Design Simulation and Finite Element Analysis of Piezoresistive Microcantilever for Human Stress Measurement

N. K. Madzhi, A. Ahmad, L. Y. Khuan and R. A. Rani

***Abstract*—This work focuses on the design and finite element analysis of a polysilicon-based piezoresistive microcantilever beam for application in human stress measurement. In principle, adsorption of saliva amylase on a functionalized surface of the microfabricated cantilever induces surface stress and consequently the bending of the cantilever beam. In this paper, the microcantilever beam is constructed and bending analysis is performed so that tip deflection of the beam can be predicted. The device is modeled using CoventorWare^{TM}. The structural variation in the piezoresistor design on the cantilever beam is investigated to increase the sensitivity of the microcantilever sensor. The stress distribution and the vertical displacement of the piezoresistive microcantilever designed are studied through simulation. The relative resistance changes in the piezoresistors as a function of the vertical displacements of the microcantilever beam is also investigated. It is found that the deflection displacement increases with length and thickness of piezoresistor. The relative resistance change in the piezoresistors is found to increase when the thickness decreases.**

***Index Terms*—Deflection, Piezoresistive Microcantilever ,Saliva Amylase, Surface Stress.**

I. Introduction

In modern societies, stress has become a major cause of health problems. With the complexities of life, stress inducing events have increasingly become common. The amount of stress one feels depends on both internal factors such as one's personality and mental state, as well as external factors such as the amount of stress at home or work. Stress in one can become so severe that it not only affects their mental state, but also the physiologic state. Hence, there is a need for monitoring systems to assist individuals in managing stress.

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Several biological indicators for stress response such as cortisol and catecholamines (norepinephrine (NE) and epinephrine (EP)) have been found to reliably indicate reactivity of physiologic systems, i.e. the hypothalamic-pituitary-adrenal (HPA) axis and the sympathetic-adrenal-medullary (SAM) system [1]. However, the concentration of cortisol in blood or saliva is very much lower than that of salivary alpha amylase, between 10-20% of protein content produced by the salivary gland. Moreover, the use of hormonal regulation such as cortisol suffers 20-30 minutes of delay [2] due to the slow change in serum NE. On the other hand, it has also been reported that salivary alpha amylase reacts within one to a few minutes in response to stress since its secretion is stimulated by direct innervations [2-7]. This is a markedly quicker and more sensitive response than that induced by hormonal regulation in respond to psychological stress than cortisol. Hence, salivary amylase activity is suggested as an excellent index for psychological stress[2].

Until a few years ago, the technology to analyze minute quantities of genetic materials and protein in saliva was not sufficient for home test kits. As such, the best body fluid to analyze disease risk would soon be saliva, and not blood. The advantage of saliva sampling is that it is non-invasive, easy for multiple sampling and stress free.

Alpha-amylase is one of the major salivary enzymes in humans which hydrolyse starch into maltose. It is secreted from the salivary glands in response to sympathetic stimuli. It has been shown that the secretion of amylase is dominated by the SAM system. Hence, the salivary amylase activity which indicates the working rate of amylase in saliva, can be used as an index to evaluate the neural activity of the SAM system [8]. It has been reported in 1996 that salivary alpha-amylase concentrations are predictive of plasma catecholamine levels, particularly NE under a variety of stressful conditions, and may be a more direct indication of catecholamine activity than changes in heart rate [9]. Quantitatively the alpha-amylase enzyme can be analyzed either by measuring the concentration or the enzyme activity. Several studies using Kraepelin Test and Roller Coaster ride as stressors have confirmed informative changes in salivary amylase activity caused by stress, even though salivary amylase is just a digestive enzyme [2-5].

Biosensor offers promising solution to sensitive, fast, repetitive and cheap measurements. It is comprised of two components i.e. a bioreceptor and a transducer. The bioreceptor is a biomolecule that recognizes the target analyte while the transducer converts the recognition event into a measurable signal. The uniqueness of biosensor is that the two components are integrated into one single sensor, enabling one to measure the target analyte without using reagents. In addition, the simplicity and speed of measurements that require no specialized laboratory skills are also attractive features of biosensor. In particular, micro-machined cantilever sensors are found ideal for biosensor application with their high degree of parallelization, which is desirable by the pharmaceutical industry for high-throughput screening. An increasing number of reports confirm the potential of microcantilever sensors for environmental application such as gas detection, mass effect and gas sensitivity [10] and biomedical application [11]. Yet, there has not been any work on the adsorption of saliva amylase on a functionalized surface of microcantilever as sensor for human stress measurement .

Our work focuses on the design of a piezoresistive poly-Si covered microcantilever beam for application in human stress measurement, which includes theoretical design, simulation study and finite element analysis. In this paper, the microcantilever beam is modeled using ConventorWare™, a commercial finite element analysis tool designed specifically for MicroElectroMechanical system (MEMS) applications. Bending analysis is performed to estimate tip deflection of the beam. The structural variation in the piezoresistor design on the cantilever beam is investigated to increase sensitivity of the microcantilever sensor since the forces generated is very small. The stress distribution and the vertical displacement of the piezoresistive microcantilever are studied through simulation. The relative resistance changes in the piezoresistors as a function of the vertical displacements of the microcantilever beam is also investigated.

II. PIEZORESISTIVE MICROCANTILEVER

Quoting a definition from [10], "A simple cantilever beam can be used as a sensor for biomedical, chemical and environmental application." With reference to Fig. 1, the changes in the surface properties of the microcantilever through absorption or adsorption of analytes to receptor molecules will influence its surface stress. This causes the deflection of microcantilever, which is proportional to the analyte concentration [12]. Usually the deflection is in micrometers and can be detected by several methods such as optical [13] and capacitive detection [14]. However, these methods require external devices for deflection measurements such as lasers, optical fibers or capacitors, which have the disadvantages of requiring alignment and calibration. However, these disadvantages can be prevented by integrating piezoresistive material. Resistivity of piezoresistive microcantilevers changes with the surface stress due to cantilever deflection upon adsorption or absorption of analytes. By arranging the piezoresistors in a balanced Wheastone bridge configuration, the voltage output of the bridge is directly proportional to the amount of cantilever bending, which reflects the amount of biochemical molecular adsorption on the top surface.

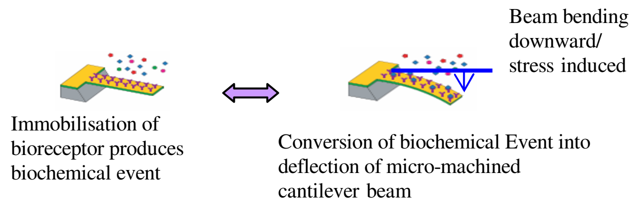

Fig. 1: A target biochemical species adsorbing on a functionalized surface of the cantilever beam

A. *Device Description*

The cantilever beam designed here comprises of three-structural layers as shown in Fig. 2.

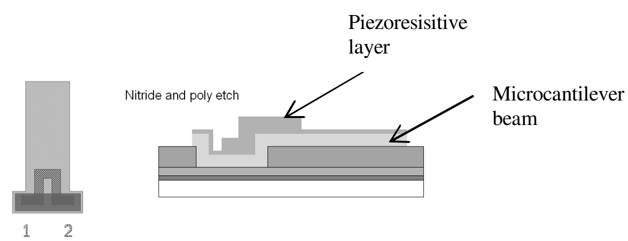

Fig. 2 Top and side view of piezoresistive microcantilever structural layers

The nitride layer fills the gap between polysilicon cantilever beam and top layer. The top polysilicon layer acts as a reactive or adsorption layer. The piezoresistor is approximately 195 micrometers long and 75 micrometers wide, while the top piezoresistive polysilicon layer thickness is varied as shown in Table 1.

TABLE 1
VARIATION OF PIEZORESISTOR LENGTH AND THICKNESS

PZR lengths (um)	PZR thickness (um)	PZR thickness (um)	PZR thickness (um)
110	0.3	0.6	1
140	0.3	0.6	1
170	0.3	0.6	1

PZR Width: 75 (um)

When the biochemical sample (biological molecules such as proteins or biological agents) is applied to the MEMS cantilever sensor, some of molecular sample binds with the top layer, causes the MEMS cantilever to deflect. Its deflection is found to be proportional to the biochemical concentration. When the top layer expands, the bending of the beam subjects the polysilicon layer to tensile or compressive stresses. Since polysilicon is piezoresistive, the stress can be determined by measuring the resistance of the polysilicon wire.

B. Device Modeling and Simulation

The bending of MEMS cantilever is due to mechanical force generated by the molecular adsorption. The adsorption-induced stress sensor has a sensitivity range based on the adsorbed mass. The adsorbed mass is proportional to the molecular size.

With reference to the simple solid 3-dimensional (3-D) model of piezoresistive microcantilever as shown in Fig. 3, the surface stress on the deflected microcantilever can be calculated using Stoney's equation,

$$\Delta\sigma_s = \frac{Eh^2}{6(1-v)r} \tag{1}$$

where $\Delta\sigma_s$ is the differential surface stress on the surface of the microcantilever, E is the Young's modulus, v is the Poisson's ration, r and h are the radius of curvature and thickness of microcantilever beam.

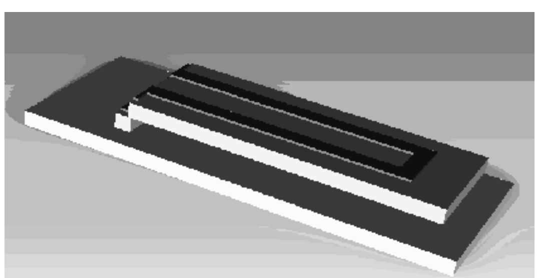

Fig. 3 A solid 3-D model of piezoresistive microcantilever

For the 3-D model of piezoresistive cantilever beam as shown in Fig.4, the relationship between the change in resistance and the existent stress field is more complex, as follows,

$$\{\Delta R\} = [\pi]\{\sigma\} \tag{2}$$

$$\{\Delta R\} = \{\Delta R_{xx}\ \ \Delta R_{yy}\ \ \Delta R_{zz}\ \ \Delta R_{xy}\ \ \Delta R_{xz}\ \ \Delta R_{yz}\}^T \tag{3}$$

$\{\Delta R\}$ represents the change in resistance of the piezoresistive element with a corresponding stress components as in (4).

$$\{\sigma\} = \{\sigma_{xx}\ \ \sigma_{yy}\ \ \sigma_{zz}\ \ \sigma_{xy}\ \ \sigma_{xz}\ \ \sigma_{yz}\}^T \tag{4}$$

Of the six independent stress components in the stress tensor $\{\sigma\}$; σ_{xx}, σ_{yy}, σ_{zz} are the normal stress components and σ_{xy}, σ_{xz}, σ_{yz} are the shearing stress components.

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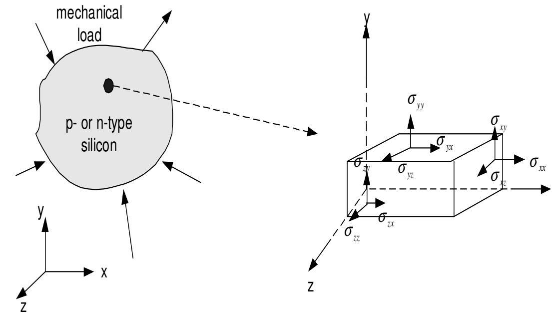

Fig. 4 A silicon piezoresistance subjected to a stress field

The vector $[\pi]$ in (2) is referred to as piezoresistive coefficient matrix which has the following form,

$$[\pi] = \begin{bmatrix} \pi_{11} & \pi_{12} & \pi_{12} & 0 & 0 & 0 \\ \pi_{12} & \pi_{11} & \pi_{12} & 0 & 0 & 0 \\ \pi_{12} & \pi_{12} & \pi_{11} & 0 & 0 & 0 \\ 0 & 0 & 0 & \pi_{44} & 0 & 0 \\ 0 & 0 & 0 & 0 & \pi_{44} & 0 \\ 0 & 0 & 0 & 0 & 0 & \pi_{44} \end{bmatrix} \tag{5}$$

By expanding the matrix equation in (3) with the appropriate piezoresistive coefficients in (5), the followings are obtained.

$$\begin{aligned} \Delta R_{xx} &= \pi_{11}\sigma_{xx} + \pi_{12}(\sigma_{yy} + \sigma_{zz}) \\ \Delta R_{yy} &= \pi_{11}\sigma_{yy} + \pi_{12}(\sigma_{xx} + \sigma_{zz}) \\ \Delta R_{zz} &= \pi_{11}\sigma_{zz} + \pi_{12}(\sigma_{xx} + \sigma_{yy}) \\ \Delta R_{xy} &= \pi_{44}\sigma_{xy} \\ \Delta R_{xz} &= \pi_{44}\sigma_{xz} \\ \Delta R_{yz} &= \pi_{44}\sigma_{yz} \end{aligned} \tag{6}$$

For the application of a piezoresistor microcantilever in Fig. 3, the relationship between the surface stress and the relative change in resistance $\Delta R/R$ for a piezoresistor is therefore given by the following[11],

$$\frac{\Delta R}{R} = -K\left(\frac{1}{E_1h_1 + E_2h_2} + \frac{Z_T^2}{E_1h_1\left(\left(Z_T - (h_1+h_2) + \frac{h_1}{2}\right)^2 + \frac{1}{3}\left(\frac{h_1}{2}\right)^2\right) + E_2h_2\left(\left(Z_T - (h_1+h_2) + \frac{h_2}{2}\right)^2\right)}\right) x\Delta\sigma_s \tag{7}$$

where E_1,h_1 are the Young's modulus and thickness of the polysilicon cantilever beam while E_2, h_2 are the Young's modulus and thickness of the piezoresistor; Z_T is the distance from neutral axis to top of the cantilever beam containing piezoresistor; K is the gauge factor of piezoresistor but only applies when the length of the Si piezoresistor is the same as the cantilever length.

III. RESULTS AND DISCUSSION

The solid 3-D model in Fig. 3 was created using thin film material property data as shown in Table 2. The finite element analysis (FEA) technique was used to solve the differential equations of each physical domain by discretizing the 3-D model into a mesh that consists of a number of elements with a specified number of nodes. Having generated the mesh model, the variable stress simulation at the same conditions of analyte concentration and the analyte capturing area was analyzed.

Table 2 Thin film material properties as used in the solid 3-D model

Thin Film Properties		Polysilicon
Young Modulus (MPa)		1.6000e+005
Poisson ratio		2.2000e-001
Density (kg/µm^3)		2.32000e-015
TCE Integral Form (1/k)		2.8e000e-006
Thermal Conductivity (pW/umK)		1.48000e+008
Specific Heat (pJ/jgK)		7.12000e+014
Piezoresistive Coefficient	Pi_11	1.000e-009
	Pi_12	1.000e-011
	Pi_44	1.000e-004

In order to study the beam deflection due to the molecule mass adsorption on the 3-D solid model, the stress for the capturing area was varied from 2 Pa to 10 Pa as shown in Fig.5, for different structural dimension of piezoresistive microcantilever.

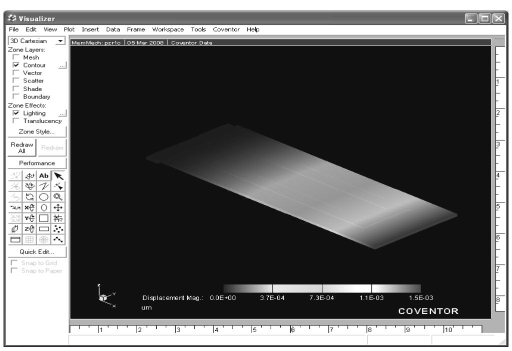

Fig. 5 PZR Microcantilever tip deflection at capturing area with stress varying from 2 Pa to 10 Pa

First, the variation in PZR cantilever beam tip deflection (in micrometer) with stress of capturing area for different PZR cantilever length is investigated. From Fig. 6a, it can be observed that when a surface stress is applied onto a cantilever, the deflection amplitude increases from the support end to the free end of the cantilever. By comparing the beam tip deflection behavior at different sensor lengths, it can be observed that the longer the piezoresistor, the larger is the deflection (see Figure 5a).

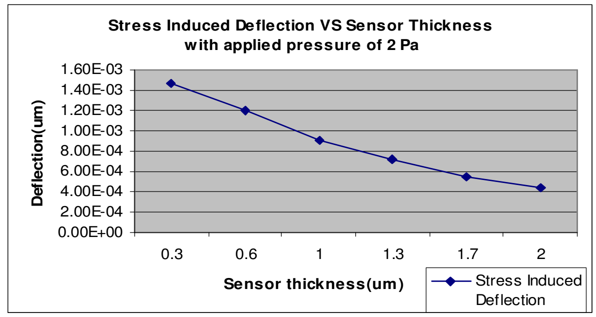

Fig. 6a Variation in PZR cantilever tip deflection with different PZR length

Then, the variation in PZR cantilever beam tip deflection (in micrometer) with stress of capturing area for different PZR cantilever thicknesses is studied. Thickness is a crucial factor that affects detection sensitivity. From Fig. 6b, it can be observed from the PZR cantilever beam tip deflection behavior at different sensor thickness the thinner the piezoresistor, the larger is the deflection.

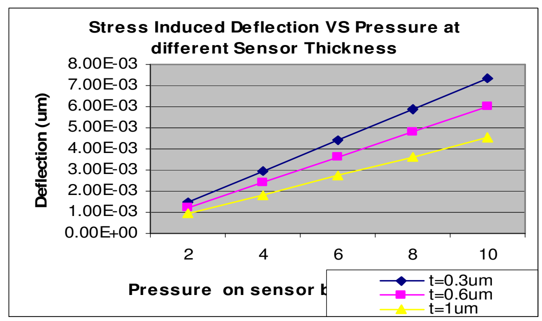

Fig. 6b: PZR cantilever tip deflection with different PZR thickness

Finite element analysis shows that cantilever deflection displacement increases when the thickness of piezoresistor decreases from 2 to 0.3 µm when a surface stress of 2 Pascal (Pa) is applied onto the surface of the cantilever and piezoresistor (see Fig. 6c). A surface stress of 2 Pa in particular is chosen here because this is the value of surface stress typically used in many microcantilever chemical or biosensors.

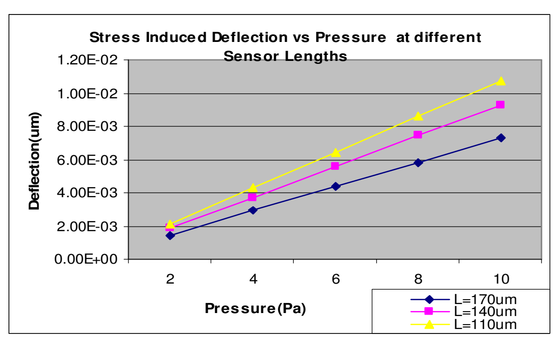

Fig. 6(c) PZR cantilever deflection with applied surface stress of 2 Pa

Results from the above simulation results show that the beam deflection is linearly proportional to the molecular mass adsorption. Linear change in all the parameters is assumed within the boundary conditions of finite element analysis simulation.

In examining the resistivity and output voltage of the polysilicon piezoresistor, the relative change in resistance, $\Delta R/R$, is found to increase when the thickness decreases (see Fig. 6d), in compliance with (1.7). This is due to the larger displacement in bending when the polysilicon piezoresistor is thinner.

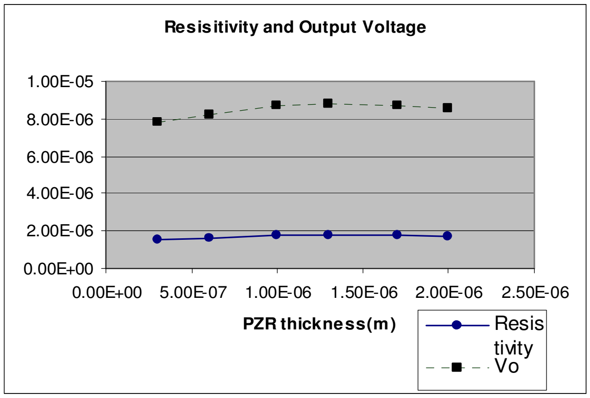

Fig.6d: Resistivity changes and output voltage of polysilicon piezoresistor

IV. FABRICATION PROCESS

The surface micromachining process using 0.5µm CMOS technology for the fabrication of a cantilever beam with a piezoresistive element is proposed. The workflow which consists of 6 steps, is illustrated as in Fig. 7.

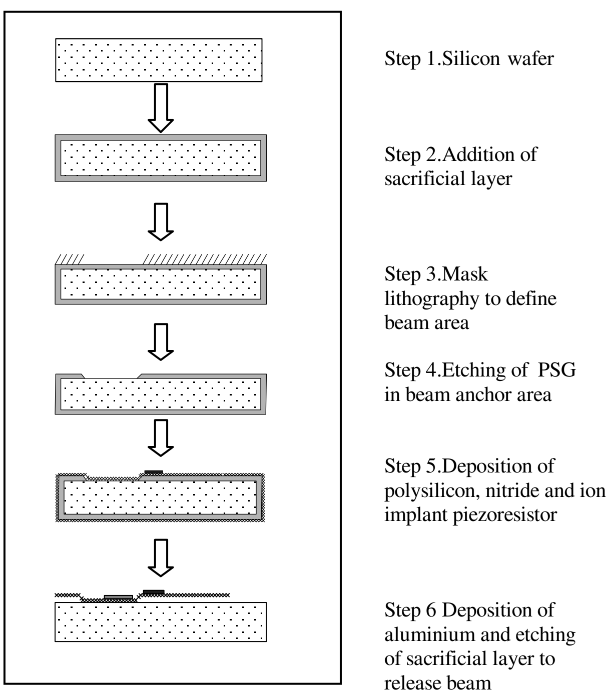

Fig. 7 Workflow of fabrication process

Starting with a silicon wafer, a 0.2 µm thickness layer of silicon nitride and 3µm thickness layer of boron phosphor silicate glass (BPSG) is deposited (step 1-2). After defining and etching the beam and anchor area, a polysilicon layer is deposited with 195µm x 75 µm u-shape resistor pattern and blanket implanted to achieve a resistor value of 1.2kΩ as in (step 3-5). Then, the electrode pad is patterned and deposited with aluminum. Finally the cantilever beam is to be released by wet etching procedure.

VI. CONCLUSIONS

The performance of polysilicon-based piezoresistor microcantilever is analysed with finite element technique. The fractional change in resistance of the microcantilever denotes sensitivity of the piezoresistive microcantilever. The microcantilever thickness and length affects the resistivity change. Under applied stress, it is found that the deflection displacement increases with length and thickness of piezoresistor. The relative resistance change in the piezoresistors is observed to increase when the thickness decreases from 1µm to 0.3µm, which is the proposed range of thickness for fabrication process later. Results here provide the basis to fabrication of the polysilicon-based piezoresistive microcantilevers.

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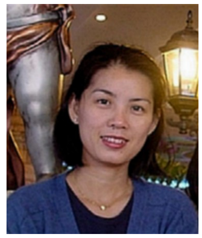

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