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RESEARCH ARTICLE

Design and Experimental Studies for a PDMS Microfluidic Sensor Detecting Low Blood Flow Rate

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ABSTRACT: This study investigates whether Lorentz-force–based sensing—well established at the macro scale—remains valid at the micro scale within a PDMS microfluidic environment. We hypothesize that when an ionic solution flows through a PDMS microchannel in a magnetic field, the induced voltage will follow the same proportional relationship to flow velocity predicted by the Lorentz force equation in macro systems. To test this, we fabricated a PDMS microfluidic sensor with gold capacitor electrodes and introduced a potassium chloride (KCl) solution at controlled flow rates. Measured voltages were compared with theoretical predictions from Lorentz-force equations derived for macroscopic conductive fluids. Experimental results confirmed a strong linear correlation between measured and predicted voltages, with an average flow-rate error of 9.0% and the detectable flow rate minimum (0.028 mL/s). These results indicate that Lorentz-force behavior scales consistently down to microfluidic dimensions, validating the physical model's applicability at small scales. The findings demonstrate that Lorentz-force–based sensing can accurately quantify micro blood flow, providing a foundation for developing wearable cardiovascular monitoring technologies. This microfluidic sensor could provide valuable insight for future work.

KEYWORDS: PDMS, Microfluid Sensor, Lorentz Force, Flow Rate, Microfluidic Channel.

Introduction

Cardiovascular disease is the leading cause of death worldwide, responsible for 17.9 million deaths annually. Studies show that nearly 75% of these deaths could be prevented with early and continuous monitoring. Blood flow rate is an important physiological parameter to indicate the health of the cardiovascular system, especially related to blockages, and could provide insight into a variety of pathological conditions. Measuring blood flows allows for real-time monitoring of cardiovascular health, helping to indicate conditions such as thrombosis, arterial stenosis (narrowing), vessel occlusion, and arterial hardening.[1] Traditional methods for detecting blood flow rate include Doppler and pulse Doppler flowmetry (for large vessels) and laser Doppler flowmetry (for small vessels).[2] However, these methods require a clinical setting, are expensive and complex, and cannot offer real-time monitoring in patients' daily lives. This research focuses on microfluidic technologies for real-time, cost-effective, minimum-invasive blood flow rate monitoring using PDMS-based chips integrated with capacitors.

The development of PDMS microfluidic sensors has been seen as a significant breakthrough in the advancement of lab-on-a-chip technology, providing the ability to manipulate and analyze fluids at the microscale in real time with high precision and low cost.[3] PDMS is an elastic polymer that is both flexible and biocompatible, and such high mechanical and chemical stability makes it popular in the microfabrication and bioengineering fields.[4] In addition, PDMS is also a cost-effective material, with an average cost of $3 per 1000 grams. PDMS's flexibility also allows the fabrication of deformable structures that are essential for creating biomedical devices.[5] Given all these qualities, we conclude that PDMS is highly suitable for the development of microfluidic sensors that are sensitive and scalable for human blood flow rate detection.

This study aimed to use PDMS material to develop a microfluidic sensor that could detect micro blood flow rate in contrast to traditional methods to measure blood flow rate, such as using large medical devices like Laser Doppler Flowmetry (LDF) and Pulse Doppler flowmetry (PDF). Utilizing a PDMS-based microfluidic sensor could provide a more cost-effective, convenient, and real-time monitoring for patients, making life easier for people suffering from cardiovascular diseases, reducing the number of follow-up visits after they have been diagnosed, and helping doctors to make corresponding adjustments to medications more conveniently. Existing microfluidic sensors for blood flow detection, each with limitations such as limited spatial resolution, high sensitivity to environmental changes, or expensive equipment, are mostly based on optical methods, capacitive sensing, or thermal anemometers.[6] To overcome these barriers, recent works have suggested using resistive sensing techniques where changes in electrical fields are measured due to mechanical deformation or flow-induced changes. It uses flexible microchannels filled with conductive composites to provide a high response to changes in flow rate.[7] Previous research has also shown that microfluidic sensors could be used to detect human vital signs such as blood pressure, as they have a biocompatible stability and are easy to wear.[8] Therefore, we hypothesize that a flexible PDMS microfluidic chip embedded with conductive composite sensors can accurately and continuously measure blood flow rate, offering a practical tool for cardiovascular monitoring.

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Inspired by previous research, we conclude that the PDMS-based flexible microfluidic sensor with integrated conductive composites could offer the most precise detection of flow rate changes; and due to the characteristics of microfluidic sensors and PDMS—such as biocompatibility, cost-effectiveness, ease of analysis, and precision—this system presents a highly promising platform for real-time, noninvasive cardiovascular monitoring and early disease detection. Based on this conclusion, we develop a flexible, PDMS-based microfluidic sensor combined with a conductive gold-circular capacitor following the Lorentz force law, as shown in Figure 1, which quantifies blood flow rate using electrical voltage measurements.

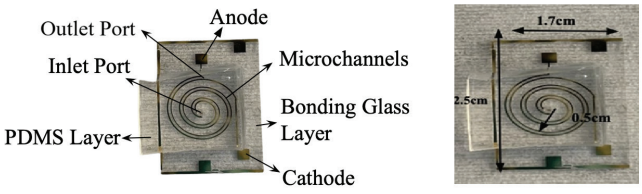


Figure 1: A flexible, PDMS-based microfluidic sensor combined with a conductive gold circular capacitor for micro blood flow detection. (a) The appearance and structure of the PDMS microfluid sensor. (b) The dimensions of the sensor.

Methods

The fabricated sensor is placed above a permanent magnet with a constant and uniform magnetic field. The blood analog test solution, potassium chloride (KCl), is dissolved in deionized water at a density of ~1.060 g/cm^3 (blood density) to ensure adequate conductivity. The solution is pumped into the microfluidic sensor with a syringe pump at a constant flow rate. Figure 2 shows the electric force and magnetic force on the flowing ions of the solution in the microfluidic channel. Both the electric field and magnetic field can be defined from the Lorentz force equation:

$$\boldsymbol{F} = q\boldsymbol{E} + q\boldsymbol{v} \times \boldsymbol{B} \tag{1}$$

where $\boldsymbol{F}$ is the Lorentz force, q is the charge of the ions, $\boldsymbol{E}$ is the electric field, $\boldsymbol{v}$ is the flow velocity, and $\boldsymbol{B}$ is the magnetic field.

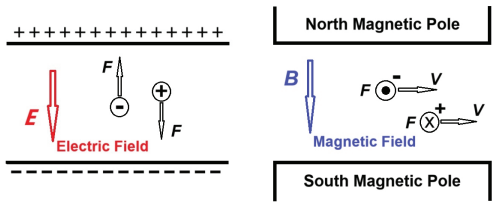


Figure 2: The electric force and magnetic force on flowing ions in the chip channel. (a) Electric force $F=qE$, a view from above of the flow channel. (b) Magnetic force $F=qv\times B$ perpendicular to both v and B, front view of flow channel.

Initially, the small flow channel consisting of two conductive walls and a non-conductive top and bottom is an uncharged coil capacitor. After the flow starts, the magnetic force pushes the positive and negative ions in opposite directions, and as the positive and negative ions gather on the walls respectively, an electric field is formed.

$$E = -\frac{v}{w} \tag{2}$$

where V is the voltage across channel plates, and w is the channel width.

The volumetric flow rate Q is given by

$$Q = vA = vWh \tag{3}$$

where A is the cross-sectional area of the flow channel, and h is the height of the flow channel.

When the flow is in a steady state, the electric force is equal to the magnetic force in magnitude, but in opposite direction, so the Lorentz force in Eq. (1) is zero. Plugging Eq. (2) and Eq. (3) into Eq. (1), we obtain:

$$V = \frac{QB}{h} \tag{4}$$

In the experiment, the channel height h is 50 μm, and the magnetic field B is about 0.4 T. Using Eq. (4), the voltage across channel plates can be calculated after the volumetric flow rate is measured, or the volumetric flow rate can be predicted after the voltage across channel plates is measured.

Eq. (4) shows that the voltage is inversely proportional to the channel height. In the application of the chip channels as flow rate sensors, in order to obtain a stronger voltage signal, the channels should be made thinner. In this case, to avoid generating a large flow resistance, a wider channel can be used because the channel width does not affect the output voltage. On the other hand, if the chip channels are used as micro-electromagnetic pumps, higher height channels can produce higher flow rates at a given input voltage.

Wafer Preparation:

The PDMS microfluid sensor is fabricated with a standard photolithography and soft lithography process. As shown in Figure 3(a), a clean silicon wafer was prepared by rinsing thoroughly with deionized water, drying in a stream of clean air, and baking at 120 °C for 10 minutes to remove any remaining moisture, allowing the SU-8 photoresist to stick better on the substrate.

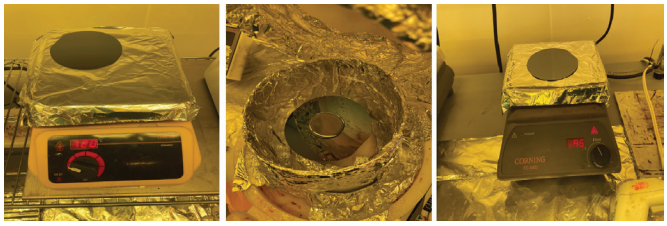
Spin Coating SU-8 Photoresist.

The spin coating process is used to create the photoresist layer, which is going to be the mold. By applying SU-8 photoresist and by setting up the spin coater ramp at 500 rpm for 10 seconds, as shown in Figure 3(b), it generates the desired thickness of the photoresist film and lets the SU-8 photoresist spread over the silicon wafer homogeneously. The acceleration of the spin coater and the viscosity of the SU-8 photoresist will define the thickness of the layer we want.

Soft Bake:

After spreading the photoresist, the spin coated wafer is then subjected to a three-step soft bake on the hot plate at 65 °C for 2 minutes, 95 °C for 8 minutes, followed by again 65

°C for another two minutes, which helps to evaporate some of the solvents in the photoresist and also solidifies the wafer to prepare for UV exposure, as shown in Figure 3(c).



(a) (b) (c)

Figure 3: Wafer preparation and soft bake. (a) The wafer is rinsed with deionized water, dried with clean air, and baked at 120 °C for 10 min. (b) The wafer is coated with SU-8 photoresist using a spin coater. The spin coater's acceleration will control the thickness of the photoresist. (c) The wafer is soft-baked on a hot plate, 65 °C for 2 min, 95 °C for 8 min, and 65 °C for 2 min.

UV Exposure:

In the photolithograph process, a photomask with circular microfluidic channel patterns is aligned above the wafer, and exposed to SU-8 under UV light at 365 nm wavelength. As SU-8 is a negative photoresist, the areas exposed to electrons are cross-linked and thus become solid.

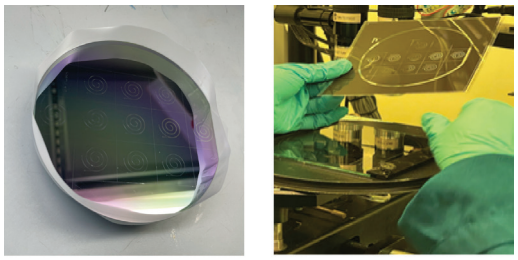
Post-Exposure Bake:

After UV exposure, the wafer is baked at 65 °C for 2 minutes, 95 °C for 8 minutes, and 65 °C for 2 minutes, with the same pattern as the soft bake. The post-exposure bake brings energy to continue the reaction activated by the UV exposure. The wafer was put into the SU-8 developer for 4–5 minutes to remove regions not exposed, and then rinsed with deionized water.

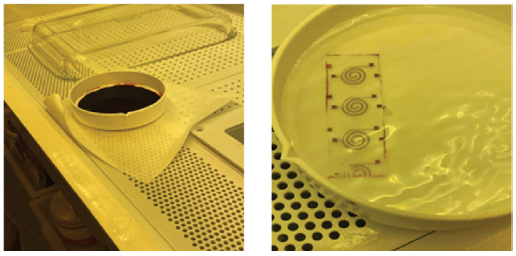
Development of the SU-8 Photoresist Mold and Bonding:

The wafer was put into the SU-8 developer for 4–5 minutes to remove non-linked photoresist, and then washed with isopropyl alcohol.

After the UV exposure of the silicon wafer, the PDMS is poured over the SU-8 mold and cured to produce the microchannel layer. Concurrently, a gold layer was deposited and patterned into the capacitor-based sensing circuit on a glass substrate. The gold was chosen for the capacitor walls because it has good biocompatibility, conductivity, and corrosion resistance. The cured PDMS chip is peeled off the mold and bonded with the glass substrate under a microscope after plasma treatment to make sure that the microchannels are accurately aligned with the circular golden capacitor. Figure 4 shows the PDMS layers and the development of a golden circular capacitor.



(a) (b)



(c) (d)

Figure 4: PDMS layers and the development of a golden circular capacitor. (a) Silicon wafer mold with PDMS on top. (b) The photomask with a circular microfluidic channel pattern is used in the photolithography process. (c) A gold circular capacitor before development. (d) The gold circular capacitor after development.

Experimental Methods:

The fabricated sensor is placed above a permanent magnet with a constant and uniform magnetic field of 0.4 T, measured with a Tesla meter. The blood analog test solution, potassium chloride (KCl), is dissolved in deionized water with a density of 1 g/cm^3 to ensure adequate conductivity. The solution is pumped into the microfluidic channel by a syringe pump at a constant flow rate, as shown in Figure 5.

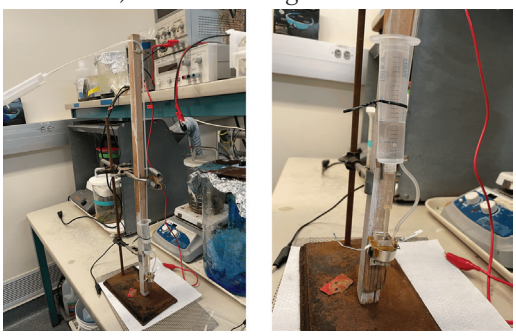


Figure 5: Experimental setup for the Lorentz-force measurements. The PDMS sensor is mounted above a permanent magnet while a syringe pump delivers ionic KCl solution at controlled flow rates. The induced voltage is recorded using a high-impedance differential electrometer.

■ Results and Discussion

To test the effectiveness of our PDMS-based microfluid sensor in detecting micro blood flow rate, we conducted a series of experiments designed to test the sensor's accuracy in measuring fluid flow rates similar to those found in the human cardiovascular system. The rationale for these experiments was to determine whether the sensor could detect and quantify flow rates within a physiologically relevant range, while maintaining sufficient accuracy (<15% of error).

In the experiments, we use a syringe pump to introduce potassium chloride (KCl) solution dissolved at a density of ~1.060 g/cm^3 (blood density) at known flow rates into the microfluidic channel of the sensor. The flow channel size is 0.4 mm wide and 0.05 mm high. The voltage between the anode and cathode was recorded during each trial. Using the Lorentz force law, we calculate the flow rate based on the detected voltage. We compared the measured flow rate from the sensor and the flow rate from the syringe pump and calculated the flow rate error for each trial (Table 1). The flow rates measured by

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the PDMS microfluidic sensor have a good agreement with the pump flow rates, as shown in Figure 6.

Our results show that the given flow rate by the pump is from 0.028 mL/s to 0.286 mL/s, and the measured flow rate by the sensor is from 0.025 mL/s to 0.263 mL/s. This range covers the average blood flow rate within the human body, such as the average blood flow rate in arteries (median of 0.24 mL/s) and in veins (median of 0.05 mL/s), showing the biomedical applicability of the device.9 Meanwhile, the average flow rate error detected by the sensor is 9.0% (Table 1), which is within the 10% margin commonly accepted in biomedical engineering for wearable sensors for healthcare monitoring.[10] This level of accuracy indicates that the sensor can potentially serve as a reliable wearable device for monitoring blood flow rate.

Table 1: Comparison of measured and reference flow rates. The table presents the pump flow rate, corresponding measured voltage, calculated flow rate from the Lorentz-force equations, and percentage error. The calculated voltage is derived from Eq. (4) using the given pump flow rate.

Test #	Given flow rate by pump [mL/s]	Measured voltage [mV]	Calculated voltage [mV]	Measured flow rate by sensor [mL/s]	Voltage error	Flow rate error
1	0.028	0.20	0.22	0.025	-10.0%	11.1%
2	0.067	0.50	0.53	0.063	-6.3%	6.7%
3	0.125	0.90	1.00	0.113	-10.0%	11.1%
4	0.214	1.60	1.71	0.200	-6.7%	7.1%
5	0.286	2.10	2.29	0.263	-8.1%	8.8%
				Average error	-8.2%	9.0%

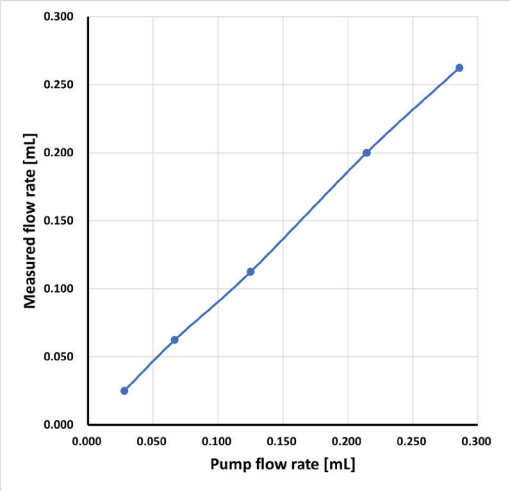

Figure 6: Correlation between measured voltage and flow rate. The plot shows linear dependence between voltage output and volumetric flow rate, demonstrating agreement with Lorentz-force theoretical predictions.

Additionally, the sensor's compact dimensions (2.5 cm × 1.7 cm × 0.5 cm) meet the spatial constraints to be a wearable device for blood flow monitoring (Figure 1). Furthermore, the average cost per sensor is approximately $22, indicating the cost-effectiveness of the design and the potential for mass production.

The results of these experiments confirm that the device successfully meets its design criteria: physiological relevance, measurement accuracy, compactness, and cost-effectiveness. Given the biocompatibility and flexibility of PDMS, along with the sensor's consistent accuracy in flow rate relations, our data support its potential for real-time monitoring of blood micro flow to aid in cardiovascular monitoring.

While the results confirm the sensor's accuracy, several limitations must be acknowledged. First, testing was performed using a KCl solution in standard human blood density under controlled laboratory conditions, which does not fully guarantee the sensor's performance in the more complex rheological and biochemical properties of blood, such as non-Newtonian viscosity and cellular components. These factors could affect flow dynamics and electrode response, causing a larger inaccuracy. Second, the experiments were conducted at room temperature; temperature variations in a clinical or wearable setting could affect PDMS elasticity and sensor performance. Third, while the average error was within the accepted biomedical threshold, the voltage signal measurement process requires clean cathode and anode surfaces; even tiny ions adhering to the electrodes could alter voltage measurements, causing the sensor's performance to vary and result in a larger error in fluid flow rate reflection.

As the voltage signal detected is inversely proportional to the height of the channel, reducing the channel height could potentially improve signal strength, resulting in a stronger voltage signal and a more accurate fluid flow rate detected. Furthermore, as Eq. (4) shows, the width of the channel does not affect the voltage signal detected; expanding the width of the channel will reduce the resistance inside the sensor, increasing the accuracy of the overall detected flow rates.

These findings have a significant impact on the development of minimally invasive biomedical devices for real-time blood flow monitoring. The PDMS microfluid sensor's ability to detect analog blood flow rates in the physiological range with less than 10% error suggests that the sensor could serve as a foundation for wearable health monitoring technologies. The small dimensions and low-cost characteristics of the design further demonstrate the sensor's potential for large-scale production in both hospital and home-care environments, reducing the cost for cardiovascular disease monitoring and the redundant follow-up diagnosis to measure human vitals such as blood flow rates around a certain location in the human body. Beyond blood flow monitoring, similar sensor designs could be adapted for intravenous infusion rate tracking, early detection of vascular blockages, or industrial applications involving microfluidic devices such as micro-electromagnetic pumps. The Lorentz force-based voltage-flow relationship can also be applied to Magneto Hydro Dynamics (MHD) pumps,[11] chaotic stirrers,[12] and minute mixers.[13]

■ Conclusion

This study evaluates a PDMS microfluidic sensor that uses the Lorentz-force equation to detect micro flow rate according to the correlation between flow rates and voltages to estimate micro blood flow rates, with the ultimate goal of developing a wearable device for real-time blood flow monitoring. The testing procedure involves using a KCl solution at a controlled blood density at controlled flow rates using a syringe pump and comparing the measured flow rates to the syringe pump's

given flow rates. The experiments demonstrated that the device detects volumetric flows between 0.025–0.263 mL/s with a mean flow error of 9.0% compared to given syringe-pump values. These results support the hypothesis that a compact, low-cost PDMS sensor can measure physiologically relevant microflow rates with biomedical-level accuracy (under 10% error) while keeping costs suitable for mass production.

Future work should focus on addressing the limitations, developing corresponding implementation methods, and implementing animal testing. This research only uses an analog human blood solution (KCl); therefore, future experiments need to employ human blood or blood analog fluids under physiological conditions to test the sensor's performance in a more physiologically relevant environment. Testing across a wider temperature range would help determine the thermal stability of the PDMS and circular gold circuit system. Currently, the measured voltage sensitivity is about 7.5 mV per mL/s. To improve voltage sensitivity without excessive flow resistance, microchannel dimension optimization—specifically reducing height and increasing width—should be investigated under a variety of experiments. Meanwhile, integration with wireless data transmission and the programming of corresponding applications need to be developed to achieve continuous remote monitoring of blood flow rates and relevant human vitals.

From a biomedical standpoint, animal testing using various blood densities and patient conditions needs to be implemented to further support and adjust PDMS microfluid sensors for blood flow monitoring. Statistical analysis of repeated measurements with larger sample sizes will provide a more detailed and accurate assessment of the sensor's reliability. Ultimately, translating this technology to clinical use would require relevant medical device regulations, extensive human trials that account for various cardiovascular conditions, patient comfort, and integration methods that relate to corresponding circumstances.

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