Proceedings of PowerMEMS 2008+ microEMS 2008, Sendai, Japan, November 9-12, (2008)

BIOMECHANICAL ENERGY HARVESTING

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Abstract: We recently developed a biomechanical energy harvester that generated substantial electricity during walking while requiring little extra effort. It took advantage of the fact that much of the displacement during walking occurs at body joints and harvested energy from knee motion. It selectively engaged power generation to assist the body in performing negative work, analogous to regenerative braking in hybrid cars. As muscle is ultimately the origin of energy available for biomechanical energy harvesting, the main purpose of this paper is to explain the physiological principles that guided our design process and to present a brief description of our device design and its performance.

Key words: human walking, muscle physiology, generative braking

1. INTRODUCTION

Human power is an attractive energy source. Muscle converts food into positive mechanical work with peak efficiencies of approximately 25%, comparable to that of internal combustion engines [1]. The work can be performed at a high rate, with 100 W mechanical easily sustainable by the average person [2]. Food, the original source of the metabolic energy required by muscles, is nearly as rich an energy source as gasoline and approximately 100 fold greater than batteries of the same weight [3]. Given these attractive properties, it is not surprising that a number of inventions have focused on converting human mechanical power into electrical power. These include hand crank and bicycle generators as well as windup flashlights, radios, and cell phone chargers [4]. One major drawback of these devices is that they require dedicated power generation by the user. This serves to limit the time available to produce power and, thus, the amount of useful energy that can be generated.

In contrast, biomechanical energy harvesters generate electricity from people as they go about their activities of daily living [5]. This results in power generation over much longer durations. An exemplary energy harvesting device is the self-winding watch which produces enough electricity to power the device without requiring the user to wind it but is insufficient for most of our portable power needs [4]. There are a number of devices based on the same fundamental principle as the self-winding watch—using an external load to drive a generator. The most successful design to date is the spring-loaded energy harvesting backpack that converts the pack's linear motion relative to the user into rotational motion of a rotary-magnetic generator producing as much as 7 W [6]. A second group of energy harvesters use the body's own inertia to generate electricity from the compression of the shoe sole harvesting as much as 0.8 W [7].

We recently developed a biomechanical energy harvester that generated substantial electricity during human walking with little extra effort required from the user [8]. Our device differed from previous devices in two main ways. First, the device took advantage of the fact that much of the displacement during walking occurs at body joints and harvested energy from knee motion rather than from an external load or the compression of the shoe sole. Second, the device selectively engaged power generation to assist the body in performing negative work, analogous to regenerative braking in hybrid cars. Our previous paper that focused on the results of our human subject testing required an understanding of the physiology of walking and a novel device design to best take advantage of the underlying physiological principles. As muscle is ultimately the origin of all energy available for biomechanical energy harvesting, our main purpose here is to explain the physiological principles that guided our design process. We also present a brief description of our device design and its performance.

2. WALKING MECHANICS

The inherent uneconomical nature of walking provides an opportunity for economical energy harvesting. During walking at a constant speed on level ground, zero net mechanical work is performed on the body since there is no net change in kinetic or potential energy—the body is not speeding up or slowing down and it is not being raised or lowered. This is accomplished with the summed contribution of a number of sources—including muscle, tendon, clothing and air resistance—performing equal amounts of positive and negative mechanical work [9]. Selectively engaging energy harvesting at the right times and in the right locations could assist with the negative mechanical work, replacing that normally provided by other sources such as muscle. This is similar to how regenerative braking generates power while decelerating a hybrid car [10]. We have termed this "generative braking" as the electricity is not reused to power walking but is available for other uses [8].

In principle, generative braking can produce electricity while reducing the metabolic cost of walking. When performing positive mechanical work, active muscle fibres shorten while developing force, converting chemical energy (i.e. metabolic energy) into mechanical energy. The peak efficiency of positive muscle work is approximately 25% [1]. That is, a muscle producing 1 W mechanical requires 4 W metabolic and dissipates 3 W as heat. When performing negative work, muscle fibres develop force but are compelled to lengthen by an external force. This braking system is not passive—muscles require metabolic energy to perform negative work. The peak efficiency of negative work production is approximately -120% [1]. That is, a muscle producing -1 W mechanical requires 0.83 W metabolic and dissipates 1.83 W heat. Methods of generating electricity that require an increase in positive muscle work—as is the case with conventional generators such as hand cranks and cycle ergometers—will cause a relatively large increase in effort while electricity generation that results in a decrease in negative muscle work will result in a relatively small decrease in effort.

While muscles are the ultimate source of positive work in walking, 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. These are considered passive sources of negative work in that, unlike muscle, they don't require metabolic energy to dissipate mechanical energy. While the contribution of air resistance and shoe sole damping are thought to be small during walking [11, 12], the quantitative contribution of soft tissue movement to negative work is not yet clear [13]. While muscles do not

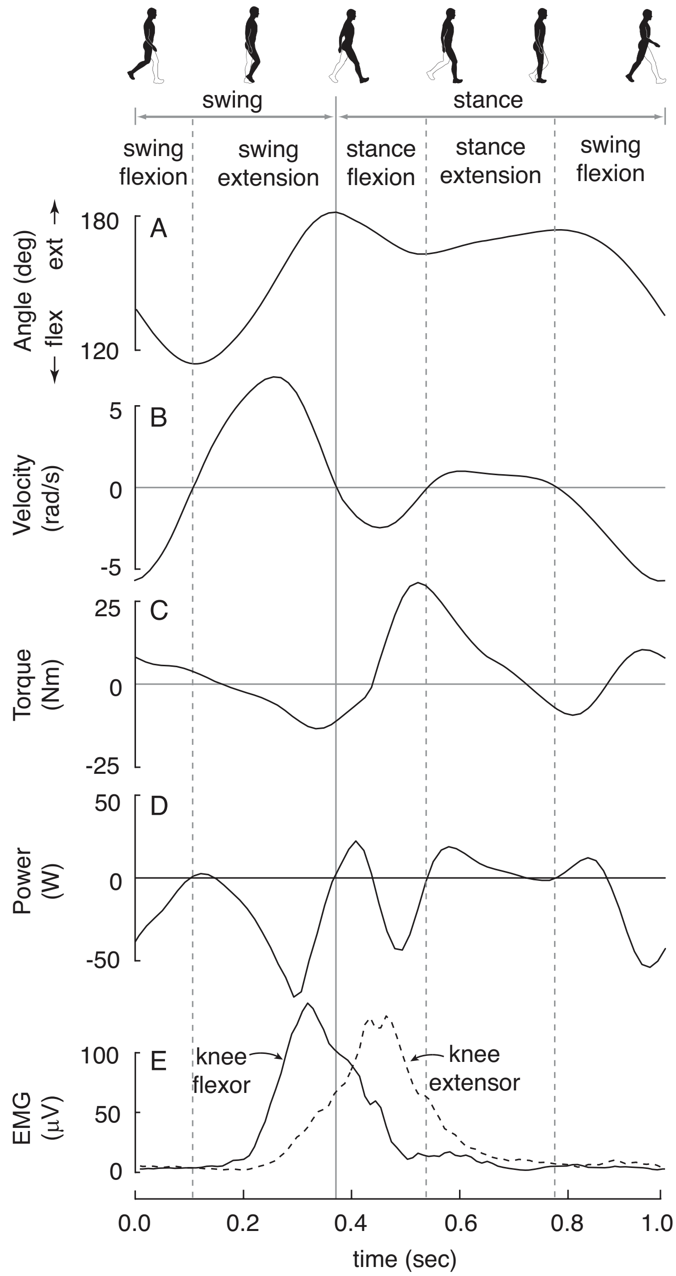

Fig. 1: Typical knee joint mechanics and muscle activity during walking (subject mass = 58 kg; speed = 1.3 m/s; step frequency = 1.8 Hz. Data are from Winter, 1990 and Hof et al. 2002). A) Knee joint angle where 180 degrees is full knee extension. B) Knee joint angular velocity using the convention that positive angular velocity is motion in the extension direction. C) Knee joint torque with the convention that extensor torques are positive. D) Knee joint power. E) Rectified and filtered electromyograms (EMG) from one knee flexor muscle (solid line) and one knee extensor muscle (dashed line).

perform all of the required negative mechanical work during walking, it is believed that they perform a substantial fraction [14-16]. Nevertheless, it is possible for negative work by an energy harvesting device to replace negative work by a passive source resulting in no change in metabolic cost to the user.

Muscles do not act on the environment directly. Instead, muscles act on the body's skeleton which functions as a system of levers to transmit the muscle work to the rest of the body. As a consequence, rates of performing positive and negative muscle work are measured externally as positive and negative joint power [17]. Figure 1 presents knee joint power data for a single subject walking at a comfortable speed [17, 18]. Mechanical power outputs at other joints can demonstrate very different patterns and power generation at all joints depends on many parameters including walking speed and the mass of the subject [17, 19]. Regardless of joint or condition, joint power is typically intermittent, bi-directional, time-varying, and relatively low speed and high torque. These characteristics represent a significant challenge for energy harvesting around joints.

It is difficult, however, to interpret muscle function from joint kinematics and kinetics alone [20]. This is for three main reasons. First, all joints are spanned by muscles that generate forces to oppose each other and these muscles can be simultaneously active. Thus, net positive joint power can result from positive and negative power production by opposing muscles. Resisting the motion of a joint may usefully assist the negative power producing muscles, even in the presence of net positive joint power. Second, muscles often cross multiple joints. An isometric muscle, or even one that is generating net positive power, may contribute to negative joint power at one joint while it simultaneously generates positive joint power at other joints [20]. Resisting the motion of a negative power producing joint may ultimately increase the positive mechanical power required of the muscles that span that joint. Third, tendons and other connective tissue can store and return elastic energy [21]. Negative joint power may be due this elastic tissue storing mechanical power for later use. Resisting joint motion may interfere with this storage and ultimately increase the positive work required of muscle. As a consequence of the complicated physiology, claims regarding the appropriate joint and timing for exploiting generative braking are best viewed as predictions until tested experimentally.

While there must be an equal amount of positive and negative work performed on the body by all sources during level-ground constant-speed walking, this is not necessarily true of any individual joint. In particular, the knee primarily performs negative work during walking making it a good candidate for generative braking. Figure 1 illustrates four main phases of knee kinematics, each delineated by a change in direction of motion: stance flexion, stance extension, swing flexion and swing extension. Beginning with foot contact, the muscles that act to extend the knee are active (E) producing an extensor moment (C) during stance flexion. However, the knee is flexing (B) as the leg accepts the weight of the rest of the body, resulting in negative joint power (D). During stance extension, the knee extensor muscles are still generating an extensor torque and have redirected the joint motion resulting in a period of positive joint power. It is important to note that there is a delay between the measured muscle activity and the corresponding muscle force resulting in activity that precedes force generation and force generation that continues after activity ends [22]. The knee flexes towards the end of stance and continues flexing into the swing phase. For convenience, we refer to this period as swing flexion while recognizing that it begins during stance. There is primarily negative joint power production during this swing flexion due to the dominant knee extensor moment. The activity patterns of the muscles responsible for this extensor moment are not shown in Figure 1. The fourth region, and the most important one for our current purpose, is swing extension. Knee joint power is primarily negative due to the flexor moment produced by the knee flexors to slow down the extending knee prior to foot contact. Of the four phases of knee kinematics, three primarily generate negative joint power.

To harvest energy using generative braking, we selectively engaged power generation during swing extension. The physiological reasons for targeting swing extension are threefold. First, a large amount of negative joint power is performed during this phase. At a comfortable walking speed, for example, each leg performs approximately -8.4 J in swing extension compared to – 6.3 J during stance flexion (Fig. 1) [23]. Second, the swing phase negative work does not depend strongly on walking speed when compared to other phases. For example, swing extension work decreases by only 19% between 1.5 m/s and 1.0 m/s while stance flexion work decreases by 56% [23]. This suggests that swing extension has greater potential than stance flexion to produce useful amounts of electricity during the slower walking speeds typical of pathological gait. The third reason is that the negative joint power during swing extension is likely due to actual negative muscle work rather than net positive work by muscles that cross more than one joint or the storage of useful elastic energy. This is because the knee is extending and the hip is flexing, forcibly lengthening the active knee flexor muscles that also act to extend the hip. In contrast, the hip and knee are both flexing during parts of swing flexion. Knee extensors can also cross the hip joint and thus it is not immediately clear if these muscles are lengthening, shortening or remaining isometric. While some of the swing extension negative work may be due elastic tissue like tendons, it is unlikely that this is returned in a useful manner because it is followed by a negative work flexion phase. In contrast, elastic tissue during stance flexion may store elastic energy and return it during the positive work of stance extension.

3. DEVICE DESIGN

The biomechanics of walking presented four main challenges for designing a device to harvest energy from the motion of the knee joint. The first challenge was to determine an effective mechanism for converting biomechanical power into electrical power. This generator had to be worn on the body so it needed to be small and lightweight. The second design challenge was to determine a mechanism for converting the knee joint power into a form suitable for efficient electrical power generation. As described in the previous section, knee joint power is intermittent, bi-directional and has particular speed and torque characteristics. The third challenge was to optimize the system parameters in order to maximize the electrical power generation without adversely affecting the walking motion. At any given point in the walking cycle, there is only a certain amount of mechanical power available for harvesting from the knee—attempting to harvest too much power will cause the user to limp or stop walking while harvesting too little results in less electrical power generated. The final design challenge was to determine a mechanism for selectively engaging power generation during swing extension to harvest energy using generative braking.

To meet these design challenges, our device operated about the knee to take advantage of the large amount of negative work that muscles perform about this joint (Fig. 1). It used a one-way clutch to transmit only knee extensor motions, a spur gear transmission to amplify the angular speed, a brushless DC rotary magnetic generator to convert the mechanical power into electrical power, and a control system to determine when to open and close the power generating circuit based on measurements of knee angle (Fig. 2). A customized orthopedic knee brace supported the hardware and distributed the device reaction torque over a large leg surface area (Fig. 3). For convenient experimentation, the control system resided on a desktop computer and resistors dissipated the generated electrical power. The device was efficient and the control system was effective at selectively engaging power generation.

4. EXPERIMENTAL RESULTS

We tested the energy harvesting performance on six male subjects walking on a treadmill at 1.5 $m \cdot s^{-1}$ while wearing a device on each leg. We estimated metabolic cost using a standard respirometry system and measured the electrical power output of the generator. In the generative braking mode, the control system selectively engaged and disengaged power generation to target the swing extension negative work region. Subjects generated 4.8±0.8 W of electricity with a 5.0±21 W increase in the metabolic cost compared to the walking while wearing the device but not generating power (P=0.6). The cost of harvesting in generative braking—the additional metabolic power required to generate one

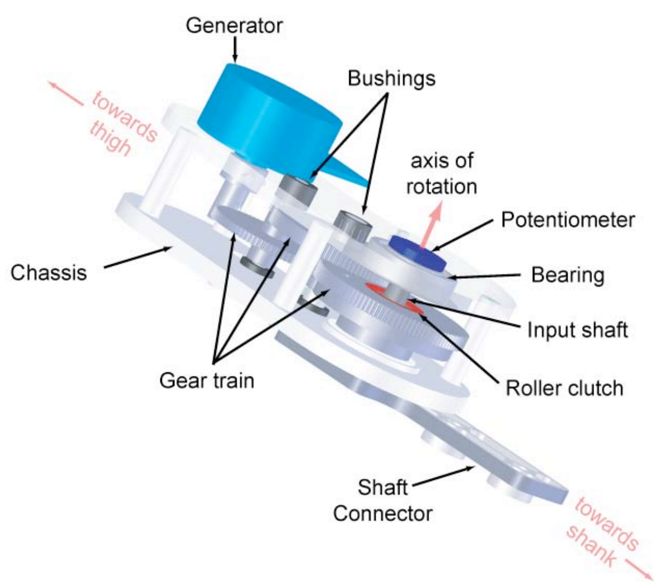

Fig. 2: A schematic of the chassis illustrates the location of transmission, generator and sensing components.

Watt of electrical power—was only 0.7±4.4 indicating that much of the electricity was generated from the deceleration of the leg [8].

4. DISCUSSION

While we have focused on harvesting energy from swing extension, power generation is possible from other periods of the gait cycle. At the beginning of the stance phase, for example, the knee flexes while the knee extensor muscles generate an extensor torque performing substantial negative work to aid in the redirection of the centre of mass velocity (Fig. 1) [24]. The amount of available energy at moderate walking speeds is only slightly less than that at the end of swing but it increases strongly with speed [23]. Consequently, our initial device design attempted to harvest energy from stance flexion in addition to swing extension. To accomplish this, we used two oppositely oriented roller clutches on the input shaft and an extra stage of gearing for knee flexion. This caused the generator to spin in the same direction regardless of the direction of knee motion and increased the gear ratio during flexion. While the higher gear ratio was required to better match the low angular velocity and high torque characteristics of stance flexion mechanical power (Fig. 1), it had a major drawback. Despite the fact that the control system opened the power generation circuit during swing flexion, the high angular velocity and acceleration during this period resulted in awkwardly large resistive forces due to the transmission and generator friction and inertia. This was not an issue for knee extension where power generation was engaged during swing extension, when knee angular velocity is high, and disengaged during stance extension, when knee angular velocity is low. While this drawback forced us to disregard power generation during stance flexion in the current design, future energy harvesting devices could approximate double power generation should a suitable mechanism for disengaging the transmission be found. Whether generative braking can be effectively accomplished during stance flexion will depend upon how much of the negative work during this period is stored and subsequently returned during stance extension. For now, generative braking during stance flexion is best considered a hypothesis that must be tested empirically.

Fig. 3: The device consists of an aluminum chassis and generator mounted on an orthopedic knee brace. The entire unit weighs 1.6 kg. While the subject in this image is wearing the device only on his left leg, all subjects were tested with devices worn bilaterally.

While future versions of this technology may prove useful to the general public for powering their portable devices, people whose lives depend on portable power will embrace it more quickly. Energy harvesting to trickle charge batteries in current computerized and motorized prosthetic limbs, for example, would allow amputees to walk further and faster. It would also enable future prosthetic and orthotic technologies to become more sophisticated by alleviating some of the limitations that batteries currently place on their design. The key principles are considerably more general than the current embodiment—they extend to joints other than the knee and to movements other than walking. The principles could also be embodied in a fully implanted energy harvester to power implanted devices, such as neurostimulators and drug pumps, increasing their duration of operation and enabling new power-demanding technologies. Irrespective of if they are embodied in a wearable or implanted design, energy harvesters that operate about body joints and selectively engage power generation have the potential to improve the quality of life for the user without increasing their effort.

ACKNOWLEDGEMENTS

We are grateful to Art Kuo, Andy Hoffer and Doug Weber for their dedicated collaboration. This research was supported by an MSFHR Postdoctoral Trainee award to QL, an NSERC USRA to V. Naing and an MSFHR Scholar Award, a CIHR New Investigator Award, and CIHR Operating Grant 159895 to JMD.

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