A NEW INTRACRANIAL PRESSURE SENSOR ON POLYIMIDE LAB-ON-A-TUBE USING EXCHANGED POLYSILICON PIEZORESISTORS

Z.Z. Wu[1], N. Bhattacharjee[1], C. Y. Li[1, 2], J. A. Hartings[3], R. K. Narayan[2] and C. H. Ahn[1]
[1]Microsystem and BioMEMS Laboratory, University of Cincinnati, Cincinnati, USA
[2]Cushing Neuromonitoring Laboratory, Feinstein Institute for Medical Research, Manhasset, USA
[3]Department of Neurosurgery, UC College of Medicine, Cincinnati, USA

ABSTRACT

A new intracranial pressure (ICP) sensor on polyimide lab-on-a-tube using exchanged polysilicon piezoresistive film has been designed, fabricated and successfully characterized in this work. Flexible polyimide film (Kapton) was used as the substrate for a lab-on-a-tube type microcatheter. Aluminum induced crystallization (AIC) process with low annealing temperature was utilized for the development of polysilicon film on polyimide substrate as the piezoresistive sensing element for a membrane based pressure sensor. The pressure sensor integrated with polyimide lab-on-a-tube shows a linearity of 0.996 and a sensitivity of 0.17 mV/mmHg in the low pressure range of 0~45 mmHg, which is very suitable for measuring the ICP.

KEYWORDS

Intracranial pressure (ICP), Piezoresistive pressure sensor, Polysilicon, Aluminum induced crystallization (AIC), Pressure sensor on polyimide

INTRODUCTION

Intracranial pressure is considered as one of the most important monitoring parameters for the patients with stroke or traumatic brain injuries (TBI). Increased intracranial pressure always leads to secondary brain injury due to the decreased cerebral blood flow and results in further morbidity and mortality. Therefore, the treatment of elevated ICP has been a central focus of many neurosurgical patients in critical condition [1]. Various microtransducer-tipped ICP probes, which are based on fiber optic (Camino ICP monitor) [2], strain gauge (Codman microsensor and Neuroven-P ICP monitor) [3, 4], or pneumatic (Spiegelberg ICP monitor) [5], are available for measuring ICP. However, these probes either need expensive and complicate detector or are composed of rigid silicon-based pressure sensor which needs complicated assembling and wiring to fit in the catheter for measuring, which makes them costly or difficult to fabricate or use. In this paper, we present a polyimide lab-on-a-tube integrated with pressure sensor. Piezoresistive polysilicon film was developed on polyimide substrate (Kapton film) and was used for the pressure sensing elements. Polysilicon film is formed on polyimide substrate using the aluminum-induced crystallization (AIC) process which is very suitable for the polymer material which requires the low processing temperature less than 400°C. By developing the piezoresistive film and electrical lead directly on the substrate, the sensor is free from the complex assembling and wiring problems. Furthermore, this sensor can be integrated with other sensors on the same tube. This flexible pressure sensor ultimately can be integrated with a lab-on-a-tube device [6] along with other microsensors to achieve multimodality neuromonitoring of injured brain.

Fig.1 shows the concept and schematic of the proposed device. The pressure sensor consists of two piezoresistors: an active one placed on the membrane and a passive one on the rigid substrate. When applying pressure, the active resistor will be under the pressure, so its resistance will be changed with the pressure, but not for the passive one.

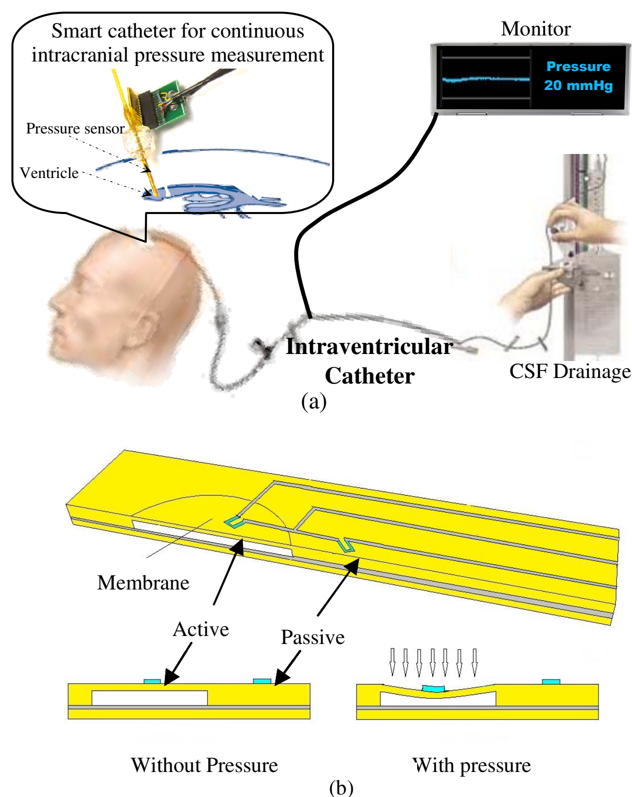

Fig. 1. Smart catheter with in situ neuromonitoring and therapy functions: (a) Flexible pressure sensor was integrated on the catheter. When its outputs indicate higher than 20 mmHg, excess CSF will be drained and (b) Polysilicon thin layer was formed on Kapton substrate and patterned into two piezoresistors.

POLYSILICON FILM

The piezoresistivity of polysilicon has been investigated and utilized so long for pressure sensing applications [7]. Comparing to metal alloys, polysilicon film has higher gauge factor, but its high processing

temperature prevents its fabrication on polymer or flexible polyimide films. However, the technique of AIC for polysilicon by Nast et al. [8] showed the development of polysilicon on the flexible polyimide substrate using a low annealing temperature. When annealed, the aluminum is embedded into the crystallized silicon film, and P-type polysilicon film is simultaneously formed. Thus, high temperature doping process can also be avoided.

In this work, aluminum and amorphous silicon thin layers with its individual thickness of 200 nm were sputtered on Kapton film. The sputtering power was carefully selected to achieve the desired surface roughness and the right deposited-grain size of aluminum layer, since the grain size of aluminum layer affects on the time required for the annealing process and also the grain size of the formed amorphous polysilicon [9]. The sample was then annealed in a diffusion furnace at 400°C for 1 hour in N_2 gas. During the annealing process, the exchange of the layers was occurred, so that a new polysilicon layer was formed under the aluminum layer. The top aluminum film was etched away, and then finally the newly formed polysilicon film was appeared on the flexible polyimide film.

XRD measurements have been done to verify the crystallization of polysilicon using X Pert Pro MPD system. Fig. 2(a) and (b) show the X-Ray results of the Kapton film with and without polysilicon layer, respectively.

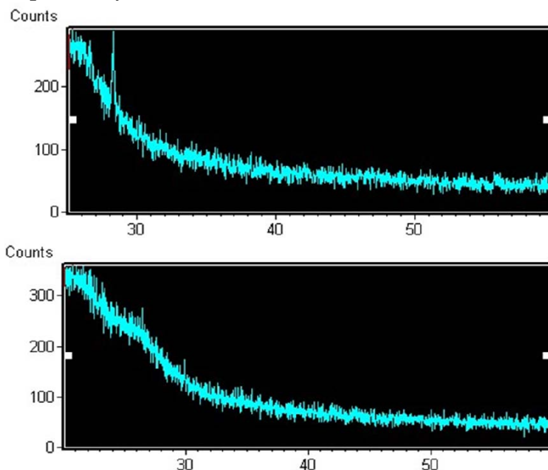

Fig. 2. Polysilicon characteristics: (a) XRD results of polysilicon layer on Kapton film: the peak at 28.5°<111>, 47.3°<220>, 56.3°<311> for polysilicon layer and (b) XRD results of Kapton film.

As shown in Fig. 2(a), the peaks were appeared at 2θ = 28.5°<111>, 47.3°<220> and 56.3°<311> for the Kapton film with polysilicon layer. Fig. 2(b) shows the amorphous nature of Kapton film, which is the background of Fig. 2(a).

The crystallite size can be estimated from the high resolution X-ray diffraction data using the Scherrer relation [10] as:

$$B = 0.9\lambda/t\cos\theta, \qquad (1)$$

where B is the full width at half maximum <FWHM>, λ is the X-ray wavelength, and t is the diameter of the crystal. Peak <111> was considered for estimating the grain size of the polysilicon film. With the FWHM of 0.144 ° in this work, the average size of the crystals of polysilicon film is about 49 nm.

Fig. 3 is the AFM pictures of surface morphology of the developed polysilicon film. As shown in the picture, a continuous polysilicon film with extremely low surface roughness has been successfully attained.

The polysilicon film was examined with energy dispersive spectroscopy (FEI XL30 ESEM) to verify the aluminum doping concentration. The results show the film has aluminum and silicon content with the At. % ratio of 2.75%: 97.25%. From the analysis, it could be seen that a small amount of aluminum was embedded in the final crystallized film which can acts as a p-type dopant in the polysilicon film.

The sheet resistance of the film was further measured to ensure the activation status of the embedded aluminum and the result of 2.5 KΩ/□ for 200 nm thick polysilicon film was attained, which clearly shows the conductivity of doped polysilicon film. This eliminated the need for an additional doping step.

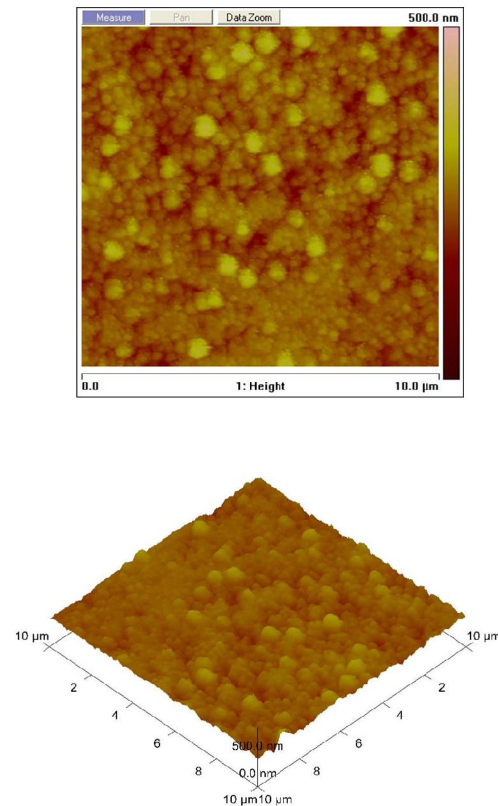

Fig. 3. AFM picture showing the continuous and densemicrocrystals of polysilicon film.

DESIGN AND MICROFABRICATION

Kapton film HN type (25 um thick) was chosen as the substrate for the pressure sensor because of its high glass transition temperature, biocompatibility, and chemically inert to etchant. The pressure sensor based on a half Wheatstone bridge was designed, with an active piezoresistor placed at the center of the membrane, while

a passive one placed on the rigid surface. When pressure is applied, the piezoresistor on the membrane is deformed. Utilizing the piezoresistivity of polysilicon, the resistance of the resistor will be changed. Since the passive one is placed outside of membrane, the resistance will not be affected by the pressure. Pressure will be measured by tracking the difference between the resistance of active and stationary resistor, in order to eliminate the possible effect brought by temperature variations. This is of essential importance for *In Vivo* testing due to the complicate working environment.

Fig. 4 shows the summary of fabrication steps. During the whole process, Kapton film was held flat by attaching to a PDMS coated silicon wafer. After annealing, the newly formed polysilicon and aluminum layer was patterned to form the sensing piezoresistors. Aluminum evaporation and etching was then performed to make electrical leads and simultaneously etched out the aluminum layer on the top of polysilicon to have polysilicon resistor. Due to the unique layer exchange, the polysilicon film can be protected from exposing to air which causes to form a harmful silicon oxide. Annealing was carried out at 250 °C for 1 hour to have Ohmic-contact between the piezoresistor and aluminum lead. The sample was then coated with 2 um thick parylene layer as the protective layer.

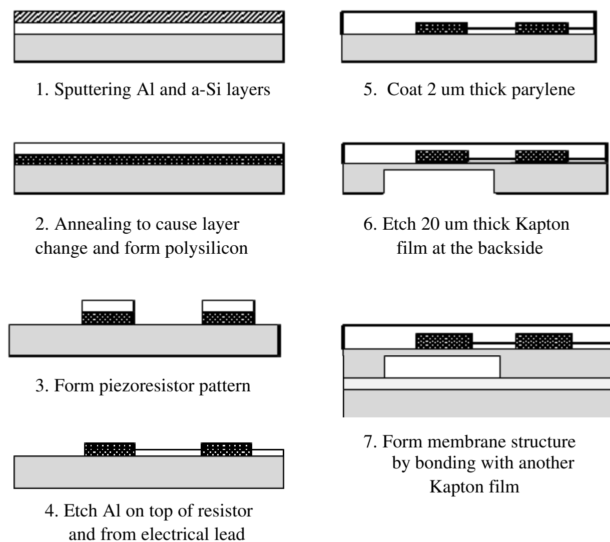

Fig. 4. Microfabrication Procedures.

To construct the membrane structure, the Kapton film with piezoresistors was flipped over, and 20 um of the Kapton film was etched out under the membrane area using reactive ion etching with oxygen and CF_4 plasma, and then sealed with another Kapton film.

The bonding was achieved using the step polymerization process of Kapton film. By curing the intermediate polymer, polyamic acid, ring-type bonding closure was produced between Kapton films. In the bonding process, Polyamic acid was spin coated on a plain Kapton film first, and then half cured to have a solid but uncured surface. The Kapton film with piezoresistor and membrane structure was then attached and further cured. During the curing process, the ring closure was constructed between the polyamic acid layer and the Kapton films, producing a strong bonding. After the completion of the curing process, two Kapton films were tightly bonded together, and the intermediate layer of polyamic acid was also turned to Kapton film. The advantages of this bonding method are its low processing temperature (100 °C) as well as excellent bonding strength.

Once the planar microfabrication steps were completed, the film with microsensor was cut to the desired sizes, detached from the supporting silicon wafer, and then spirally rolled around a metal rode to make the intraventricular catheter.

Fig. 5(a) shows the real device in both plain and rolled forms. Fig. 5(b) shows the fabricated pressure sensor on the membrane. . The radius of the membrane is 500 um.

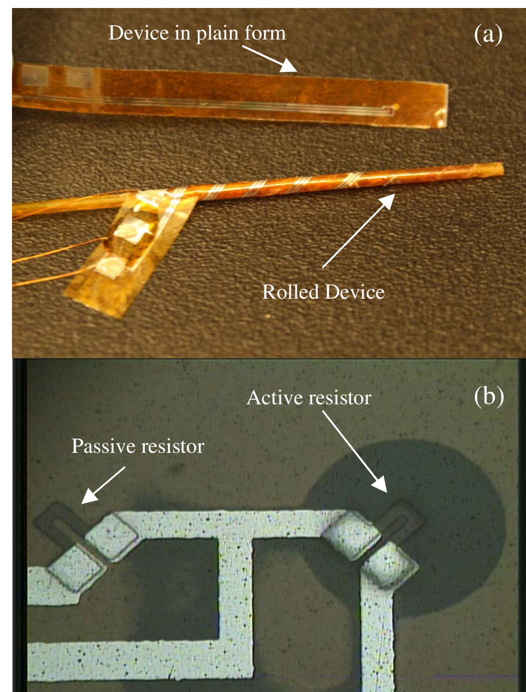

Fig. 5. Photographs of the fabricated device: (a) Real device in both plain and rolled form and (b) Membrane part under microscope.

EXPERIMENT AND RESULTS

The electrical leads of the pressure sensor were bonded on the contact pads by silver epoxy and covered with UV-cured adhesive as a protective layer. In order to characterize the fabricated pressure sensor, the pressure sensor was mounted between a cylinder to contain water and a rigid plate. The cylinder and the plate were then bonded together by Epoxy to make a container of water column. The reference static pressure was produced by changing the level of the water column. While the applied pressure was varied in the range of 0- 45 mmHg, the variations of resistance from the piezoresistors were measured.

The measured resistances for the developed intracranial pressure (ICP) sensors were shown in Fig. 6. As clearly shown in Fig. 6, the resistance of passive piezoresistor was changed very little. However, the active piezoresistor shows a linear variation in terms of the changes of pressure. A constant DC current of 30 uA was

applied to the sensor and voltage differences between the two resistors were measured as the output. In Fig. 7, the output voltage from the half Wheatstone bridge shows a linearity of 0.996 and a sensitivity of 0.17 mV/mmHg for the pressure of 0-45 mmHg.

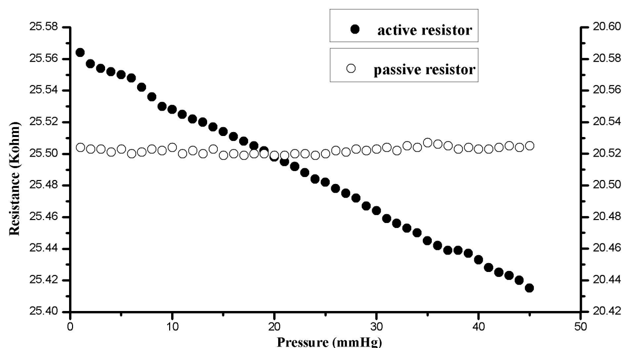

Fig. 6. Resistance change with pressure for both active resistor and passive resistor.

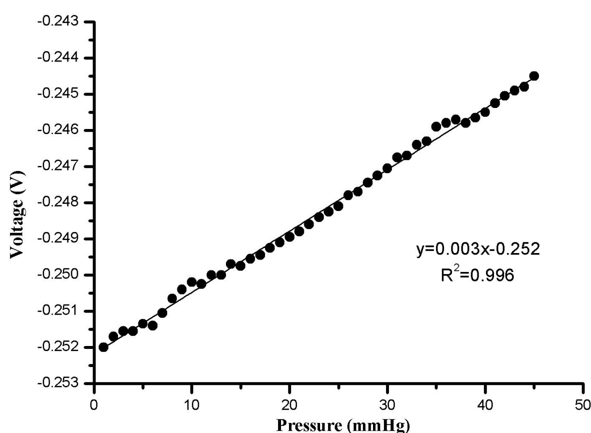

Fig. 7. Output voltage vs. applied pressure.

In addition, temperature variability tests were also performed in the temperature range of 25 - 45°C under the pressure of 12 mmHg to characterize the effect of temperature on the device performances. The results are shown in Fig. 8. As shown in the graph, the output voltage is very stable with respect to the wide variation of temperature.

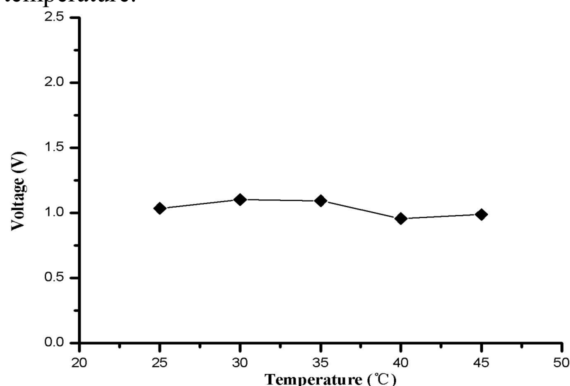

Fig. 8. Output voltage vs. temperature.

CONCLUSION

In this work, polysilicon film on Kapton film has been successfully developed and applied to the new intracranial pressure (ICP) sensor integrated with a polyimide lab-on-a-tube. The crystallization and electrical property of the developed polysilicon film were fully characterized. The pressure sensor utilizing the polysilicon film as the piezoresistive sensing component was also fully characterized for the pressure rang of 0-45 mmHg, and the results show a sensitivity of 0.17 mV/mmHg and a linearity of 0.996 within the range.

The polysilicon film and pressure sensor developed in this work envisages the development of new biomedical devices on polyimide substrate for numerous neurosurgical monitoring or health care applications.

ACKNOWLEDGEMENT

This work was supported by the US Department of Defense under project No. PT090526P4 (W81XWH-10-1-0978).

REFERENCE

[1] M. Smith, “Monitoring Intracranial Pressure in Traumatic Brain Injury”, Anesth. Analg., Jan 2008 106: 240~248

[2] E. Munch, Weigel R, Schemiedek P, Schurer L, “The Camino intracranial pressure device in clinical practice: Reliability, handling charateristics and complications”. Acta Neurochir (Wien) 140: 1113~1120, 1998

[3] H. M. Fernandes, K.Bingham, I.R. Chambers. “Clinical evaluation of the Codman microsensor intracranial pressure monitoring system” Acta Neurochir, Suppl. (Wien) 71, 44, 1998

[4] Citerio G, Piper I, Cormio M, Galli D, Cazzaniga S, “Bench test assessment of the new Raumedic Neurovent-P ICP sensor: a technical report by the BrainIT group”. Acta Neurochir (Wien) (2004)

[5] I. Piper, L. Dumn, C. Contant, Yau Y, Whittle I, Citerio G, “Multi-centre assessment of the Spiegelberg compliance monitor: preliminary results.” Acta Neurochir Suppl. (Wien) 76, 491, 2000

[6] C. Li, L.A. Shutter, P.M. Wu, C.H. Ahn, R.K. Narayan, “Potential of a simple lab-on-a-tube for point-of-care measurements of multiple analytes”, Lab on a Chip, Vol. 10, pp. 1476-1479, 2010.

[7] P. J. French, A.G.R. Evans, “Piezoresistance in polysilicon and its applications to strain gauges”, Solid-State Electronics, Vol. 32, NO.1, pp. 1~10, 1989

[8] Oliver Nast, Tom Puzzer , Linda M. Koschier, Alistair B. Sproul, and Stuart R. Wenham. “Aluminum-induced crystallization of amorphous silicon on glass substrates above and below the eutectic temperature”, Appl. Phys. Lett. 73, 3214, 1998

[9] Oliver Nast and Andreas J. Hartman, “Influence of interface and Al structure on layer exchange during aluminum-induced crystallization of amorphous silicon”, J. Appl. Phys. Vol. 88, 2000

[10] S. Vijayalakshmi, Z. lqbal, M.A. George, J. Federici, H. Grebel, “Characterization of laser ablated silicon thin films”, Volume 339, Pages 102-108, 1999

CONTACT

Z.Z. Wu +5139237896 wuzn@mail.uc.edu