Piezoresistive Microcantilevers for Biomedical Applications

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Abstract

The present work discusses the application of piezoresistive microcantilevers for two important biomedical applications, namely, biosensor and hotplate. Owing to different temperature and its distribution characteristics, the design principles for piezoresistive micro-cantilevers for biosensor and hotplate applications are different. The self-heating phenomenon which is considered a source of inaccuracy in biosensors is vital in hotplates. This study discusses the design requirements in view of self-heating associated with both applications. Numerical analysis is used to study the effect of applied voltage, operating medium, piezoresistor element size and substrate material on temperatures and deflections produced in the cantilevers. Results show that cantilevers with long element and silicon dioxide substrate are more suitable for biosensors, whereas, those with long element and silicon substrate are better for hotplates. In addition, the silicon dioxide cantilevers with short piezoresistor are good choice for actuators.

Keywords - Microcantilevers, Biosensor, Thermal cell lysis, Hotplate, Self-heating, Piezoresistivity.

Introduction

Piezoresistive sensors are among the earliest micro-machined silicon devices. A comprehensive review on theory and applications of piezoresistor can be found in Barlian *et al.*, (2009). Such sensors have been successfully used in many novel and diverse applications, including strain sensor, accelerometer, cantilever force sensor, pressure sensor, inertial sensor and flow sensor. The sensors measure the change in electrical resistance of its piezoresistive element, i.e., piezoresistor, to determine the external source. Recent studies have showed that piezoresistive cantilever biosensors are attractive option to optical cantilever biosensors because of their low power, compact and robust features (Zhou *et al.*, 2009). However, the sensitivity and resolution of these sensors is less than the optical one. Low performance of piezoresistive cantilever biosensors can be attributed to low gauge factor of the piezoresistor and thermal drifting caused by self-heating in the cantilevers (Thaysen *et al.*, 2000). The gauge factor can be improved by using a piezoresistor made of single crystalline silicon. Minimizing the adverse effects of thermal drifting is, however, more challenging.

Thermal drifting is a major source of inaccuracy in piezoresistive microcantilever biosensors. The heat generated by the electrical current passing through the piezoresistor, i.e., self-heating, increases the temperature of the cantilever and produces thermal drifting in form of bimetallic bending and changes the electrical and piezoresistance properties of the cantilever. Self-heating action in piezoresistive microcantilever can be adapted to useful means such as microhotplates or microheaters. In these applications, the cantilever is supplied with electrical current to produce temperatures required to provide controlled microenvironment for different processes to occur or to actuate/sense the cantilever tip movement by means of bimetallic bending. Using thermal actuation and sensing, Chui *et al*. (1998) and Binnig *et al*. (1999) reported piezoresistive microcantilevers for ultra-high density atomic force microscopy data storage. Lee and King, (2007) studied thermal characteristics of microcantilever hotplates of different designs. Privorotskaya *et al*. (2010) successfully used self-heating and self-sensing characteristics of piezoresistive microcantilever to study thermal cell lysis. Thus, we see the design principles for biosensors and hotplates are fundamentally different. Biosensors exploit the piezoresistive property of the cantilever, whereas, the hotplates use the electrical resistance property. Further, the self-heating which is undesirable in cantilever biosensors is essential in hotplates.

Low temperature rise and high temperature uniformity are the main requirements of cantilevers for use in biosensors and hotplates, respectively. The temperature rise and its distribution strongly depend on the voltage applied, the thermo-electrical properties of the cantilever and piezoresistor material, and the environment the cantilever is operated (Ansari and Cho, 2010). In addition, the shape and size of the cantilever and the piezoresistor are also critical. The temperature rise is directly proportional to the square of the voltage applied and inversely to the thermal conductivity of the cantilever and electrical resistance of the piezoresistor, respectively. Further, the temperatures increases with decrease in cantilever size because of its high volumetric rate of heat generation. The temperature distribution along the cantilever length is mostly one-dimensional and increases with the square of the distance from cantilever base.

The present study investigates the temperature behaviour of piezoresistive microcantilevers for biosensor and hotplate. The rectangular cantilevers are made of silicon and silicon dioxide substrate with a u-shaped piezoresistor made of p-doped silicon inside. The cantilevers are assumed to be used in air and liquid environments. A finite element analysis software ANSYS is used to study effect of applied voltage on temperature and deflection behaviour of cantilevers having different thermo-electrical properties and different piezoresistor size is studied. Finally, the design principles to select the most suitable cantilever design for biosensors and hotplates are discussed.

Theory and Modeling

Piezoresistivity is the change in bulk electrical resistivity of a material caused by an applied mechanical stress. The resistivity can increase or decrease depending on the material type and the load condition. Many materials exhibit piezoresistivity when stressed, but the effect is most prominent in semiconductors. The piezoresistivity of semiconductors is more than an order of magnitude higher than that of metal. In semiconductors, piezoresistivity strongly depends on dopant type and its concentration. Electrical resistivity is the property of a material to resist the flow of electrical current through it. Both piezoresistivity and electrical resistivity are temperature dependent. For semiconductors, piezoresistivity decreases with temperature but the electrical resistivity increases with it. Microcantilever biosensors exploit the surface stress-induced deflections to assay the analyte. When the target molecules attach onto the functionalized top surface of the cantilever, the surface stress distribution on its surface is changed. In case of piezoresistive microcantilever biosensors, the stress variation also changes the electrical resistance of the cantilever as $\Delta R/R = \pi\sigma$, where π is piezoresistivity coefficient and σ is applied stress. The measurement of the resistance change gives information on type and concentration of the analyte causing the stress. In contrast, microcantilever hotplates use the electrical resistance, and not its variation, for heating the cantilever to the desired temperature.

Figure 1 shows the schematic design of the microcantilever containing a u-shaped element. A thin film of gold is deposited on top surface of the cantilever. This design can be used in both biosensor and hotplate applications by changing the element type. The element is piezoresistor in biosensors but electrical resistor in hotplates. This is achieved by doping the silicon element with proper dopant and its concentration depending on the application. In this work, both piezoresistor and electrical resistor are made of doped silicon and the cantilever substrate is made of silicon and silicon dioxide, respectively. Cantilevers in two configurations were used in this study. In Model#1, the element and substrate are made of silicon, whereas, in Model#2 the element is silicon but the substrate is silicon dioxide. For both models, the element length (l) was changed to 90 and 180 µm but the width (b) was fixed at 45 µm. The cantilever size was fixed 200×100×0.75µm. The thicknesses of the layers from the top are 0.05, 0.1, 0.1 and 0.5 µm. The applied voltage was 10 V and 20 V. The cantilevers were assumed to be operated in air and water with heat transfer coefficients 200 and 1000 W/m2 °C, respectively. The cantilever base temperature (Tb) and the ambient fluid temperature (Tf) were fixed at 25°C.

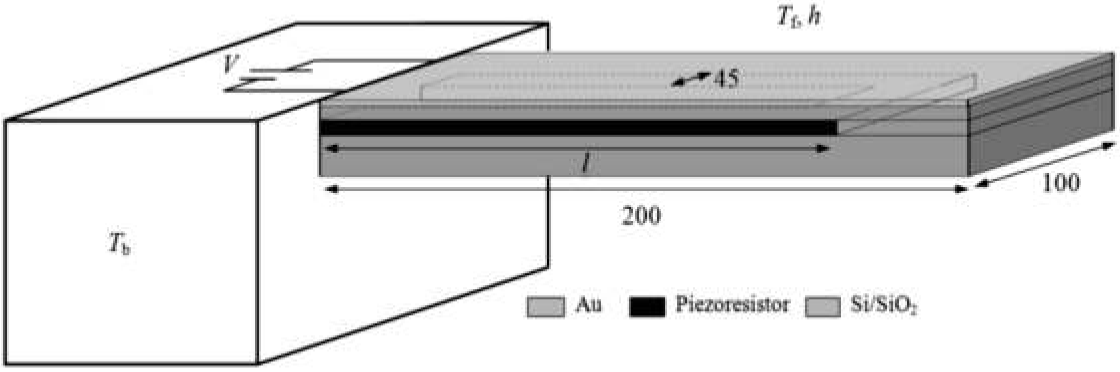


Figure 1: Schematic of a microcantilever with u-shaped element (µm unit).

A finite element analysis software ANSYS Multiphysics was used to study the temperature and deflection in cantilevers produced by self-heating. The finite element model of the cantilever was meshed by three-dimensional coupled field 8-node scalar SOLID5 elements and was solved under steady-state condition. The model did not include the base of the cantilever. These elements have capability to solve coupled-field problems involving mechanical, thermal, electrical and piezoresistive effects. About 100,000 elements were used in each analysis. Table 1 lists the thermo-mechanical properties of the cantilever model shown in Fig. 1.

Table 1: Thermo-mechanical properties of the cantilever materials.

Property	Au	SiO2	Si
Thermal conductivity ($\times 10^6$ pW/µm°C)	317	1.38	150
Thermal expansion coefficient ($\times 10^{-6}$/°C)	14.2	0.5	2.8
Specific heat ($\times 10^{12}$ pJ/kg°C)	129	745	712

Elastic modulus ($\times 10^3$ MPa)	80	70	160
Poisson's ratio	0.42	0.20	0.23
Mass density ($\times 10^{-15}$ kg/μm3)	19.3	2.22	2.32
Electrical resistivity (Tohm−μm)	----	----	10^{-9}

Results and Discussions

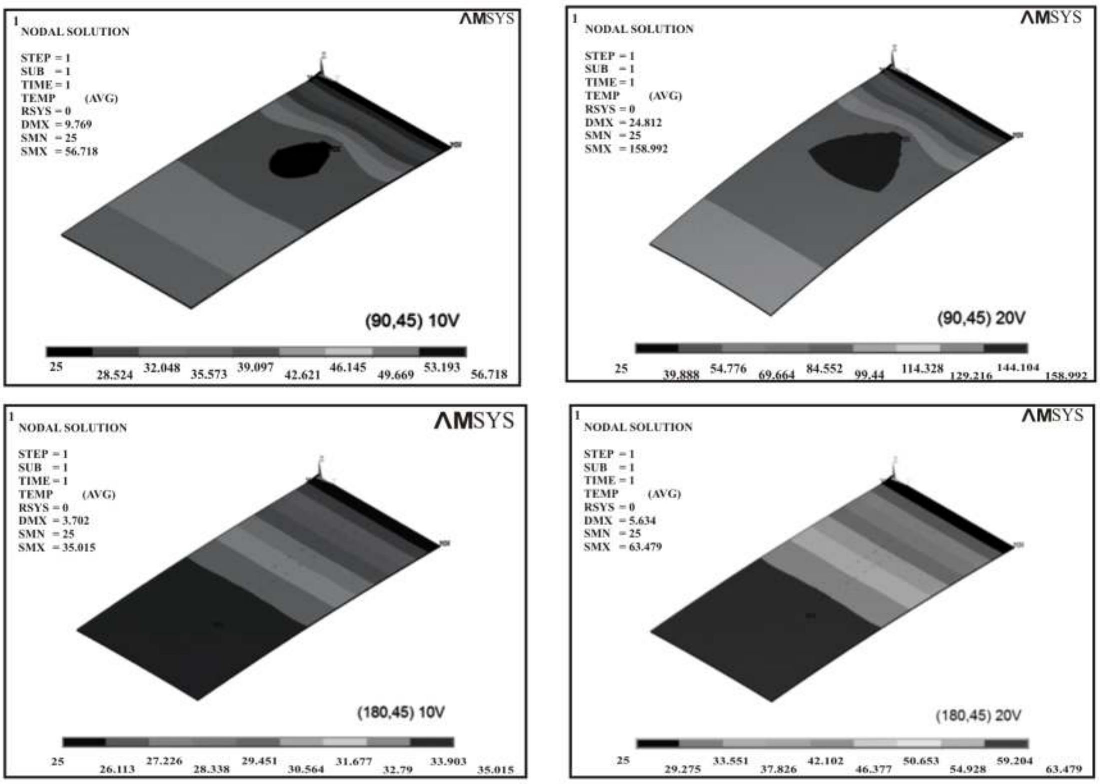
Table 2 presents the maximum temperature and thermal deflection results for Model#1 and Model#2 cantilevers of different (l,b) element size operated in air and water at applied voltages 10 V and 20 V, respectively. Figure 2 and 3 show the temperature distribution in these cantilevers. In general, the qualitative behaviour of temperature and deflection are similar. It can be observed in the table that the maximum temperatures produced in the cantilever depend on voltage applied, operating environment, element length, and cantilever model. A similar behaviour can also be observed for thermal deflections. The deflections are closely related to temperature because of strong thermo-mechanical coupling in form of bimetallic bending deflections.

Results indicate the temperatures and deflections increase with applied voltage. This is understandable because high voltages increase the amount of current passing through the current-carrying element. Since self-heating is directly proportional to the square of electric current, increase in current results in much higher heating. The increase in temperatures increases the cantilever deflections because of bimetallic bending. Thus, high temperatures and high deflections are very closely related.

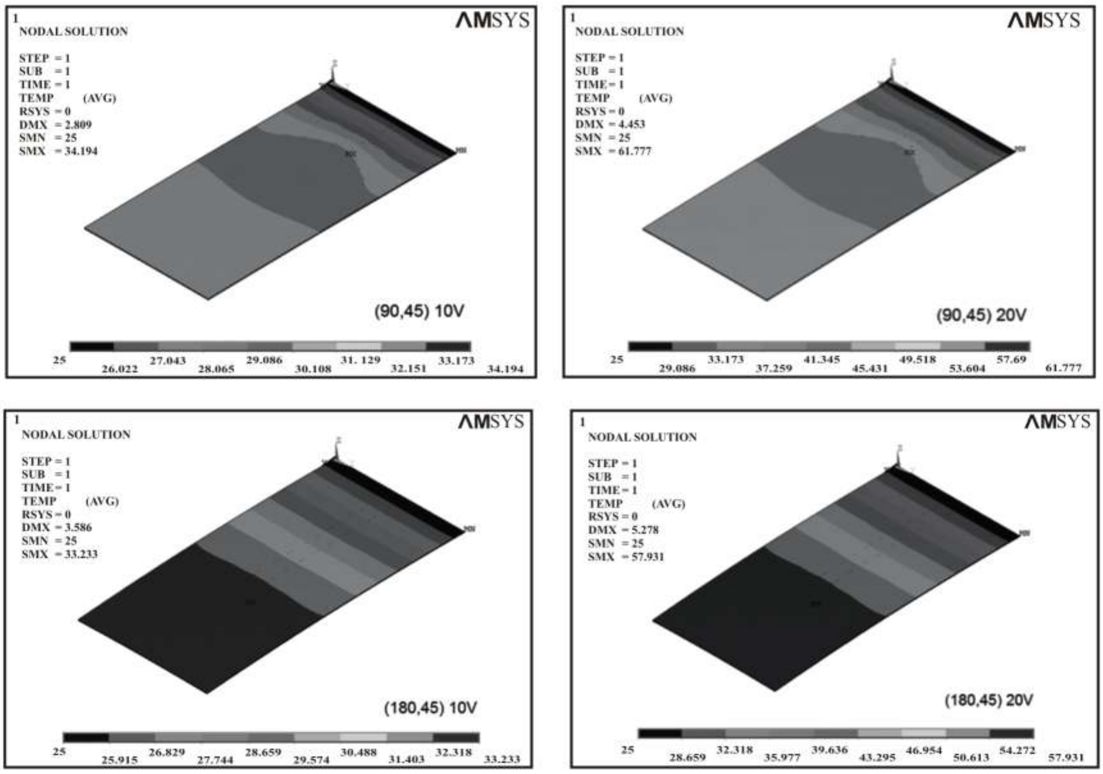
Table 2: Maximum temperature and deflection results for cantilevers operated in air and water.

Cantilever		Air				Water			
		10 V		20 V		10 V		20 V	
		Temp. (°C)	Defl.(μm)	Temp. (°C)	Defl.(μm)	Temp.(°C)	Defl.(μm)	Temp.(°C)	Defl.(μm)
Model#1	(90,45)	56.72	9.77	158.99	24.81	34.19	2.81	61.78	4.45
	(180,45)	35.01	3.70	63.48	5.63	33.23	3.59	57.93	5.28
Model#2	(90,45)	59.09	10.37	161.36	25.41	52.72	8.83	135.89	19.26
	(180,45)	55.72	11.76	147.97	26.16	43.68	9.96	99.72	18.99

Table 2 suggests that temperatures and deflections depend on operating environment as well. Results show that cantilevers having same configuration and same applied voltage have lower temperatures and deflections when operated in water than in air. The heat generation in these cantilevers will be same because their configuration and applied voltage are same. The low temperatures can be attributed to the relatively higher heat loss from the cantilever to the ambient via convection heat transfer (Ansari and Cho, 2012). The convective heat transfer coefficient for water is assumed five times of air, i.e., 1000 versus 200 W/m2 °C. The higher the heat transfer coefficient, the greater the heat will be carried away from the cantilever. Thus, the low temperatures observed for cantilevers operated in water are mainly due to high heat loss from these. Since the temperatures are lower, the deflections are also reduced.



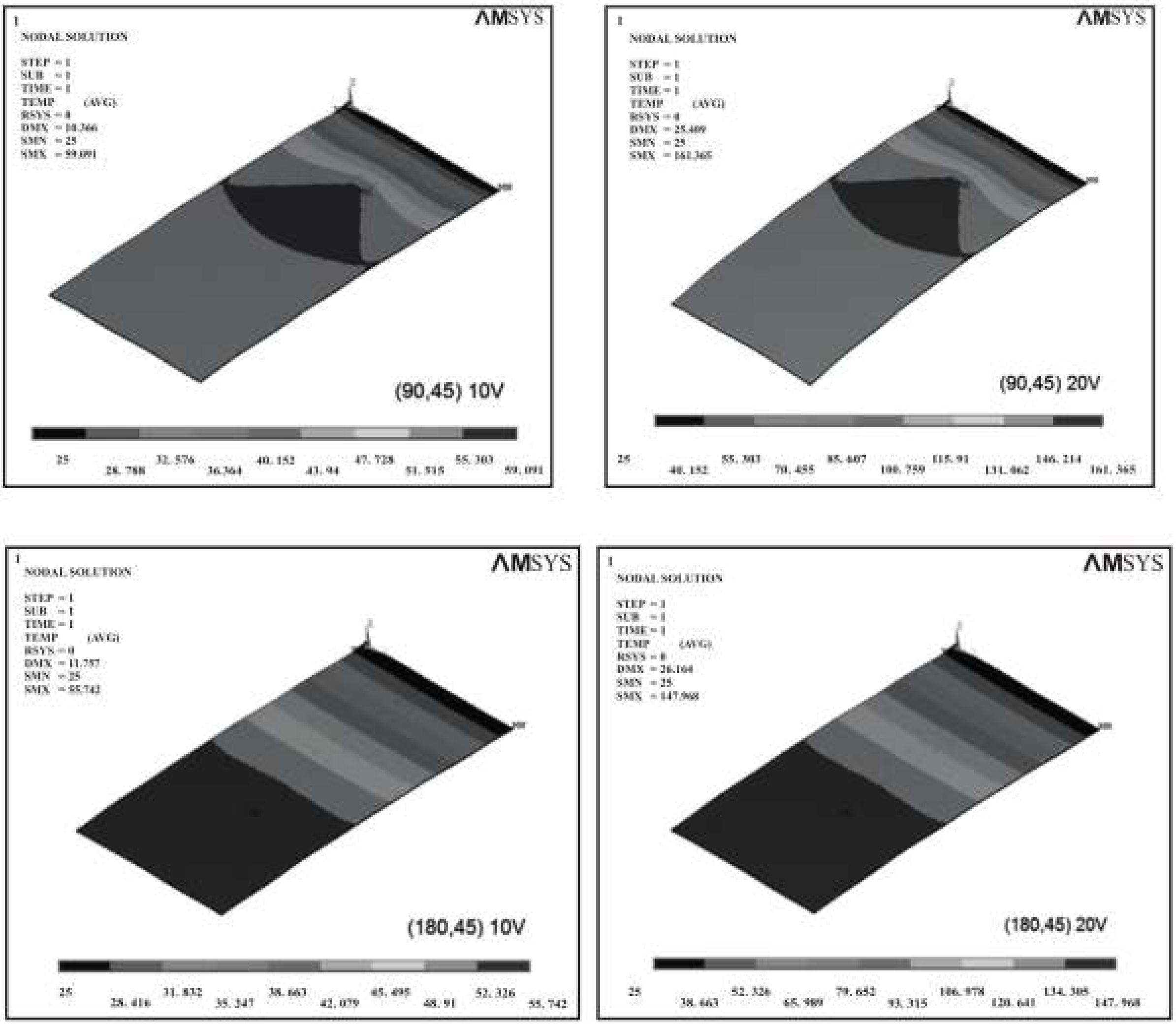
(a) air



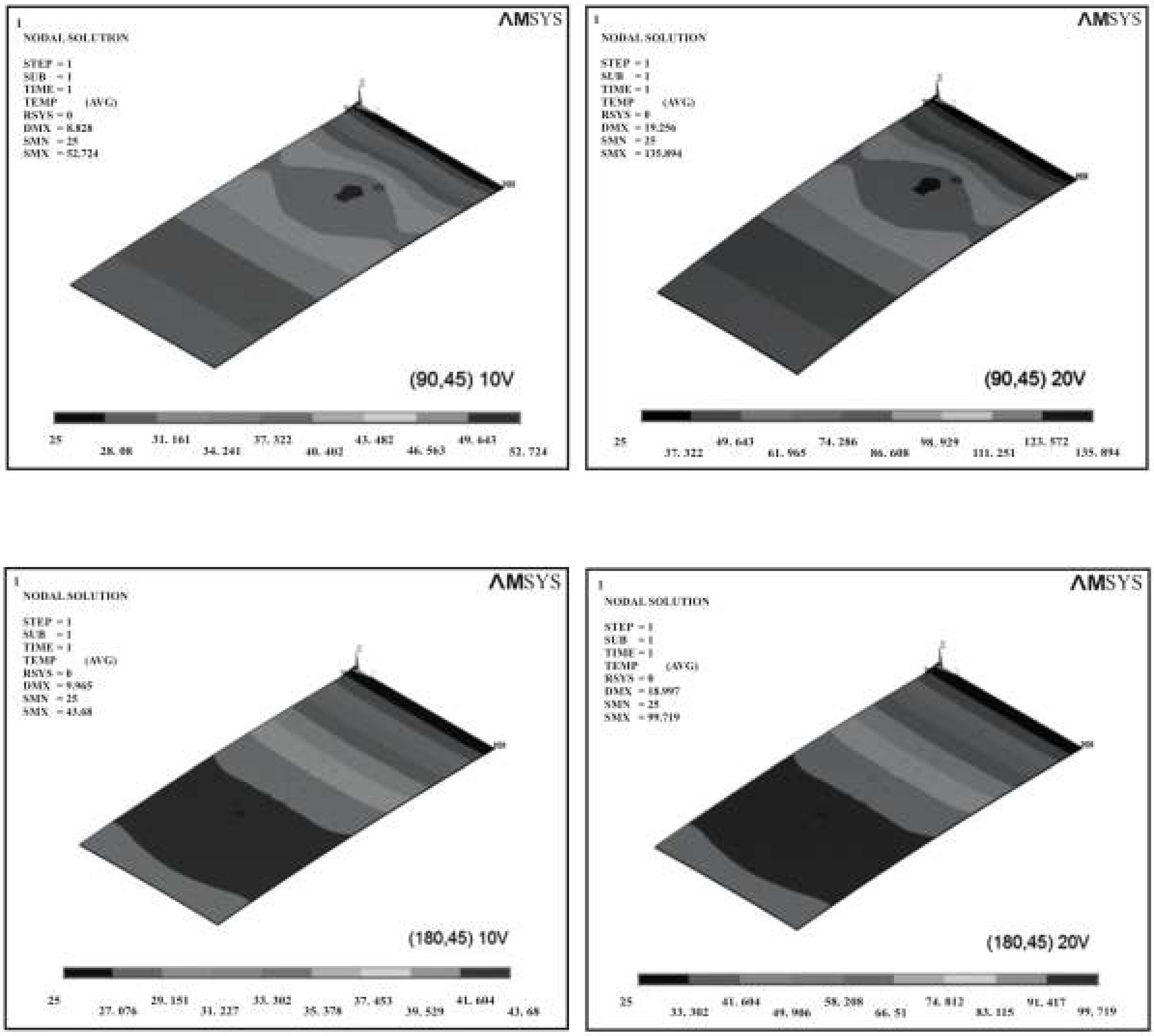
(b) water

Figure 2: Temperature distribution in Model#1 cantilevers operated in different environments.

The temperatures and deflections in Table 2 are found to decrease with increase in element length. The total current-carrying length in (90,45) and (180,45) are 190 μm and 370 μm, respectively. The explanation to this observation is not straightforward. Since electrical resistance is directly proportional to the total length of the conductor, the increase in element length increases its resistance. This will result in larger self-heating and therefore higher temperature rise in (180,45) cantilevers. But, the results in Table 2 show the inverse effect. This anomaly can be explained by the different temperature distribution patterns observed in (90,45) and (180,45) cantilevers, shown in Figure 2 and 3. The temperature distribution is nearly one-dimensional for (180,45) cantilevers and the maximum temperature zones are much bigger than those in (90,45). In other words, since the maximum temperatures in (180,45) cantilevers are spread over a large area, their overall values will be lower.



(a) air



(b) water

Figure 3: Temperature distribution in Model#2 cantilevers operated in different environments.

A comparison between temperatures and deflections results for Model#1 and Model#2 shows that under same conditions Model#1 have lower temperature than Model#2. This can be explained by the different thermal properties of the two models, see Table 1. Though the amount of self-heating is same in both models, the temperatures are different because of their different substrate materials. Model#1 cantilevers have silicon substrate, whereas, Model#2 have silicon dioxide. The temperature increase in cantilevers due to self-heating is inversely proportional to their thermal conductivity (Ansari and Cho, 2010). The thermal conductivity of silicon is more than 100 times silicon dioxide. Since the conductivity of Model#1 is much higher than Model#2, for same amount of self-heating the temperature increase will be lower in Model#1. The higher the thermal conductivity, the greater the uniformity in temperature distribution will be.

The primary function of a piezoresistive biosensor is to measure the electrical resistance change across the piezoresistor due to analyte-receptor binding. The higher the resistance change, the better the sensitivity and resolution of the biosensor will be. Piezoresistors made of doped single crystalline silicon have better gauge factor. Therefore, this is a common practice to fabricate cantilevers made of silicon piezoresistor with a substrate normally made of silicon or silicon dioxide. This study considered both cases in form of Model#1 and Model#2. The second model is more popular in biosensor applications because of it has low flexural stiffness than the first. The elastic modulus of silicon is more than twice the silicon dioxide, see Table 1. Low stiffness allows the cantilever to bend more and induce larger stress in the piezoresistor. High stresses will produce high resistance change in the cantilever. Thus, Model#2 cantilevers are more popular in biosensor applications. In case of biosensors, rise in temperature should be minimal to avoid thermal drifting. This is normally achieved by applying low bias voltages to the cantilever. Based on temperature results shown in Table 2, we can conclude that Model#1 cantilevers of size (180,45) are better in reducing temperature increase in cantilevers.

The purpose of a hotplate is to create a temperature-controlled environment for chemical or biological reactions to occur. Therefore, creating a large, uniform temperature area on the cantilever surface is the fundamental requirement. The deflections are irrelevant. This can be easily observed in Figure 2 and 3 that (90,45) cantilevers have discontinuous temperature profile. The maximum temperature areas are highly localized and are observed near the element length tip. Therefore, we can conclude that (90,45) cantilevers are unsuitable for hotplate wherein a large, uniform temperature field is required. In contrast, nearly all (180,45) cantilevers have large, uniform temperature field under different operating conditions. In these cantilevers, the temperatures increase continuously from the base temperature value of 25 °C to the maximum ones normally achieved at their tips. Due to absence of heating element and higher heat losses, the cantilever tip temperatures are lower for Model#2 operated in water. Thus, we can conclude that (180,45) cantilevers are better choice for hotplates. The selection between Model#1 and Model#1 depends on the temperature required.

The high deflection characteristics shown by (90,45) cantilevers indicate their potential application as actuators. Model#2 cantilevers show better actuation characteristics than Model#1. Thus, we can conclude that Model#2 cantilevers of size (90,45) are best suited for actuator application.

Conclusions

This study investigated piezoresistive microcantilevers for ascertaining their temperature behavior under different operating and material conditions. Cantilevers made of silicon piezoresistor and silicon substrate (i.e., Model#1) and silicon piezoresistor and silicon dioxide substrate (i.e., Model#2) were studied under different applied voltage, element lengths, and operating environments of air and water. Results found that temperatures increase with applied voltage but decrease in mediums with high convective transfer coefficient. It was also found that temperatures decrease with increase in piezoresistor length from (90,45) to (180,45). Model#2 cantilevers produced higher temperatures than Model#1. Finally, we can conclude that (i) element size (180,45) is most suitable for both biosensor and hotplate applications, (ii) Model#1 is more suitable for hotplate, (iii) Model#2 is more suitable for biosensors, and (iv) Model#2 with element size (90,45) size are more suitable for actuator applications.

References

Ansari, M. Z., Cho, C. 2010. An analytical model of joule heating in piezoresistive microcantilevers. *Sensors*, 10, 9668–9686.

Ansari, M. Z., Cho, C. 2012. A conduction-convection model for self-heating in piezoresistive microcantilever biosensors. *Sensors and Actuators A,* 175, 19–27.

Barlian, A., Park, W.T., Mallon Jr., J. R., Rastegar, A. J., Pruitt, B. L. 2009. Review: Semiconductor piezoresistance for Microsystems. *Proceedings of the IEEE,* 97, 3, 513–552.

Binnig, G., Despont, M., Drechsler, U., Haberle, W., Lutwyche, M., Vettiger, P., Mamin, H.J., Chui, B.W., Kenny, T.W. 1999. Ultrahigh-density atomic force microscopy data storage with erase capability. *Applied Physics Letters,* 74, 1329.

Chui, W., Stowe, T.D., Ju, Y.S., Goodson, K.E., Kenny, T.W., Mamin, H.J., Terris, B.D., Ried, R.P. 1998. Low-stiffness silicon cantilever with integrated heaters and piezoresistive sensors for highdensity AFM thermomechanical data storage. *Journal of Microelectromechanical Systems*, 7, 69.

Lee, J., King, W.P. 2007. Microcantilever hotplates: design, fabrication, and characterization. *Sensors and Actuators A,* 136, 291–298.

Privorotskaya, N., Liu, Y., Lee, J., Zeng, H., Carlisle, J.A., Radadia, A., Millet, L., Bashir , R., W.P., King. 2010. Rapid thermal lysis of cells using silicon–diamond microcantilever heaters. *Lab on Chip,* 10,1135–1141.

Thaysen, J., Boisen, A., Hansen, O., Bouwstra, S. 2000. Atomic force microscopy probe with piezoresistive read-out and a highly symmetrical Wheatstone bridge arrangement. *Sensors and Actuators,* 83, 47–53.

Zhou, Y., Wang, Z., Wang, C., Ruan, W., Liu, L. 2009. Design, fabrication and characterization of a two-step released silicon dioxide piezoresistive microcantilever immunosensor, *Journal of Micromechanics and Microengineering,* 19, 065026.

