Symmetrical PolyMUMPs-Based Piezoresistive Microcantilever Sensors With On-Chip Temperature Compensation for Microfluidics Applications

Adisorn Tuantranont, Tanom Lomas, Kata Jaruwongrungsee, Apichai Jomphoak, and Anurat Wisitsoraat

***Abstract*—Microelectromechanical systems (MEMS)-based cantilever beam sensors for microfluidics applications with on-chip temperature sensors for temperature drift compensation were developed. The stress induced on gold surface with polysilicon piezoresistive sensing is demonstrated. In principle, adsorption of biochemical species on a functionalized surface of the microfabricated cantilever will cause surface stress and, consequently, cantilever bending. The sensing mechanism relies on the piezoresistive properties of the doped polysilicon wire encapsulated in the beam. The beam is constructed through multiusers MEMS Process (PolyMUMPs) foundry with postprocessing silicon etching. Bending analysis is performed so that the beam tip deflection can be predicted. The piezoresistor designs on the beams were varied, within certain constraints, so that the sensitivity of the sensing technique could be measured by external read-out circuit. The mass detection of 0.0058–0.0110 g is measured by the beam resistor series as a balanced Wheatstone bridge configuration. The voltage output of the bridge is directly proportional to the amount of bending in the MEMS cantilever. The temperature dependency and sensor performance have been characterized in experiments. Compensation by resisters on the substrate significantly reduces the temperature dependence.**

***Index Terms*—Gas and chemical sensor, microcantilever, piezoresistive, PolyMUMPs.**

I. Introduction

MICROELECTROMECHANICAL SYSTEMS (MEMS) technology has generated a significant amount of interest due to the potential performance and cost advantages with microscale devices fabricated based on a silicon processing technology.

MEMS devices utilize numerous transducing mechanisms for both actuation and sensing. A microcantilever beam with piezoresistive sensing is applied in the control of computer disk drive read/write heads, the control of microscope heads, and on/off switches for fiber optical devices. The MEMS cantilevers such as those used by atomic force microscopy (AFM) have been applied to measure physical and biochemical properties. The deflection of these cantilever beams can be detected using various techniques same as the techniques for AFM technology such as optical reflection, piezoresistive, piezoelectric, capacitive, and electron tunneling [1].

Alternatively, this device acts as a surface stress sensor which sensitivity of detection based upon the adsorption-induced force and resonant frequency shift when a specific biochemical species adsorbed on a functionalized surface of the cantilever beam [2]. These biochemical sensors have the potential be applied as a patient classifier for the presence of various diseases with faster responses, higher accuracy, and lower cost [3].

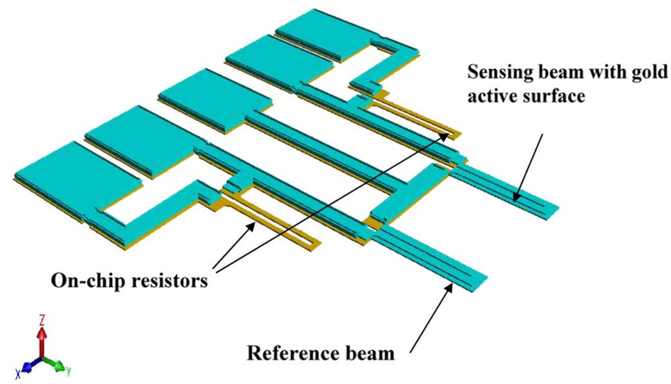

Fig. 1. Model of our MEMS piezoresistive cantilever sensor with an integrated full Wheatstone bridge on a chip.

II. Device Fabrication

The cantilever-based sensors have been developed with the integrated piezoresistive read-out with the integration of a full and symmetric Wheatstone bridge on chip with two resistors placed on the cantilevers and two resistors on chip, as shown in Fig. 1.

The researchers have applied a commercially available MEMS process, PolyMUMPs (multiuser MEMS processes) to design for fabrication. This process will follow by the post-processing etching steps allowing the structure released from the bulk of the silicon wafer after the PolyMUMPs process [4]. A deep trench under the cantilever was created by a wet etching process using ethylene diamine pyrochatechol (EDP). All layers of PolyMUMPS and silicon substrate are exposed to

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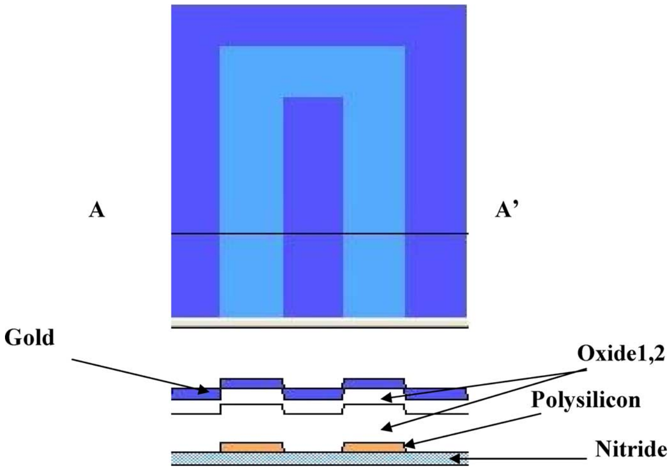

Fig. 2. Cross section of MEMS cantilever $(\mathrm{A}-\mathrm{A}')$.

chemical etchant using the superimposition of "Anchor1," "Anchor2," "Poly1-poly2 via," "Dimple," "Hole1," and "Hole2" layers on top of each other to create a deep trench mask. By this method, the nitride and all the oxide layers above nitride are opened through during standard PolyMUMPs fabrication [5]. The polysilicon structures are not affected by EDP etching because the etching rate of polysilicon in this chemical etchant is much slower than single-crystal silicon and can be negligible.

Our piezoresistive cantilever can be functionalized with biochemical materials to perform antigen-antibody reaction or DNA hybridization while the other bare cantilever can be applied as a reference beam. Thus, there is a possibility of performing differential measurements through the subtraction of signals from two cantilevers with the minimization of background noises from thermal drift and fluid turbulence effects.

The cantilever beam is comprised of four structural layers, silicon nitride, polysilicon, silicon dioxide, and gold, as shown in Fig. 2. The polysilicon wire forms a resistive element that runs down the length of the beam and back. The silicon dioxide, Oxide1 layer fills the gap between the polysilicon wires, while the Oxide2 layer fills the gap between polysilicon and top gold layer. The AFM picture of top gold layer acts as a reactive or adsorption layer, as shown in Fig. 3. The beam has a length of 200 μm with 40 μm width. All layers dimension including thickness was shown in Table I.

When the biochemical sample (biological molecules such as proteins or biological agents) is applied to the cantilever sensor, some of molecular sample is binding with the gold layer, the gold surface is either tension or compressive. This causes the cantilever to deflect and its deflection was found to be proportional to the biochemical concentration. When the gold layer expands, the bending of the beam subjects the polysilicon layer to tensile or compressive stresses. Since polysilicon is piezoresistive, it acts as a sensing element; thus, the stress can be determined by measuring the resistance of the polysilicon wire. The deflection principle was shown in Fig. 4. Fig. 5 shows SEM pictures of the MEMS cantilevers array fabricated device with a deep trench silicon substrate. The cantilevers were bending downward after etching.

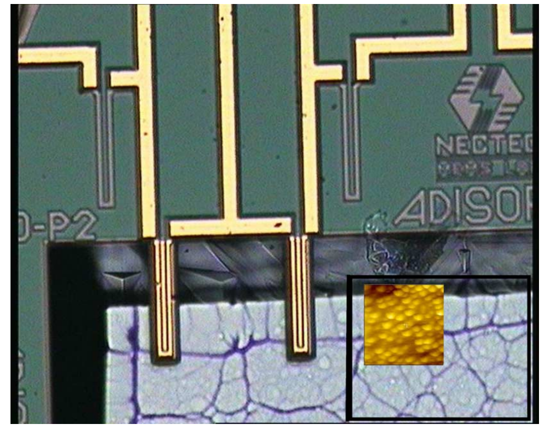

Fig. 3. Top gold layer acts as a reactive or adsorption layer, inlet shows surface morphology by AFM.

TABLE I
ALL DIMENSIONS OF MEMS CANTILEVER BEAMS

Material layer	Wide (μm)	Long (μm)	Thickness (μm)
Nitride	40	100	0.5
Polysilicon	10	400	0.5
Silicon dioxide	40	200	2.25
Gold	40	200	0.5

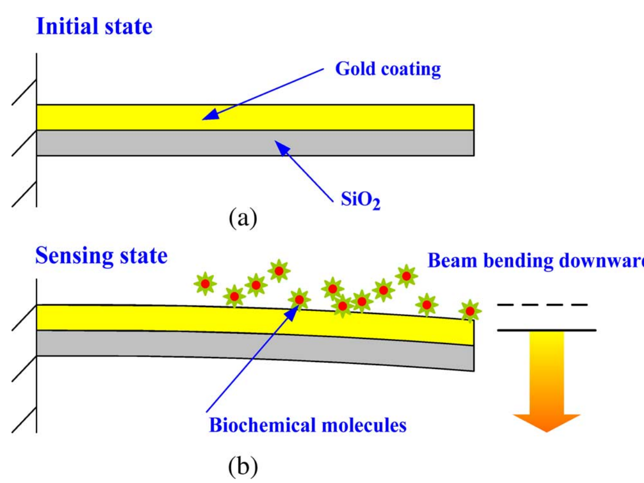

Fig. 4. Cantilever beam response. (a) Initial state. (b) Sensing state.

III. Device Modeling and Simulation

Device Modeling: This methodology is conducted to analyze the beam deflection due to surface stress after the device was adsorbed by biochemical molecules. The adsorbed layer can attempt to expand or contract (known as a compressive surface stress and a tensile surface stress, respectively) [6].

MEMS cantilever is bending due to mechanical force generated by molecular adsorption. Adsorption-induced stress sensors have a sensitivity range based on adsorbed mass which proportional to molecular size. In addition, MEMS cantilever bending is also ideal for liquid-based biosensor applications.

Using Stoney's formula [7], the radius of curvature of cantilever bending due to adsorption is expressed as

$$\frac{1}{R}=6\frac{(1-v)}{Et^2}\delta\sigma \qquad (1)$$

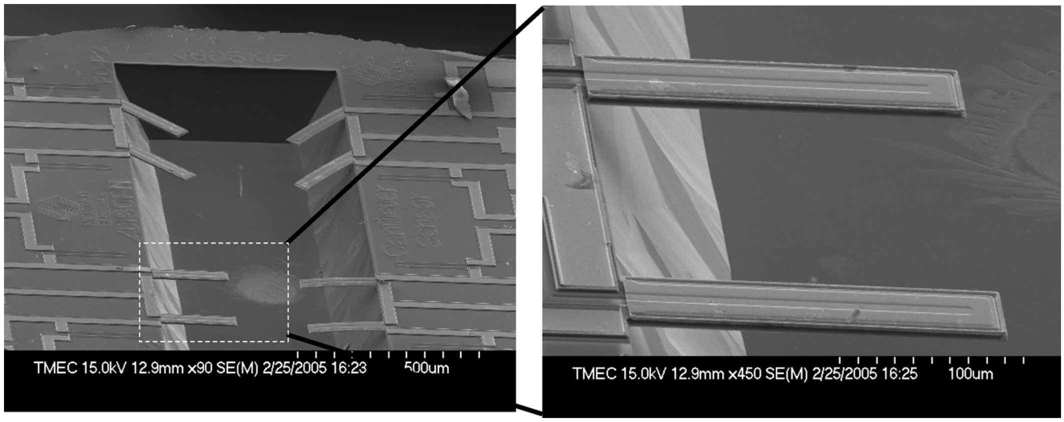

Fig. 5. SEM of fabricated cantilever array sensors.

where R is the cantilever's radius of curvature; v and E are Poisson's ratio and Young's modulus for the substrate, respectively; t is the thickness of the cantilever; and $\delta\sigma$ is the differential surface stress, which is the difference between the surface stress of the top and bottom surfaces of the cantilever beam in units of N/m or $\mathrm{J/m^2}$. A relationship between the cantilever tip displacement and the differential surface stress can be expressed as [6]

$$z = 3L^2 \frac{(1-v)}{Et^2} \Delta\sigma \tag{2}$$

where L is the length of cantilever and z is the deflection at the tip of cantilever. This equation shows a linear relation between cantilever bending and differential surface stress.

The mechanical design of piezoresistive polysilicon wire encapsulated in MEMS cantilever is optimized for detecting changes in surface stress upon analyze surface adsorption. The fractional change of resistance $(\Delta R/R)$ in a piezoresistive wire is described by the following expression [8]:

$$\frac{\Delta R}{R} = 3\beta\pi_L \frac{(1-v)}{t} \Delta\sigma \tag{3}$$

where π_L is the piezoresistive coefficient of polysilicon, $\Delta\sigma$ is the difference of the longitudinal stress and the transverse stress, t is the thickness of the cantilever, and β is a factor that is adjusted for the thickness of the piezoresistor [9]. From the above expression, the $\Delta R/R$ ratio is proportional to the stress difference, $\Delta\sigma$ is the stress difference distribution depending on the geometric factors of the layers and the reaction forces between the gold surface and the biochemical molecules. Therefore, maximizing the stress difference in the way of changing the geometric factors can increase the deflection signals.

Stress Induced Simulation: The cantilever model was created using the process simulator and CIF mask import to CoventorWare. Table II shows the thin-film material property data used in this modeling and numerical simulation.

A variable stress induced simulation is performed using MemMech solver. First, the gold functionalized area had varied stress from 0.001 MPa to 0.020 MPa to verify the beam deflection due to the stress induced on solid model, as shown in Fig. 6. This stress is corresponding to surface stress along 40-μm-wide cantilever beam of 40 mN/m to 800 mN/m, respectively. Thus, the simulation result suggests that the cantilever beam system has sufficient resolution for bio/chemical reactions, which typically give surface stress response in the order of tens of mN/m [3]. The amount of tip deflection shows in the color bar. The plot of stress variable versus beam tip deflection was shown Fig. 7. The simulation results show that the beam

TABLE II
THIN FILM PROPERTIES USED FOR SIMULATION

Property	Unit	Polysilicon	Oxide	Gold
Elastic Constants	MPa	1.65e05	7.0e04	7.72e04
	Poisson ratio	2.3e-01	1.7e-1	4.2e-01
Density	kg/µm^3	2.23e-15	2.1e-15	1.9e-14
CTE	1/k	3.5e-06	0.35e-6	1.42e-5
Thermal Conductivity	pW/µm.K	5.0e007	1.42e06	3.0e08
Specific Heat	pJ/kg.K	1.0e14	7.1e-14	1.28e14
Electrical Conductivity	pS/µm	7.0e010	-	3.4e13

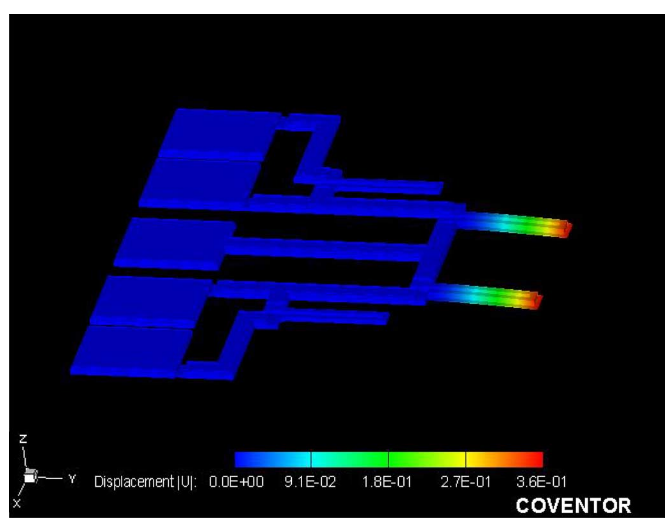

Fig. 6. Simulation of stress induced on cantilevers when cantilevers are bent.

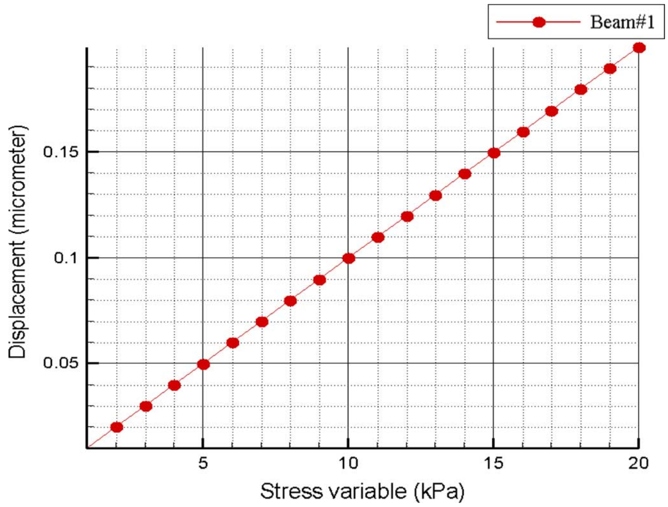

Fig. 7. Active area was varied stress 20 times of initial state.

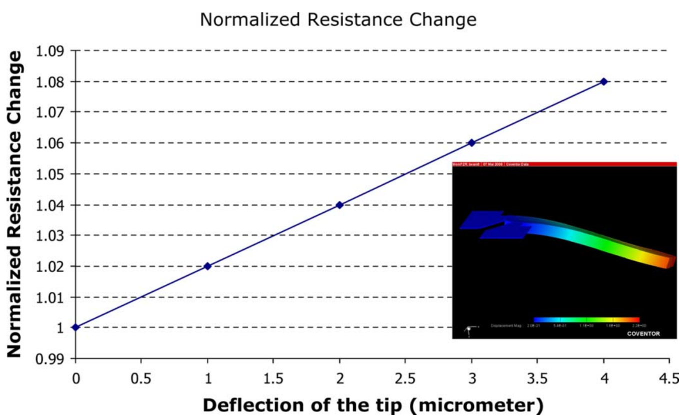

Fig. 8. Plot of normalized resistance change in relative with beam deflection.

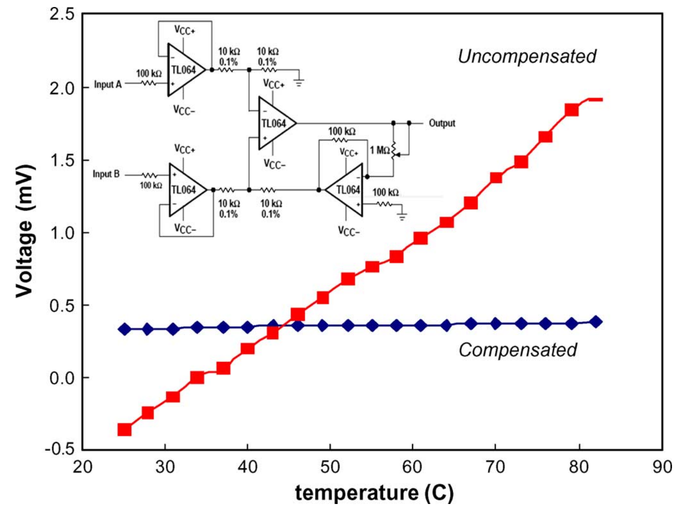

Fig. 9. Plot of temperature dependence of cantilever beam (a) noncompensate temperature and (b) compensate temperature.

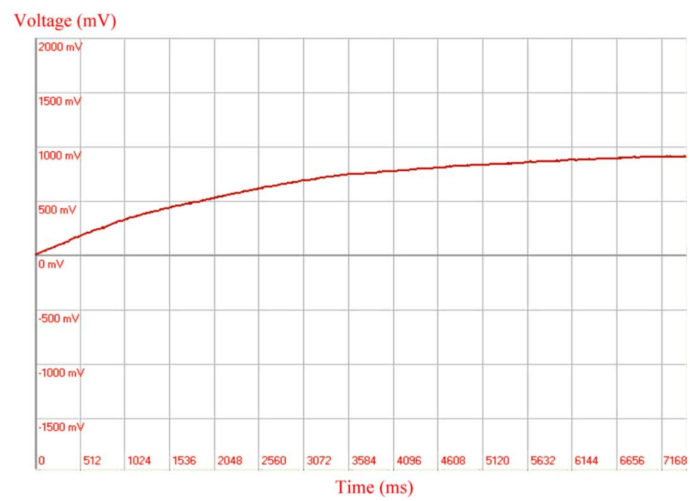

Fig. 10. Plot of output voltage (mV) versus measured time (millisecond) when the cantilever has being functionalized by SAM of aminoethanethiol.

deflection is proportional linearly with induced surface stress. The maximum displacement of 0.2 μm is observed when stress of 20 kPa was applied.

Piezoresistive Effect Simulation: A piezoresistive simulation is performed using Coventorware. First, the tip deflection was varied from 0 μm to 4 μm to verify the resistance change due to beam deflection. The simulation result of tip deflection shows in inset. The plot of normalized resistance change versus beam tip deflection was shown in Fig. 8.

IV. Device Testing and Characterization

Temperature Dependence Characterization: The micro-cantilever beam has been tested at room temperature to study the change of piezoresistive by various external temperatures. Using the Wheatstone bridge with built-in offset and consisting of four identically behaving resistors, the output signal will vary linearly with supply voltage. To characterize temperature dependence, the beam was placed on an adjustable PID hot plate in chamber and varied temperature up to 80°. The output voltage of uncompensated and compensated beam varied with temperature, as shown in Fig. 9. From this result, compensation by resisters on the substrate significantly reduces the temperature dependence. The amplification circuit was shown in an inset of Fig. 9.

Cantilever Characterization: The resistivity of the polysilicon piezoresistive wire is found to be about 1 kΩ. The series resistance as a Wheatstone bridge configuration was fabricated on sensor chip and made the cantilever very suitable for cantilever-based biochemical sensing [10]–[13]. The bridge has been fed by a constant current of 2 mA. The temperature effect has been calibrated to minimize the thermal drift effect. The voltage output has been amplified by an instrument amplifier and read by the oscilloscope. Fig. 10 shows the preliminary measurement result of output voltage versus measured time when the cantilever has being functionalized by a self-assembled monolayer (SAM) of aminoethanethiol for use as receptor molecules on the cantilever surface. These preliminary results show that the mass detection of the cantilever is capable in the range of 0.0058–0.0110 g.

V. Conclusion

This paper shows that induced surface stress from biochemical absorption will cause the beams to curl downward. The possibility of detecting the biochemical concentration with a

MEMS cantilever was demonstrated. Finite element modeling was used to simulate the mechanical behavior of a MEMS cantilever. The fabricated cantilever was characterized to verify the cantilever performance piezoresistive effect with varied absorbed mass and drifting temperature was experimentally measured. From the experimental result, compensation by resisters on the substrate significantly reduces the temperature dependence. The mass detection of 0.0058–0.0110 g is measured by the beam resistor series as a balanced Wheatstone bridge configuration. The voltage output of the bridge is directly proportional to the amount of bending in the cantilever. The resolution of this device is on the operation range of convention biochemical sensor applications. Therefore, the amount of biochemical molecules on the active surface can be verified.

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