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A Sensing Tube with an Integrated Piezoelectric Flow Sensor for Liver Transplantation

Wooseok Jung[1], Chunyan Li[1,2], Dong-Sik Kim[3], and Chong H. Ahn[1]
[1]Microsystems and BioMEMS Laboratory, Department of Electrical and Computer Engineering
[2]Department of Neurosurgery, [3]Department of Surgery
University of Cincinnati, Cincinnati, OH, USA

***Abstract*— An innovative sensing tube with an integrated piezoelectric flow sensor to measure flow rate in portal vein during liver transplantation has been developed and characterized in this work. This smart tube aims to measure flow rates in the portal vein in a real-time continuous format with high reliability. The PVDF-TrFE piezoelectric flow sensor integrated in a tube was able to measure flow rates with the sensitivity of 0.02 mVpp per 20 mL/min in the range of portal vein flow rate. It also showed excellent operational stability for 120 minutes with a standard deviation of 0.084 mVpp. The tube can be used as a clinical tool for assessing the need for portal vein inflow modification during liver transplantation.**

I. INTRODUCTION

PATIENTS with cirrhosis often develop portosystemic shunts (i.e. extensive venous collateral formation) to redistribute flow away from the main portal vein due to hypertension in the liver. This leads to portal vein atrophy and/or thrombosis. When the patients with liver cirrhosis undergo liver transplantation, the newly transplanted liver can be damaged due to the shunts that prevent blood flow into the liver if the shunts are very large and well established. On the other hand, if the patient receives a partial liver with marginal volume, the grafted liver can be damaged by the excessive blood flow. Thus, the ligation or creation of portosystemic shunts in the recipients, who are undergoing liver transplantation, is very desirable to avoid injury to the newly transplanted liver. So, flow measurement at the intraoperative portal vein should be performed to accurately determine the need for inflow modification [1].

The measurement of portal vein flow has typically been accomplished with a Doppler effect-based ultrasound [2-3]. But, with this technique a precise flow measurement at the portal vein is considered as one of the difficult tasks due to its operator-dependent nature.

Portal vein flow can also be measured by a thermodilution catheter [4]. In this method, a thermal temperature sensor located at the distal end of the catheter measures the temperature change due to the cold saline solution ejected through a proximal hole in the shaft. Thus, it can measure the flow velocity from the time required to achieve a certain change in portal vein temperature. However, this method is unable to measure portal vein flow continuously because the volume of cold saline needed for an effective measurement is hard to be sustained.

Thus, there is a large demand for the development of a new flow measurement method with a good reliability and long-term stability while minimizing possible disturbances in blood vessels through minimally-invasive flow sensors.

The flow rate of the portal vein ranges from 700 mL/min to 2,000 mL/min [1]. Other requirements, such as the power consumption of less than 10 mW/cm^2 and the heating temperature of less than 5 K above blood temperature for invasive applications in human vessels [5], restrict the ability of thermal microsensors to measure portal vein flow rate. Additionally, low flow pressures (5~10 mmHg) and negligibly low pressure drop along the portal vein also restrict the use of conventional sensors based on the differential pressure-based measurement. So, a new sensor approach is needed to achieve 20 mL/min flow resolution in the range of portal vein flow rate, which is recommended by liver transplant surgeons. In this paper, we describe a new sensor that uses a polymer piezoelectric material, PVDF-TrFE [6], which detects the output amplitude of the driving signal that is proportional to the flow rate of the portal vein.

Figure 1 illustrates the portal vein structure where the new sensing tube integrated with a piezoelectric flow sensor is placed. Since the flow sensor is integrated on a tube in a diameter of less than 1 mm, the sensing tube can measure the portal vein flow rate without a surgical incision. Additionally, the new design minimizes the disturbance of the blood flow due to its structural compatibility with the portal vein. Spiral-rolling of the flexible PVDF-TrFE thin film, which contains a micro flow sensor, allows the realization of a sensing tube.

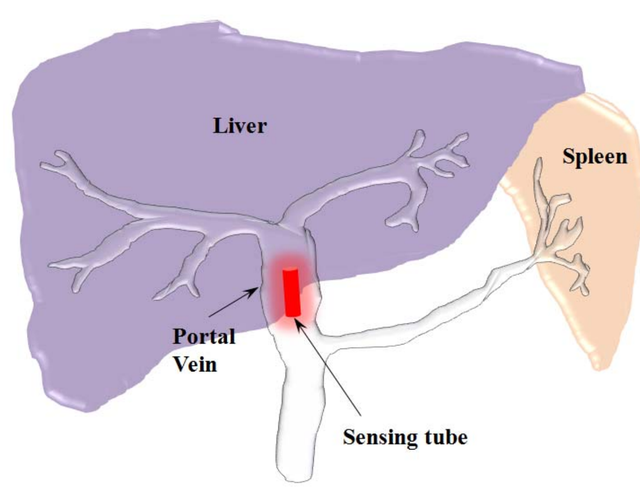


Fig. 1. A structural illustration of the portal vein where a sensing tube integrated with a piezoelectric flow sensor will be located.

II. DESIGN AND FABRICATION

A. Sensing Principle

The device has two electrically isolated electrode pairs that are used as input and output electrodes, as described in Figure 2. The regions, in which electrode pairs are deposited, have a high piezoelectricity that can be achieved through a selective DC-poling technique. The sensor region, where input and output electrode pairs are deposited, is fabricated as a diaphragm type, as shown in Figure 2 (a). This maximizes the magnitude of the deformation and, correspondingly, improves the sensitivity of the sensor.

PVDF-TrFE based piezoelectric sensors intrinsically can sense AC signals and its output signals usually have a poor resolution at below 2 Hz. However, the flow rate of blood in portal vein is varied at the breathing frequency of approximately 0.25 Hz. In order to sense this quasi-static blood flow in the portal vein, a driving AC driving signal at a fixed frequency is constantly applied to the input electrodes, so that a constant mechanical vibration is produced on a diaphragm through the converse piezoelectric effect [7]. PVDF-TrFE piezoelectric thin film is deformed along the mechanical length direction in less than 100 kHz with the converse piezoelectric effect [8]. Additionally, the diaphragm vibrates up and down in the length mode with the driving AC signal as shown in Figure 2 (a) because the region, where electrodes are located, is anchored around its perimeter as a diaphragm. Subsequently, the blood flow in the portal vein affects the mechanical vibration of the diaphragm. The changed mechanical vibration is then converted back to the electrical signal through the direct piezoelectric effect and detected at the output electrodes [9] as shown in Figure 2 (b). Since the input and output electrode pairs are deposited on the same diaphragm, the output signal shows a change only in the amplitude without a frequency change compared to the input signal.

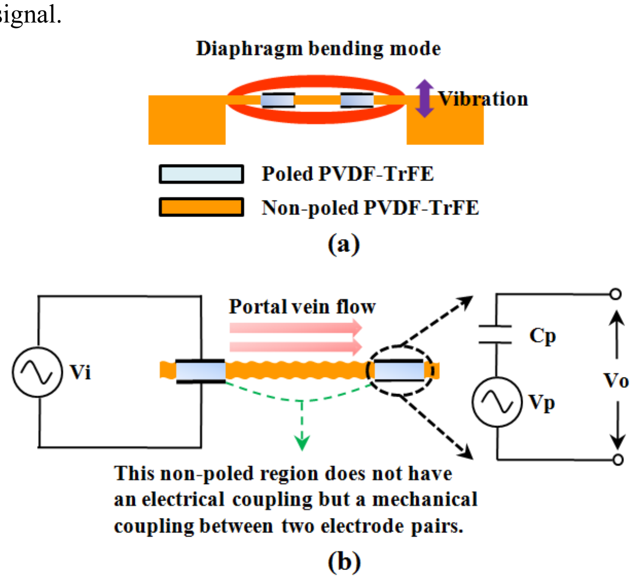


Fig. 2. Sensor structure and driving principle: (a) The regions where input and output electrodes are located are poled and fabricated as a diaphragm type and (b) Illustration of the sensing principle. The contents of the PVDF-TrFE piezoelectric thin film at the output electrode are represented as an electrical equivalent composed of Vp, the piezoelectric voltage source which is directly proportional to the applied deformation on the diaphragm and Cp, the capacitance of the film [8].

B. Fabrication

Figure 3 summarizes the fabrication procedures to make a piezoelectric flow sensor in-plane. After fabrication was complete, the PVDF-TrFE piezoelectric thin film was spirally-rolled following the method described in [10] to make the device as a tube.

PVDF-TrFE solution was made by melting 30 gram of PVDF-TrFE powder (75/25 molar ratio, Measurement Specialties, Inc.) into 100 ml of dimethyl acetate. The solution was spin-coated on a 3 inch silicon wafer and cured at 120 degree Celsius to achieve a 15 µm thick PVDF-TrFE thin film. Then, 1,000 Å of copper was evaporated on top of the PVDF-TrFE thin film and patterned to form the input and output bottom electrodes. After patterning the bottom electrodes, the PVDF-TrFE thin film was peeled off from the silicon wafer and reversed. The reversed film was metalized by copper evaporation to form the input and output top electrodes. After patterning of the top electrodes, a selective DC poling [11] process was used in order to increase the piezoelectricity of the PVDF-TrFE thin film. The material was poled at 90 °C oven with 70 MV/m of DC voltage for 30 minutes [12]. For the electrical insulation and biocompatibility of the sensor, 2.5 µm of parylene was coated on the device. Finally, the PVDF-TrFE piezoelectric flow sensor was spirally-rolled with an inner diameter of 1 mm and a length of 1 cm. This structure can not only minimize the disturbance of the portal vein flow during a measurement but also ensure its insertion stability.

Figure 4 shows the sensing tube with a luminal piezoelectric flow sensor fabricated on a flexible PVDF-TrFE substrate with spiral-rolling. Figure 4 (a) shows the top view of the fabricated device. The top and bottom electrodes of each inner and outer electrode pair are aligned on the PVDF-TrFE thin film. Figure 4 (b) illustrates how the plane PVDF-TrFE film with a flow sensor is spirally-rolled to form the tube shown in Figure 4 (c).

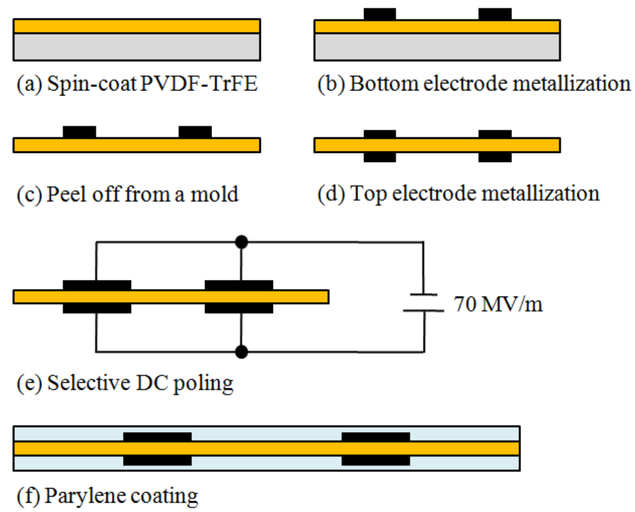
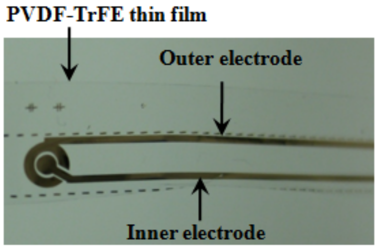
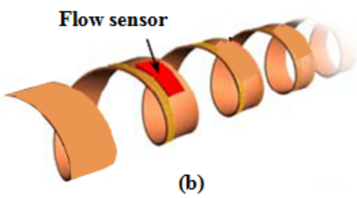


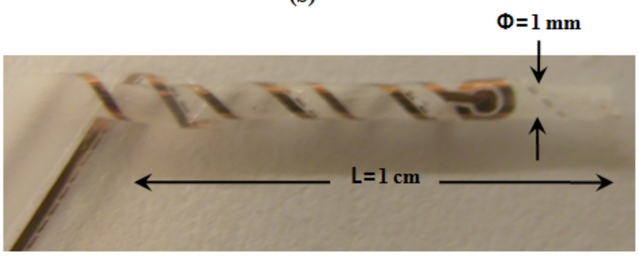
Fig. 3. Brief summarization of the fabrication procedures to make a piezoelectric flow sensor with a PVDF-TrFE thin film through microfabrication methods [12, 13].



(a)



(b)



(c)

Fig. 4. Fabricated sensors: (a) Top view of the PVDF-TrFE thin film after completion of the plane fabrication. Inner electrode has a diameter of 0.75 mm and the outer electrode has a diameter of 1.7 mm. The gap between inner and outer electrodes is 0.1 mm. The width of the PVDF-TrFE thin film is 2.225 mm to get a tube diameter of 1 mm after spiral-rolling; (b) Illustration of spiral-rolling [10]; and (c) Spirally-rolled PVDF-TrFE thin film with a tube form which has a diameter of 1 mm and a length of 1 cm.

III. Experimental Results

In this work, the amplitude changes of output signal, which has same frequency as input signal, were measured at different flow rates. The piezoelectric flow sensor was placed in a PTFE tube with 9 mm diameter and 6 cm length, which mimics the physical feature of the portal vein.

In order to determine a suitable frequency for the input signal, frequencies from 1 kHz to 10 MHz were swept through a function generator (Agilent, 33120A). As the frequency of an input signal was increased from 1 kHz to 10 kHz, the output signal amplitude was increased. However, the output signal amplitude remained almost the same from 10 kHz to 1 MHz. Moreover, the output signal amplitude started to decrease at frequencies above 1 MHz as shown in Figure 5. In addition, the fundamental resonance frequency (f_r^B) for the bending mode can be written as [14]

$$f_r^B = \frac{\lambda^2 t}{4\pi r^2}\sqrt{\frac{E}{3\rho(1-\nu^2)}},$$

where λ^2(=10.22) is the natural frequency constant of a circular diaphragm, t and r are the thickness and radius of the diaphragm, and ρ, ν, and E are the density, Poisson's ratio, and Young's modulus of the piezoelectric thin film under open circuit condition, respectively. The calculated fundamental resonance frequency of the fabricated piezoelectric sensor is around 8.5 kHz.

Overall, the input signal frequency of 10 kHz was selected to make a simple and inexpensive device while maximizing the bending amplitude of the film.

Concerning the input signal amplitude, when the input signal amplitude was less than 1 Vpp, the output signal was not measurable because of relatively high 60 Hz common mode noise. The amplitudes of output signal were almost linearly proportional to the amplitudes of input in range of 1-4.8 Vpp. However, once the amplitude exceeded 4.8 Vpp, the output signal became unstable.

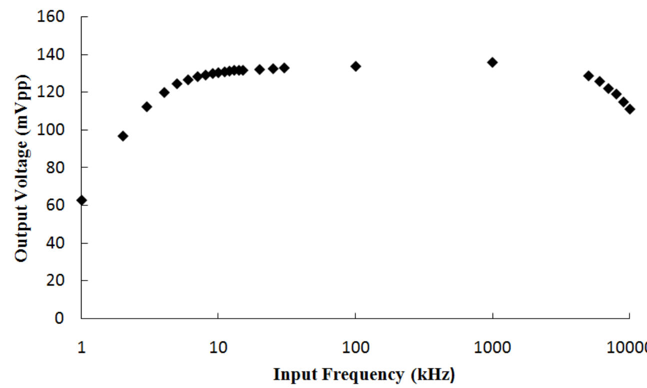


Fig. 5. The output voltage corresponding to the change of the input frequency from 1 kHz to 10 MHz. The abscissa is plotted in log scale.

Thus, the signal with a frequency of 10 kHz and amplitude of 4.5 Vpp was chosen and constantly applied to the input electrode. Then, the amplitude and frequency of the signal detected at the output electrode was measured while varying the flow rate of the water.

The frequency of the output signal exactly followed the frequency of the input signal at all flow rates. This verifies that the input and output electrode pairs vibrate at the same frequency as they are in the same piezoelectric diaphragm.

The amplitude of the output signal decreased as the flow rate was increased. This showed that the strength of the mechanical vibration induced from the input frequency signal was decreased by the disturbance from the flow. This disturbance was increased as the flow rate increased.

The piezoelectric flow sensor realized in this work has two pairs of electrodes. Figure 6 shows the experimental result when the input signal was applied to the inner electrode and the output signal was measured at the outer electrode. Because the frequencies at the output electrode were all identical at all flow rates, only the amplitudes of the output signal are plotted on the graph.

The flow rate was varied with a liquid pump (Cole-Parmer, No. 7553-70) and a flow controller (Cole-Parmer, No. 7553-71) in the range of the portal vein flow rate and the output signal was measured without amplification. When the input signal was applied to the inner electrode, the output showed sensitivity of 0.02 mVpp per 20 mL/min with the linear coefficient of R^2=0.983.

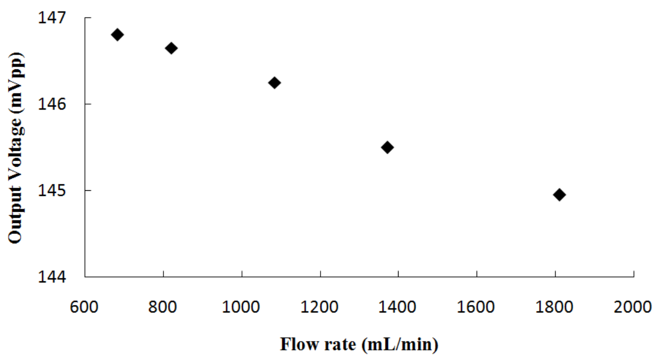


Fig. 6. Measured flow rates when an input signal was applied to the inner electrode and an output signal was measured at the outer electrode.

The sensitivity of the fabricated sensor can be improved using signal amplification and noise filtering circuits. The resonant frequency of the fabricated PVDF-TrFE thin film can be applied [15] to further increase the sensitivity by maximizing the mechanical vibration of the diaphragm-type piezoelectric flow sensor.

The flow rate needs to be monitored during liver transplantation. Thus, the operational stability of the sensor is also characterized. Figure 7 shows the amplitude changes of output signal for the monitoring of 120 minutes at a fixed flow rate of 1,000 mL/min. Little changes on the amplitude of output signal were observed with a standard deviation of 0.084 mVpp.

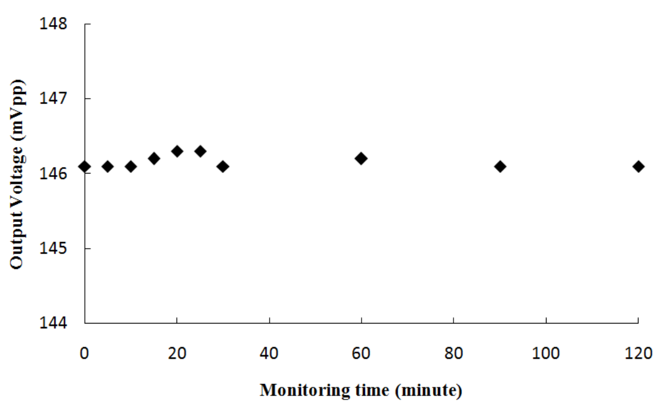


Fig. 7. Operational stability of the fabricated PVDF-TrFE piezoelectric flow sensor. The output was continuously measured for 120 minutes at 1,000 mL/min.

IV. CONCLUSION

A new sensing tube integrated with a piezoelectric flow sensor was developed and fully characterized in this work. In order to satisfy the unique characteristics of portal vein flow, a PVDF-TrFE piezoelectric thin film was utilized for the development of a piezoelectric sensor. Flow rates were measured by applying an excitation signal to the sensor input in order to induce vibrations on a sensor diaphragm. The corresponding output signal was affected by liquid flow and detected by a direct piezoelectric effect.

The developed flow sensor successfully measured flow rates with the sensitivity of 0.02 mVpp per 20 mL/min in the range of portal vein flow rate, achieving excellent operational stability for 120 minutes with a standard deviation of 0.084 mVpp.

These experimental results showed that this sensing tube can envisage a simple but innovative flow sensor for liver transplantation. With the development of a reliable measurement of the portal vein flow using this sensor, an operative strategy that assures adequate portal flow can be developed and planned as well.

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