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# Numerical Study of Flow inside the Micro Fluidic Cell Sense Cartridge

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**Abstract**

Biosensor is a device which utilizes the biological element as a recognition element for the detection of analytes. The sample receiving unit is a critical integral part of the biosensor through which the sample is supplied to the biological element. In this study, the flow pattern inside the fluid cartridge was simulated using COMSOL multiphysics software with an objective of optimizing the shape and size of cartridge elements. The velocity distribution, flow induced shear stress and pressure drop were simulated as a single phase incompressible laminar flow under no slip condition. Influence of the shape and geometry of cartridge on flow patterns were studied by changing the shape and geometry of the cartridge. At a flow rate of 500 μl/min, the velocity and flow induced shear stress are in the range of 22 to 24 mm/sec and 0.76 to 0.85 $N/m^2$ respectively.


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*Keywords:* Microfluidic, Simulation, Biosensor, COMSOL

## 1. Introduction

Biosensors, integrated systems, are meant for detecting very minute quantity of analytes. This integrated system comprises sample receiving unit attached to analyte recognition system consisting of bioreceptor coupled to transducer. The transducer detects the biological signal produced by bioreceptor. Various types of transducers such as electrical, electronic, optical, magnetic, temperature based are used and their selection depends on case to case basis [1, 2]. Bioreceptor could either be any component of live system like enzymes,

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immunoglobulins, oligonucleotides, or could be natural or recombinant version of live system likes viruses, bacteria, yeast, plant, mammalian cells [3]. Specificity and sensitivity aspect and detection of bioavailable part of the target molecule makes these sensors preferential amongst other sensors in pharmaceutical industry, medical diagnostics, environmental monitoring, and food safety assurance [4].

Computational fluid dynamics (CFD) is a powerful tool to simulate and visualize fluid flow. This tool has been employed here to study the influence of various non-rectangular geometrical configurations for biosensor sample receiving unit (cartridge) on the flow distribution and optimize the cartridge geometry. CFD approach has been employed by various researchers to predict the fluid stress, force, and torque on an adherent cell as a function of the fluid viscosity, channel height, cell size, and flow rate per unit width as well as to study the strength of cell adhesion in microchannels [5-7].

Navier-Stokes (NS) equations that govern flow in macroscale are valid only under continuum assumption i.e., fluid is considered to be continuous rather than discrete molecules. Since the microchannel dimensions are in the order of several microns, the continuum assumption may or may not be valid, depending on the Knudsen Number ($K_n$). $K_n$ is defined as the ratio of the mean free path of the gas ($\lambda$) to a characteristic length scale of the flow ($L$) (usually the channel hydraulic diameter).

$$K_n = \lambda / L \tag{1}$$

For liquids, the mean free path term is replaced by the lattice spacing of the molecules ($\delta$) given by $\delta = [V/N]^{1/3}$, where V is the molar volume and N is Avogadro's number. The continuum assumption is valid when $K_n$ is less than $10^{-3}$. The working fluid in the present study is water. With hydraulic diameter of 1µm, $K_n$ is found to be $3 \times 10^{-4}$, indicating that continuum assumption is valid for the length scales considered (which are in the order of 100µm) and therefore, Navier-Stokes equations without any additional modifications may be used to study the flow in microchannels. Smallest channel size before the need for modified slip boundary conditions for water is ~0.3 nm [6].

## 2. Numerical procedure and validation

The governing equations and numerical methods used for the simulations were validated against the published literature by simulating the shear stress in a flat microchannel and on a single cell attached to a microchannel wall. The COMSOL multiphysics code was employed for modeling and simulation. Mesh was adequately resolved for the boundary layers and grid-independence studies were conducted.

A study on shear stress distribution in a flat microchannel reported by Lu et al [9] was taken as test case. Flow induced shear stress at a flow rate of 1 ml/min inside a rectangular channel with a dimension of 2000×500×25 µm (L×B×D) was simulated as single phase incompressible laminar flow with no-slip at walls. The wall shear stress is defined by the normal velocity gradient at the wall, given as:

$$\tau = -\mu \, (du/dn) \tag{2}$$

The magnitude of shear stress obtained from the present study (Fig. 1) was found to be in good agreement with the result of reported by Lu et al (2004). The predicted maximum shear stress at the centre of channel is 3250 dyne/cm$^2$ where as the estimated value by Lu et al is 3200 dyne/cm$^2$. The maximum error between the simulated values and analytic values at any point across the breadth of the channel was about 1.9%.

Gaver and Kute [10] reported a theoretical model for fluid stresses on a cell adhering to a microchannel wall. This study was taken as second test case for the validation of present numerical approach. The radius of cell attached to the wall was 7.5 µm and the dimension of 2-D channel were 500×25 µm. Fig. 2 shows the predicted shear stress distribution on the cell normalized with shear stress on the flat wall. Maximum dimensionless shear stress on the cell reported by Gaver and Kute [10] was 3.0 where as it was found to be 3.14 for the present case. With the validated model, parametric studies were further conducted.