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Biomechanical Energy Harvesting: Apparatus and Method

Q. Li, V. Naing, J.A. Hoffer, D.J. Weber, A.D. Kuo and J. M. Donelan

Abstract— **A biomechanical energy harvester is presented that generates electricity during human walking. The key feature of this device is that the power generation adds only a minimal extra effort to the user. The knee-mounted devices accomplish this by selectively engaging power generation at the end of the swing phase when knee flexor muscles act to brake knee motion. Analogous to regenerative braking in hybrid cars, the device assists deceleration of each leg within each stride while generating electrical power. We developed a control system to engage/disengage power generation based on the measured knee kinematics during a gait cycle. Experimental results show that generative braking generated 4.8±0.8 W of electrical power with a minimal increase in metabolic cost.**

I. Introduction

In recent years, we have become more and more dependent on portable electronic devices. These devices range from biomedical devices such as pacemakers, electromechanical or neuroelectric prostheses, to consumer products such as cellular phones, personal digital assistants, and global positioning systems. At present, all of these devices are powered by batteries, which add weight, size, and inconvenience to the user. There is a need to develop alternative sustainable power sources.

Biomechanical energy harvesting allows electrical power generation from human movement during everyday activities [1] such that the power generation is relatively transparent to the user. An exemplary energy harvesting device is the self-winding watch that utilizes the motion of the user's arm to accelerate a small internal mass which produces 5 μW [2]. The recently invented energy harvesting backpack used similar principles to harvest mechanical energy by converting the pack's linear motion relative to the user into rotational motion of a rotary-magnetic generator [3, 4]. It generates 7.4 W electrical from a 38 kg load when users walk fast and approximately 0.5 W electrical from more modest loads at more comfortable speeds. For energy harvesting that does not require an obligatory load, much of the effort to date has focused on the development of shoe-mounted technologies [2]. The design that has produced the most power uses an electroactive polymer generator inserted into the shoe heel to harvest 0.8 W electrical [2].

Human muscle is the origin of the mechanical power available for biomechanical energy harvesting. Muscles require metabolic energy to perform both positive and negative work. If a biomechanical energy harvesting device could reduce the demand for either positive or negative mechanical work from muscle, it will benefit the user by decreasing metabolic cost. While the energy harvesting backpack operates with an impressively high efficiency, the energy was harvested at a cost as it is easier to carry the load without simultaneously generating power [4]. It is likely that the energy harvesting shoe also increases user effort because normal shoe soles typically store and return 40-60% of the mechanical energy applied during a typical walking step [5]. If any of this energy is harvested rather than returned to the body, muscles would have to perform more positive work to replace it thereby increasing metabolic cost. Our purpose is to develop wearable energy harvesters that generate substantial electrical power without requiring substantial effort from the user. We hypothesized that by selectively harvesting electrical energy during periods when muscles normally produce negative mechanical work, the generator will assist muscles in braking the motion, generating substantial electrical power without a concomitant increase in metabolic cost. We term this *generative braking*.

To test this hypothesis, we designed a knee-mounted biomechanical energy harvester with a control system that selectively engages power generation only during periods when muscles normally produce negative work. In our current study, we target power generation at the end of swing phase when knee flexor muscles act to brake leg motion. In this paper, we first discuss walking biomechanics and energy harvesting methods. We then present the design of the biomechanical energy harvester and the control system. Finally, we present the experimental results from ergometer and human subject testing.

II. Walking Biomechanics and Energy harvesting

A. Walking biomechanics

To effectively harvest energy from walking, it is necessary to first understand walking mechanics and the underlying

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Q.li, V. Naing, J.A. Hoffer, and J.M. Donelan are with School of Kinesiology, Simon Fraser Univ., Burnaby, BC, Canada, V5A 1S6 (e-mail: qlib, vnaing, hoffer, mdonelan@sfu.ca).
D. J. Weber is with Dept. of Physical Medicine and Rehabilitation, Univ. of Pittsburgh, Pittsburgh, PA, 15213(e-mail: djw50@pitt.edu)
A.D. Kuo is with Dept. of Mechanical Engineering, Univ. of Michigan, Ann Arbor, MI, 48109 (e-mail: artkuo@umich.edu)

muscle function. During walking at a constant speed on level ground, no net mechanical work is performed on the body since there is no net change in kinetic energy (i.e. speed) or potential energy (i.e. slope of the ground). This means that equal amounts of positive and negative work are being performed on the body by all sources. While muscles are the only source of positive work, there are other sources of negative work in addition to muscle. These include air resistance, damping within the shoe sole and movement of soft tissue. The first two are known to be small during walking [6, 7]; it is believed that muscles must perform a substantial fraction of the required negative work [8-10].

Muscles do not act on the environment directly. Instead, muscles act on the body's skeleton which functions as a system of levers to perform the required power. As a consequence, positive and negative muscle power is seen externally as positive and negative joint power. Fig. 1 presents net joint power data, calculated using inverse dynamics [11] for the human knee measured during walking at a moderate speed (subject mass = 58 kg; speed = 1.3 m/s; step frequency = 1.8 Hz. Data from [11, 12]). For the angle plot, 180 degrees is full knee extension and knee flexion is <180 degrees. Positive angular velocity is motion in the knee extension direction. Positive joint moment is a net knee extensor torque. The area under the power curve, the integral with respect to time, is mechanical work. The bottom plot is rectified and filtered EMG signals from knee flexor and extensor muscles. EMG stands for electromyogram and is a measure of the electrical potential generated by muscles when they are active. Note that the EMG signals precede the negative and positive joint moments because there is a delay between when a muscle is activated and when it begins to generate force. Mechanical power outputs at all leg joints can be much greater than in Fig. 1 when walking faster, during activities like knee bends or in heavier people [13].

While there must be an equal amount of positive and negative work performed by all sources, this is not true of any joint. The knee, for example, primarily generates negative power during walking making it a good candidate for generative braking. Fig. 1 illustrates three main negative joint power regions. During stance flexion, the muscles that act to extend the knee are active producing an extensor moment. However, the knee is flexing as the leg accepts the weight of the rest of the body, resulting in negative joint power. There is also negative joint power production during the swing flexion phase due to the extensor knee moment. The activity of the muscles responsible for this extensor moment is not shown in Fig. 1. The third region, and the most important one for our current purpose, occurs during the latter half of swing extension. Knee joint power is negative due to the flexor torque produced by the knee flexors to slow down the extending knee prior to heel-strike.

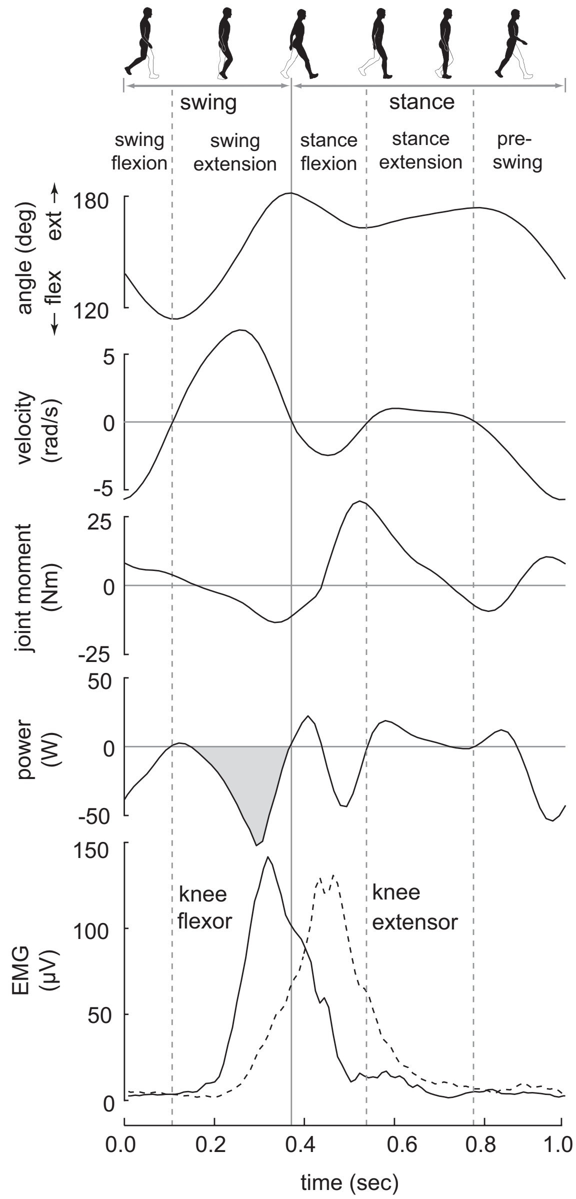

Fig. 1 Typical walking mechanics and muscle activity.

Due to the complexity of muscle function, measured negative joint power is not necessarily a consequence of negative muscle power. Many muscles cross each joint and each muscle can cross multiple joints. At any one time, negative power at one joint may be due to the coordination of net positive muscle power production that is distributed throughout the leg. Attempting to harvest energy at these times will interfere with the coordination and result in a net increase in positive muscle power and metabolic cost. In addition to the power generating muscle fibres, muscles have elastic elements such as tendons. This provides a mechanism to store and return elastic energy saving on positive muscle fibre power production. Regions of negative joint power may actually be times at which elastic energy is being stored and attempting to harvest energy will increase the total amount of positive work and metabolic cost. In short, regions of negative joint power are best viewed as potential regions for energy harvesting and determining their appropriateness requires experimentation. We focused on the swing phase extension because a) there is a large amount of negative joint power performed, b) the knee flexors, which act also to extend the hip, are lengthening because the knee is extending and the hip is flexing suggesting that they are indeed performing negative work, and c) The energy harvester acts as a rotary damper element in which the reaction torque is proportional to the angular velocity. This property favors energy harvesting during the end of swing phase where the angular velocity is large, allowing efficient power generation with a miniature generator and small gear ratio gear train.

B. *Energy Harvesting Methods*

In light of the distinct functions of muscle, we distinguish between two general methods of harvesting energy: parasitic and mutualistic. For parasitic energy harvesting, the electricity is harvested at the expense of metabolic energy of the user. In this method, the energy is harvested during the periods when muscles normally perform positive work, causing muscles to perform more positive work than they would otherwise. On the other hand, mutualistic energy harvesting is accomplished by selectively harvesting energy at times and in locations when muscles normally decelerate the body. Rather than braking entirely with muscles, a generator would perform some of the required negative work converting the mechanical energy of the body into electrical power. In this manner, mutualistic energy harvesting would be similar to regenerative braking in hybrid cars [14].

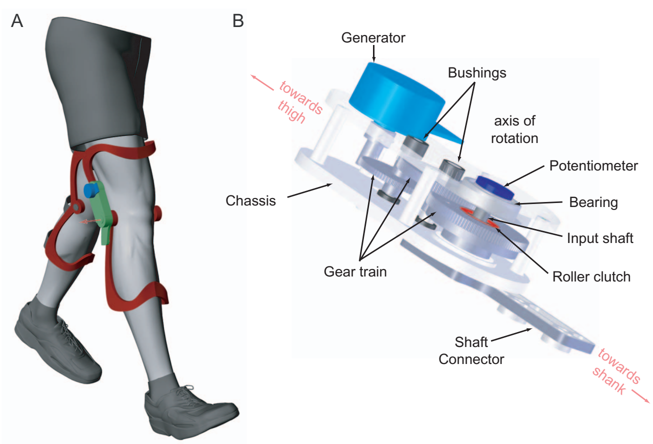

Fig.2 Biomechanical energy harvester. (A) harvesters are worn on both legs. (B) mechanical design

III. ENERGY HARVESTER DESIGN AND CONTROL

We built a wearable energy harvesting device that couples generator motion to knee motion (Fig. 2 A). A Solidworks model of the energy harvester prototype is shown in Fig. 2 B. An aluminum chassis was CNC machined and mounted on the lateral aspect of a customized orthopaedic knee brace. Each device weighs 0.79 Kg.

The energy harvester selectively converts intermittent biomechanical low velocity/high torque knee power into high velocity/low torque mechanical power for efficient power generation by a DC magnetic generator. The harvester consists of three subsystems; a mechanical system, control system, and power generation system. The overall structure is shown in Fig. 3. The mechanical system captures knee motion and regulates the mechanical power into a form that is suitable for power generation. The power generation system consists of a miniature DC generator that converts mechanical energy in to electrical energy. The control system provides the power generation engagement/ disengagement commands for completing the circuit between the generator and the loads.

A. *Mechanical design of harvester*

At the end of swing phase, knee angular velocity is less than 100 rpm, and the peak knee joint torque is around 20 N·m (Fig. 1). The energy harvester accepts the input to generate electrical power and generates enough reaction torque to match the joint torque normally produced by muscles. The matching of the joint torque and the harvester braking torque is critical since too much braking torque will interfere the normal walking and too small torque will not assist the knee flexors enough. The correct torque is achieved by a properly designed mechanical system.

The mechanical system consists of a chassis and a transmission. The input shaft accepts the knee motion at 1:1 ratio through a single hinge knee brace. A roller clutch on the input shaft couples the harvester with knee motion during knee extension phase, and decouples the harvester from knee motion during the knee flexion phase.

The normal knee joint mechanical power is computed as:

$$P_k = M_k \times \omega_k \qquad (1)$$

Where P_k is the knee joint mechanical power, M_k is the knee joint torque from inverse dynamics, ω_k is the knee angular velocity.

After feeding through the roller clutch, the input angular velocity to the gear train is zero during knee flexion phases.

The angular velocity is then amplified by the gear train before being applied to the generator.

$$\omega_g = \omega_k \cdot r_t \qquad (2)$$

Where r_t is the transmission gear ratio.

The gear train will spin the generator at a speed of ω_g and the generator will convert the input mechanical power into electrical power. The generated voltage is computed by the following equation

$$V = K_g \omega_g \qquad (3)$$

Where K_g is the back electromotive force (EMF) constant which gives the voltage per unit of rotational velocity. This design parameter of generator depends on the total number of turns in the armature winding, the number of parallel paths, the number of poles, and the magnetic flux per pole. A generator with more coils, poles, and stronger flux density normally gives a larger K_g. For a motor, the speed constant is the reciprocal of the back EMF constant.

When connecting a load to the generator, there will be current I in the complete circuit.

$$\begin{aligned} V &= V_l + I \cdot R_g \\ V_l &= I \cdot R_l \end{aligned} \tag{4}$$

Where R_g is terminal resistance of the generator. R_l is the external load we connect to the generator. V_l is the output voltage of the generator. The output electrical power is

$$P_e = V_l^2 / R_l \tag{5}$$

The power dissipated by the generator is computed as

$$P_g = I^2 \cdot R_g \tag{6}$$

When generating electrical power, the generator produces a reaction torque that acts on the gear train,

$$M_g = K_m \cdot I \tag{7}$$

Where K_m is the torque constant which equals to the back EMF constant.

The reaction torque is amplified by the gear train before being applied to the input shaft and knee joint. The reaction torque applied to the joint is

$$M_r = M_g \cdot r_t / \eta_t \tag{8}$$

Where η_t is the efficiency of the gear train.
The mechanical power absorbed by the harvester is the product of the reaction torque and the knee angular velocity,

$$P_m = M_r \cdot \omega_k \tag{9}$$

The efficiency of the harvester is the ratio between the generated electrical energy and required mechanical energy,

$$\eta_h = \int P_e dt / \int P_m dt \tag{10}$$

For the energy harvester, we want to maximize the electrical power output of (5) and the mechanical to electrical efficiency of (10) while producing a reaction torque (8) matching the joint torque normally produced by muscles at the end of swing phase. The design parameters are gear ratio r, output resistance R_l, the speed constant K_g, and the terminal resistance R_g. Regardless of the choice of gear ratio and output resistance, a generator with a smaller speed constant and a smaller terminal resistance will result in higher power output and higher efficiency. However, reducing the speed constant and the terminal resistance means an increase in the weight of the generator. There is a tradeoff between the weight and the preferred generator parameters. We selected a motor with a speed constant of $285 rpm/V$, the terminal resistance $R_g = 1.03\Omega$, and a mass of 110 g. After choosing the generator, we found the output resistance and the gear ratio to maximize the efficiency and match the reaction torque with the knee joint torque. Through simulation, we found an optimal combination of gear ratio as 113:1 and output resistance of 5 Ω such that the electrical power output and mechanical to electrical efficiency are maximized. We considered the friction of the gear train but neglected the inertia of the transmission and the generator.

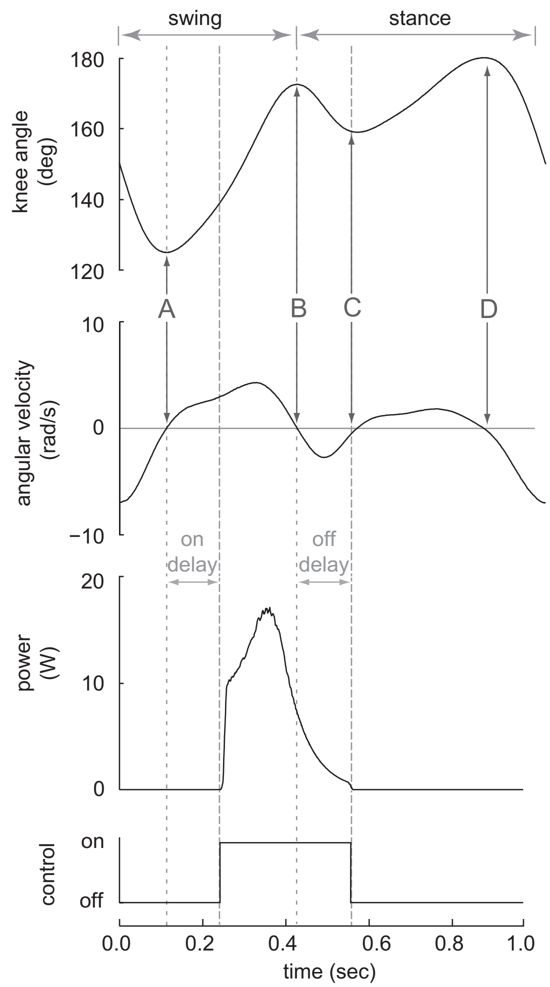

Fig. 3 Control signals based on knee angle and angular velocity.

B. Control system design

In order to achieve generative braking, we designed a control system that selectively engages/disengages power generation during a gait cycle. The control system consists of a potentiometer that measures knee angle, an algorithm that generates the control commands, and an electrical switch that accepts the control command to open/close the circuit between the generator and resistors. When the circuit is closed, the harvester produces a braking torque that acts on the knee joint. The control system is implemented on Simulink, compiled using Real Time Workshop and executed at 1 KHz using Real Time Windows Target on a desktop computer. This allows for rapid prototyping of the control system. Data acquisition of the potentiometer data and the control commands to the switch is accomplished by an A/D and D/A board through the computer.

The potentiometer mounted on the input shaft measures knee joint angle in real time. The knee joint angle signal is first filtered by a low-pass filter, and then differentiated to get the angular velocity. The control algorithm uses knee angle and angular velocity to distinguish different phases of the gait cycle. The logic of the control algorithm is as the following:

(1). If the angular velocity goes across zero upward and the knee angle is small, it is the start of swing phase knee extension (Point A on Fig. 3).

(2). If the angular velocity goes across zero upward and the knee angle is large, it is that start of stance phase knee extension (Point C on Fig.3).

(3). If the angular velocity goes across zero downward between stance phase knee extension and swing phase knee extension, it is the start of the pre-swing knee flexion (Point B on Fig.3).

(4). If the angular velocity goes across zero downward between swing phase knee extension and stance phase knee extension, it is the start of stance phase knee flexion (Point D on Fig.3).

Since we target the end of swing phase to harvest electrical power, the power generation engagement signal is generated by adding a 70-90 ms delay to the detected start of swing phase knee extension, which is approximately when knee flexor muscles normally become active to brake knee extension. The disengagement signal is generated by adding a delay to the start of the stance phase flexion. Instead of turning off energy harvesting at the beginning of the stance phase, we keep the harvester on during the stance flexion phase to allow the generator to harvest the kinetic energy remaining in the transmission and the generator inertia from the swing phase knee extension. This does not generate extra resistance for the stance flexion phase since the harvester is decoupled from knee motion during knee flexion by the roller clutch.

An example result of the control system is shown in Fig. 3. The results indicate that the control system effectively engages power generation at the middle of the swing extension phase and disengages at the end of stance flexion phase. Human subject testing demonstrates that the control system is robust with respect to the variation in knee profile between subjects. The control system correctly engaged/disengaged power generation for over 50,000 gait cycles without a failure.

IV. EXPERIMENTAL RESULTS

We operated the harvester in there modes: disengaged mode, continuous generation mode and generative braking mode. In *disengaged mode*, the roller clutch is manually disengaged so that the transmission is never in motion. This mode serves as a control condition for human subject experiments to account for any physiological changes that result from carrying the added mass independent of physiological changes resulting from energy harvesting. In the *continuous generation mode*, energy harvesting is not selective as the power generation circuit is always completed. In the *generative braking mode* the control system selectively engages and disengages power generation to target the negative work region at the end of walking swing phase.

Ergometer testing served two purposes. One was to evaluate the harvester efficiency of converting mechanical power to electrical power, which was used later to determine the relationship between the amount of generated electrical power and the metabolic cost. The other purpose was to determine the amount of braking torque produced by the harvester.

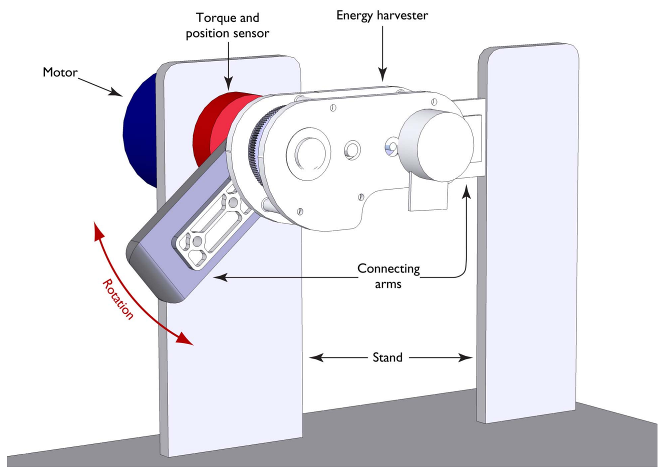

Fig. 4 Test ergometer for the efficiency evaluation

We designed a test ergometer to drive the harvester with a specified kinetic profile. The kinetic profile was set as the average knee angle measured during our human subject trials (Fig. 4). By measuring the angular velocity, reaction torque and electrical power generation, we calculated the efficiency as the ratio between the generated electrical power and the input mechanical power. To determine the reaction torque produced by friction, the inertia of the generator and gear train, we performed a test under an *open switch condition* where no electrical power was generated. Typical measurements of velocity, torque, computed mechanical power, and measured electrical power in generative braking, continuous generation, and open switch modes are shown in Fig 5. The harvester has an efficiency of 63% in continuous generation mode and 56% in generative braking mode. The efficiency in generative braking mode is lower because the harvester spends a greater amount of time dissipating mechanical energy without producing electrical power. To determine the sensitivity of the calculated efficiency to the variation of knee kinematics, we scaled the input angular velocity profile by ±10% and found only small changes in the efficiency (< 3%).

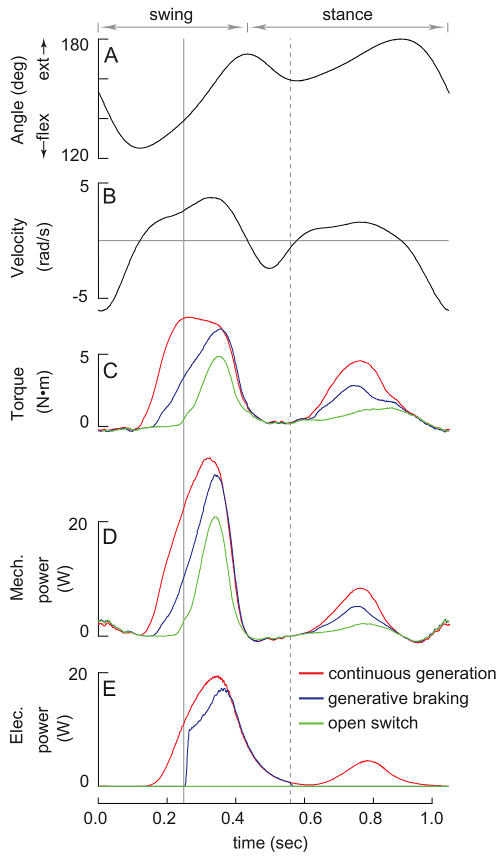

Fig. 5 Test ergometer data from three simulated walking stride cycles.

The energy harvester achieves a peak reaction torque over 5 N·m and a peak mechanical power over 20 W. These results indicate that the designed energy harvester can produce a reasonable amount braking torque during the end of swing phase to help reduce knee flexor muscle activity. The measured reaction torque under open switch condition is shown in Fig. 5. The braking torque experienced by the user is both from inertia and electrical power generation. The braking torque increased by about 30% when we closed the circuit, indicating the use of an electrical switch to engage the power generation is effective.

In order to make a comparison between different harvesting modes, we introduce a measure named as cost of harvesting (COH), a dimensionless quantity defined as the additional metabolic power required to generate one Watt of electrical power. With continuous harvesting mode efficiency as 63%, if muscles perform positive work (conventional generation) to harvest electrical power, the estimated COH will be 6.4, calculated from the reciprocal of the product of the device efficiency and the peak muscle efficiency for positive work.

The COH in generative braking is 0.7±4.4; less than one Watt of metabolic power was required to generate one Watt of electricity. The COH is 2.3±3.0 for the continuous harvesting mode. The detailed results of human subject testing can be found in [15].

V. Conclusion

In this paper, we present a novel biomechanical energy harvester for generating electricity during walking with minimal user effort. The experimental results indicate that by harnessing the characteristics of walking, we achieve a very small COH. The high power produced by these devices, and the low effort required by their user, makes them well suitable for charging powered prosthetic limbs and other portable medical devices.

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