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Piezoelectric rubber films for autonomous physiological monitoring systems

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ABSTRACT

We have successfully demonstrated the fabrication of piezoelectric rubber films and their applications in heartbeat sensing and human energy harvesting. To realize the desired stretchability and electromechanical sensitivity, cellular polydimethylsiloxane (PDMS) structures with micrometer-sized voids are internally implanted with bipolar charges, which function as dipoles and respond promptly to diverse electromechanical stimuli. The resulting composite structures behave like rubber (with an elastic modulus about 300 kPa) and show strong piezoelectricity (with a piezoelectric coefficient d_{33} higher than 1500 pC/N). In the prototype demonstration, an autonomous heartbeat monitoring system utilizing stacked piezoelectric PDMS films is demonstrated. While integrated with a 12-μA h solid-state energy storage device and a 552-nW nano-power charge amplifier, the energy harvested from human body is stored and employed to monitor heartbeat. With a working area of 10 cm^2 and a compressive load of 10 kg, it is estimated that a charge of 0.1 μC/cycle can be collected using a single-layered piezoelectric rubber film. Furthermore, multiple layers can be stacked and connected in parallel to magnify the charge output. As such, the demonstrated piezoelectric rubber films, which function as both stretchable energy harvesters and highly-sensitive impact sensors, possess a great potential for the realization of wearable and implantable human physiological monitoring systems.

1. Introduction

In recent years, wearable or skin-attachable electronic sensors have been demonstrated based on a variety of detection schemes [1–3]. These devices could potentially be incorporated into clothing or attached directly to the body, and provide the functions of personal health monitoring and therapeutics. To realize the desired functional and human-friendly devices, it is required to combine the attributes of sensitivity and stretchability—the ability to conform to and cover movable and arbitrarily shaped objects [4–6]. In general, this can be achieved by integrating a number of sensing and detecting elements on various stretchable substrates [7–9]. Despite the potential and high performance of these devices, wearable or skin-attachable sensors still present challenges because the fabrication of these integrated devices would be, in many cases, complicated, expensive and limitedly reproducible. Compared to resistive and capacitive type sensors, sensors made of piezoelectric materials can be considered as active ones, which require no external power source for operation. Furthermore, it is possible to power the devices utilizing the energy harvested from the human body via everyday activities [10,11]. However, materials with high piezoelectricity are often rigid and brittle, while piezoelectric polymers, which are flexible but still not stretchable, usually have low piezoelectricity.

The rapid progress in semiconductor technologies contributes to great reduction of power consumptions in today's electronic products and less reliance on the power supplies from typical rechargeable batteries. Consequently, a variety of energy-harvesting techniques have been developed so that bulky batteries can be replaced with more portable and sustainable power sources for electronic devices [12–14]. Humans are complex biomechanical generators which produce significant power output through their physiological motions [15]. For example, a 68-kg man walking at 2 steps per second can deliver 67-W power to his heel, and the maximum power from heat emission of human body can reach up to 2.8–4.8 W. Therefore, the wearable sensors can be made autonomous by utilizing these available powers from humans, and many strategies for power extraction have been also reported. Thermoelectric generators (TEG) convert heat into electrical energy via the temperature difference between heat sources and the environment on thermopiles [16]. With the assistance of MEMS technologies, miniaturized thermoelectric generators are successfully realized and integrated into wireless body sensors [17,18]. Nevertheless, compared with other heat sources such as boiler and fire, the heat flow from human body is not only limited but also strongly dependent on its surrounding

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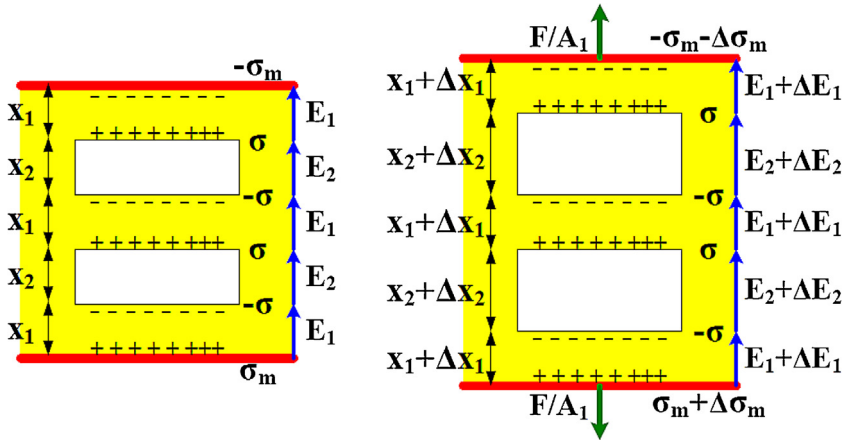

Fig. 1. A simplified model for the piezoelectricity of a charge-implanted cellular structure.

temperature and thus may degrade the power-conversion efficiency of TEG. Besides, the fabrication process of micro-TEG with high power-density is also complicated and expensive. Piezoelectric transducers possess the advantage of simple structure and ease of manufacturing, and the linear relationship between charge generation and applied force with little dependence on temperature variation make them reliable electrical energy sources [10,19–21]. Wearable and implantable piezoelectric generators have been presented [11,22] which can provide sufficient power output for commercial electronic devices. However, these generators, fabricated with typical piezoelectric materials such as PZT or PVDF, lacks elasticity and sensitivity and hence work only when large mechanical stresses are applied. To scavenge energy from slight yet ubiquitous human physiological motions such as respiration and limb movements, properties of stretchability, skin-attachment and high-sensitivity are desired for piezoelectric transducers which impose little load on users and have significant power generation even under the applications of forces with low magnitude and frequency.

To realize the desired stretchable and electromechanically sensitive materials, this work employs a multilayer PDMS fabrication process that can readily produce piezoelectric rubber films with desired cellular structures and electromechanical properties [23]. By utilizing the low elastic modulus of cellular PDMS structures and the high charge density of PTFE films, high piezoelectricity is demonstrated. Furthermore, the piezoelectricity of the PDMS films can be tailored by adjusting the dimensions of the cellular structures. In the prototype demonstration, an autonomous heartbeat monitoring system utilizing stacked piezoelectric PDMS films is demonstrated [24]. While integrated with a 12-μA h solid-state energy storage device and a 552-nW nano-power charge amplifier, the energy harvested from human body is stored and employed to monitor heartbeat. With a working area of 10 cm^2 and a compressive load of 10 kg, it is estimated that a charge of 0.1 μC/cycle can be collected using a single-layered piezoelectric rubber film. As such, the piezoelectric rubber films could function as both sensing and powering elements, and potentially realize the integration of human physiological monitoring and energy harvesting, which is desired for various applications.

2. Operating principle

As illustrated in Fig. 1, the structure is composed of top and bottom electrodes, and alternating three solid and two void layers in between. The application of a force to a charge-implanted cellular PDMS film is expected to change the thickness of the film, and therefore the induced charge density on the top and bottom electrodes as well. Having been demonstrated previously by the authors [23], the effective elastic modulus, $(c_{33})_{\text{eff}}$, of the film is estimated to be

$$(c_{33})_{\text{eff}} = \frac{s_r(3+2t_r)}{[3s_r+2t_r]}c_{33}, \tag{1}$$

where c_{33} is the elastic modulus of PDMS, s_r is the ratio of A_2 divided by A_1 (A_1 and A_2 are the effective areas of solid and void layers, respectively) and t_r is the ratio of x_2 divided by x_1 (x_1 and x_2 are the thicknesses of solid and void layers, respectively). For this structure, the electric fields in the solid layers (E_1) and void layers (E_2) can be determined by Gauss' law for the interfaces

$$E_1 = \frac{\sigma_m}{\varepsilon_1\varepsilon_0} \quad \text{and} \quad E_2 = \frac{\sigma_m - \sigma}{\varepsilon_0}, \tag{2}$$

where σ_m is the charge density on the electrodes, σ is the charge density on void surfaces, and ε_0 and ε_1 are the dielectric constant of air and the relative dielectric constant of PDMS, respectively. Under short-circuit conditions (i.e., $\int Edx = 3x_1E_1 + 2x_2E_2 = 0$), the charge density on the electrodes (σ_m) is

$$\sigma_m = \frac{n\varepsilon_1\sigma x_2}{(3x_1 + 2\varepsilon_1 x_2)} \tag{3}$$

Therefore, the variation of the charge density on the electrodes ($\Delta\sigma_m$) can be expressed as a function of the thickness variations (Δx_1 and Δx_2) as

$$\Delta\sigma_m = \frac{\partial\sigma_m}{\partial x_1}\Delta x_1 + \frac{\partial\sigma_m}{\partial x_2}\Delta x_2 = \frac{6\varepsilon_1 t_r(1-s_r)\sigma x_1}{s_r(3x_1+2\varepsilon_1 x_2)^2}\Delta x_1 = \frac{6\varepsilon_1(1-s_r)\sigma x_2}{t_r(3x_1+2\varepsilon_1 x_2)^2}\Delta x_2. \tag{4}$$

By its definition, the piezoelectric coefficient (d_{33}) of the multilayer structure can be expressed as the applied stress divides the variation of the effective charge density on the electrodes

$$d_{33} = \frac{6\varepsilon_1 t_r(1-s_r)^2\sigma}{s_r(3+2\varepsilon_1 t_r)^2 c_{33}} \tag{5}$$

Assuming that σ, c_{33} are constant, it is found that the piezoelectricity (d_{33}) can be tailored as desired, simply by adjusting the dimensions of the cellular PDMS structure. Moreover, the cellular structure can be considered as capacitors in series and then