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Simulation of MEMS Based Flexible Flow Sensor for Biomedical Application

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Abstract: Arterial disease, especially Coronary Artery Disease (CAD) is one of the leading causes of premature morbidity and mortality. During the flow, blood not only interacts with vessel wall mechanically but also chemically which modulates the plaque formation in blood vessel thus leading to coronary artery diseases. Here we propose to simulate a MEMS based flexible flow sensor based on anemometer principle which is designed to integrate at a catheter tip. This sensor will be useful for on-line monitoring of body fluid during catheter based operation. Simulation work has been carried out in COMSOL 4.1 which includes optimal resistive heater design and identification of suitable substrate material for realization of the heater on a flexible polymer followed by its steady as well as transient CFD analysis.

Keywords: Coronary Artery Disease, MEMS, flexible polymer, flow sensor.

1. Introduction

Development of micro-fabrication technology for realization of MEMS (Micro-Electro-Mechanical Systems) has been possible by exploiting the well advanced silicon integrated circuit technology [1]. Efforts are being made to realize MEMS devices for its use in biomedical sensor applications for low cost clinical healthcare [2,3]. In medical instrumentations, Bio-MEMS devices such as pressure sensors, temperature sensors, flow sensors, etc and actuators like micromotor, microvalve and microfluidic devices, etc are utilized in implantable systems and in drug delivery at point of care [4].

Coronary Artery Disease (CAD) is one of the leading causes of premature morbidity and mortality due to formation of plaque. Onset of plaque formation in blood vessel starts due to mechanical as well as chemical interaction of the blood particles with the vessel wall. In the present study COMSOL Multiphysics is used to simulate a MEMS based flexible [3] flow sensor proposed to be mounted near a catheter tip to measure the in situ blood flow velocity. Present work attempts to model the thin nichrome heater film on the flexible elastomer wrapped around a catheter surface using the shell model approximation for thin geometries. Optimization of heater design and identification of suitable substrate material along with steady state as well as transient CFD analysis of the sensor inserted inside the fluid vessel has also been performed in order to study the changes in fluid velocity, pressure and temperature distribution.

2. Theory

The proposed flow sensor is based on the principle of thermal anemometer, an indirect technique for measurement of fluid velocity. In this method, fluid velocity is determined by the amount of heat dissipated in the fluid from the electrically heated sensing element exposed in the fluid medium. The temperature and hence the resistance of the sensing element is maintained at a higher value than ambient conditions by passing an amount of current through the heater. When the sensing element is introduced inside any fluid flow, the temperature of the sensing element reduces due to heat dissipation from the sensing element into the fluid. In constant temperature mode, the temperature of the sensing element is maintained constant by increasing the input current to the sensing element in order to compensate the heat loss due to fluid flow. Thus, a different heater current is produced for each different flow condition. The heat loss from the sensing element causes a potential drop due to reduction in its resistance due to fluid flow across the sensing element which is a measure of the fluid velocity.

For correct and reliable detection of fluid velocity using the above technique the heat generated in the sensing element should completely dissipate in the fluid with minimum leakage or absorption in the bulk of the sensor substrate. In this case heat transfer will take place in two ways: conduction of the heat through the underlying catheter substrate & blood and through convection by blood only. Heat loss by conduction is proportional to the temperature gradient over a length of the material, *L* and cross-sectional area, A_{Cond} and is given by

$$Q_{Cond}=\frac{k*A_{cond}}{L}*(T_2\text{-}T_1) \qquad (1)$$

where $(T_2 - T_1)$ is the differential temperature across the volume and *k* is the thermal conductivity of the material.

Heat convection takes place through the net displacement of a fluid, which transports the heat content from a solid surface to the fluid through the fluid's own velocity [5]. The equations pertaining to convective cooling is same as above in addition to a velocity field.

3. Use of COMSOL Multiphysics

Simulation of the heater element using COMSOL multiphysics involves appropriate substrate selection for the sensor fabrication, choice of suitable heater design and verification of simulated data with that of practically measured one. The thin heater element of nichrome was modeled using 'Electric Currents, Shell interface (ecs)' which enabled effective meshing of very thin elements present over a very thick substrate. 'Heat transfer (ht)' physics was used along with this model to solve the amount of heat generated by the heater elements as discussed below.

The Heat Transfer interface with the Heat Transfer in Fluids feature solves the following equation for the temperature, *T* [5].

$$C_p\frac{\partial T}{\partial t} - \Delta.(k\Delta T) + \rho C_p U.\Delta T = Q$$

where ρ is the density, C_p is the heat capacity, *k* is the thermal conductivity, *U* is the fluid velocity and *Q* is the heat source (or sink). The second term denotes the heat loss by conduction in the underlying substrate as well as in the fluid medium. For a steady-state problem the temperature does not change with time and the first term disappears and the heat loss per unit area due to conduction becomes

$$Q = -\Delta.(k\Delta T) \quad W/m^2$$

which is similar to equation 1. Finally the third term denotes the convective heat loss due to fluid flow over the heated surface.

Moreover transient analysis of temperature rise, velocity and pressure distribution of the fluid inside the vessel in the presence of the catheter has also been performed.

3.1 Substrate selection

Heater elements of various shapes were fabricated over 1cm x 1cm x 1mm substrate area with the nichrome heater thickness of 0.2 µm. Different substrate materials like SiO_2/Si, Glass and PDMS (Polydimethlysiloxane) polymer was used for simulation. All the material properties were defined in the material properties database as per requirement. Electrical power required to raise the temperature of the heating elements by about 6 K above the normal body temperature was calculated for all. Ideally heater substrate should have minimum thermal conductivity to maximize the convection loss through fluid and minimize the loss through substrate. The external temperature in these simulations was kept at room temperature of 300K. Joule heating was used to determine the temperature generated in the heater and calculate the changes in resistance due to varying in input voltage by using linearized resistivity in the electric shell physics, since direct use of 'Joule Heating (jh)' physics was not possible for the resistors using shell model approximation. This model was solved using two way coupling between the Electric shell and Heat transfer physics, with the surface resistive losses (ecs.Qrsh) coupled as boundry heat source for the Heat transfer and the heat generated being coupled back to the temperature field for current conservation (ht/solid1) in the Electrical shell physics. The simulations were performed without any fluid flow over the sensor. Hence the heat loss was simulated to the default air domain above the heater element. The results of heater simulation over different substrates have been shown in Fig.1.

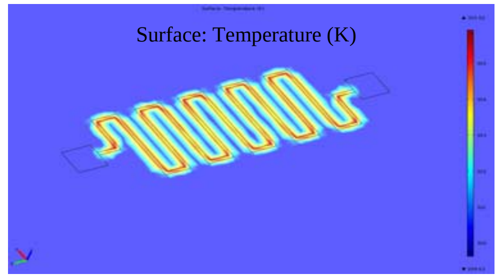
Fabricated Heater structure over SiO_2/Si substrate was tested at 300K (room temperature) and compared with the simulated results of a heater of similar dimensions which showed good correlation between them.

3.2 Heater design selection

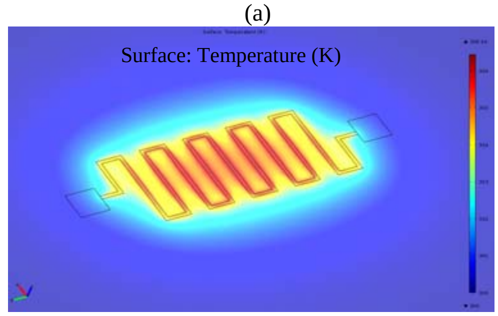
Various heater designs were tested for effective heat distribution over the entire catheter circumference. Heater designs were varied from straight line to meanderline shape with an optimum resistance of around 2 KΩ. Resistance value below this resulted in reduced sensitivity and above this resulted in high power consumption to heat the sensor to approximately 6 K above the normal body temperature. Heater structures of varying meanderline width with rounded corners was also tested and optimized to obtain uniform heat distribution over the surface. The effective heater length was fixed according to the circumference of the catheter around which it is proposed to be wrapped. In the present case a catheter of 3mm diameter was chosen which resulted in a total heater length of about 9 mm.

3.3 Transient analysis of heater simulation

A meanderline heater structure was designed around the cylindrical catheter of diameter 3 mm and was placed about 2.5cm from its tip. The diameter of the blood vessel was taken to be approximately 2cm. The external fluid has been taken as blood with an initial linear velocity of 0.2 m/s flowing in a direction opposite to that of the catheter insertion. The substrate material used for the fabrication is PDMS and the catheter element has been taken as polyethylene. Transient analysis of the heater structure was performed using ‘Electric Currents, Shell (ecs)’ along with ‘Conjugate heat transfer (nitf)’ physics. The coupling between the two physics has been done in a similar way as explained above for the substrate selection. Simulation with meanderline heater design has been performed and time required for the heater to reach the steady state value has been achieved.



(a)



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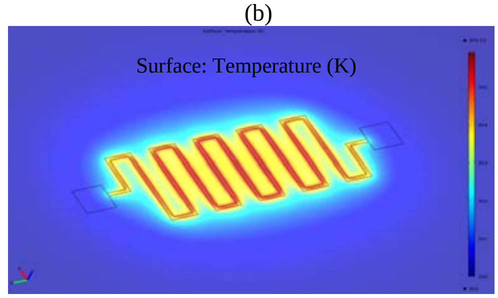


(c)

Fig. 1: Temperature distribution over a nichrome heater structure showing 6 K rise at different input voltages of (a) 22V for SiO_2/Si, (b) 9V for glass and (c) 4.2 V for PDMS substrate.

4. Results and Discussion

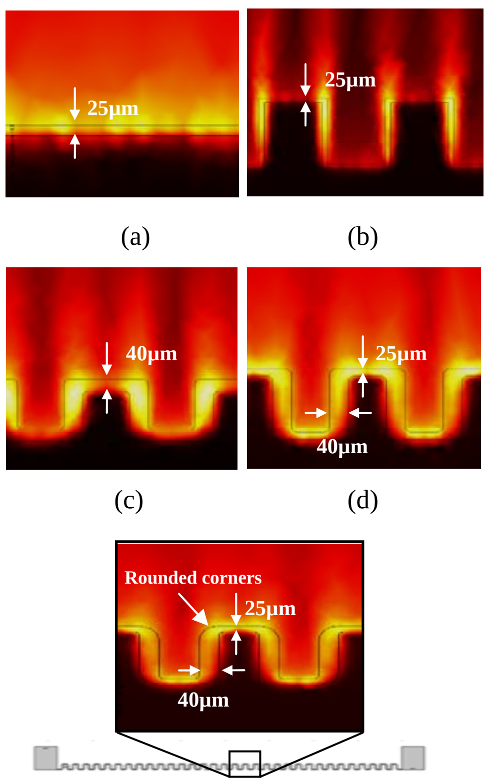
4.1 Substrate selection

A meanderline heater structure (~ 2 KΩ) was tested for suitable substrate selection based on the power required to raise the heater temperature by about 6 K. The results of heater simulation over different substrates have been shown in Fig.1. Table 1 shows that the PDMS elastomer required the least power of only about 7.98 mW as compared to 217.8 mW of power for the same heater structure over a SiO_2/Si substrate. Another reason for selection of PDMS as the substrate was due to its flexible and

Table 1: Power requirement for 6K rise in temperature for different heater/substrate simulation.

Substrate material	Average temp.(K)	i/p Voltage (V)	Power (mW)
SiO2/Si - Nichrome	305.9	22	217.8
Glass - Nichrome	306.1	9	36.9
PDMS - Nichrome	305.85	4.2	7.98

* Ambient temperature = 300 K

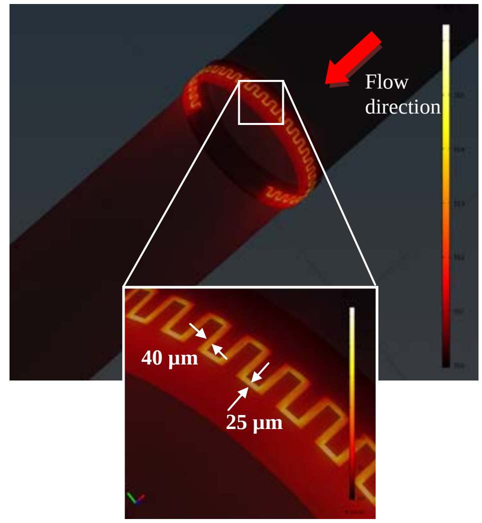


(e)

Fig. 2: Temperature distribution of different heater patterns simulated. (a) straight (width= 25µm), (b) meanderline structure (width= 25µm), (c) meanderline structure (width= 40µm), (d) meanderline structure (width= 25 & 40µm) and (e) meanderline structure (width= 25 & 40 µm) with rounded corners for uniform temperature distribution.

biocompatible [6,7] nature for utilization in bending of the sensor around the catheter surface.

The tested heater structure after fabrication also showed good correlation with a similar simulated structure. The fabricated sensor over SiO_2 deposited on Si wafer was found to show an



(b)

Fig.3: Steady state temperature profile of meanderline heater with 25 & 40 µm heater width.

increment of 6 K from the ambient room temperature of 300 K with an input voltage of ~ 25 V whereas the simulated result required ~ 22 V for the same. The increment in resistance for the simulated data is about 8.3 Ω for a temperature increment of 6.3 K whereas the experimentally measured resistance change is about 7 Ω for a temperature increment of 6 K. This observation verifies that the simulation result matches well with the practical data and might be extended for future models incorporating wrapping of the heater around the catheter tip and insertion of the same inside a blood vessel.

Various heater structures were simulated to achieve approximately 2 KΩ resistances for a resistor of length of 9.5 mm to be wrapped around a catheter of 3 mm diameter. Meanderline heater structure provides optimum increase of resistance of the sensor within the specified circumference of cathter with total resistance of about 2 KΩ. Heater structures of varying meanderline width (25/40 µm) and a pitch of 120 µm was also optimized along with rounded corners to obtain better heat distribution over its surface as shown in Fig. 2.

Steady state analysis of the meanderline heater structure without rounded corners having heater width of 25 and 40 µm was performed as shown in Fig.3. The voltage required to attain a

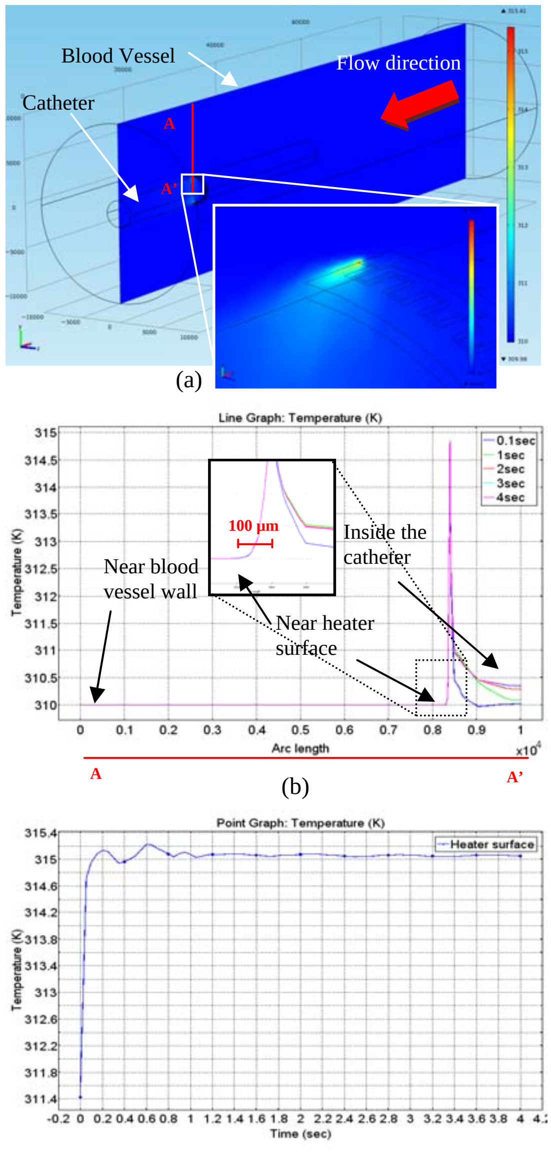


Fig. 4: Simulation of Meanderline heater (a) Temperature profile, (b) variation of temperature profile across the blood vessel, (c) Transient analysis of temperature profile.

steady state temperature of 315 K is about 12.8V and requiring about 60 mW of electrical power.

Transient analysis for the meanderline heater element was also performed as described earlier. Initial simulation was performed for meanderline heater structure without rounded corners. The temperature profile under transient condition is obtained as shown in Fig 4. From Fig 4c it is observed that the temperature of the heater

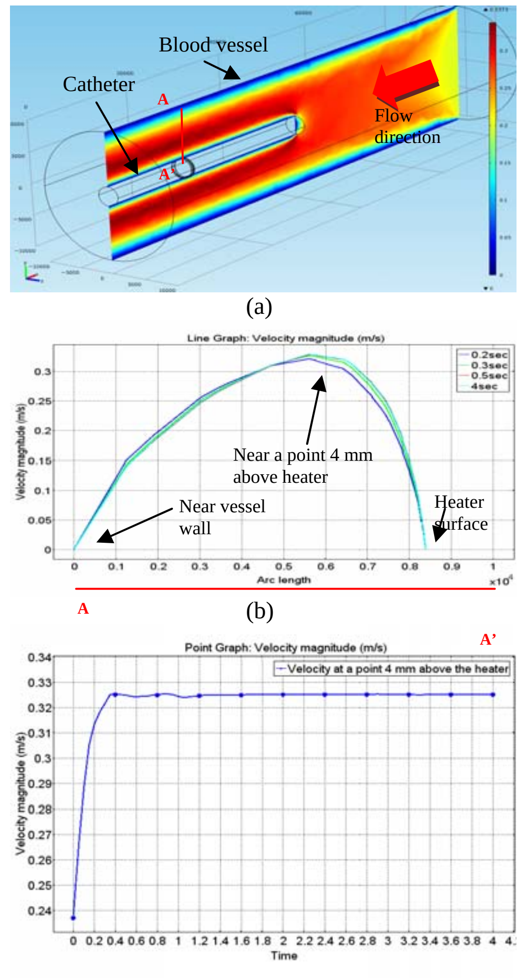
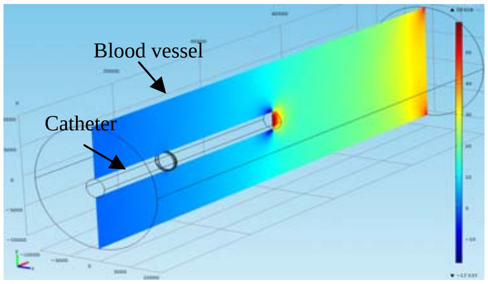
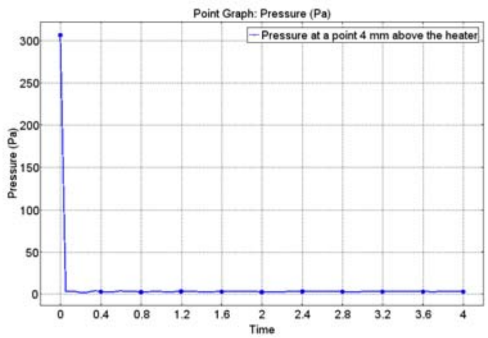


Fig. 5: Simulation of Meanderline heater (a) Velocity profile, (b) variation of velocity profile across the blood vessel, (c) Transient analysis of velocity profile at a point 4 mm above the heater surface.

reaches its steady state value of 315 K within about 0.2 sec. The temperature above the heater surface rises to about 5K and again settles to the normal body temperature of 310K within about 100 µm from the heater surface as shown in the inset of Fig. 4b. Thus the sensor does not practically heat the blood volume considerably and might be considered safe for practical usage. Fig. 5 shows the velocity distribution after the sensor insertion. From Fig. 5c it may be observed that the velocity reaches its steady state within about 0.4 sec. The velocity distribution shows an inflow of about 0.2 m/s at the inlet and a laminar flow of the blood over the sensor region which is situated about 2 cm from the



(a)



(b)

Fig. 6: Simulation of Meanderline heater (a) Pressure profile, (b) Transient analysis of pressure profile at a point 4 mm above the heater surface.

catheter tip in this simulation. Pressure variation and its transient analysis have also been shown in Fig.6. Further analysis of the sensor, by varying its position along the catheter tip, simulating the actual meanderline heater structure with rounded corners and incorporating a pulsating flow at the inlet needs to be performed.

5. Conclusion

This paper presents the simulation results of the temperature distribution of a nichrome heater with a varying input power considering different substrate materials. This result helped to choose the proper substrate for development of a flow sensor for a specific application. Various resistor layouts like straight line, meanderline with/without varying pitch have been designed and simulated to optimize the heater geometry for achieving uniform temperature distribution. Finally transient analysis of the sensor wrapped around the catheter tip and inserted in the blood vessel was also simulated. Initial results indicate a response time of about 0.2 sec for the heater temperature to rise 5 K above the normal body temperature. The test results needs to be further optimized to simulate the actual sensor with a pulsating blood velocity and increase the sensitivity of the sensor.

6. References

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