Development of a Self-Assembled Muscle-Powered Piezoelectric Microgenerator

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ABSTRACT

As technology advances into a new era of N/MEMS, powering these tiny devices becomes a significant issue. To address this issue, a self-assembled living/nonliving microgenerator has been developed through seamless integration of PVDF-TrFE and neonatal cardiac muscle cell. This microgenerator system can covert glucose into electrical energy, which is of particular interest in engineering applications. Since glucose is ubiquitous inside our body, the microgenerator is especially suitable for powering implantable devices. The calculated peak voltage is approximately 0.45V for each hybrid device with the dimensions 1 mm length, 50μm width, and 10μm thickness. Through monolithic integration with integrated circuits (IC), an integral system of novel BioMEMS can be built that carries potential applications in both health monitoring and biosensing.

Keywords: piezoelectric microgenerator, self-assemble, muscle, PVDF-TrFE

1 INTRODUCTION

The increasingly smaller dimensions of MEMS challenges scientists to develop novel techniques of powering these devices. Current methods of powering these devices become problematic once they become free standing systems and are no longer tethered to a silicon wafer. This is not just an issue of energy, but also one of integration and miniaturization, all of which must be addressed on the basis of compatibility of the components. This is especially challenging when dealing with MEMS devices to be used in biological systems. On the other aspect, when dealing with the medical applications of MEMS devices, it would be ideal to develop a system capable of utilizing glucose as power source. However, this proposal is complicated by the fact that most current inorganic microdevices are driven by electrical energy.

Piezoelectric materials have been extensively utilized because of their unique capability of converting mechanical energy into electrical energy. As components in engineered systems, biological muscles have unique advantages over other inorganic actuators, such as high power density, utilization of biochemical fuels, and self-assembly from single cells. Seamless miniaturized integration of these two components would ultimately solve the above-mentioned problem and promise us to create novel implantable power generators.

2 BACKGROUND

Piezoelectric materials have been utilized to build a power generation system using, as energy harvest becomes more and more important. Polyvinylidene fluoride (PVDF) and its copolymer PVDF (Polyvinylidene fluoride)-TRFE (trifluoroethylene) are of great interest because of their strong piezoelectricity. PVDF is a semicrystalline homopolymer, which was shown to have a strong piezoelectric coefficient in 1969 by Kawai [1]. Its copolymer PVDF-TrFE crystallizes to a much greater extent than PVDF, yielding a higher remanent polarization, lower coercive field and much sharper hysteresis. After applying a high electric field, called poling, the Coulomb interactions between injected, trapped charges and oriented dipoles in crystals make the polarization stable [2].

When the PVDF-TrFE film is stretched in transverse direction, the dipole crystalline structures realign due to the applied stress and an internal field is created, causing charges to build up on the electrodes. This produces a voltage potential between two electrodes. The generated charge Q and voltage V be written as

$$Q=\sigma A d_{31} \quad (1)$$

$$V=Q/C \quad (2)$$

Where σ is the stress, A is the area of cross section, d_{31} is the piezoelectric coefficient, and C is the capacitance of the piezoelectric film.

Mechanical force, vibration, thermal gradient, and fluid flow have been used as energy sources to activate piezoelectric films to generate electrical energy [3]. However, these devices are not suitable for implantable applications due to their demanding operation conditions. In our research, we provide an alternative and compatible way to activate the piezoelectric film - the utilization of muscle power.

As microcomponents in engineered systems, biological muscles have unique advantages such as high efficiency, utilization of biochemical fuel, and self-assembly from single cells, over other inorganic actuators. Successful integration of muscles with inorganic MEMS promises the capability of creating self-assembled autonomous implantable structures, powered by ubiquitous glucose in the human body.

However, the use of mature muscle tissues in these devices is impractical and inefficient, as the tissues must be dissected and incorporated into each device by hand thus creating crude interfaces between the biological tissues and

inorganic materials [4-8]. Integration of muscle with fabricated structures would be optimally achieved through patterned growth *in situ*.

A novel system for fabricating self-assembled muscle-powered microdevices has been recently developed in our laboratory [9]. This system has enabled us to selectively direct myocyte growth and differentiation into a single muscle bundle *in situ*, attach these functional bundles to MEMS devices and controllably release hybrid devices. Mechanical studies of cardiac myocytes and fabrication of self-assembled hybrid microrobots have been performed. Here we extend this system to develop self-assembled muscle-powered piezoelectric microgenerators.

3 METHODS AND RESULTS

3.1 Cell Isolation and Culture

The neonatal ventricular myocytes of 1-3-day-old Sprague-Dawley rats (NRVMs), which were prepared and isolated as previously described[10], were kindly provided by Dr. Robert S. Ross at UCLA. The cell cultural medium (NRVMs) and conditions also were followed to the previous description except moderate modification [11]. Briefly, prior to the isolated myocytes being plated, the fabricated devices already glued to ordinary culture dishes were warmed to 37°C. The plated myocytes density is 4.6-6.1 million/cm2 and this culture would be kept at a 37°C incubator supplied with 5% CO2 for 2-3 days.

3.2 Device Fabrication

A novel muscle-powered piezoelectric microgenerator has been fabricated. The creation of the device consists of three main steps: microfabrication of the bendable PVDF-TrFE unimorph, selectively poling of the PVDF-TrFE layer, and self-assembly of muscle cells. The bendable PVDF-TrFE unimorph is a multiple-layer beam structure, which rested on a supporting anchor (Figure 1.). For the microfabrication of process, we determine the shape and thickness of each layer using surface micromachining techniques.

After the fabrication of a Si anchor, a thin layer of poly-N-isopropylacrylamide (PNIPAAm) was coated and dried on the wafer as a sacrificial layer. After patterning PNIPAAm through a shadow mask, a 0.7μm-thick parylene layer was deposited as an electrical insulator. A 200 nm thick Au layer was deposited through a shadow mask to form the first electrode. Following the electrode layer, the 10μm PVDF-TrFE was then deposited. A 200nm-thick Au layer was deposited as the second electrode followed by the deposition of a second 1.5μm parylene. A final 300 nm Au layer was deposited, which was utilized as a mask for the subsequent plasma etching to pattern the unimorph, which is the last step of the microfabrication process.

Unlike the conventional procedure for poling PVDF-TrFE, a 10μm PVDF-TrFE layer were poled at 40 V/um electric field, at 65 ^{o}C for 2 hours because of the low glass temperature of PNIPAAm. The poled devices were fixed on a petri dish and followed by the addition of the cell culture medium mixed with the neonatal cardiac myocytes. After 2~3 days culture, the devices were moved to room temperature. Once PNIPAAm was dissolved, the unimorph powered by muscle began to bend, and the generated voltage was measured.

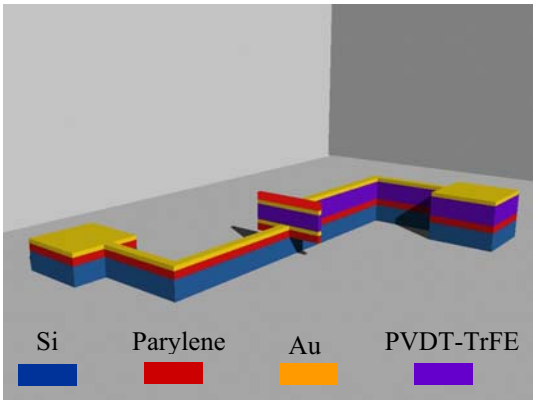

Figure 1. Schematic diagram of a piezoelectric generator. Note: The third Au layer is not shown; and both electrodes are insulated with PDMS prior to the addition of the culture medium.

PNIPAAm, a thermally responsive polymer, has been previously considered as an intelligent substrate to harvest intact cells. A solid at temperatures greater than 32°C, PNIPAAm undergoes a solid-liquid phase transition as it is cooled to lower temperatures and can dissolve in a surrounding liquid medium. Therefore, after the removal of PNIPAAm, our device was a double-ended cantilever beam, which is enclosed by parylene and unpoled PVDF-TrFE (Figure 2).

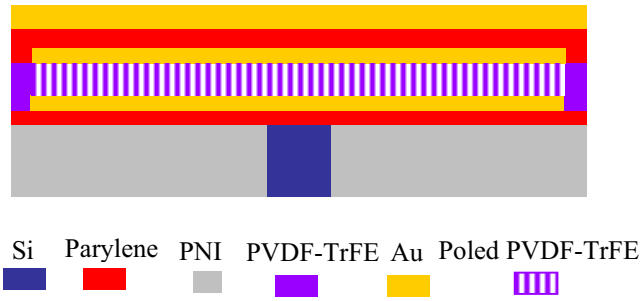

Figure 2.Cross-section of a piezoelectric generator. Muscle bundle self-assembled on the top of Au film is not shown.

Parylene was chosen as an insulation layer because of its unique capabilities of water-resistance, chemical and electrical inertness, large yielding strain, and low Young's modulus [12]. Since our device is the unimorph and requires an elastic layer to shift the stress distribution, the

second parylene layer was deposited thicker as part of the elastic layer. When the self-assembled muscle bundle contracts, it introduces a bending moment and cause the unimorph to bend up. The upper layer of the unimorph is subjected to a compressive stress while the bottom layer is tensional stressed. Due to the stress change, the piezoelectric layer will parasically converts the mechanical energy into electrical energy.

3.3 Results

When applying an external moment M from the muscle, the unimorph will generate the charge and voltage calculated by the following equations [13-16]:

$$Q = \frac{6d_{31}L}{t_p^{\,2}} \frac{AB(B+1)}{1+4AB+6AB^2+4AB^3+A^2B^4} M \quad (3)$$

$$V = \frac{6d_{31}}{\varepsilon_{33}^X w t_p} \frac{AB(B+1)}{1+4AB+6AB^2+4AB^3+A^2B^4-k_{31}^2 AB(1+AB^3)} M \quad (4)$$

where

$$A = \frac{E_m}{E_p}, \quad B = \frac{t_m}{t_p}$$

E_m and E_p are Elastic modulus of the elastic layer and piezoelectric layer; t_m and t_p are thickness of the elastic layer and piezoelectric layer; w and L are the width and length of the unimorph; ε_{33}^X is the dielectric constant of the piezoelectric material under a free condition, k_{31} is the transverse piezoelectric coupling coefficient.

Based on the assumption that the electrode layers and parylene layers are very thin and can be negligible, we can calculate the charge and voltage generated, around 8.3E-15C and 0.48V respectively, by our device. The device itself is 1000μm long, 50um width and 20um thick. A group of such hybrid devices with various dimensions have been fabricated and the preliminary experimental results matched the theoretical calculation very well.

4 DISCUSSION AND FUTURE WORKS

A novel system for the self-assembly of muscle-powered MEMS devices have been previously reported. Through the manipulation of material interfaces and phase, we are able to effect the self-assembly and growth of muscle fibers from individual muscle cells on micro-size lithographically fabricated structures. Using this system, a number of muscle-powered MEMS devices have been created. In this work, we extended this system to develop a self-assembled muscle-powered piezoelectric generator. Like the other muscle-MEMS devices, this novel self-assembled generator system has the following advantages: non-tethered, simple fabrication, and biologically friendly operating conditions.

Using surface micromachining techniques, we can easily alter the pattern of the gold film, which subsequently determine the dimensions and orientation of the self-assembled muscle bundle. Therefore, the geometries and configuration of the devices can be tailored in series or parallel to produce higher voltage and energy. Finally, since all integration steps including MEMS fabrication, Parylene and PVDF-TrFE deposition, PNIPAAm and Au patterning, and cell self-assembly are relatively independent, a variety of MEMS devices, electronic circuits, and cells can be integrated and packaged, thus showing high versatility and flexibility of our devices. For example, with monolithic integration with IC, an integral system of BioMEMS can be built up, which carries potential applications in health monitoring, and biosensing.

The potential limitation of this piezoelectric generator is its low energy density produced. There are various ways to solve this limitation. The first is to build a piezoelectric bimorph structure, which will reduce the mechanical losses and increase the output power. In addition, the higher crystalline the annealed piezoelectric film has, the higher polarization it can achieve, thus the higher output [15]. Therefore, we can use the poled PVDF-TrFE films to solve the limitation of the low glass temperature of PNIPAAm.

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