MICRO ENERGY HARVESTER USING ACOUSTICALLY RESONATING BUBBLE FOR BIOMEDICAL IMPLANTATION

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ABSTRACT

We present a micro acoustic energy harvester utilizing a micro bubble resonator, which is much smaller than existing acoustic resonators. PVDF-TrFE is used as piezo material on a PDMS diaphragm that encapsulates a micro bubble amplifying the structural resonance and thus energy harvesting efficiency. Geometries of the diaphragm and bubble are carefully designed to match and maximize the structural resonance of piezo layer, diaphragm, and bubble. The maximum harvested power is measured at 1.8-µW, which is 7 times higher than the case without the bubble. The current harvester can be applied to implantable biomedical devices since acoustic waves of low frequency (< 9-kHz) can reach deeper spots in the human body than other types of waves (ultrasound, electromagnetic wave, etc.).

KEYWORDS

Oscillating Bubble; Resonator; Piezoelectric.

INTRODUCTION

Energy harvesting has seen much interest in the past two decades as the power consumption of MEMS devices is steadily reduced. Harnessing energy from ambient or wirelessly applied sources is attractive as a way to negate the need for batteries in applications where replacement of a battery would be inconvenient, expensive, or impossible, such as in the case of medical implantable devices [1-3]. There are typically two methods to harvest energy biomedically, passive energy harvesting involves utilizing human movement such as heart movement or lung movement as energy source, while active energy harvesting involves using external energy source to transfer energy into human body [4]. Since human body movement usually has a fixed, relatively low frequency range, power generated can be low, and the energy level and frequency is uncontrollable, it is natural to seek active methods. Typically, acoustic energy harvesting has less attenuation compared to other active harvesting methods, and it can reach a further distance within human body. In the case of human tissue, there is a clear trend of reduced attenuation at lower frequency in all ranges studied, from ultrasound down to the audible range [5]. Thus, it should be clearly beneficial for acoustic energy harvesting applications to target lower frequencies.

Piezo material selection

To harvest acoustic wave energy, piezoelectric films are usually used. The equation describing the direct piezoelectric effect, the ability of the material to convert mechanical strain into electrical charge is

$$D_i = d_{ij}\, T_j + \varepsilon^T_{ik}\, E_{k,} \qquad (1)$$

where i, j, and k represent 3 spatial dimensions, D is the electrical displacement, d is the piezoelectric stress coefficients tensor, T is the mechanical stress vector, ε^T is the permittivity tensor, and E is the electric field vector [6]. It is clear from this equation that increasing the stress in the material will increase the power output. There are typically 2 modes under which piezoelectric materials operate, d_{31} and d_{33} mode. Basically, the film is under d_{31} mode when applied stress and harvested electric field is perpendicular to each other, and is under d_{33} mode when they are parallel. Another performance metric of interest is the electromechanical coupling coefficient, k which in essence takes into account that a more elastically compliant material will strain more under a given load, increasing conversion efficiency when compared to a material with equivalent piezoelectric coefficient, d [3].

Several thin film piezoelectric materials have been used for energy harvesting in the past. One of the most popular options is ceramics such as lead zirconate titanate (PZT). The typical advantage of the ceramic materials tends to be much higher piezoelectric d_{33} and d_{31} coefficient [6]. Current research progresses on biomedical acoustic energy harvesting are done mostly with PZT as material [7]. However, the power output of these devices is quite small to this day, and despite consistent improvement over the years, it is still necessary to improve the efficiency of these devices in order to increase viability for use in commercial devices. A common strategy to improve the efficiency is exploring different structural designs beyond the common and easy to implement cantilever [5] or membrane [8]. Different designs may seek to improve the efficiency by inducing greater strain for a given input as in the case of a cymbal design [9] and bi-stable beam [10], or lower the resonant frequency to target ambient vibrations in the case of the arc-based cantilever [11] to list some examples. In addition, PZT contains lead, which is not favorable to implantation devices.

Another typical material usually used for energy harvesting is polymer such as polyvinylidene fluoride (PVDF) and its copolymers [11]. Advantages of the polymer materials tend to be easier processing, biocompatibility, and flexibility. Though the polymers have much lower d_{33} and d_{31} coefficients, their greater flexibility leads to a still lower, but more comparable k_{33} and d_{31} electromechanical coupling coefficients when compared with the ceramics [12].

However, PVDF as a material used for biomedical acoustic energy harvesting has not been developed before, majorly because its energy level is low compared to that harvested from PZT so for the already low acoustic energy level using this material can be challenging. It is noticed by the authors that there have been no or few attempts as to include a resonator in the energy harvesting devices. It is not surprising as in every case found by the authors, the size of the resonator is significantly larger than the energy harvesting device itself, often one or two orders of magnitude when comparing the largest dimension and several orders of magnitude when comparing the smallest dimension (thickness). Typical acoustic resonators often take the shape of a tube resonator [13] or a Helmholtz resonator [14]. More recently, the sonic crystal has seen some interest for this purpose [15]. It is then critical to have a small sized resonator to improve power generation for biomedical use. Therefore, this study targets on using polyvinylidene fluoride trifluoroethylene (PVDF-TrFE) and developing a small-scale resonator to improve power efficiency while keeping its biocompatibility, flexibility and ease of fabrication which are deemed important for the intended application in medical implantable devices [16,17].

Acoustic bubble

Acoustic bubble is a common tool used in various microfluidics applications. Under different frequencies, bubbles go through volumetric or translational motion while generating an acoustic streaming field around its vicinity.[18] Acoustic bubbles were used to pick up objects [19], rotate them [20], induce strong mixing in flow [21], pump liquid [22], deliver drug [23] and generate propulsion [24]. It was observed in these previous experiments that polymer bodies of bubble undergo oscillation upon application of acoustic field just as the liquid-air interface. It is then natural to wonder whether or not this vibration on bubble surfaces can be used as vibration source needed for acoustic energy harvesting applications. Previous study has demonstrated on a commercial cantilever beam that bubble oscillations can be used as energy harvesting vibration source [25]. But no comparison was done on whether including a bubble in acoustic harvesting system would enhance power harvesting performance. In this article, an attempt is made toward developing a small-scale bubble-based resonator that operates under low frequency range (~5kHz) with PVDF-TrFE as harvesting material. A comparison is made between harvesters with or without bubble resonator.

DESIGN AND FABRICATION

Design idea of the device is to oscillate the piezoelectric film by an oscillating bubble so as to enhance the strain experienced by film and thus the power output. The system consists of several mechanical system: the acoustic bubble, the polymer diaphragm and the ceramic transducer. In order to maximize the harvested power, it is necessary to predict each resonant frequency according to geometry and make them match. For the bubble, resonance is predicted with Minnaert equation [26]:

$$f_0 = \frac{1}{2\pi}\sqrt{\frac{3\kappa P_0}{\rho}\frac{1}{R_0}} \qquad (2)$$

where f_0 is the resonant frequency of bubble, κ the heat capacity ratio, P_0 the pressure the bubble is under, ρ the density of fluid, and R_0 the radius of bubble. Bubbles are built into cylindrical shape with the similar diameter and height to spherical bubble radius so that resonant frequencies of these bubbles are assumed to be similar to that of spherical bubbles with the same volume. The resonant frequency of diaphragm is predicted with Ansys modal analysis while the transducer resonance characteristics is given by the manufacturer product datasheet. The designed resonant frequency is then aimed at 4.6 kHz, the round diaphragm dimension is calculated to be 1 mm in diameter and the bubble dimensions are designed to be 1 mm in diameter and 1.5 diameter in depth as shown in Fig. 1. In order to distinguish the difference of having a bubble on backside of diaphragm or not, a comparison is done between different cases of bubble arrangement. As is shown in Fig 1. (a) three bubble setups are tested and compared.

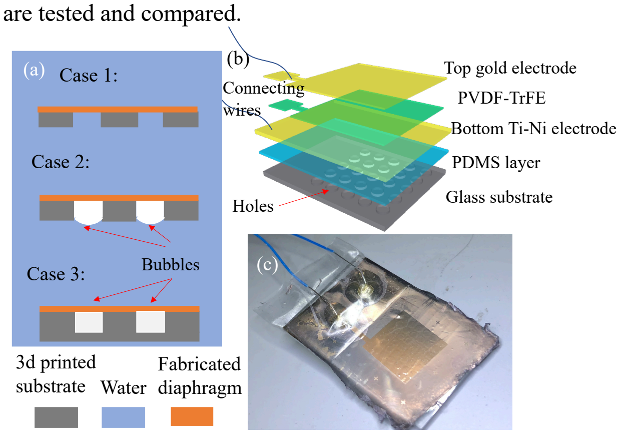

Figure 1 Device designs. (a) 3 different bubble configurations for comparison. (b) 3-D layer-by-layer device structure. (c) fabricated device.

Fabrication of the device consists of two parts, the substrate with cavities and a flexible PDMS film with PVDF-TrFE harvesting layer on top. Substrates was fabricated by laser cutting a glass substrate for case 1 and SLA 3d printing for cases 2 and 3. Process steps for flexible membrane are shown in Fig. 2, a silicon wafer was first coated with silane for easy PDMS detachment. Sylgard 184 PDMS and curing agent were mixed at ratio of 10:1 and spin coated on wafer at 4000 rpm, which gives 20 µm of PDMS layer. PDMS layer was then treated with oxygen plasma for 60s for better electrode adhesion. Then a layer of titanium and a layer of gold was e-beam evaporated onto PDMS. PVDF-TrFE (Solvene 300/P300 Sigma Aldrich) was dissolved in MEK solution (78-93-3 Sigma Aldrich) at 1:10 mass ratio and spin coated at 1500 rpm, thickness of this film is 3 µm. Another layer of gold is then e-beam evaporated and patterned with lithography. Excess PVDF-TrFE and photoresist were etched away with oxygen plasma using RIE. Electrodes were connected to lead wires with silver paste. The whole device is then coated with 2 µm parylene as insulation layer. Flexible membrane layer is then peeled off from wafer and placed on top of plastic substrate.

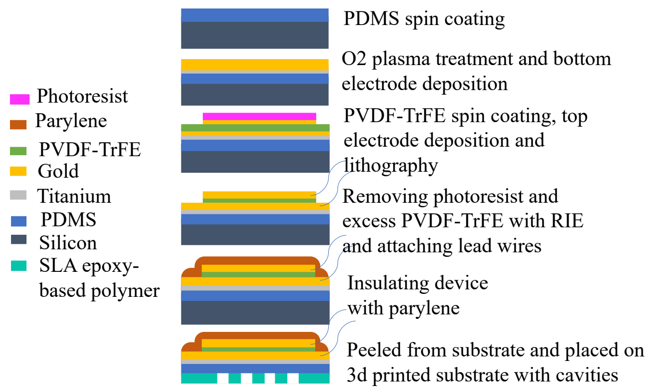

Figure 2 Fabrication process of the device.

RESULTS

Experimental setup is shown in Fig. 3. Comparison experiments are done in an acrylic tank, and the function generator outputs a 100 $mV_{p\text{-}p}$ sinusoidal signal with varying frequencies. The amplifier amplifies the signal 200 times, which becomes 20 $V_{p\text{-}p}$ and is fed to a ceramic piezoelectric actuator suspended in the tank. The fabricated energy harvesting device is put into the tank 5 cm away from the input actuator. The output from the harvester is read with a customized circuit with a load resistor and an oscilloscope.

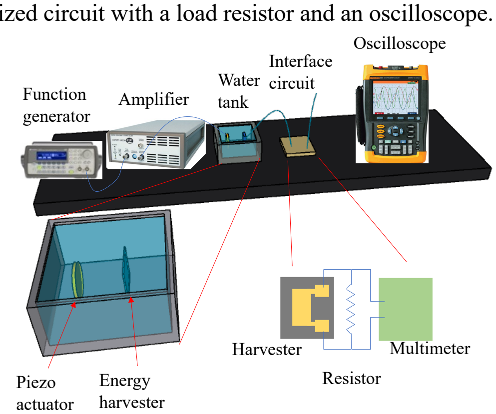

Figure 3 Experimental test set up.

A comparison study is carried out with 3 different cases shown in Fig. 1 (a), with all other parameters fixed. The power output is shown in the Fig. 4. Adding a bubble at the backside of the diaphragm significantly increases the power output, and whether the bubble is open to liquid or closed does not affect the power output level.

In order to verify that the increase in harvested power originates from amplified diaphragm oscillation, an LDV (laser doppler vibrometer) system is then used to measure the vibration amplitude of the diaphragm.

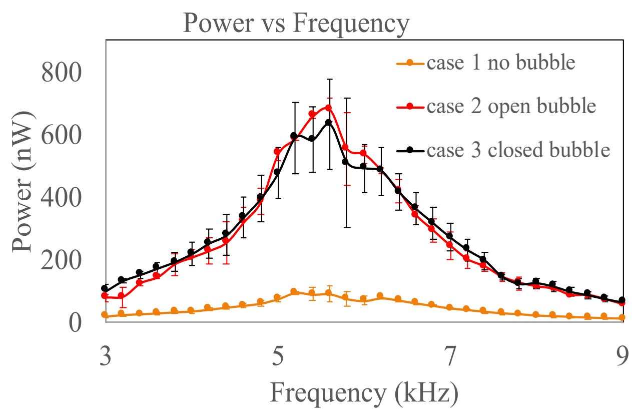

Figure 4 Power output comparison result.

The deflection of the diaphragm is measured to be increasing from 110 nm to 350 nm (Fig. 5 (a)) when a bubble is added (from case 1 to case 2 or 3). It is clear that the increase in oscillation amplitude is responsible for the increase in voltage output and thus the electrical power being harvested. A numerical calculation is carried out to verify that the deflection can indeed generate the voltages measured in our experiments. Ansys package is used to find the stress on the film when the diaphragm deflection is 111 nm and 350 nm. Deflections and the corresponding stress distributions are shown in Figs. 5 (b) and (c). Since the deflection amplitudes are similar for cases 2 and 3, they are considered as the same scenario in further simulation where the deflection of the diaphragm is large. The stress is simulated to be 0.5 MPa and 1.5 MPa, respectively when the deflections are 110 and 350 nm. Then, the obtained stress is taken into equation: $V = g_{31}\ \sigma_3\ t$ to calculate for the generated voltage. In the equation, V, g_{31}, σ_3, t represents the voltage output, the piezoelectric voltage constant, the applied mechanical stress, the thickness of film, respectively. The piezo voltage constant is 216 mV·m/N. The voltage output from the film is calculated to be 49 mV and 166 mV, respectively for small and large deflection. These matches our experimental results, which further confirms that the increase in voltage is due to the increase in the amplification of oscillation amplitude.

In order to find out the maximum output power of the device, impedance matching process is carried out by sweeping load resistor value from 1.5 to 172 kΩ. The voltage and power output is shown in Fig. 6 (a). It is found that 120 kΩ is the impedance of the device, and maximum power can reach up to 1.8 µW.

Furthermore, the effect of distance between the input actuator and harvester is also studied, power output as a function of distance is shown in Fig. 6 (b). The harvested power rises slightly as harvester moves further away from the transducer. In this case, it is found that the distance between the input actuator and harvester is not a critical parameter since the power output remains at same order of magnitude. When the wavelength is much smaller than the present experiment, the wave interference significantly affects the power output, and thus the distance is a critical parameter. However, In the present experiment, the acoustic wavelength is in the range of 10 centimeters, meaning that the acoustic oscillation pressure is a bulk oscillation in the tank. The variation in power output may possibly result from the different acoustic level at different tank locations.

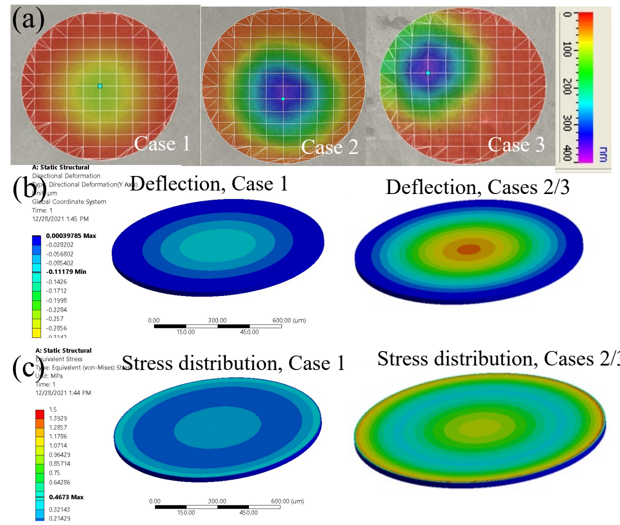

Figure 5 Validation of power output. (a) LDV results of the diaphragm oscillation amplitude of i) no bubble ii) open bubble and iii) closed bubble cases maximum deflection is measured to be 111 nm, 359 nm and 350 nm respectively. (b) Ansys simulation results, i) and ii) is deflection of small/large deflection diaphragm iii) and iv) is stress distribution on the small/ large deflection diaphragm.

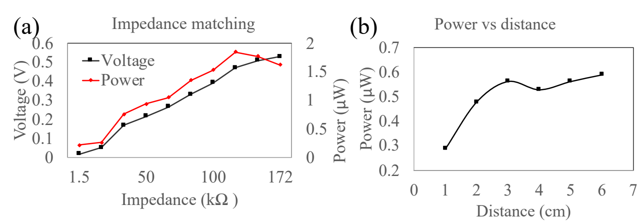

Figure 6 Power output level with (a) impedance (b) and distance between the input actuator and harvester.

The present microscale resonator for acoustic energy harvesting improves the harvested power by 7 times. In order to characterize the space utilization efficiency compared to the existing results, the area power density of the device is calculated by dividing the power density by the pressure level experienced and volume of the whole device. The present harvesting device has the space efficiency of 7.742×10^{-5} µW/cm²·Pa·mm³ while other existing work has the space efficiency of 3.1×10^{-7} µW/cm²·Pa·mm³, 9.6×10^{-10} µW/cm²·Pa·mm³, and 3.6×10^{-11} µW/cm²·Pa·mm³. [13-15] This clearly distinguishes our work in space utilization as bubble size is much smaller than other types of acoustic resonators. We hope that this method will not only improve acoustic energy harvesting process, but also improve the acoustic energy transduction and receiver field.

CONCLUSION

In this paper, we have demonstrated the possibility of an acoustic bubble resonator with much smaller size than typical acoustic resonators. PDMS is used as substrate while PVDF-TrFE as piezo material. With this resonator, the power output level of this diaphragm-based polymer energy harvester was improved by 7 times than the case without the bubble resonator. This bubble resonator design can be further utilized in biomedical acoustic harvesting devices.

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