Flexible Porous Piezoelectric Cantilever on a Pacemaker Lead for Compact Energy Harvesting

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Self-sustainable energy generation represents a new frontier to greatly extend the lifetime and effectiveness of implantable biomedical devices, such as cardiac pacemakers and defibrillators. However, there is a lack of promising technologies which can efficiently convert the mechanical energy of the beating heart to electrical energy with minimal risk of interfering with the cardiovascular functions. Here a unique design is presented based on existing pacemaker leads tailored for compact energy harvesting. This new design incorporates flexible porous polyvinylidene fluoride-trifluoroethylene thin film within a dual-cantilever structure, which wraps around the pacemaker lead with two free ends sticking out for harvesting energy from the heart’s motion. Under various anchor methods of the lead, the maximum electrical output yields 0.5 V and 43 nA under the frequency of 1 Hz. It is found that adding a proof mass of 31.6 mg on the dual-cantilever tip results in a 1.82 times power enhancement. The scalability of the design is also demonstrated, e.g., by connecting two such units in parallel, their simultaneous vibration can together contribute to energy conversion. Collectively, this study implies that sufficient electrical energy can be converted from the kinetic energy of a pacemaker lead especially at low frequencies to sustain operations.

1. Introduction

The development of implanted biomedical devices for health monitoring and deficiency treatments is essential since they directly affect the lives and safety of humanity. For example, pacemakers and automated implantable cardioverter defibrillators (AICDs) provide effective treatments for arrhythmias, ventricular dysrhythmias, and congestive heart failure.[1] Currently, lithium-based batteries provide the power for operations of implantable biomedical devices, such as cardiac pacemakers and AICD. Although the advances in microelectronics technology reduce the internal current drain concurrently allowing for a smaller volume and greater reliability of implantable biomedical devices, the batteries used in those devices only have a few years operating lifetime. Patients are still exposed to health risks associated with doing periodic surgeries to replace the depleted lithium-based batteries of the implantable biomedical devices. Therefore, energy consumption and battery replacement are key to the lifetime and effectiveness of implantable biomedical devices. This work represents an effort in tackling the challenges in extending the lifetime of the batteries for such biomedical devices as cardiac pacemakers and AICDs.

Given the advances in low power consumption in implantable biomedical devices (such as 0.3 μW for cardiac activity sensing,[2] 10–100 μW for pacemakers,[3,4] 100–2000 μW for cochlear implant,[5] and 1–10 mW for neural recording[6]), it is desirable to make them self-sustainable by having their own renewable power supply. One promising way to provide this alternative power source is by means of energy harvesting, i.e., to convert the source energy to electrical energy in order to power implantable biomedical devices. Emerging new approaches on energy harvesting for powering biomedical devices are discussed in the literature[7–11]. In recent development, in vivo studies have been conducted to further advance the energy harvesting capabilities of these devices in living systems.[12,13] For example, a piezoelectric energy harvesting device using single crystalline (1-x) Pb(Mg1/3Nb2/3)O3-xPbTiO3 (PMN-PT) was implanted into the heart of a live rat and the device was then used to show functional electrical stimulation of the heart.[14] Heartbeats of pigs were also used to power wireless communication systems,[15] of which the integration with energy harvesting devices would allow for further implementation into implantable medical devices. Moreover, implantable triboelectric nanogenerators (ITENG) have been employed to convert the mechanical energy from the contraction from breathing of a rat to electricity, which was then used to power a pacemaker.[16] In another application,
iTENG made use of heart beats to power wireless communication for real-time cardiac monitoring. In addition to the piezoelectric and triboelectric based energy harvesting devices, biofuel cells have been investigated as a method for harvesting energy in vivo. For instance, enzyme modified carbon nanotubes have been used to create a biofuel cell based on the oxidation of glucose within a live rat. Most of the reported in vivo energy harvesting devices were tested in open chest conditions; however, a recent study indicated that significant decreases in energy outputs were found in open versus closed chest heart-powered devices. Therefore, further studies are needed to yield a more comprehensive understanding of the effects of in vivo versus in vitro harvesting capabilities.

Reported implantable energy harvesters are designed to convert various energy sources in the body into electrical power output. Examples include an implantable energy harvester utilizing the pulsation of ascending aorta, a muscle-driven in vivo nanogenerator for harvesting energy from breathing, a piezoelectric nanogenerator that collects the mechanical energy from the legs’ motion, and an energy harvester based on the beating heart by a mass imbalance oscillation. A particularly compelling energy source to power the implantable biomedical devices is the motion of the heart. For instance, a nonlinear energy harvester using the induced magnetic forces was proposed to power pacemakers from the heartbeat. Although their design was suggested to be able to meet the power requirement, the rigid substrate made of brass, and the use of magnets may not be favorable and can interfere with the functions of implantable biomedical devices. Another candidate, nanogenerators based on piezoelectric ZnO nanowires opened new venues for the design of flexible energy harvesting devices and inspired the discovery of improved materials with properties for efficient and high performance of energy conversion. More recently, a flexible lead zirconate titanate (PZT) energy harvesting system was developed to harvest and store energy from motions of the heart and lung. Experiments were performed by affixing the energy harvesting devices onto the epicardium using suturing techniques. While no detectable change in cardiac contraction or epicardia motion was found following this procedure, the need for suturing such devices to the heart inevitably introduces potential risks and unnecessary costs.

Therefore, there is a lack of promising technologies which can effectively covert the mechanical energy of heart to the electrical energy while imposing minimal risks of interfering with the cardiovascular functions. The present work proposed an interdisciplinary approach, combining thin-film energy conversion materials development with mechanical design, to harness the kinetic energy from the lead of a cardiac pacemaker or AICD toward conversion into an electrical power output for biomedical devices. Instead of suturing the energy harvesting device, mounting an “inductively wireless pacing system” directly onto the heart, or using series of magnets placed inside the chest, this new design does not introduce additional loads on the heart and hence will be (or close to) minimally invasive. The energy harvesting device (encapsulated within a soft tube) is compatible with the design of existing pacemaker leads placed for clinical indications or can be developed as a stand-alone device for powering implantable biomedical apparatuses. A shaker-based experimental method was conducted to simulate the heart’s motion based on clinical analyses of heart’s systole and diastole dynamics and anatomical geometries. By using porous polyvinylidene fluoride-trifluoroethylene (PVDF-TrFE) thin films, a flexible dual-cantilever energy harvester is developed to harvest the mechanical energy from the motion of the cardiac pacemaker or AICD lead. Ultimately, such energy harvesting devices will remove the need for additional surgery in order to replace the battery for implantable biomedical devices such as pacemakers.

2. Results and Discussions

A schematic diagram of AICD lead within cardiovascular system by incorporating anatomically based parameters is shown in Figure 1A. The AICD lead moves with the motion of the heart during the cardiac cycle (Movie S1, Supporting Information). To characterize the performance of pacemaker and AICD leads under in vivo conditions, an animal surgical protocol was developed and approved by the Institutional Care and Use Committee at The University of Texas (UT) Health San Antonio. The protocol involves the implantation of prototype leads into a large animal model of congestive heart failure, followed by at least 24 h of continuous data collection. Fluoroscopy images and videos (Movie S1, Supporting Information) were acquired from dogs with chronically implanted pacemaker leads as shown in Figure 1B. During diastolic phase of the cardiac cycle, an angle of $69.4 \pm 1.1^\circ$ was found between the direction along the AICD lead at anchor and longitudinal direction. The AICD lead moves with the motion of the heart during the cardiac cycle, and an angle of $72.85^\circ$ with a lead distal tip displacement of $0.461 \pm 0.065$ cm was observed during the systolic cycle. The AICD lead distal tip travels along the path in the direction of $44.6^\circ \pm 0.68^\circ$ to the longitudinal direction as shown in Figure 1B.

On the other hand, cantilever geometry-based structures for energy harvesting have been explored in the literature given the advantage for targeting low frequency vibrations. Moreover, those piezoelectric cantilever configurations are prone to undergo large strains corresponding to high electrical energies. In the current work, a porous PVDF-TrFE thin film ribbon is wrapped around the AICD lead to form a helical shape but with two cantilever ends sticking out. The helical part of the film is firmly taped on the AICD lead, therefore, the boundary conditions of the dual-cantilever structure are assumed to be fixed at one end of each cantilever in both FEM simulations and experimental tests. A schematic diagram of a dual-cantilever energy harvester on an AICD lead within cardiovascular system is shown in Figure 1A. By utilizing the kinetic energy of the AICD lead, the dual-cantilever piezoelectric energy harvester bends with the vibration of the AICD lead (see Movie S2, Supporting Information). Practical implementation of the energy harvester on a pacemaker lead will certainly require a way to encapsulate the device to be isolated from the body fluids. One effective way to enclose the energy harvester is to use a soft tube with two caps fixed on the AICD lead as shown in Figure 1C. An additional consideration
in practical terms is to make sure that the two cantilever ends move freely with the AICD lead without interfering with the wall of the soft tube.

For the cantilever geometry-based energy harvester, finite element analysis using ABAQUS was performed to study the stress distribution and corresponding bending curvature changes of a cantilever structure. In FEM simulations, a piezoelectric cantilever is modeled as a composite shell with three layers of Kapton film, solid and porous PVDF-TrFE thin films with the boundary condition of one fixed end. Details of the parameters used in the FEM study are given in Table 1. The S4R elements were employed in the FEM simulations. The cantilever structure is subjected to a vibration excitation in the out-of-plane direction, and a dynamic animation of stress distribution of piezoelectric cantilever is provided in Movie S3 in the Supporting Information. The analysis results of stress distributions based on this FEM approach (Figure 2A) and corresponding curvature changes (Figure 2B) quantitatively characterize the deformations in the device. Simulation results suggest that the bending curvature of the piezoelectric cantilever can be increased by putting an additional mass on the free end and tuning the mass location, thus allowing the intrinsic frequency of the cantilever to match the vibration frequency of the lead motion.

Using the experimental methods outlined above, the performance of the energy harvester was evaluated. A schematic of testing conditions of the shaker was shown in Figure 3A. Given that the clinical studies found a lead distal tip displacement of 0.461 ± 0.065 cm during the cardiac cycle (Figure 1B), three different displacements of 3.43, 6.36, and 9.35 mm were examined.

![Diagram with text](image-url)

**Figure 1.** A) Concept of piezoelectric thin film energy harvester for implantable cardioverter defibrillator and a flexible porous PVDF-TrFE dual-cantilever energy harvester on the AICD lead. B) Video analysis of chronically implanted pacemaker lead from a dog. C) A dual-cantilever energy harvester within a soft tube on the AICD lead.

**Table 1.** Material and geometric parameters of piezoelectric cantilever used in FEM model.

<table>
<thead>
<tr>
<th>Description</th>
<th>Value</th>
<th>Units</th>
</tr>
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<tbody>
<tr>
<td>Young’s modulus of Kapton film</td>
<td>2.5</td>
<td>GPa</td>
</tr>
<tr>
<td>Young’s modulus of solid PVDF-TrFE film</td>
<td>1.45</td>
<td>GPa</td>
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<tr>
<td>Young’s modulus of porous PVDF-TrFE film</td>
<td>145</td>
<td>MPa</td>
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<td>Thickness of each layer (three layers in total)</td>
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<td>µm</td>
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<tr>
<td>Width of a cantilever</td>
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<td>mm</td>
</tr>
<tr>
<td>Length of a cantilever</td>
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<td>cm</td>
</tr>
<tr>
<td>The proof mass added on one cantilever end</td>
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<td>g</td>
</tr>
<tr>
<td>Vibration frequency (sinusoidal)</td>
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<td>Hz</td>
</tr>
<tr>
<td>Vibration amplitude (sinusoidal)</td>
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<td>mm</td>
</tr>
<tr>
<td>Angle between cantilever orientation and lead longitudinal direction</td>
<td>45°</td>
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under the excitation frequency of 1 Hz (Figure 3B). The corresponding results of the open circuit voltage and the short circuit current are presented in Figure 3C,D, respectively. The maximum electrical output for one dual-cantilever energy harvester was found; the open circuit voltage (peak to peak) was 0.5 V and the short circuit current (peak to peak) was 43 nA under Case 3 of 9.35 mm displacement. Since the performance of implantable energy harvesters under higher frequencies (>1 Hz) have been pursued frequently in the literature,[22,24,27] few recently reported studies[28,38] focused on low frequencies on the order of 1 Hz. For example, it was reported that the cell generator with 40 cantilevers can output peak voltage of 70 mV,[38] while a 120 beams were connected to generate an open circuit voltage of 3.7 V and a short circuit current of 150 nA.[28] Since the results given in Figure 3 were based on a single element of the dual-cantilever energy harvester, further optimization using multiple harvesting units was expected to generate higher electrical outputs.

The orientation of the anchor for pacemaker or AICD lead on the heart is also important as illustrated by the results in Figure 4. Measurements of the anchor effect were performed by changing the anchor angles, which were defined as the angles between the direction along the AICD lead at anchor and longitudinal direction. Given that the limit of anchor angle is 69.4° based on clinical data of cardiac motion (Figure 1B), three different anchor angles (70°, 80°, and 90°) were examined.

Figure 2. Simulation results from ABAQUS of A) stress distribution as a function of time and B) bending curvatures of a piezoelectric cantilever model subjected to various vibration excitations.

Figure 3. Evaluation of a flexible porous PVDF-TrFE dual-cantilever energy harvester performance: A) schematic of energy harvesting device and testing parameters including the displacements from the shaker motion and the electrical output from PVDF-TrFE energy harvester; B) various displacements of the shaker motion at 1 Hz; experimental results of C) open circuit voltage and D) short circuit current as a function of time under various displacements.
Figure 4 shows the analysis of energy harvesting performance of the device at various anchoring angles but under the same load displacement of 6.36 mm. The 90° case produces larger voltage and current than those at 70° and 80° anchoring angles. While a complete explanation for the anchor effect on the changes of the motion of AICD lead in bodily fluids is outside the scope of the current work, it appears that those changes can further affect the electrical outputs of energy harvester. In practical terms, implantation techniques and anchor methods (actively or passively attached to the interior heart wall) have significant impact on the orientation of anchor for pacemaker or AICD leads.

As is shown in Figure 5, the cycling stability of the energy harvesting device is also tested. Under an input from the shake of 6.36 mm displacements at 1 Hz, the corresponding voltage output of energy harvesting is very stable without degradation over 10^4 cycles.

In addition, an experimental study of the effect of the heartbeat on energy harvester performance has been performed. The frequencies were sweeping from 0.5 to 2 Hz, corresponding to the heartbeat of 30 beat per minute (bpm) to 120 bpm. As can be observed in Figure 6, both voltage and current increase with the rise of frequencies even at the same displacement of 6.36 mm. This trend is because a higher frequency at the same displacement corresponds to an augmented kinetic energy from the AICD lead; as a result, at a higher frequency within the studied range the cantilever-based energy harvester has an enhanced energy output featuring both increased voltage and current.

Additional characterization of charging ability of the proposed dual-cantilever energy harvester is also performed. By using a Schottky bridge rectifier (MB12S, Micro Commercial), the alternating current generated from the piezoelectric energy harvester converts to a direct current signal as shown in Figure 7A,B, and a peak voltage output of 0.14 V after rectifying can be obtained. A sawtooth voltage curve is measured by adding a 10 µF capacitor (in Figure 7C), since the capacitor only has time to discharge briefly before the next DC voltage recharges it back up to the peak value. Furthermore, the charging ability is tested by connecting the energy harvester with a 10 µF capacitor and 0.91 MΩ resistor load, and it takes about 20 s for the dual-cantilever energy harvester to charge 0.13 V from the kinetic energy of the AICD lead, implying a significant potential for charging microbatteries.
Expanding upon the evaluation of energy harvester performance, two enhanced methods for the proposed dual-cantilever energy harvester were further investigated: 1) two harvesting units were connected in parallel to increase the electrical output (Figure 8A–C) and 2) an added mass is attached at the free end of cantilever-based energy harvester (Figure 8D–F). The results in Figure 8C show that the strategy of using two harvesting mechanisms in parallel connection contributed to a higher (doubled current in this case) current output. From a design perspective, a mix-pattern (both in series and parallel) connection could be further pursued using multiple energy harvesting units \((n > 2)\). On the other hand, by appending an additional proof mass on the free end of cantilever, the intrinsic frequency of vibration structure of energy harvesting can be lowered, especially when the target frequency is on the order of 1 Hz. The oscillation of a cantilever with an added mass allows for a larger bending curvature, which results in a higher electrical output of the piezoelectric energy harvester. This expectation is confirmed by the results shown in Figure 8E,F, i.e., both voltage and current outputs were increased by adding a 31.6 mg proof mass at each free end of dual-cantilever compared to the results without an added mass (Figure 3C,D in Case 2), resulting a 1.82 times enhancement for the power output. Moreover, a coupled mechanical and piezoelectric analysis of the cantilevers will be conducted in future work to optimize the length, width, and electrode area to

Figure 6. The effect of the heartbeat on energy harvester performance: experimental results of A) open circuit voltage and B) short circuit current as a function of time under a shaker displacement of 6.36 mm but with various frequencies from 0.5 to 2 Hz.

Figure 7. The charging ability of the dual-cantilever energy harvester: A) experimental results of open circuit voltage without rectification under a shaker displacement of 6.36 mm at 1 Hz; B) experimental results of voltage as a function of time with rectification; C) experimental results of voltage with a bridge rectifier and a capacitor; and D) charging curve for the energy harvester with a bridge rectifier and a resistor-capacitor (RC) circuit.
3. Conclusions

The rapid development of implantable biomedical devices such as cardiac pacemakers and AICDs demands sustainable power sources to eliminate the need of repeated surgeries for replacing the batteries. However, no existing technologies can perform efficient conversion of the mechanical energy from the beating heart to electrical energy without any significant risks of interfering with the cardiovascular functions. Here we developed a unique design tailored for energy harvesting from low frequency vibrations suitable for powering pacemakers and AICDs. By using porous PVDF-TrFE thin film, a flexible dual-cantilever structure of energy harvester was developed to harvest the mechanical energy from the motion of a pacemaker/AICD lead. The reported results demonstrated the capability of providing electrical energy to assist pacemaker battery functions directly from the kinetic energy of a pacemaker/AICD lead.

4. Experimental Section

Materials and Fabrication Methods: PVDF-TrFE, a polymer-based piezoelectric material, has a higher tensile strength and lower stiffness compared to ceramic-based piezoelectrics, making PVDF-TrFE attractive in biomedical applications. The piezoelectric coefficients of β-phase of PVDF-TrFE were found 20–30 pC N⁻¹ (d₃₃) and 16–18 pC N⁻¹ (d₃₁), which are sufficient to generate useful levels of electrical outputs.[39] Recent studies showed the piezoelectric output has been enhanced by threefold through the optimization of PVDF-TrFE porous structure and electromechanical coupling efficiency compared to solid PVDF-TrFE thin film.[40,41] Specifically, the mechanical flexibility of the porous PVDF-TrFE thin film can be tuned by controlling the porous structure, such as the pore diameter and porosity, through the water vapor phase separation method.[41] Later, a kirigami PVDF-TrFE film was developed to exhibit an extended strain range while still maintaining significant voltage generation when compared with films without cuts.[42] In addition, the excellent biocompatibility of PVDF-TrFE has already been confirmed and verified through in vivo tests in the earlier work.[43] Here, a porous PVDF-TrFE thin film structure is developed to achieve high energy conversion and mechanical flexibility for flexible energy harvesters that can be incorporated into pacemakers for long-term functioning.

Polyvinylidenedifluoride–trifluoroethylene powder (molar ratio: 75:25) was dissolved in N,N-dimethylformamide solvent under magnetic stirring for 8 h at a 15 wt%. The first layer of solid PVDF-TrFE thin film was made through consecutive spin coating (1000 r min⁻¹, 30 s) on a 10 nm gold coated of 25 µm thick Kapton film. The cast sold film was
evaporated in an oven for 10 min at temperature of 50 °C. The spin coat process was repeated to produce the second layer of PVDF-TrFE thin film directly on the first layer of solid PVDF-TrFE film. Then the device was moved into a chamber with 90% relative humidity at temperature of 37 °C for 6 h in order to allow complete phase separation between solvent and nonsolvent. After evaporation, a 10 nm gold electrode was sputter coated on the top layer. The overall thickness of the device was estimated to be 60 µm based on the previous experiments.\textsuperscript{[40,41]} Then the sample was kept in an oven for 2 h at a temperature of 135 °C for annealing in order to increase the material crystallinity. After that, the electrical poling process was performed at 100 °C by applying an electrical field of 80 V µm\textsuperscript{-1} for 30 min. A schematic of porous piezoelectric energy harvester and porous PVDF-TrFE ribbon before wrapping on the AICD lead are shown in Figure 9A,B, respectively. The SEM cross-sectional image of the film is shown in Figure 9C and an average diameter of 8 µm was found for the pores within the second layer porous PVDF-TrFE thin film.

**Experimental Testing Methods:** By incorporating anatomically based parameters,\textsuperscript{[30–33]} a test setup was designed to closely approximate the in vivo conditions (Figure 1A and Figure 10A). A 21 cm long soft tube was used to represent the superior vena cava (SVC) given the distance of 21 cm from the insertion point at the subclavian vein to the tricuspid valve.\textsuperscript{[30,31]} The soft tube had an approximate inner diameter of 16 mm based on anatomical values of SVC diameters of 16–20 mm.\textsuperscript{[32]} A full length of right ventricle (RV) of 75 mm was used in the setup according to the anatomical parameters of 71–79 mm from tricuspid valve to the apex of the heart during diastolic phase of the cardiac cycle.\textsuperscript{[31–33]} However, the RV shortens to around 50 mm during the systolic phase of the cardiac cycle, thus imposing a length constraint on the energy harvesting device to be placed inside the RV.\textsuperscript{[31]} This constraint also limits the position of the energy harvester on the AICD lead as illustrated in Figure 10A.

On the other hand, it is noted that a limitation of the clinical studies is that those clinical data also depend on the slack of the lead and implantation techniques during the surgeries. Furthermore, a recent study also concluded that the RV constraint and load cell placement had the largest impact on the pacemaker and defibrillator leads tips, and thus providing a new insight on how various parameters contribute to possible perforation of the heart tissue.\textsuperscript{[44]} Nevertheless, the clinical studies provide guidance and motivate efforts on in vitro experimental method to evaluate the energy harvester performance. A shaker-based test platform was successfully built as shown in Figure 10B to simulate the motion of the myocardium and the corresponding deformation of a pacemaker lead. The platform was consisted of a shaker (2025E from The Modal Shop), a fixture frame, and an AICD lead (Model 7122Q/52 from St. Jude Medical). This AICD lead was partially within a soft tube in order to mimic the SVC environment, while the distal tip of AICD lead was connected to the shaker to produce a certain displacement and frequency by recapitulating the motion of heart diastole. The input excitation was provided by using a function generator (Keysight 33 500 B) with a power amplifier (2100E21-400 from The Modal Shop). The real time voltage output from the energy harvester was collected using a data acquisition card (National Instruments, model NI USB-6008) and LabView software. This shaker-based test configuration (Figure 10) is considered to be the closest to the in vivo conditions.

**Supporting Information**

Supporting Information is available from the Wiley Online Library or from the author.
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Conflict of Interest
The authors declare no conflict of interest.

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