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Lower-Limb-Driven Energy Harvesting: Preliminary Analysis

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Abstract— **In this paper, we present a new lower limb driven biomechanical energy harvester and its preliminary performance analysis. An estimate of the mechanical available power, estimated user felt resistance, and preliminary testing were conducted in this study. The estimated total available mechanical power and user felt resistance are based on the kinematic motion data and the mathematical model of the energy harvester prototype. Two key advantages of the new model are: generation of a higher mean power and application to a wider range of subject motion. The device is mounted on a backpack with lower limb attachments. Power generation occurs during the swing phase where negative power occurs. The new energy harvester prototype is capable of harvesting power on the same order of magnitude as previous models.**

I. Introduction

The availability and portability of electrical power is a topic that is rarely a concern in more developed countries. Yet it plays a huge role in our daily lives and has considerable impact on developing nations as well as populations in remote locations. In our daily lives we can see evidence of an increasing dependence on electrical power, from consumer products such as cell phones, to biomedical devices such as pacemakers, even basic amenities such as the lighting in our rooms requires electricity. Such amenities are rarely available in remote areas or in developing nations. In such environments, there is a demand for portable, light weight, and sustainable power supplies. Such requirements are met with the recent development of biomechanical energy harvesters which are capable of generating the electrical power, from basic human motion during daily activities.

Currently there are two well known biomechanical energy harvesting systems that can generate a substantial amount of electricity for applicable purposes: an electricity generating backpack [1] and a knee brace device [2], [3]. The electricity generating backpack uses the vertical displacement of the centre of mass of the user during regular activities such as walking or hiking, to drive a spring-loaded motor to generate electricity. This system is capable of producing a maximum of 7.37 W with a 38 Kg backpack while walking over a flat surface at approximately 6.4km/h [1]. The disadvantages of this system are its' dependence on the vertical displacement of a mass, which moves relative to the backpack. Not only does this limit the types of activities for which the device can generate power, but its' energy output is directly proportional to the obligatory mass which increases the load on the user. In addition, the backpack system does not support selective energy harvesting meaning that the system harvests during both positive and negative work phases of gait. This indicates that the backpack system will only increase user effort and metabolic cost, as selective energy harvesting has been identified as a means of minimizing user metabolic costs [2], [3]. Moreover, the up-and-down oscillating mass may negatively affect gait pattern and walking stability. On the other hand, Li et al developed a 1.6 Kg knee-mounted energy harvester that comprised of a gear train with a one-way clutch, transmitting only knee extension motions, during a phase of negative muscle work, to provide speeds suitable for a brushless DC rotary magnetic generator for electricity generation [2], [3]. A pair of harvesters generated 5 W of electricity (dissipated in resistors) during walking at a speed of 1.5 m/s. However, the much-lighter knee-mounted device [3] suffers from carrying a mass distally on the knee. Because the metabolic cost of carrying a given mass distally is considerably more expensive than carrying it proximally [4], wearing the knee device without power generation requires 20% more metabolic energy expenditure when compared to walking without wearing the device.

Attaining a substantial amount of electricity generation, while avoiding substantial load and sacrificing user comfort, necessitates a new energy harvesting technology. To achieve this objective, we propose a new energy harvester design and provide a preliminary evaluation of its performance herein.

II. Method

A. Biomechanics of Walking

The biomechanics of walking, from an energy harvesting perspective, have been previously mentioned by [3], the body experiences periods of negative and positive work during normal walking on level ground, however no net mechanical work is performed on the body. Muscles act on the body's skeletal structure and is seen to function as a system of levers to perform the required power. Thus the positive and negative muscle power is seen externally as positive and negative joint power [3]. The periods of negative power are viewed as potential regions of energy harvesting. The swing phase has been previously validated as a preferred region for energy harvesting due to relatively

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large generation of negative joint power [4]. The knee-mounted harvester is designed to generate power during the swing phase where large amounts of negative joint power is performed and where the knee flexors also act to extend which indicate that they are indeed performing negative work [2]. By selectively engaging electrical power generation during periods of negative work through a control system, the harvester should exhibit mutualistic energy harvesting characteristics which is analogous to generative braking [2].

B. *Lower-Limb Mechanical energy during Walking*

In order to select harvester parameters, we first calculated the total mechanical energy that is available during walking. The methodology implemented for the estimation of the instantaneous mechanical energy of the lower limbs is similar to that found in [6]. For a given limb, the total instantaneous mechanical energy is the summation of the mechanical energy of the thigh, shank and foot segments. The total instantaneous mechanical energy of the i^{th} lower limb segment is the sum of its translational kinetic energy, rotational kinetic energy, and potential energy, which is determined using the following equation,

$$E_i = \frac{1}{2} m_i v_i^2 + \frac{1}{2} I_i \omega_i^2 + m_i g h_i \qquad (1)$$

Where m_i is the mass of the segment, v_i is the instantaneous linear velocity of the centre of mass of the segment, I_i is the segments' mass moment of inertia about the centre of mass, ω_i is the segments' angular velocity about the centre of mass, and h_i is the height of the centre of mass above a datum. Using the subjects' measured segment lengths, the inertial properties of the segments are determined using Dempster's anthropometric data from [5].

Analysis of the subject's motion is performed only in the sagittal plane as majority of energy generated or absorbed during walking is in the sagittal plane.

The total estimated mechanical energy available from the lower limbs is obtained by summing the estimated energy of the lower limb segments, per respective leg. The estimated mechanical power available from the lower limbs during swing phase is defined as the differential of the total available mechanical energy during that phase,

$$P_{Mech} = \frac{dE(t)}{dt} \qquad (2)$$

Lower-limb position data is collected using a motion capture system with reflective markers. All motion data are filtered using a 2nd order Butterworth filter with a cut-off frequency of 8 Hz, prior to numerical differentiation. The analysis is performed using a custom MATLAB program.

Specific joint power is not derived for the newly proposed energy harvester as its application does not focus on a joint centre but on the entirety of the lower limb during its swing phase of the gait cycle. Therefore the entire energy of the lower limbs is required.

C. *Design of Energy Harvester*

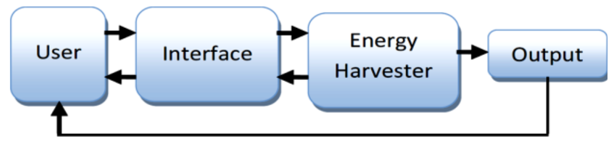

Fig. 1. Block diagram of the energy harvester system depicting the hierarchy of the system and the feedback between each substructure e.g. the relationship between the user and the interface substructures is the input motion from the user to the interface and the resistance feedback from the interface to the user.

We designed a new lower-limb-driven energy harvesting device that generates electricity during walking. Fig. 1 illustrates the basic hierarchy of the harvester system.

The premise of the energy harvester design process is to develop an energy harvesting system which does not require the user's focus, with additional benefits of generative braking. The designed energy harvester utilizes the swinging motion of the legs to generate power. As a result of the design, only the relative motion between the lower limbs and the trunk is considered to be a source of mechanical power generation. This is the energy and power that are derived in the previous section.

D. *Interaction between User and Harvester*

An important consideration of the energy harvester design is the impact of the harvester on the user. A preliminary simulation of user-felt resistance is developed in a custom Matlab program to estimate the resistance that the user experiences at various walking speeds. The user-felt resistance comprises of two components, the resistance from electricity generation and the inertial resistance,

$$T = T_g(\theta(t), \omega(t)) + T_I(\alpha(t)) \qquad (3)$$

Where T is the total reflected torque of the harvester system, $T_g(\theta(t),\omega(t))$ is the net reflected resistance of the harvester system from power generation, $T_I(\alpha(t))$ is the reflected inertial resistance of the system's mechanical components, and t is time. Both components, the resistance from power generation and the inertial resistance, are both reflected and functions of the input angular displacement, $\theta(t)$, velocity ,$\omega(t)$, and acceleration, $\alpha(t)$. These input terms are estimated from the kinematic data as follows,

$$\theta(t) = \frac{L(t)}{d} \qquad (4)$$

$$\omega(t) = \frac{d\theta(t)}{dt} \qquad (5)$$

$$\alpha(t) = \frac{d\omega(t)}{dt} \qquad (6)$$

Where $\theta(t)$, $\omega(t)$, $\alpha(t)$ are the input angular displacement, velocity and acceleration from (3). $L(t)$ is the change in the distance between the heel and the hip of the respective leg, which serves as the arc length for the angular displacement of the input pinion, with radius d. The input pinion is the first stage input of the energy harvester system. Angular velocity and acceleration inputs are subsequently determined as shown in (5) and (6), by taking the first and second derivative of the input angular displacement, respectively.

The following equation governs the resistant torque that the generator applies on the user during power generation

$$T_r = \frac{N_t^2}{\eta_t} \frac{K_g K_t}{R_g + R_l} \omega(t) \qquad (7)$$

Where T_r is the resultant torque, N_t is the gear ratio of the gear train, η_t is the mechanical efficiency of the gear train, K_g and K_t are the speed and torque constants of the generator, respectively, R_g and R_l are the resistance of the generator and resistance of the electrical load, respectively, and $\omega(t)$ is the angular velocity input as defined by (5).

The resultant, resistance force is determined from (3) by dividing T, the reflected torque, by the radius of the input pinion, d, as a cable between the input pinion and heel connects the user to the energy harvester system.

$$F = \frac{T}{d} \qquad (8)$$

E. Mechanical Energy - Kinematics

An analysis of the kinematic data during unabated walking is performed to determine the energy and power profiles during a gait cycle. All segment lengths and the joint centers for each segment are determined by a camera-based motion capture system. The preliminary trial of estimating available mechanical energy in the lower limbs is conducted using motion data from one male subject, 27 years of age, with a mass of 85 kg, and a height of 1.96 m. The subject is asked to perform three walking trials on a treadmill at the following speeds: slow (2.1 km/hr), normal (3 km/hr), and fast (3.9 km/hr). The total available mechanic energy of each leg is determined as previously described.

F. Harvester Performance

A preliminary evaluation of the energy harvester prototype power generation performance is conducted using one male subject, age 23 years, height of 1.78 m, and weight of 84 kg. The subject is asked to walk at speeds ranging from 2 km/hr to 4 km/hr at increments of 1 km/hr, per trial, on a treadmill while wearing the harvester prototype. A multivariable oscilloscope is used to record the voltage and current from the prototype harvester. The electrical power generated by the harvester is calculated as the product of the measured voltage and current. This test provides a comparative benchmark to previous harvester designs.

III. Results

Although data for multiple walking speeds are available, the data is represented by Fig. 2. The knee flexion angle, total lower limb energy, user experienced resistance, mechanical power, and tested power output at a walking

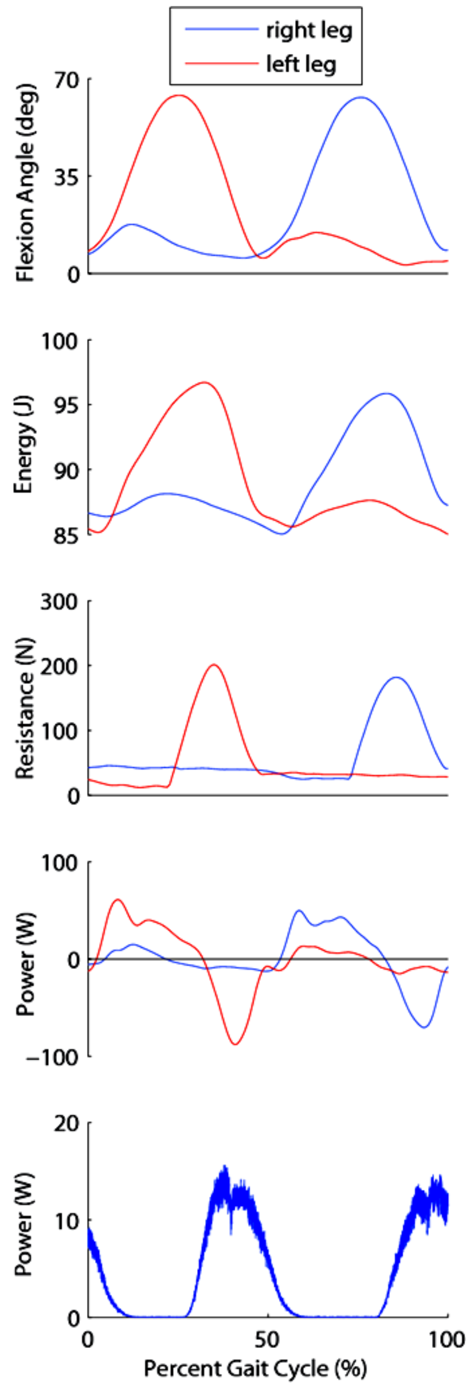

Fig. 2. Plot a) is the knee flexion angle, b) is the estimated mechanical energy of the lower limbs, c) is the estimated resistance on user, d) is the estimated available mechanical power, and e) is the test of mechanical power normalized to percent gait. The final plot e) was not from the same subject or trial as the motion data collected (a-d). Plot e) is also manually synchronized to provide an illustration of when power generation occurs during normal gait.

Table I. The maximum and mean powers of the harvester prototype during preliminary test are presented for each walking speed

Walking Speed (km/hr)	Max. Power (W)	Mean Power (W)
2	12.6	4.4
3	15.6	5.6
4	19.6	7.6

speed of 3 km/hr are presented respectively in Fig. 2. All plots are normalized to percentage of the gait cycle, where zero and 100% are the heel strike events of the right leg. Plots of Fig. 2(a) to Fig. 2(e) are derived from kinematic data as previously explained. The kinematic data collected for these results are when the subject is under free and normal conditions, not wearing the harvester prototype.

The energy profile shown in Fig. 2(b) identifies that more energy is available during swing phase relative to the stance phase. However, during the swing phase, Fig. 2(d) reveals that the body experiences a period of positive work in the first half of the swing phase during knee flexion, and negative work in the second half during knee extension. Since the harvester design is intended to exhibit mutualistic characteristics, power generation should only occur during the period of negative work, when the knee extends during swing phase.

The results of the preliminary testing on the harvester prototype are summarized in Table I. The preliminary performance results indicate the potential power generation of the harvester. At 3 km/hr, the harvester prototype is capable of generating a peak power of 15.6 W while sustaining a mean power of 5.6 W, which is comparable to the knee brace model that produces 5 W. Fig. 2(e) depicts the power profile generated by the harvester for walking at 3 km/hr.

Table II. Summary of estimated user-felt resistance at increasing walking speeds, derived from kinematic analysis and the model of the harvester prototype as explained by (3) and (8)

Walking Speed (km/hr)	Max. Resistance (N)	Mean Resistance (N)	
2	188.37	47.17	42.04
3	215.10	48.78	60.77
4	255.72	85.86	54.33

From the estimation of user-experienced resistance, it is estimated that the user will experience a maximum resistance of 215 N, acting at the heel, at time of maximum velocity of the foot in swing phase. Table II summarizes the estimated resistance at three progressively increasing walking speeds. Variation of estimated resistance between the legs is likely due to physiological variation between the subjects' dominant and non-dominant leg. The resistance to the user is expected to increase with walking speed.

IV. Discussion

Although the system offers a maximum estimated resistance of 215 N during normal walking at 3 km/hr, this resistance acts to assist the muscles in braking the downward motion of the leg which reduces the amount of negative muscle work, thus reducing the metabolic cost. From Table I of the results, it is shown that the harvester prototype is capable of generating power on a similar order of magnitude as current state of the art harvesters. However, an improvement from the current design is the increase of user comfort and increase of lower limb mobility.

Fig. 2(c) shows that the user experienced resistance does not reach zero at any point in a gait cycle. Much like the knee brace harvester, a transition period is required in which the user become accustomed to the baseline mechanical resistance of energy harvester. For the current prototype, the baseline resistance is ~50 N (11 lbs), where baseline is measure from stance phase. The preliminary testing did not incorporate testing for results under varying electrical loads. As such, the plot in Fig. 2(e) presented is the resulting power profile under an 8 Ohm electrical load.

To conclude, the performance of a lower limb biomechanical energy harvester prototype is presented in this paper. The RMS powers generated by the harvester prototype are 4.4 W, 5.6 W, and 7.6 W, with maximum estimated mechanical resistance exerted on the user, from the prototype to be 188 N, 215 N, and 255 N, at 2 km/hr, 3 km/hr, and 4 km/hr walking speeds, respectively. Maximum power generation occurs during the mid to late swing phase where the lower limb experiences maximum velocity. Maximum resistance is also estimated to occur at this time.

Future testing is required to validate the simulation of the resistance experienced by the user. Further investigation of metabolic energy expenditure would prove insightful and will be investigated in future work. The interface between the user and the energy harvester and its impact on the users' gait is also of interest and will be investigated in future work. Additional investigation of the system performance and user experience during varying activities and electrical loads will be conducted along with development of a control system for storing the electrical energy generated such that it can be implemented to power other useful devices.

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