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Fundamental Study on Medical Tactile Sensor Composed of Organic Ferroelectrics

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***Abstract*—In the present study, we investigated the piezoelectric properties of organic ferroelectrics for use in medical tactile sensor applications, especially for intravascular surgery, by means of both experimental analyses and mathematical simulations. First, we measured the piezoelectric properties of the organic ferroelectrics. Second, we investigated the palpation *in vivo* by the tactile sensor composed of organic ferroelectrics using a computer-based surgical simulator to simulate a catheter and a guidewire in blood vessels for treatment of the brain which we have previously developed. We used the experimental results for the simulation parameters. Our catheter and guidewire simulator shows that when the mechanical properties of the blood vessel was changed partially assuming that these changes occur by a disease, the sensor outputs changed. The findings obtained confirm the feasibility of this sensor for improving catheter manipulation and palpation of living tissue.**

***Index Terms*—Electromechanical properties, Organic ferroelectrics, Piezoelectricity, Tactile sensors**

I. Introduction

Organic ferroelectrics, such as poly(vinylidene fluoride) [PVDF] [1]–[11], have been used in a number of commercially available products, such as hydrophones, pressure sensors, pacemakers, pipes for the chemical industry, and speakers, and also as a copolymer with trifluoroethylene [P(VDF/TrFE)] [12]–[14]. They are promising materials for use in tactile sensors due to characteristics such as: 1) high piezoelectric voltage sensitivity; 2) flexibility, thinness and low weight; 3) responsiveness over a wide frequency range; 4) durability and inertness to chemical agents; and 5) lead-free. Based on these attractive features, a number of tactile sensors have been proposed for industrial robots. Moreover, tactile sensors using organic ferroelectrics are promising from the viewpoint of palpation because PVDF tactile sensors have already been developed to evaluate human skin and prostate abnormalities [1].

VDF oligomer is a novel substance with fewer VDF units and a lower molecular weight than PVDF [15]–[22]. A thin film can be prepared by the vacuum evaporation method and the piezoelectric constant of VDF oligomer is larger than that of P(VDF/TrFE) [18], [19]. Therefore, it may be possible to miniaturize PVDF-tactile sensors by replacing PVDF with VDF oligomer and glue the sensor on a catheter tip for the application of minimally invasive surgery. *In vivo* measurements would bring very large advantages because most living tissue changes its mechanical property once it is removed from human body. If an untouchable part can be measured using the miniaturized tactile sensor quantitatively, it will become possible to obtain a new knowledge about the living tissue and establish a new diagnosis.

Therefore, in the present study, we investigated the piezoelectric properties of organic ferroelectrics for use in medical tactile sensor applications, especially for intravascular surgery, by means of both experimental analyses and mathematical simulations.

II. Experiments

A. Materials and Methods

First, we measured the piezoelectric properties of the organic ferroelectrics. The lower aluminum electrode, the VDF oligomer, and the upper aluminum electrode were evaporated, in this order, through a shadow mask onto a substrate. The areas in which the three materials were placed over one another acted as capacitors, in which the charge was produced in proportion to the stress that was applied to this overlapped region. After evaporation, the VDF oligomer film was poled in order to obtain a piezoelectric effect.

The piezoelectric coefficient inherent to each material determines the relationship between the mechanical input and the electrical output. Each direction within the film has a different constant, and the output charge of the sample in the present study is due to a combination of the piezoelectric constants along all directions. Therefore, the output current of a ferroelectric material (I) is expressed as follows:

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$$I = A\left(d_{31}\frac{d\sigma_1}{dt} + d_{32}\frac{d\sigma_2}{dt} + d_{33}\frac{d\sigma_3}{dt} \right) , \quad (1)$$

where A (= 0.062 mm^2) is the area of overlap of the two electrodes, (d_{31} d_{32} d_{33}) is the piezoelectric coefficient for the material, σ_1 is the applied tensile stress in the drawn direction for PVDF, σ_2 is the applied tensile stress in the transverse direction, and σ_3 is the normal stress to the plane of the film. In this study, we directly measured the piezoelectric effect. i.e., applying a stress and measuring the induced charge.

In our previous studies, we measured the d_{33} of the VDF oligomer evaporated on a glass plate using pneumatic pressure [18], [19]. There, as the sample was supported by a metal plate, we assumed that the contribution of sample bending (specifically, elongation and compression along the surface) to the piezoelectric response could be neglected. In this study, therefore, in order to utilize the flexibility of the organic film, we used flexible polyethylene naphthalate (PEN) film as the substrate and measured the d_{31} by applying a stress in the lateral direction. Note that, for a given applied force, the output charge from the film in the lateral direction is much higher than that of thickness direction [2]. This is because the extreme thinness of the film results in much higher stresses being applied to the film and the absolute values of d_{33} and d_{31} for PVDF are similar.

The experimental apparatus used to evaluate the piezoelectricity of the samples is illustrated in Fig. 1. We attached the VDF oligomer sample and a strain gauge on the both sides of a stainless steel plate. The electrodes of the sample were connected to a charge amplifier (gain: 10^8 [V/A]). The edge of the stainless steel plate was clamped between two plates by screws. When the steel plate was deformed by the application of sinusoidal displacement by a vibrator (Asahi factory Corporation), the output of the sample and the strain gauge were acquired using an oscilloscope. By the measurement of the tip displacement, the strain of the plate can be estimated. But, as the frequency increased, several nodes occurred on the steel plate. Therefore, we used the strain gauge in this study.

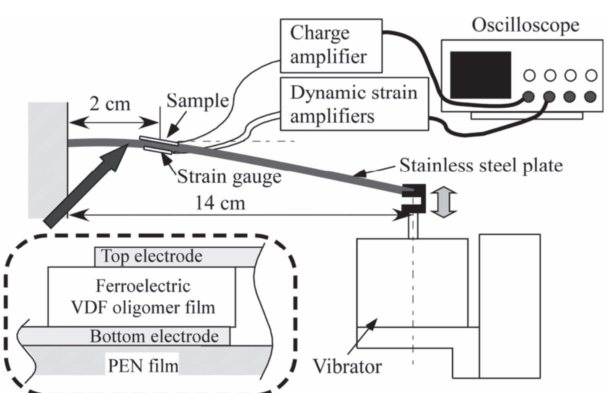

Fig. 1. Experimental apparatus.

B. Results

The resulting piezoelectric characteristics of the VDF oligomer were similar to those reported in our previous study (Fig. 2). Namely, the sample output is proportional to the stress rate applied to the sample. In this experiment, Young's modulus of the VDF oligomer is necessary to calculate the d_{31}. Therefore, it may be more proper to use not d_{31} but e_{31}, because e_{31}, which represents the induced stress by applied voltage, is almost independent of the Young's modulus of the ferroelectric film [23].

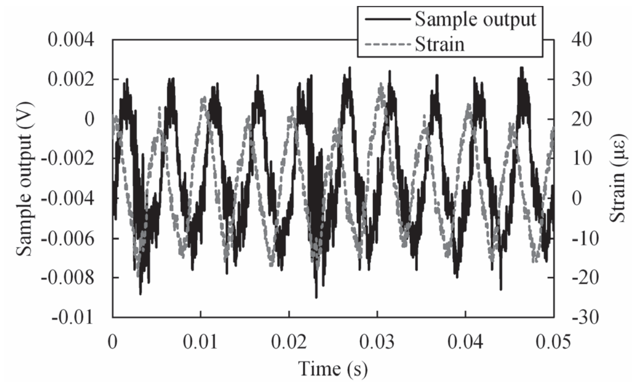

Fig. 2. Example of sample outputs and applied strain.

We can make the prototype by cutting the sample film, wrapping on a guidewire and wiring (Fig. 3). In the following section, we evaluated the performance of the sensor and its measurement principle *in vivo* using our computer-based surgical simulator to simulate a catheter and guidewire in blood vessels [24], [25].

III. Simulation

A. Sensor model

Catheters and guidewires are used in the treatment of infarctions and aneurysms. As the point of insertion is often the thigh, the catheters and the guidewires must be 1 m in length for treatment of the brain. However, the procedure is very difficult due to the small diameter and tortuosity of blood vessels. In addition, surgeon sensory perception (visual and tactile) is severely reduced during manipulation in such surgery, as these tools are long and flexible, and have few degrees of freedom. One method to improve the manipulation ability of the medical device is the measurement of the tactile force between the medical device and the caliber wall. Tactile sensors are also useful to detect the stiffness of tissues, which suffers changes due to diseases.

Furthermore, in our previous study, in order to make intravascular treatment safer, we have also developed a system to simulate a catheter and a guidewire in blood vessels [24], [25]. This system was developed in order to predict the course of approach to a lesion and to present numerical results and animations for surgical planning, intra-operative assistance, and the design of new catheters. On the other hand, to date, few data on the specification of the catheter type tactile sensor have been reported. In this study, we investigated the palpation *in* *vivo* by the tactile sensor composed of organic ferroelectrics using our catheter and guidewire simulator.

Typical catheter-type tactile sensors measure the force by piezoresistence effect, capacitance and the reflection of laser [26]–[28]. Comparing the other methods, one advantage of the use of the organic ferroelectrics is flexibility. As the vessel is tortuous, the intravascular sensor should be flexible. Moreover, when the sensor is used for not the detection of the contact but the palpation, the sensor should be flexible because the contact is necessary for the palpation.

Another advantage of our sensor is that the stress rate can be measured. Measurement of not the stress itself but the stress rate would be suitable for the palpation because it is necessary to measure the difference between the normal and the abnormal diseased parts. On the other hand, it is possible to measure the stress itself by integrating the output current or measuring the charge by changing the preamplifier. But, it has several demerits. Pyroelectricity, which is concomitant with the piezoelectric effect, is sometimes regarded as a disadvantage of ferroelectric materials because pyroelectricity can cause undesired artifacts in the detection of mechanical signals. When the output current is measured, the signal caused by the pressure change can be separated from the signal caused by the temperature change, because the pressure changes much faster than the temperature [3].

As shown in Fig. 3, the detecting areas made of organic ferroelectrics are glued on the side of a guidewire and the sensor output was calculated by our catheter and guidewire simulator. The sensor is bent by the applied force. As the ferroelectric film is much thinner than the radius of the guidewire (for example, the thickness of the VDF oligomer is 192 nm in our previous study [19] and the thickness of the PVDF is 28 [4], 25 [2], and 80 [5] μm), the strain is assumed to be uniform and equal to the strain on the surface of the guidewire. Namely, the uniform stress is applied on the cross section of the ferroelectric film as follows:

$$\sigma_1 = \frac{E_f M_i R_w}{E_w I_w} \quad , \tag{2}$$

where M_i is bending moment of each joint (i), E_f and E_w are Young's modulus of the organic ferroelectrics and the guidewire, respectively, R_w (= 0.17 mm) and I_w are the radius and the area moment of inertia of the guidewire, respectively. In this study, we assumed that A was equal to the cross section of the guidewire (0.091 mm^2). We used the characteristics of PVDF for d_{31} (= 23 pC/N) and E_f (= 3 GPa) [3].

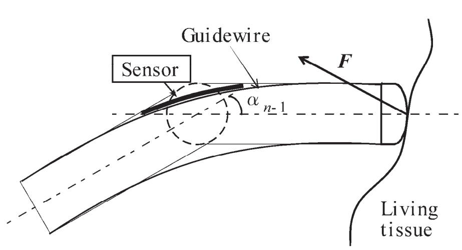

Fig. 3. Sensor model.

B. Catheter/guidewire/blood vessel model

Simulation models used in this study are shown in Fig. 4 [24], [25]. The guidewire model is composed of viscoelastic springs and segments (n = 17). The segment length is equal to l / 2 at both ends and l for the others (l = 2.5 mm). The motion of the guidewire is represented by the Newton-Euler equations of motion. We obtained actual data using a commercial guidewire and used the results in the simulation (density: 5.3 g/cm^3, E_w = 22 GPa). The proximal part of the guidewire is inserted into the catheter model. The inner diameter of the catheter was 0.2 mm larger than the outer diameter of the guidewire. The catheter model was fixed at y = -1 mm and x < 0, and assumed to be a rigid tube. In the guidewire model, M_i is approximated as follows:

$$M_i = \frac{E_w I_w \alpha_i}{l} + D_w \frac{d\alpha_i}{dt} \quad , \tag{3}$$

where α_i and D_w (=0.005 N· m·s/rad) are the bending angle and the damping coefficient of each joint, respectively. In this study, ignoring the second term of the right side, from (1) and (2), the output from the sensor is assumed as:

$$I = A d_{31} \frac{E_f R_w}{l} \frac{d\alpha_{n-1}}{dt} \quad , \tag{4}$$

where α_{n-1} is the bending angle of the joint where the sensor is attached.

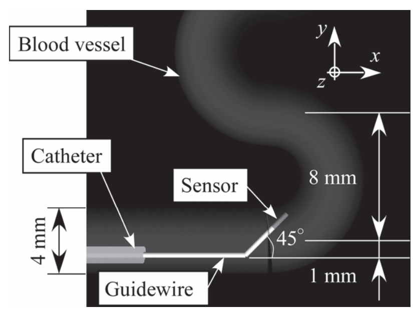

Fig. 4. Simulation model (initial position).

The vessel is a circular elastic cylinder, whose shape is defined by the centerline and the radii. The centerlines are represented by numerical data. The position of the centerline of the vessel is constant. These contact forces between the guidewire and the blood vessel are determined according to the stiffness and friction of the vessel wall. Deformation of the vessel tube in the radial direction is modeled with an elastic coefficient (= 32 N/mm$^{3/2}$). Frictional force is considered using a frictional cone (static friction coefficient: 0.08, dynamic friction coefficient: 0.04). In this study, we also used the blood vessel model where the dynamic friction coefficient was 0.08 (case 1) and the elastic coefficient was reduced by half (= 16 N/mm$^{3/2}$, case 2) at y (of the centerline) = 4 to 6, respectively. These changes may occur by a disease.

In this study, similarly to palpation of the blood vessel, the proximal node of the guidewire model was moved at a constant speed (3 mm/s) in the ± x direction as shown in Fig. 5. To determine whether there is contact between the guidewire and the vessel, the distances between the joints and the tip of the guidewire model, as well as the centerline of the vessel, were calculated.

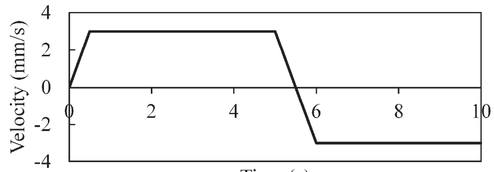

Fig. 5. Motion of proximal part of guidewire model.

C. Results and discussions

Trajectory of the guidewire tip is shown in Fig. 6. As the proximal end of the guidewire was pushed, the guidewire tip began to contact (Fig. 7 (a)). Then, as the whole side of the guidewire contacted along the blood vessel wall, the guidewire tip left the wall (Fig. 7 (b)). After that, as soon as the proximal end was pulled, the guidewire tip began to move backward and contacted at the blood vessel wall again.

The absolute values of the contact force ($|\boldsymbol{F}|$) are shown in Fig. 8. As shown in Fig. 8, when the friction coefficient was large, the contact force also became large (indicated by arrow). The tangential force is calculated by multiplying the normal force and the friction coefficient. Therefore, when the dynamic friction coefficient changed from 0.04 to 0.08 in case 1, the change of the dynamic friction would be 4 % of the normal force. As the normal and the tangential force make the total contact force, the change in Fig. 8 is proper. On the other hand, in case 2, the contact force became small by reducing the stiffness of the vessel wall (indicated by arrow in Fig. 8).

Comparison of the sensor output between basic condition and cases 1 and 2 are shown in Figs. 9 and 10, respectively. It is difficult to detect these changes of the contact force without the sensor because the contact force occurring at not only the tip but also the sides of the guidewire transmit to the surgeon. According to the contact conditions, the sensor outputs changed as indicated by arrows in Figs. 9 and 10. For example, the sensor output from the sensor became large where the friction coefficient was increased (Fig. 9).

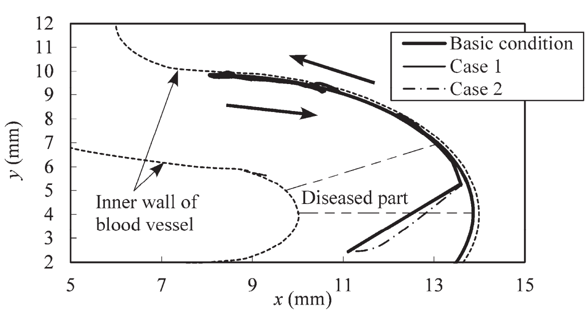

Fig. 6. Trajectory of guidewire tip.

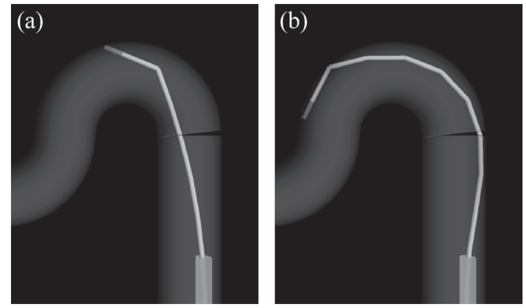

Fig. 7. Appearance of guidewire after insertion. (a) $t = 2.4$ s. (b) $t = 5$ s.

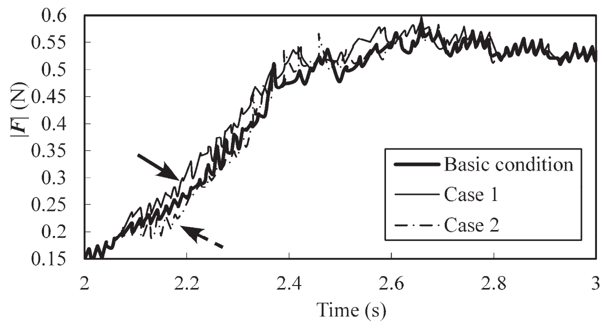

Fig. 8. Close-up view of transition of contact force at guidewire tip. When $t = 2.1$ s in cases 1 and 2, the parameters of the blood vessel contacted with the guidewire tip changed.

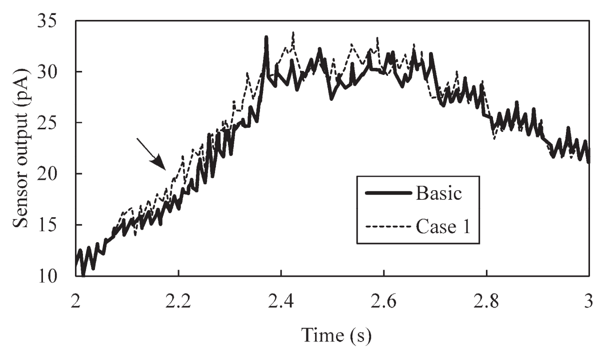

Fig. 9. Comparison of sensor output between basic condition and case 1

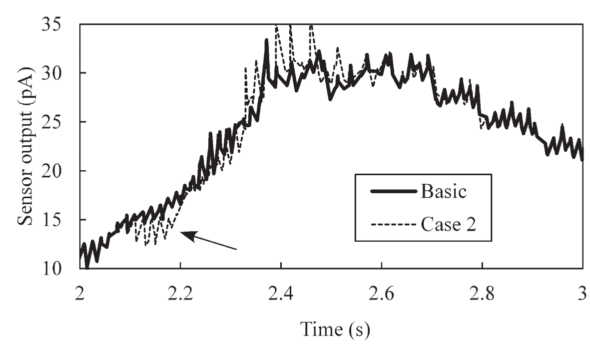

Fig. 10. Comparison of sensor output between basic condition and case 2

IV. Conclusion

In this study, we investigated the piezoelectric properties of organic ferroelectrics for intravascular surgery, by means of both experimental analyses and mathematical simulations. First, we measured the piezoelectric properties of the organic ferroelectrics. Second, we investigated the palpation *in vivo* by the tactile sensor composed of organic ferroelectrics using our catheter and guidewire simulator. The results show the feasibility of this sensor for improving catheter manipulation and palpation of living tissue.

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