A Particle Detection Modeling of Non-contact Coplanar Differential Impedance Sensor in Microfluidic System

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Abstract. Over the last few years, study of microudile flow cytometry using electrical signals has developed at a fast pace. There are several electrical detection configurations but differential impedance sensing offers a greater advantage due to its capability of cancelling common noise from the detected signal, resulting in improved signal to noise ratio. This paper presents the simulation of a differential impedance sensor by employing the finite element method to gain an insight and to find the proper range of the working excitation frequency for the detection system. A polystyrene microbead was used as a model particle flowing past the detection area that had non-contact excitation and two pickup coplanar electrodes in which an excitation electrode was positioned between the two pickup electrodes. The modeling results showed that for a 2 µm PDMS separation layer, the range of optimal excitation frequency was about 100 - 1000 Hz when DI water was a background medium. The bead size to microchannel height ratio was found to affect the sensitivity of detection, in which the closer the height ratio was to the unity ratio, the more current was detected. The results of the simulation study will be used in fabricating an actual device for microfluidic particle cytometry applications.

1. Introduction
In recent years, particle flow cytometry in microfluidic systems using the impedance sensor technique has been of increasing interest. Conventional flow cytometry exploits the optical detection principle; thus, particles have to be fluorescently labeled before detection. The traditional technique is widely used in biomedical applications. However, it has some drawbacks such as being an expensive system as well as requiring detection treatment of particles [1]. To overcome the limit of conventional flow cytometry, the non-contact impedance sensor for particle flow cytometry is a great choice due to being inexpensive and the efficiency in
cancelling common noise in the signal [2]. Differential detection has three electrodes where an excitation electrode is positioned between two pickups electrodes. The two measured signal were differentiated using differential amplifier. The aim of this work is to model the differential non-contact impedance sensor in a microfluidic system employing the finite element method of the COMSOL Multiphysics software to optimize system conditions. The results of the study will be implemented in design and fabrication of the differential impedance sensor in microfluidic devices.

2. Theory

Figure 1 (a) shows a schematic of a half of the differential detection area. The electrolyte medium can be considered as a resistor ($R_{el}$). The PDMS wall, electrical double layer, coupling between two electrodes can be consider as $C_{wall}$, $C_{DL}$, and $C_{stray}$ Capacitors respectively as shown in Figure 1 (b) [3].

![Figure 1](image)

The total impedance of the schematic can be calculated using equation:

$$\frac{1}{Z} = \frac{1}{X_{stray}} + \frac{1}{R_{el} + 2X_{wall} + 2X_{DL}}, \quad (1)$$

where $X_{stray}$, $X_{wall}$, and $X_{DL}$ are the impedances of the stray capacitor, wall, and electrical double layer respectively. Figure 1 (c) shows the total impedance of the detection area calculated as a function of excitation frequency. The working frequency will be at the plateau position since the resistances of the electrolyte medium dominates the total impedance. To find the other parameters affecting the sensitivity of the detection system, a 3D model of the detection area was designed and simulated using AC/DC module in COMSOL Multiphysics software in the frequency domain. The equations are

$$\nabla \cdot \mathbf{J} = Q_j \quad (2)$$

$$\mathbf{J} = \sigma \mathbf{E} + j\omega \mathbf{D} + \mathbf{J}_e \quad (3)$$

$$\mathbf{E} = -\nabla V, \quad (4)$$

where $\mathbf{J}$ is current density per unit area, $Q_j$ is the current source, $\sigma$ is the electrical conductivity, $\mathbf{E}$ is the electric field, $\mathbf{D}$ is electric displacement, $\mathbf{J}_e$ is the external generated current density, and $V$ is electric potential.
3. Designs and Method
The 50 µm wide electrodes were initially fixed at 50 µm apart from each other. The narrow channel had dimension of 15 µm in height, 20 µm in width, and 50 µm in length with a tapered angle 90°. A 10 µm polystyrene bead flowing in the middle of the channel plane was used for detection while DI water was used as a background medium in the channel. The applied voltage was 24 Vp-p with varying excitation frequencies. The electrical conductivity and relative permittivity of materials are shown in Table 1.

| Material     | Conductivity (S/m) | Relative permittivity |
|--------------|--------------------|-----------------------|
| DI water     | $5.5 \times 10^{-6}$ | 80                    |
| PDMS         | $2.5 \times 10^{-14}$ | 2.75                  |
| Polystyrene  | $1 \times 10^{-16}$  | 2.3                   |

In this work, the effect of the separation layer thickness and edge-to-edge space between the excitation and pickup electrodes on the impedance spectra were studied. Other factors affecting detection efficiency including spacing between the electrodes and channel height to bead size ratio were also studied.

4. Results and Discussion
The impedance spectra of half of the detection area with varying excitation frequencies is shown in Figure 2. Figure 2 (a) shows the impedance spectra of the sensor with varying separation PDMS layer thickness. It can be seen that the thinner the separation layer, the higher the impedance of the system; note that $C_{\text{wall}}$ affects the impedance only at low frequencies. When spacing between electrodes or $C_{\text{stray}}$ was decreased, the impedance increased, as evident in Figure 2 (b). The working excitation frequency of the simulated system was found to be 100 - 1000 Hz.

![Figure 2](image)

**Figure 2.** The impedance spectra of half of detection area when varying (a) separation layer thickness and (b) spacing between electrodes.

In the actual experiment, the measurable quantity of the sensor system is the current rather than the impedance. The goal of the simulation is to find optimized conditions that would result in maximum current. The typical current profile of the differential impedance sensor is illustrated in Figure 3 (a). The simulated signal agrees well with the previous experimental study [7]. From the figure, the sensor starts to pick up the current signal when the bead arrives at the outer edge of the pickup electrode and the current is highest when the bead is at the edge.
of excitation electrode. The effect of channel height to bead size ratio of the maximum pickup current is depicted in Figure 3 (b). The sensor detected the highest current when the channel height was slightly greater than the bead size. The edge-to-edge spacing between the excitation and pickup electrodes shows optimal spacing of approximately 35 µm, shown in Figure 3 (c), in which the current is at its maximum. The finding agrees to that of the similar simulation study [8] in which the authors argue that the electric field distribution for the closely spaced electrodes concentrates mostly in the area just above the electrodes, resulting in lower sensor sensitivity. For the wider spaced electrodes, the bigger detection volume when the bead volume is constant results in lower sensitivity of the sensor.

Figure 3. (a) Differential current signal from the simulation and plot of maximum current with varying (b) channel height to bead size ratio and (c) electrodes spacing.

5. Conclusion
In this study, the differential non-contact impedance sensor was modeled using the finite element method. The simulation gave out optimal working conditions for the sensor, in which for a 10 µm polystyrene bead in DI water and 50 µm wide electrodes, the optimal working excitation frequency is 100-1000 Hz when the channel height is comparable with the bead size. The optimal spacing between electrodes is 35 µm. The results of this work will be used in fabricating impedance sensors for biological cell flow cytometry applications.

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