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Short Papers

Performance Enhancement of Piezoelectric Bending Beam-Based Human Knee Energy Harvester

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***Abstract*—Biomechanical energy harvesters can capture biomechanical energy from human motion and generate electricity for powering wearable and portable electronics. Thanks to that, the widely used body-worn devices can be battery-free and eliminate battery capacity limitations. This article develops a lightweight piezoelectric bending beam-based human knee energy harvester to generate electricity during walking. To enhance the performance, especially in the stance phase, we propose a new design and introduce free vibration in the stance phase to improve the output voltage and power. Specifically, one end of the bending beam is released and moves freely, such that the bending beam experiences free vibration in the stance phase. While in the swing phase, the bending beam still undertakes forced motion. Besides, a proof mass is mounted on the bending beam to tune the free vibration. Experimental results indicate that compared with the previous design, the free vibration can improve the average output power by 52.3%. When walking at 7 km/h on a treadmill, the average power output can reach 43.4 mW. This work significantly facilitates the further development of biomechanical energy harvesters. The output power is improved by 10 to 100 times compared with the existing works.**

***Index Terms*—Bending beam, energy harvesting, free vibration, knee motion, piezoelectric.**

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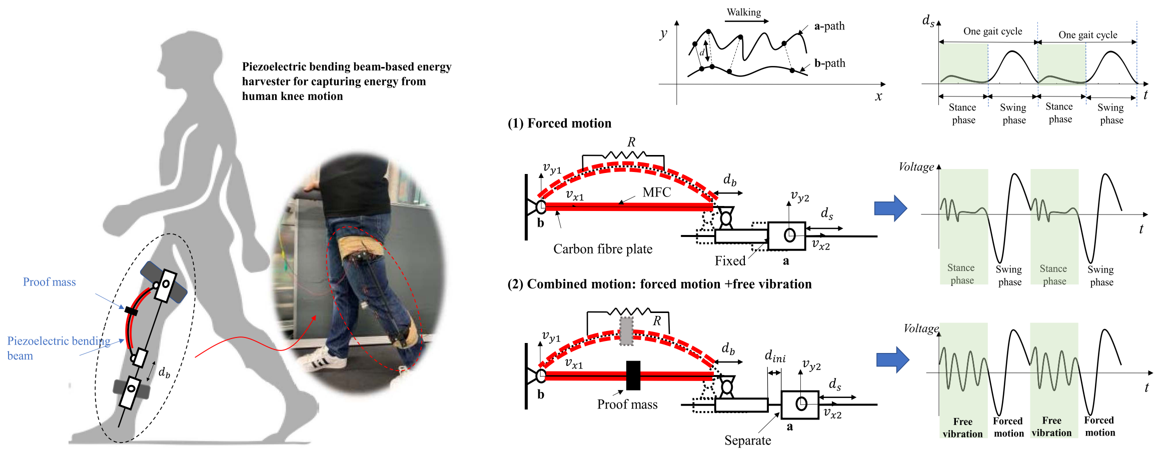
I. Introduction

Wearable and portable electronics, for instance, smartphones, smart watches, wireless headphones, healthy wireless monitors, GPS, etc., are widely used in our daily lives. As most of them are electrochemical batteries-powered devices, battery replacement, and daily charging significantly block the application explosion and decrease user experience. With substantial development in microelectronics and wireless communication technology, body-worn electronics have become smaller and require less power [1], [2], [3]. Owing to that, it becomes possible to obtain substantial electricity sources from the users' bodies or ambiance for powering body-worn devices via exploiting energy harvesting technologies [4], [5], [6]. The body-worn devices can be battery-free and get rid of battery capacity limitations, such as self-powered insole for human motion recognition [7], self-powered smartwatch [4], [8], [9], [10], battery-free neuromodulator [11], leadless intracardiac pacemaker [12], and wireless human-machine interface glove [13].

Energy sources in ambiance and human bodies are available in different forms, including solar, heat, radiofrequency, and kinetic energy [2], [4]. Researchers proposed utilizing different transducers, for instance, thermal transducers and light transducers [14], to capture this energy and transform it into electricity to power wearable electronics. This work mainly focuses on scavenging kinetic energy from human motion. It is noteworthy that the maximum power consumption can reach 60 W during arm lifts, and the power that can be harvested from the heel's fall is 67 W during walking [2]. Scavenging part of the above-mentioned biomechanical energy is a promising solution to the energy issue confronting body-worn electronics.

Biomechanical energy harvesters (BEHs) are devised to extract human bodies' biomechanical energy from human motion and then generate electricity for powering wearable electronics. BEHs can be categorized into four types: electromagnetic energy harvesters [9], [15], [16], [17], [18], [19], [20], [21], [22], [23], [24], [25], piezoelectric energy harvesters [26], [27], [28], [29], [30], [31], [32], [33], [34], [35], [36], [37], [38], [39], [40], [41], [42], triboelectric energy harvesters [43], [44], [45], [46], and hybrid energy harvesters [10], [47], [48], [49]. Shenck and Paradiso [29] proposed a shoe-integrated rotary electromagnetic generator with gear transmission and piezoelectric transducers to extract energy from foot strike and forefoot bending deformation in 2001. With the aim of improving energy efficiency, the authors in [30], [31], and [32] designed different mechanisms to increase the piezoelectric transducers' deformation, thus generating more energy from the heel strike. The authors in [33] and [34] proposed utilizing piezoelectric cantilever beams to harvest biomechanical energy from the shank motion. Kuang et al. [35] developed a rotary piezoelectric knee energy harvester in which several piezoelectric cantilever beams combined with a frequency-up mechanism were employed.



(a) (b)

Fig. 1. (a) Schematic diagram of the energy harvester developed for the human knee. (b) Working principles of the energy harvester. (1) Two ends of the bending beam are jointed to the shank brace, and one end of the bending beam moves along a linear guide. Herein, the bending beam experience a forced motion during the whole walking gait cycle. (2) One end of the bending beam is released and is joint to a slider that moves along the linear guide freely. Here, the bending beam experiences a forced motion during the swing phase. While in the stance phase, the bending beam will undergo a free vibration. Consequently, the output voltage and power can be enhanced compared to the case (1). Besides, a proof mass is employed to modulate the free vibration.

Recently, flexible piezoelectric transducers [36], [37], [38], [39], [40], [41], [42], for instance, polyvinylidene fluoride (PVDF) and micro fiber composites (MFC), are receiving increasing interest, as they can directly capture or comply with the human limbs' or joints' large amplitude motions. Proto et al. [38] integrated a flexible polymetric piezoelectric and PVDF into a tight suit to harness energy from the human body. In this case, the deformation in the PVDF is small, as the substrates (garments) possess relatively low stiffness compared to PVDF. Consequently, the average power output was small, about 0.01 mW. To improve the performance, Kim and Yun [39] developed a stretchable helical piezoelectric energy harvester (HPEH), in which the stretchable piezoelectric was configured like a helical spring, and the two ends of the HPEH were fixed to the garments. The peak power output can reach about 1.4 mW, when installed on the knee joint.

In our previous work, we developed electromagnetic and piezoelectric human knee energy harvesters for extracting biomechanical energy during walking [17], [40], [41]. We tested the harvesters' output power when wearers walked on treadmills and also measured the effect of the harvesters on wearers' walking gait and metabolic cost. The piezoelectric energy harvesters employed a piezoelectric bending beam to generate electricity. Herein, the bending beam consists of a carbon fiber plate (the substrate) and MFCs. During walking, the flexion and extension of the human knee rustled in deformation in MFCs periodically. Consequently, the biomechanical energy is scavenged and converted to electricity. In this article, to further improve the performance, we proposed a new design and introduced free vibration to the piezoelectric bending beam-based energy harvester. Specifically, noted that as the knee flexion angle is relatively small during the stance phase [41], the output voltage and power are significantly decreased with respect to those of the swing phase. To address that, in this study, we introduced free vibration in the stance phase. As a result, the deformation was not induced by the human knee flexion but by the free vibration. In this case, the output power can be enhanced. Finally, to test the performance of the new design, we measured the output power when the user walked on a treadmill.

II. Piezoelectric Bending Beam-Based Human Knee Energy Harvester

Fig. 1(a) shows the improved human knee energy harvester's schematic diagram, and Fig. 1(b) demonstrates the working principles of the energy harvester. During walking, the human knee's flexion and extension drive the shank fixture block (point a) to reciprocate along the linear guide periodically. When the end of the bending beam is jointed to the shank fixture block (in case 1), the bending beam undergoes a forced motion during the whole gait cycle. The bending beam's displacement is equal to that of the shank fixture block. While if the end of the bending beam is separated from the shank fixture block and joined to a slider that moves along the linear guide (in case 2), the end of the bending beam's displacement is no longer only associated with the motion of the shank fixture block. During the swing phase, the human knee's flexion angle is large, which results in a large displacement in the shank fixture block. The shank fixture block drives the slider to compress the bending beam. In this phase, the bending beam undergoes a forced motion. However, in the stance phase, the flexion angle is relatively small, so the shank fixture block's displacement is small. In this phase, the inertial force of the bending beam makes the bending beam undergo free vibration; therefore, the slider separates from the shank fixture block. Herein, it should be noted that owing to the free vibration, the displacement and the deformation are significantly increased. As a result, the output voltage in the stance phase can be improved compared to case 1. Furthermore, a proof mass mounted on the bending beam is employed to modulate the free vibration thus improving the energy harvesting efficiency. This novel design has been disclosed in our US patent application [50].

In this article, we fabricated a prototype, as shown in Fig. 1(a). One end of the harvester is attached to a nylon thigh brace through a pin joint, and the other is attached to a slider that can move freely along a linear guide. The linear guide is connected to a slider which is attached to the nylon shank brace through a ball joint. The prototype's weight is 290 g, including the braces' weight and excluding the proof mass's weight. The bending beam's substrate is a carbon fiber plate (width

× length × thickness: 31 mm × 309 mm × 0.5 mm), and six MFC slices (M8528-P2, Smart Materials Corp.) are bounded to the bending beam. Three of them are attached to one side of the bending beam. MFC slices are jointed to the substrate through AB glue. The thickness of the AB glue is about 0.15 mm. Through calculation, the reaction force and torque between the human body and the harvester for a whole gait cycle are limited to 8.7 N and 0.65 Nm, respectively [41].

III. Mathematical Model

Piezoelectric MFCs can convert mechanical deformation into electricity. The charge Q_h generated on the piezoelectric MFCs is associated with the size, shape, and deformation, as well as the piezoelectric properties, which is given by

$$Q_h = 2\int d_{31} w_b d\left(\sigma_{31}\right) d\left(l_b\right) = 2 l_b d_{31} w_b E_{31} \overline{\varepsilon} \tag{1}$$

$$U = \frac{Q_h - Q_l}{C_h} = \frac{Q_h - \int U/R_l dt}{C_h} \tag{2}$$

where l_b refers to the length of the piezoelectric MFCs. w_b is the width of the piezoelectric MFCs. The piezoelectric MFCs attached to the two sides of the substrate are electronically connected in parallel. d_{31} is the piezoelectric strain coefficient. E_{31} denotes the elastic modulus. $\overline{\varepsilon}$ is the average strain of the piezoelectric MFCs. U refers to the harvester's output voltage. R_l is the resistive load. Q_l denotes the charge passing through the resistive load. C_h refers to the capacitance of the piezoelectric MFCs. The relationship between the piezoelectric MFCs' average strain and the end of the bending beam's displacement can be approximated as a four-order polynomial function [41]

$$\overline{\varepsilon} = -67.11 d_b^4 + 15.29 d_b^3 - 1.319 d_b^2 + 0.0742 d_b. \tag{3}$$

It should be noted that d_b should be limited to 0.0772 m, or piezoelectric MFCs' maximum strain will exceed the maximum allowable strain 0.45% [41]. In that case, piezoelectric MFCs may be broken.

The displacement of the shank flexure block d_s is calculated as follows:

$$d_s = l_t + l_s - \sqrt{l_t^2 + l_s^2 - 2 l_t l_s \cos\left(\pi - \theta_{\text{knee}}\right)} \tag{4}$$

where l_t denotes the thigh link's length. l_s refers to the shank link's length. θ_{knee} is the human knee angle. When the human body stands straight, the knee angle is zero. When the shank flexure block drives the slider to move along the linear guide, the bending beam undertakes a forced motion. Herein, the displacement of the slider d_{b_forced} is given by

$$d_{b\_\text{forced}} = d_s + d_{\text{ini}\_d} - d_{\text{ini}\_g} \tag{5}$$

where d_{ini_d} refers to the initial deformation. In this study, d_{ini_d} is equal to zero. In other words, there is no deformation in the bending beam when the human body stands straight. d_{ini_g} denotes the initial gap between the slider and the shank fixture block.

It is noteworthy that if the slider separates from the shank fixture block, the bending beam will experience a free vibration. The calculation of the slider's displacement becomes complicated. To build the dynamic model of the system, the bending beam is simplified as a pseudo-rigid links chain (two cantilever beams), and each cantilever beam works as a pinned-guided compliant beam which is characterized as a Pseudo-Rigid-Body (PRB) model [51], as shown in Fig. 2. The dynamic equation of the damped free vibration of the PRB model can be expressed as

$$M(q)\ddot{q} + C(q,\dot{q})\dot{q} + Kq = -c\dot{q} \tag{6}$$

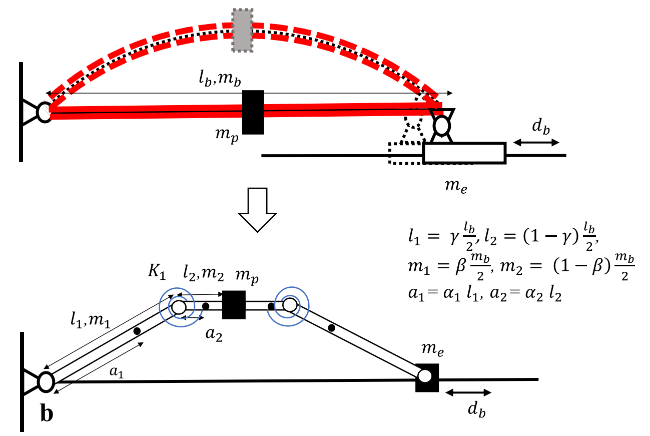


Fig. 2. Bending beam is characterized as a PRB model.

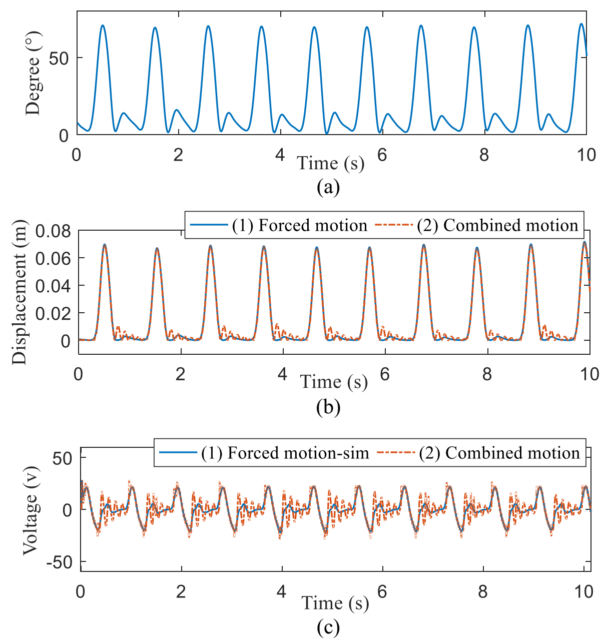


Fig. 3 (a) Human knee angle during walking. (b) Displacement of the slider. (c) Simulated output voltage.

where

$$M(q) = m_1\alpha_1^2 l_1^2 + \frac{1}{6} m_1 l_1^2 + \left(2m_2 + m_p\right) l_1^2 + m_1\left(\left(2-\alpha_1\right)^2 l_1^2 \sin q^2 + \alpha_1^2 l_1^2 \cos q^2\right) + 4 m_e l_1^2 \sin q^2 \tag{7}$$

$$C(q,\dot{q}) = \left[m_1\left(\left(2-\alpha_1\right)^2 l_1^2 - \alpha_1^2 l_1^2\right) + 4 m_e l_1^2\right] \sin q \cos q \dot{q} \tag{8}$$

$$K = 2K_1 \tag{9}$$

where $M(q)$ is the equivalent mass. K denotes the equivalent stiffness. $C(q,\dot{q})$ refers to Coriolis force and Centripetal force. c refers to the equivalent damping ratio. q is the pin joint angle. Here, compared with the inertial force, damping force, and spring force, the electromechanical coupling force of piezoelectric MFCs is relatively small and can be ignored in dynamic equations. The length of the links in the PRB

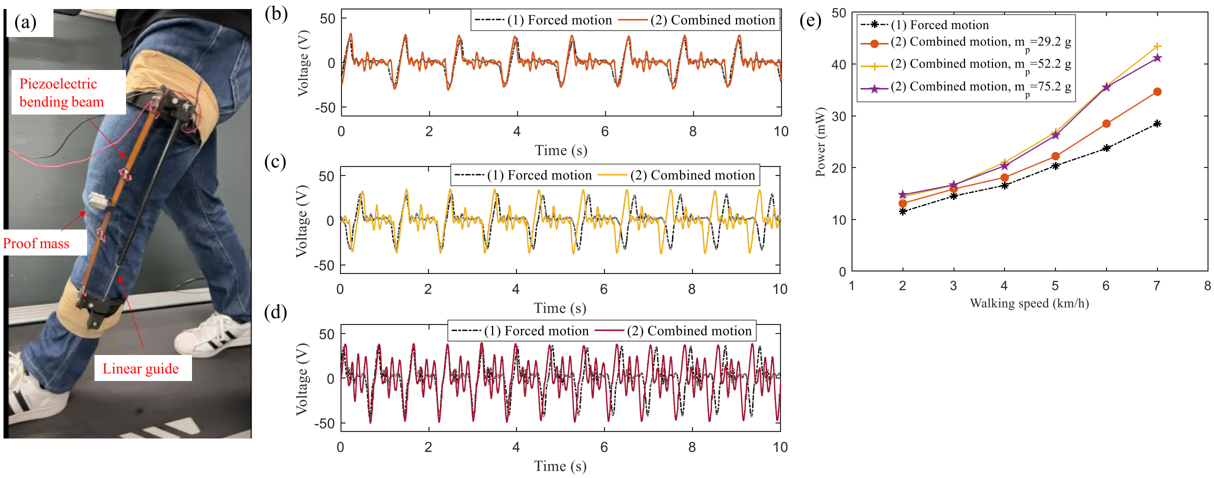


Fig. 4. (a) Subject walking with the proposed energy harvester on a treadmill. (b) Measured output voltage, $m_p = 52.2$ *g*. Walking speed: 2 km/h. (c) Walking speed: 4 km/h. (d) Walking speed: 7 km/h. (7) Average output power at different walking speeds. Resistive load is 10 kΩ.

model is $l_1 = \gamma\frac{l_b}{2}$ and $l_2 = (1-\gamma)\frac{l_b}{2}, \gamma = 0.8$ [51]. l_b denotes the bending beam's length. The masses of the links in the PRB model are $m_1 = \beta\frac{m_b}{2}$ and $m_2 = (1-\beta)\frac{m_b}{2}, \beta = 0.8$ [51]. m_b denotes the bending beam's mass. α_1 refers to the ratio of distance of the mass center to the length, $\alpha_1 = 0.56$. K_1 refers to the stiffness of the torsion spring in the PRB model, $K_1 = 2\gamma k_\theta\frac{E_b I_b}{l_b}, k_\theta = 2.68$. E_b refers to the bending beam's Young's modulus. I_b is the bending beam's moment inertia. m_p denotes the proof mass's mass. m_e represents the mass of the slider being jointed to the end of the bending beam. Herein, the slider's displacement d_{b_free} is given by

$$d_{b\_\text{free}} = 2l_1\left(1 - \cos q\right). \tag{10}$$

The following condition is utilized to check the working mode of the bending beam:

$$\begin{cases} \text{state} = 1, d_{b\_\text{free}} \le d_{b\_\text{forced}} \\ \text{state} = 0, d_{b\_\text{free}} > d_{b\_\text{forced}} \end{cases}. \tag{11}$$

When state $= 1$, the bending beam undertakes a forced motion. If state $= 0$, the bending beam experiences a free vibration. It should be noted that to simplify the calculation, here we assume that once the slider collides with the shank fixture block, they will move together later. There is no high-frequency collision.

The human knee joint angle when the subject walks on a treadmill is plotted in Fig. 3(a), measured by a motion capture system [41]. Here, the walking speed is 4 km/h, and the harvester's resistive load is 10 kΩ. The proof mass's weight is 75.2 g. Fig. 3(b) illustrates that when the slider is separated from the shank fixture block during case (2), the bending beam will undertake a free vibration in the stance phase. Consequently, the output voltage will increase significantly compared to case (1), as shown in Fig. 3(c). Here, it is noteworthy that the motion's amplitude in the stance phase is small relative to that of the swing phase, whereas the frequency is quite high. Because of that, the output voltage in the stance phase is comparable to that of the swing phase.

IV. Experimental Testing

To validate our new design and test the new harvester's performance, we conducted experimental testing on a subject (age: 35 years old, height: 162 cm, and weight: 66 kg) when he undertook walking on a treadmill, as shown in Fig. 4(a). One prototype was equipped on the left side of the subject's lower limbs. The subject walked at different speeds ranging from 2 to 7 km/h, and the resistive load is 10 kΩ. The output voltage was measured by an oscilloscope (internal impedance: 1 MΩ). With that, the output power can be calculated. For each trial, the data for a 10-s period (data plotted on a screen of the oscilloscope) was recorded. The clinical research ethics of this study was approved by the Joint CUHK-NTEC CREC in Hong Kong. In addition, the subject was provided with written informed consent before testing. During testing, three different proof masses with different weights were attached to the mid of the bending beam to investigate the effect of the proof mass's weight.

Fig. 4(b)–(d) shows the harvesters' output voltages during several sequential gait cycles when walking at 2, 4, and 7 km/h, respectively. Increasing walking speed can improve the frequency and the peak value of the harvesters' output voltage and power. When walking at 2 km/h, the peak voltage and power are 27.2 V and 74 mW, respectively, in case (1). Then raising the walking speed to 7 km/h, the peak voltage and power rose to 35.6 V and 126.7 mW, respectively. In addition, the frequency also is improved from 0.79 to 1.33 Hz when the walking speed is increased from 2 to 7 km/h. With the increase in the walking speed, the average output power is improved significantly, as shown in Fig. 4(e). When walking at 2 km/h, the average power is 11.5 mW in case (1). If the walking speed is improved to 7 km/h. the average power can reach 28.5 mW.

Because of free vibration in the stance phase, case (2) has a higher power output than case (1). Further, with the increased walking speed, the free vibration amplitude is improved significantly, as shown in Fig. 4(b)–(d). As a result, the average power output is increased considerably, as shown in Fig. 4(e). When walking at 2 km/h, the free vibration can improve the average output power from 11.5 to 14.4 mW (increased by 25.2%). If the walking speed rises to 7 km/h, the free vibration can boost the average power output from 28.5 to 43.4 mW (increased by 52.3%). Besides, it should be noted that modulating the proof mass's weight can adjust the free vibration, thus changing the average output power. When the proof mass's weight is equal to 52.2 g, the harvester has the highest average output power relative to the other two cases.

V. Conclusion

In this article, with the aim of enhancing harvesters' performance, we developed a new human knee piezoelectric energy harvester and introduced free vibration to the harvester. Specifically, one end of the bending beam was released and moved freely along the linear guide such that the bending beam experienced free vibration in the stance phase. While in the swing phase, the harvester still undertook a forced motion. Besides, to further improve the output power, a proof mass was attached to the bending beam to modulate free vibration. To validate our design, we fabricated a prototype and conducted testing on a subject when he walked on a treadmill at different walking speeds. Experimental results showed that the new design significantly enhanced the harvester's performance owing to free vibration in the stance phase. Compared with forced motion, the combined motion consisting of forced motion and free vibration can improve the average power output from 28.5 to 43.4 mW (increased by 52.3%). To the authors' best knowledge, the improved piezoelectric-based energy harvester possesses the largest power output compared to the existing piezoelectric-based energy harvesters. In future work, we will try to integrate the piezoelectric energy harvester, power regulation circuits, GPS or IMUs, and wireless transmission modules to develop real self-powered electronic devices.

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