material. The relation between the PZR and stress has been mentioned in equations (2.1-2.4).

$$(\Delta R / R) = \pi_l \sigma_l + \pi_t \sigma_t \tag{2.1}$$

Where σ indicated the stress and π indicated the piezoresistive coefficient. At first we compute the value of piezoresistive coefficient is given by,

$$\pi_l = \frac{\pi_{11} + \pi_{12} + \pi_{44}}{2} = \frac{\pi_{44}}{2} \tag{2.2}$$

$$\pi_t = \frac{\pi_{11} + \pi_{12} - \pi_{44}}{2} = \frac{\pi_{44}}{2} \tag{2.3}$$

$$\left(\Delta R / R\right) = \frac{\pi_{44}}{2} (\sigma_l - \sigma_t) \tag{2.4}$$

The equation (2.4) indicated the PZR is proportional to difference of the longitudinal and transverse stress, and keep with π_{44} is proportionality constant [9].

$$V_o = \frac{\Delta R}{R} V_a \tag{2.5}$$

$$S=0.1155\, \pi_{44} (1 - v)(a / h)^2 \tag{2.6}$$

$$GF = 1.56\ PZR \tag{2.7}$$

Where Va is the applied voltage to the piezoresistor for the sensor applied voltage is 1.5V. The PZR sensor interfaced with 130 nm CMOS technology-based ASIC developed by Texas Instruments (Ti). The CMOS IC is operated by the supply voltage of 1.5V and the typical power consumption is 676 pW [2]. The equations (2.5-2.7) are related to the sensitivity of the PZR sensor.

3. RESULT AND DISCUSSION

The parameters of MEMS-based sensors are deflection, stress and strain distribution, temperature distribution, current and voltage distribution, and sensitivity. All the parameters are calculated by the finite element method, which provides electro-mechanical analysis with more accurate results. The input parameter of a sensor is mechanical parameters, and the output for the sensor is electrical voltage or current. The energy conversion of mechanical to electrical is very fast in MEMS sensors concerning the conventional sensors. The input applied voltage of the PZR element is 1.5V dc, which is the operating voltage for CMOS IC. The deflection of the sensor towards downward direction indicated negative deflection and also the deflection of the diaphragm displacement in upward direction indicated positive deflection. Due to the differential input pressure on the diaphragm producing an oscillatory movement of the defection, in this case, resultant sensitivity is the difference between these input pressures.

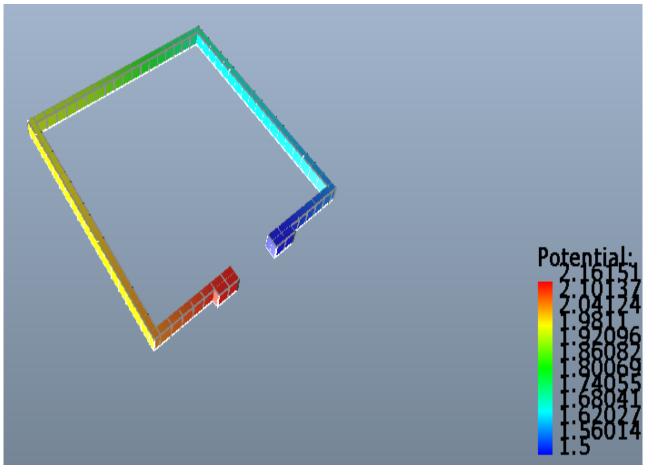

Fig.3.1. Piezoresistive surface electrical potential

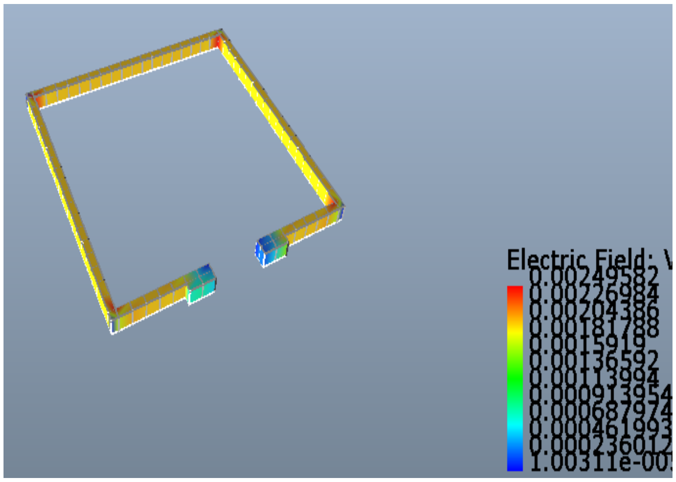

Fig.3.2. piezoresistive surface electric field distributions.

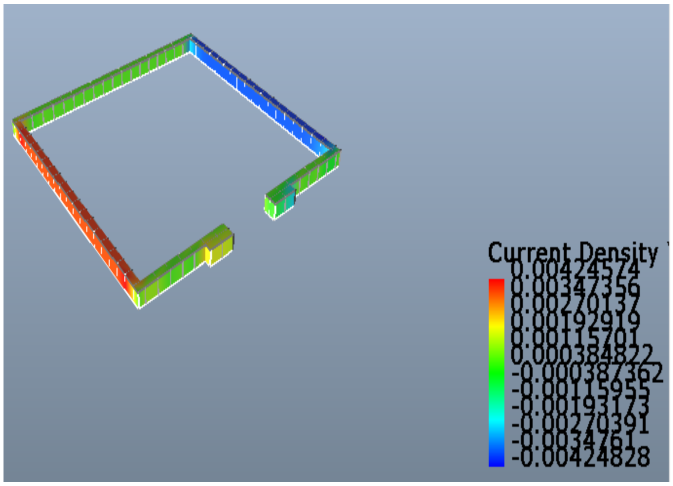

Fig.3.3. piezoresistive surface current density.

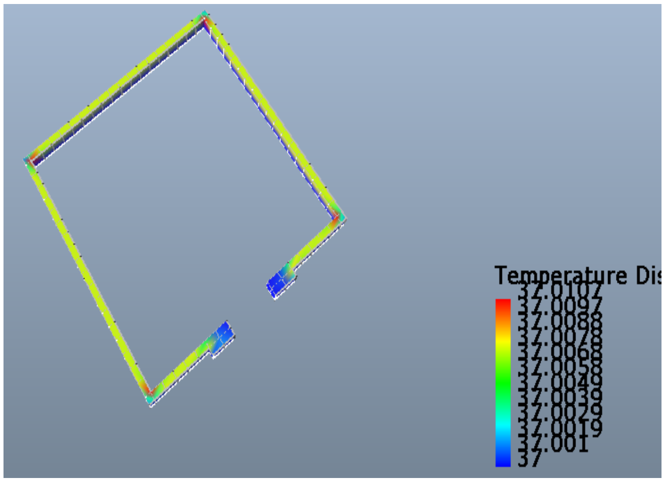

Fig.3.4. piezoresistive surface temperature distributions

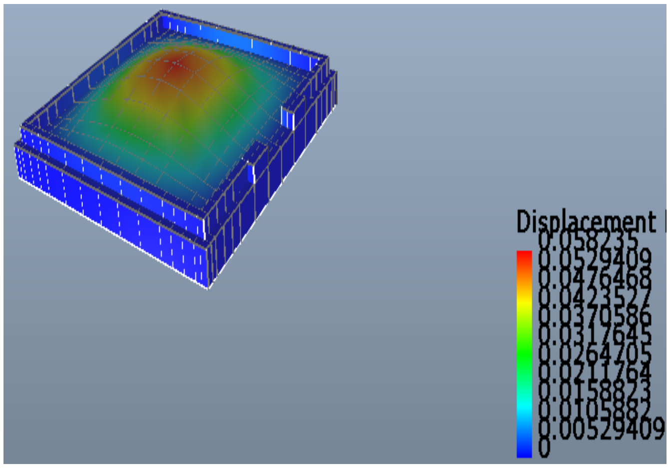

Fig.3.5. Displacement of sensor at IOP pressure range

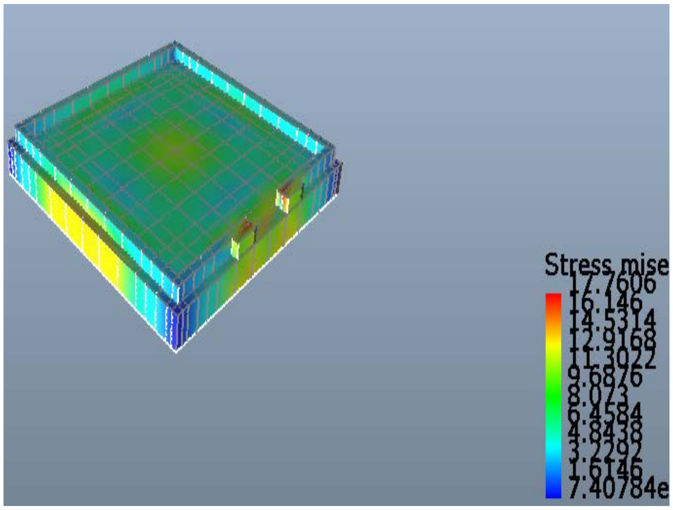

Fig.3.6. Von-mise stress at IOP pressure range

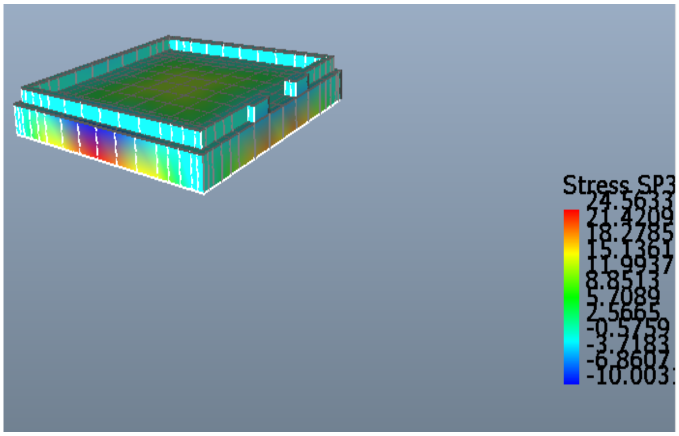

Fig.3.7.Principal Stress tensor (in Z-direction) at IOP pressure

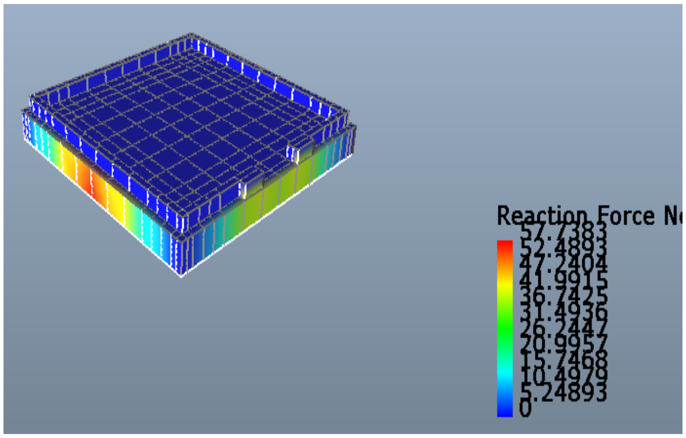

Fig.3.8. Reaction forces at IOP pressure normal to incident.

Fig. (3.1-3.4) results focused on the PZR elements at different input IOP pressure conditions. Fig. (3.5-3.8) results as shown the stress and deflection of the diaphragm-based sensor. Fig.3.1 indicated the potential distribution of the piezoresistive surface. Here the supply voltage is 1.5V dc has applied on the surface of the Piezo-resistors elements. The maximum value of electrical potential around 2V is obtained, which is sufficient for interface with CMOS IC. The potential difference exists between the two terminal points of PZR elements (highest potential indicated as red color and blue color for lower potential) i.e. gradient of potential has not a uniform along with the PZR are shown in Fig.3.2. The non-uniform distribution of the Electric field has producing the non-uniform electric current density has shown in Fig.3.3. The temperature discrepancy of the PZR sensor has shown in Fig.3.4. The temperature distribution function of the PZR sensor has almost constant. The typical range of human body temperature lies in between 36.50c to 37.50c, for this reason, the designed PZR sensor has operated at a constant temperature of 370c [5]. Fig.3.5 has indicated the deflection of the piezoresistive sensor with operating pressure in the range of IOP pressure (from 1KPa to 8KPa). The maximum value of deflection has 0.052 µm at the center of the diaphragm. The values of displacement has not equal in all directions because the four edges of the square diaphragm are fixed. The maximum value of displacement has appeared at the center of the diaphragm. The total equivalent stress has denoted by the von mise stress, has shown in Fig.3.6 Due to reason for the maximum stress induced at the edges of the diaphragm; the piezoresistive p-type material has been placed on the edges of the silicon diaphragm. Fig.3.7 indicated the principal stress distribution; principal stresses are those stresses that act on the principal surface. The principal surface the components of shear stress components are zero. Stress is an intrinsic material property; equal to the intrinsic force acting on a unit area of the cross-section of the diaphragm. This intrinsic force is also known as a reaction force of the sensor. Fig.3.8 clarified the reaction force value has maximum at the edge of the sensor.

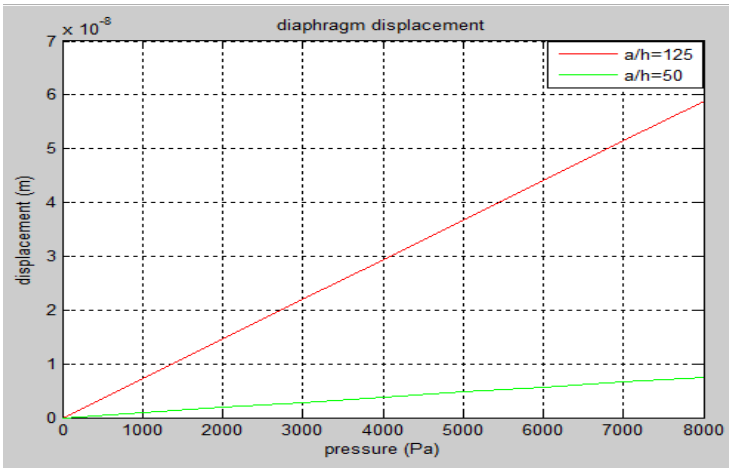

Fig.3.9.Diaphragm displacements versus aspect ratio

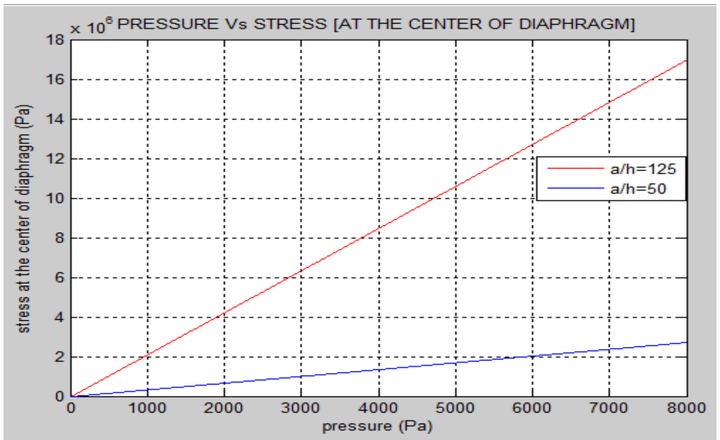

Fig.3.10.Stress at the centre of diaphragm versus aspect ratio

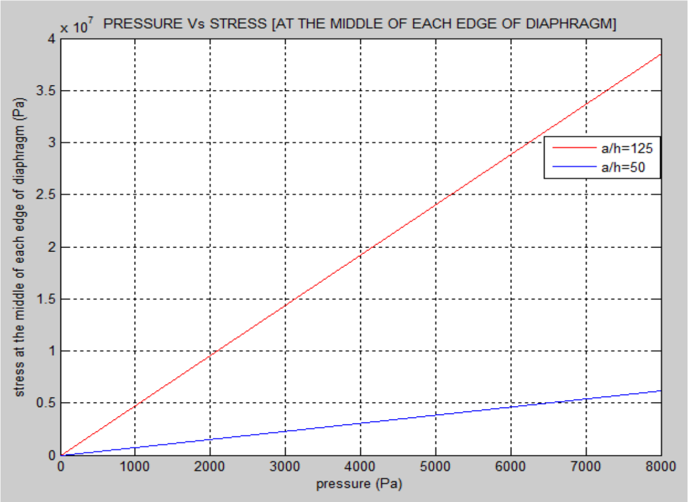

Fig.3.11.Stress value at the edges of diaphragm versus aspect ratio

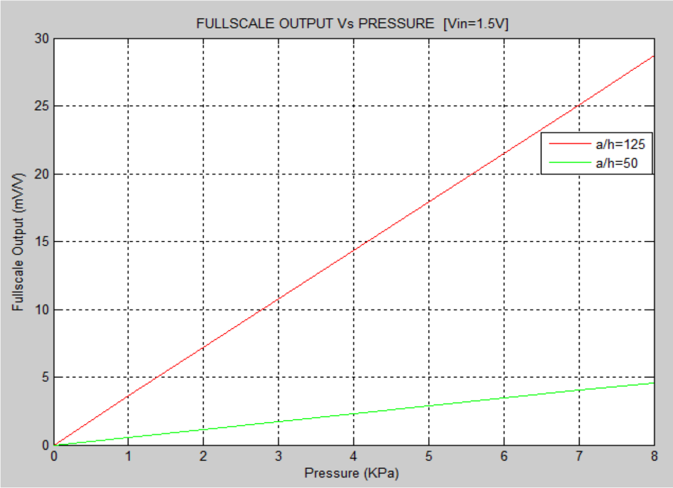

Fig.3.12.Full scale output versus aspect ratio

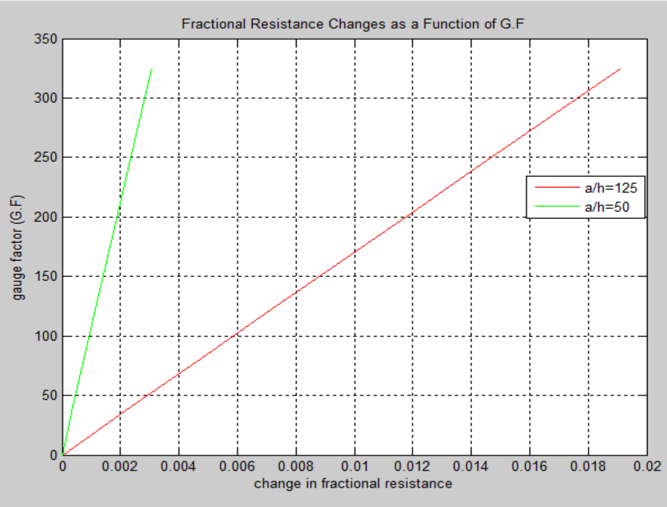

Fig.3.13. Fractional resistance changes versus gauge factor

The aspect ratio has depends the performance of the sensor. The higher value of the aspect ratio provides the higher performance of the design as shown in Fig. (3.9-3.13). Fig.9 shown the diaphragm displacements versus aspect ratio, As a result of 8KPa applied pressure, the displacement of the diaphragm has 0.0059 µm (redline) at aspect ratio 125, 0.0018 µm (green line) at aspect ratio 50. The total displacement is proportional to an applied pressure. Fig.3.10 and Fig.3.11 indicated the stress versus applied IOP pressure plots. The highest value of the aspect ratio sensor provided more induced stress concerning the lower aspect ratios with the same input pressure.Fig.3.12 shown the FSO output versus input IOP pressure plot, with respect to the different aspect ratios. The aspect ratio 125 produced FSO has 28mV/V; similarly, aspect ratio 50 produced FSO has 4.828mV/V concern the same input the pressure (8KPa). Equation (2.7) indicated the relation between the gauge factor (G.F), and the piezoresistive effect and Fig.3.13 has shown the linear plot of fractional resistance change as a function of the gauge factor (G.F). The slope of the graph increased by the decrease the values of a/h ratio. The proposed design has achieved frictional resistance change with the aspect ratios 125 and 50. The sensitivity of the PZR sensors linearly vary with input pressure.

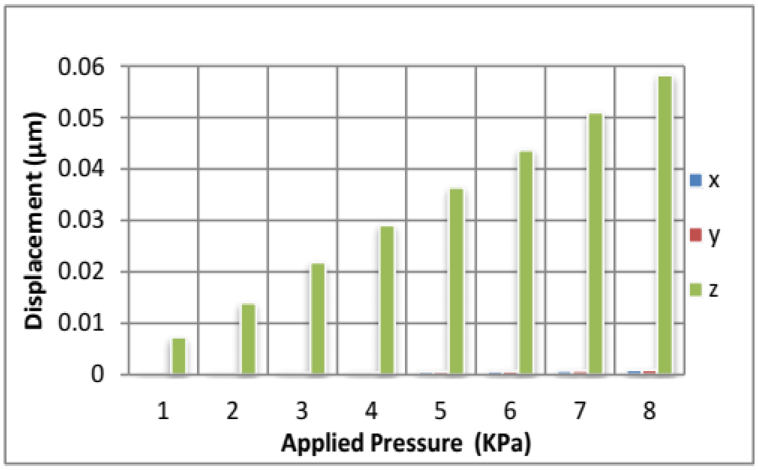

Fig.3.14.Diaphragm displacements versus applied IOP pressure

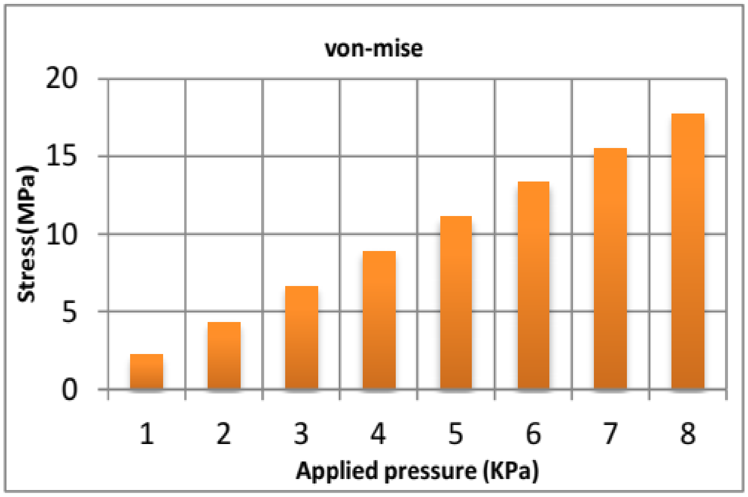

Fig.3.15. Von-mise stress versus applied IOP pressure

Fig.3.14 indicated the displacement of the diaphragm from 1KPa to 8KPa in (x, y, z) direction. The highest value of displaced sensor has occurred at z-direction. The "Von Mises stress" indicated the equivalent stress of the sensor. (The yield stress is the intrinsic stress of the sensor, considered as the failure stress condition). The total equivalent stress is often called the "von mises stress" of the sensor. If the "von mises stress" exceeds the yield stress, the material has considered a failure condition [9]. Fig.3.15 indicated von mises stress of PZR sensor with aspect ratio 125.

4. CONCLUSION

The designed MEMS-based piezoresistive sensor has highly beneficial for glaucoma monitor applications. The designed PZR sensor overall volume is 125x125x1 μm^3. The full-scale span of the sensor is 28mV. The given sensor design as based on the FEM method. The diaphragm dimensions and their values of the aspect ratios are receiving from the conventional theory of plate. Traditionally, the piezoresistive device has the highest linearity and sensitivity concerning the capacitive sensor. The PZR MEMS sensor is one of the leading sensors in the pressure sensor market, and it is very useful for biomedical applications. The design equations provided an essential guide for choosing the dimensions, shape of the diaphragm, and the placing area of sensing resistors.

References

[1] K V Menna, A R Shankar, “Biomedical Catheters with Integrated Miniature Piezoresistive Pressure Sensors: A Review”, IEEE Sensors Journal, May 2021, vol. 21, pp. 10241 – 10290.
[2] E Chow, E Chlebowski, "A Miniature-Implantable RF-Wireless Active Glaucoma Intraocular Pressure Monitor", IEEE transactions on biomedical circuits and systems, December 2010, vol.4, pp.340-349.
[3] A Barlian, W T Park, "Review: Semiconductor Piezoresistance for Microsystems", Proceedings of the IEEE, April 2009, vol.97, pp.513–552,
[4] S Sharma, A Cherty, “Nanofibrous Membrane Based Pressure Sensor with Ultra-High Sensitivity Over Broad Pressure Range for Wearable Healthcare Applications”, IEEE 34th International Conference on MEMS, Jan 2021.
[5] H Phan, D Dao, “The Piezoresistive Effect of SiC for MEMS Sensor at High Temperatures: A Review", Journal of micro electro mechanical Systems, December 2015, vol.24, pp. 1663 – 1677.
[6] M Madou, “Fundamentals of microfabrication”, CRC Press, March 1997.
[7] S.M Sze, “Physics of semiconductor device”, John Wiley &Sons, April 1988.
[8] F Khoshnoud and C W Silva, "Recent Advances in MEMS Sensor Technology-Biomedical Applications", IEEE Instrumentation and Measurement Magazine, February 2012, vol.15, pp.8-14.
[9] M A Fraga, L Koberstein, “An overview on the modeling of silicon piezoresistive pressure micro sensor”, IEEE Workshop on Engineering Applications, May 2012.
[10] M Teymoori, H Asadollahi, “MEMS Based Medical Microsensors", IEEE Second International Conference on Computer and Electrical Engineering, December 2009.

Dr.Sajith.K Received the B.Tech degree from in College of Engineering Thalassery, Kerala in 2011, and M.Tech degree in Pondicherry Central University in 2013, and Ph.D degree in 2020, Electronics and Communication Engineering, Dept. of Electronics Engineering, Pondicherry Central University, under the guidance of **Dr.T. Shanmuganantham** (Gold medalist in Antenna research from NIT Trichy), Dept. of Electronics Engineering, Pondicherry central University. He has 3 years of teaching experience in various reputed Engineering Colleges and currently he is working as Asst. Prof. in the Dept. of Electronics and communication Engg, Govt. Engineering College Wayanad, Kerala Technical University (KTU), and Kerala. He is a member of IEEE, IEEE APS society, and IEEE MTTs Kerala chapter. During his research received a senior research fellowship grant award from the Government of Kerala. During his research carrier, he developed many metamaterial loaded CPW fed on-body antennas for healthcare monitoring applications, and also he received “**Five Best paper awards”** in various IEEE conferences. He has authored 15 international journals, 3 chapters in books, and 25 International conference papers. His research interest in the area of planar monopole antennas, FSS for electromagnetic shielding, Wearable and Implantable medical antennas, Microwave and Millimeter-wave antennas, Fractal Antennas, Metamaterial inspired antennas, and RF MEMS reconfigurable antennas, MEMS Phase shifter.

Dr. T. Shanmuganantham [IEEE chairman APS madras section India] received his B.E. degree in Electronics & Communication Engineering from University of Madras in 1996, M.E. degree in Communication Systems from Madurai Kamaraj University in 2000 and **Ph.D. (Received Gold Medal)** in the field of Antennas from NIT (National Institute of Technology), Tiruchirappalli in 2010 under the guidance of **Prof.S.Raghavan**, Senior Professor, Dept. of ECE, and NIT Tiruchirappalli. He has 21 years of teaching experience in various reputed Engineering Colleges and currently he is working as Asst. Prof. in the Dept. of Electronics Engg, School of Engg & Technology, Pondicherry Central University, and Pondicherry. His research interest includes Antennas, Microwave/Millimeter-Wave Engineering and MEMS/NEMS. He has published 680 research papers in various National and International level Journals and Conference proceedings. He received many Best Paper awards. Eight Ph.D. students had awarded under his guidance. He has completed two sponsored projects worth of Rs.57 lakhs. He is potential reviewer for IEEE Trans. on Antennas and Propagation, MTT, PIER Journals, John Wiley, Elsevier, Taylor and Francis and many IEEE Conferences. He has been elected as Fellow in Antenna Test and Measurement Society (ATMS) by ISRO Scientists and a senior member in IEEE, a Life Member in ISSS, IETE, IE (India), CSI (India), Society of ISTE, EMC, ILA, OSI and ISI. He is serving as office bearer for IEEE Circuits and Systems Society (India Chapter) and also he is Member of Board of Studies in Pondicherry University, University of Madras, and Annamalai University. His biography was included in **‘Marquis who is who in the world’** USA in 2010.

Dr. D. Sindhahaiselvi received B.E. degree in Electronics and Instrumentation Engineering in 1998, M.E. degree in Process Control and Instrumentation in 2000 both from Annamalai University in first class with distinction and Ph.D. in the area of MEMS Sensor from Pondicherry University in 2016. She has 20 years of teaching experience in various reputed Engineering colleges. Currently she is working as Assistant Professor in the Department of Electronics and Instrumentation Engineering, Pondicherry Engineering College (An Autonomous Institution of Pondicherry Government), Pondicherry. Her research interest includes MEMS Sensors, MEMS SWITCHES, MEMS Devices and BIO MEMS. She has published 75 research papers in various National and International level Journals and Conferences. She received several “**Best Paper Awards”** in various IEEE conferences. She is a Life Member in ISSS, ISTE, IE (India), ISOI and ACDOS. She is a Member of Board of Studies in Pondicherry University and Annamalai University. She is a reviewer of Reputed Journal such as JESTEC.