Energy Harvesting from the Biomechanical Movements of Human Body

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Abstract: In this presentation, the subject of biomechanical energy harvesting, and the studies performed in this field are introduced. Currently used lower limb prostheses manufactured with modular components cannot properly provide the expected functions and the needs of daily living activities due to their passive structure. Although substantial effort has been made in the field of developing active prostheses, these devices have not adequately become widespread because of the necessity of carrying large and heavy batteries which must have been charged frequently. Therefore, some studies have been performed in order to generate energy by utilizing the biomechanical movements of the human body, such as a mechanism converting the mechanical energy from the vertical movement of carried suspended-load in backpack to electricity, and an energy harvester mounted at the knee joint which generates electricity, during human walking. To meet a portion of the energy requirement for amputees wearing the active lower limb prosthesis, and for individuals having high electricity demands in rural areas are aimed with the harvested energy.

Introduction

Many people with lower extremity amputations are using prostheses for restoration of their lost functions. The effective restoration of amputees' lost functions can be acquired by the use of these prosthetic devices. This is one of the most important factors improving their life quality. Passive prostheses being currently in use do not respond to the needs of daily living activities of many amputees. For example, it is difficult to climb stairs with natural posture and to adjust the stiffness of the knee joint motion during the swing phase. High metabolic energy consumption and insufficient symmetry of the gait are the consequences of non-powered artificial joints. The duplication of the kinematics and dynamics of gait patterns is limited with conventional prostheses. They do not allow knee extension after heel strike at the beginning of the stance phase. The absence of the prosthetic leg's push-off phase, which is due to the sudden contraction of the shank's back face muscles at the end of the stance phase, causes the insufficient gait symmetry, shortens the stride length and decreases the gait velocity. In order to remove these disadvantages, it is necessary to add energy producing or storing modules to the system (Kapti, 2007).

On the other hand, humans have become increasingly dependent on technology, particularly electronic devices. During the past decade, electronic devices have become more mobile, enabling people to use medical, communication, and global positioning system devices as they move around cities or in the wilderness. At present, all of these devices are powered by batteries, which have a limited energy storage capacity and add considerable weight. Although substantial progress has been made in reducing the power requirements of devices and increasing the power densities of batteries, there has not been a breakthrough in the parallel development of a portable and renewable human-driven energy source. The combination of limited energy and the large weight of batteries poses the most critical problem for individuals, such as field scientists or explorers, having high electricity demands in remote areas and who are already carrying heavy loads. At present, replacement batteries may make up a substantial proportion of the very heavy packs that such users must carry (Rome, 2005).

During terrestrial locomotion, the environment does no work on the body and humans do no work on the environment. Almost all of the mechanical work is generated and dissipated inside the body. This makes it exceedingly difficult to capture mechanical energy to drive an electrical energy conversion apparatus, because the device would need to be either surgically placed within the body or attached to the outside of the body, which would affect the person's maneuverability and comfort. Therefore, researchers in the field have focused on putting devices in the only accessible location. Although the shoe is the first thing comes to mind, such heel-strike devices have permitted only small levels of electrical energy generation. The primary reason for this limitation is that on a hard surface, essentially no mechanical work is done at the foot-ground contact point, because under normal circumstances the point of vertical force application does not move in the vertical plane. Although one can make the shoe compliant so that the foot moves a small distance because of compression of the sole and heel, this is problematic because increasing compliance leads to declining maneuverability and stability. Although considerable effort has gone into developing exotic energy-generating technologies for shoe devices, the small magnitude of the mechanical energy source remains a limitation (Rome, 2005).

In order to help solving mobile human-driven energy problem, some studies for developing energy harvesting device which extracts mechanical energy from the human body movements during daily living activities, and converts it to electricity for powering portable devices were performed in the literature. The studies performed in the field of energy harvesting from the human body movements are mostly been on the regions of back, knee joints, and foot. In this review, after giving one example from the literature for each of these classifications, the applicability of biomechanical energy harvesting approaches in the field of active lower extremity prostheses will be examined.

In the Backpack

The vertical movement of a heavy load in the backpack carried in gravitational field during walking represents a source of mechanical energy and a potential opportunity to generate substantial levels of electricity. A walking person acts like an inverted pendulum. Due to this movement causing the center of mass of the body move up and down by 4 to 7 cm, a load in a backpack has to go up and down the same vertical distance. In the case of a 36-kg load and a 5-cm vertical load displacement, 18 J of mechanical energy transfer accompanies each step, and this is equivalent to 36 W, at the walking velocity of 2 steps per second. Although this represents a large potential source of mechanical energy, it is also inaccessible if the load is rigidly attached to the body. In order to extract this mechanical energy, Lawrence C. Rome et al. (Rome, 2005) developed the suspended-load backpack device decoupling the load from the body, to allow the differential movement between the load and the body for mechanical energy extraction and ultimately electricity production. In this device interposed between the body and the load (Fig. 1), the pack frame is fixed to the body, but the load is suspended by springs from the frame. During walking, the load is free to ride up and down on bushings constrained to vertical rods. Electricity generation was accomplished by attaching a toothed rack to the load plate, which when moving up and down during walking, meshed with a pinion gear mounted on a geared dc motor, functioning as a generator, rigidly attached to the backpack frame.

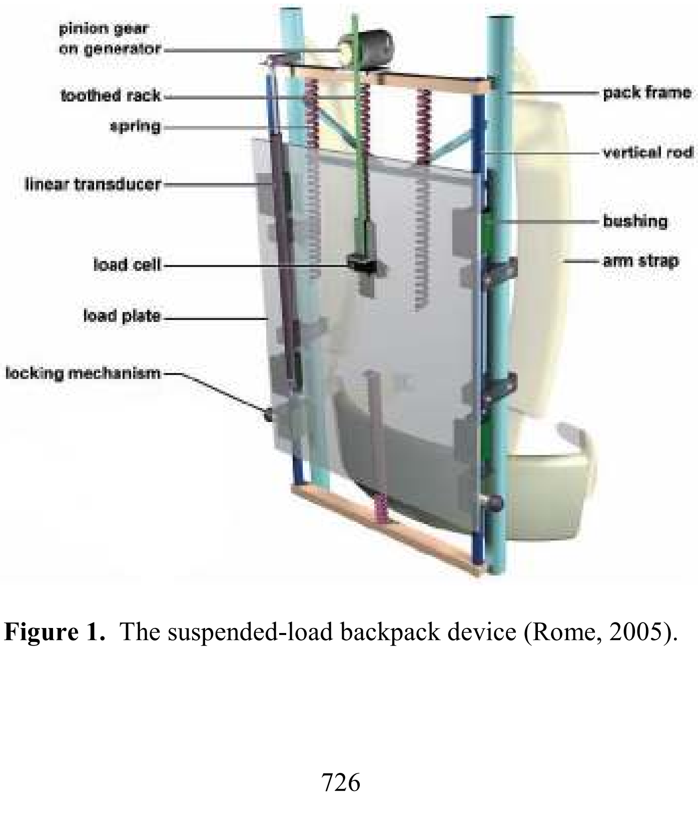

Figure 1. The suspended-load backpack device (Rome, 2005).

The average electrical power obtained by them was 5.6 W in the trial of 38-kg load and 4.5-cm relative movement of the load, and the number of revolution of 25:1 geared dc motor was reached up to 5000 rpm. Average electrical power increased with walking speed and the weight of the load. The maximum electrical power output obtained on the flat was 7.37 W. The mechanical power harvested by the generator is the product of the average force exerted on the rack, the displacement of the load, and the step frequency. The efficiency of conversion of mechanical energy to electrical energy (that is, electrical power output divided by mechanical power input) was nearly constant (30 to 40%). To power portable devices or charge batteries, the alternating polarity of the voltage and current must be rectified. Using circuitry for voltage smoothing, the suspended-load backpack can power multiple devices such as cell phones (Rome, 2005).

If generating electricity while wearing the backpack markedly increased metabolic rate, the device would be of limited use. One would expect that because mechanical energy is continuously removed from the system by the generator, the muscles would need to perform additional mechanical work during electricity generation in order to replace it. For instance, the mechanical power input to the generator is 12.15 W while walking at 5.6 km/h and carrying a 29-kg load. Because the maximum efficiency of mechanical power production by human muscle is about 25% (Margarira, 1968), if the body movement was the same, one might anticipate a minimum increase of 48.6 W in metabolic power input. They measured the rate of O_2 consumption and CO_2 production of participants walking with the backpack. They found that the metabolic rate increase compared to that with the locked backpack was only about 19.1 W, which is much less than would be predicted. These results indicate that electricity can be generated metabolically more cheaply than anticipated (Rome, 2005).

The energy-harvesting backpack is novel because it generates useful amounts of electrical power while costing less metabolic energy than would be expected. The saving only applies in comparison to a person already walking with a heavy load. The explanation may lie in the transition between pendulum-like walking steps, when the body's center of mass is redirected from one pendular arc to the next (Fig. 2). The center of mass is located near the hip joints and undergoes a small U-shaped displacement during this step-to-step transition, which occurs mainly when both legs contact the ground. Force is exerted by, and directed along, each leg, with the leading leg performing negative work on the center of mass and the trailing leg positive work. The leading leg's force is at such an angle with the direction of center of mass displacement that negative work is unavoidable, if the center of mass is to be redirected to another pendular arc. This negative work is thought to be largely dissipated as an energy loss. An equal magnitude of positive work performed by the trailing leg cancels this loss, as is needed to walk at steady speed (Kuo, 2005).

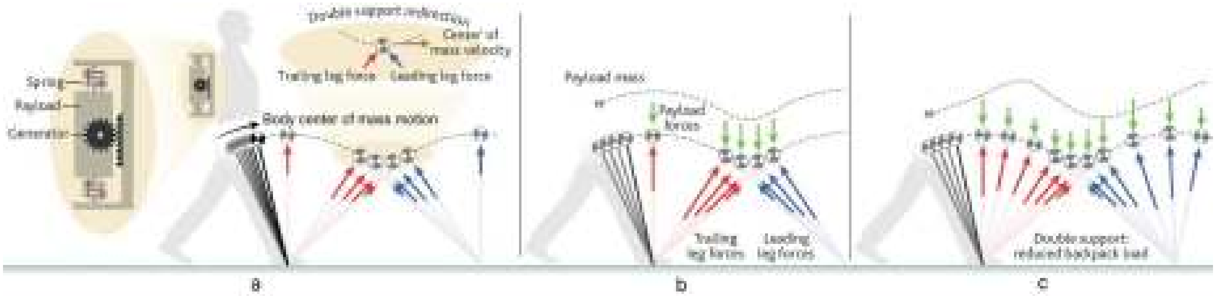

Figure 2. Simple models of an energy-harvesting backpack and its relation to human walking (Kuo, 2005).

On the Knee Joint

J. M. Donelan et al. (Donelan, 2008) have developed a device that generates electricity during human walking with little extra effort. The general view, the internal structure and the schematic diagram of this device called biomechanical energy harvester are shown in Fig. 3. The device has an aluminum chassis and generator mounted on an orthopedic knee brace, totaling 1.6-kg mass, with one worn on each leg (Fig. 3-A). The chassis contains a gear train that converts low velocity and high torque at the knee into high velocity and low torque for the generator, with a one-way roller clutch that allows for selective engagement of the gear train during knee extension only and no engagement during knee flexion (Fig. 3-B). The schematic diagram shows how a computer-controlled feedback system determines when to generate power using knee-angle feedback, measured with a potentiometer mounted on the input shaft (Fig. 3-C). For electrical power generation over longer durations, it would be desirable to harvest energy from everyday activities such as walking. Unlike conventional human-powered generators that use positive muscle work, their technology assists muscles in performing negative work. Energy-harvesting performance was tested (see Donelan, 2008) on six male subjects who wore a device on each leg while walking on a treadmill at 1.5 m/s. For convenient testing, generated electrical power is dissipated with a load resistor rather than being used to charge a battery. The energy harvester mounts at the knee and selectively engages power generation at the end of the swing phase. Test subjects walking with one device on each leg produced an average of 5 W of electricity. They estimated metabolic cost using a standard respirometry system and measured the electrical power output of the generator. In the continuous-generation mode, subjects generated 7.0 ± 0.7 W of electricity with an insignificant 18 ± 24 W increase in metabolic cost over that of the control condition. This electricity is sufficient to power 10 typical cell phones simultaneously. The results demonstrate that substantial electricity could be generated with minimal increase in user effort. Producing substantial electricity with little extra effort makes this method well-suited for charging powered prosthetic limbs and other portable medical devices (Donelan, 2008).

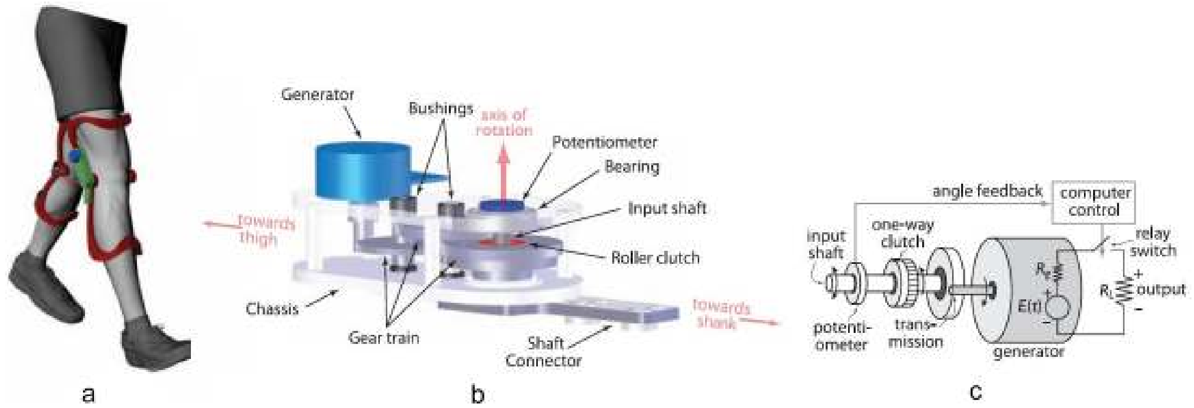

Figure 3. Biomechanical energy harvester (Donelan, 2008).
(a: the general view of the device, b: the inertial structure of the device, c: the schematic diagram of the device)

Under the Sole

The ankle normally produces a larger work than any other joint during walking (Winter, 1991). Ankle impairments following amputation, joint fusion or stroke typically reduce ankle work and increase metabolic energy expenditure by at least 20%, comparable to carrying an extra 15 kg load or walking 20% faster. Ankle function might be restored by powering the joint directly, a technique that shows promise (Sawicki, 2008, Au, 2009) but requires large motors and energy sources that are heavy and bulky. Much of the dissipation in normal walking occurs when the body center of mass velocity is redirected at the transition between steps. During each step, the stance leg behaves similarly to an inverted pendulum as it transports the center of mass along an arced path. When the other leg contacts the ground, it flexes slightly and performs dissipative negative work as it redirects the center of mass to the arced path of the next step as part of the step-to-step transition. To walk at steady speed, all dissipation must be recovered by an equal amount of positive work. Total work may theoretically be minimized if the positive work is performed by trailing leg push-off and timed immediately before heel-strike, reducing the change in center of mass velocity performed by the collision. This reduces both the dissipation and the amount of positive work needed to recover loss. Normal ankle push-off appears appropriate for this purpose, performing positive work beginning just before and in nearly equal magnitude to the collision loss. If the collision energy can be successfully recycled, it may therefore be sufficient to supplement an impaired push-off (Collins, 2009).

Steven H. Collins and Arthur D. Kuo (Collins, 2009) developed an energy-recycling artificial foot (Fig. 4) that captures collision energy and returns it for push-off. 1.37-kg weighed this device approximates the size and form of a conventional prosthetic foot, but has separate rear-foot and fore-foot components that rotate about an axis at mid-foot. When the heel contacts the ground at the beginning of a stride, the rear-foot component rotates and compresses a coil spring. At maximum compression, the rear-foot is latched by a continuous one-way clutch. Rather than releasing the spring energy spontaneously as in conventional elastic prostheses, our device stores it until sufficient load is detected on the fore-foot. It then releases the fore-foot, and the spring provides push-off as the person begins to unload the trailing leg, with timing similar to normal ankle push-off. A small return spring resets the device during the ensuing swing phase, so that the rear-foot is in position for the next step. All of the energy capture is performed passively, so that the only active elements are a microcontroller and two micro-motors that release the energy-storing spring and reset the mechanism. The device is powered by a small battery at about 0.8 W of electricity. Active control of energy storage and return distinguishes this device from conventional prosthetic feet with passive elastic elements, which have not been found to significantly reduce the metabolic energy consumption of walking with ankle impairment, while low electrical power requirements distinguish it from other robotic prostheses.

Steven H. Collins and Arthur D. Kuo (Collins, 2009) tested the artificial foot on able-bodied human subjects walking with an artificially-immobilized ankle, at a speed of 1.25 m/s. Subjects wore the device on one leg using a prosthesis simulator, a rigid boot that immobilizes the ankle and provides a prosthesis attachment beneath the foot. This allowed direct comparison between normal walking and prosthesis test conditions. Subjects also wore a lift shoe on the other foot to equalize height. The device was compared against a conventional prosthetic foot. Mechanical performance was recorded through motion capture and a force plate-instrumented treadmill. They used motion and force data to estimate the work captured and returned by the device, the work performed by the human leg and device on the center of mass, and the work performed at each biological joint. They also recorded rates of oxygen consumption to estimate metabolic energy expenditure. The conventional prosthesis reduced ankle push-off and increased metabolic expenditure for all subjects. The energy recycling artificial foot captured collision energy and returned it as positive ankle work later in stance phase, resulting in greater push-off and lower metabolic expenditure than with the conventional prosthesis. The rate of increasing of metabolic expenditure was determined as 23.1% for conventional prosthesis, and as 13.8% for the energy recycling artificial foot, and 9.3% improvement was provided (Collins, 2009).

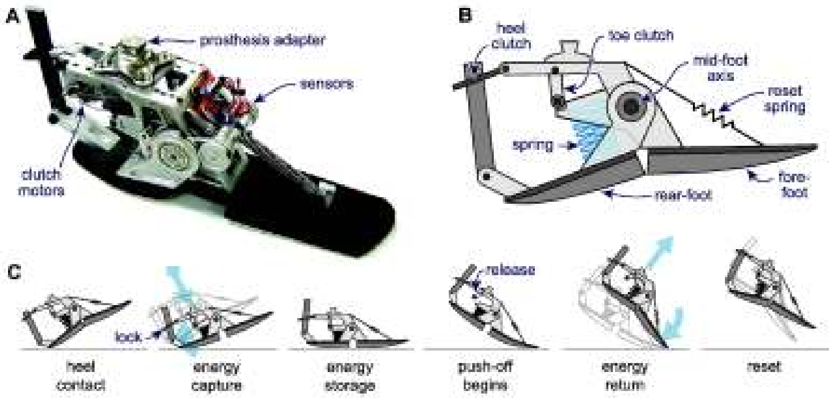

Figure 4. Prototype energy recycling device (Collins, 2009).
(A: The general view of the device, B: Schematic design, C: The energy recycling sequence)

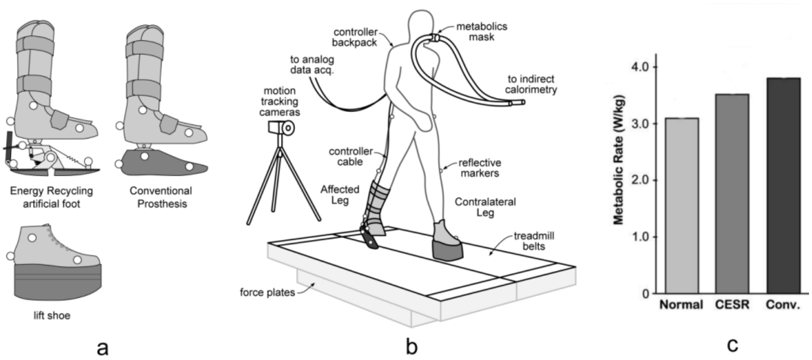

Figure 5. Experimental setup (Collins, 2009).
(A: The energy recycling device, conv. prost. and the lift shoe, B: Experimental setup, C: Experimental results)

Applications on the Active Prostheses

Currently used lower limb prostheses manufactured with modular components cannot properly provide the expected functions and the needs of daily living activities due to their passive structure. In order to contribute to the developments of new kinds of prosthetic system and to remove the insufficient properties of the prostheses, a force controlled elastic prosthesis mechanism that can be utilized as artificial ankle and knee joints for active lower extremity prostheses was designed and produced as a mechanism consisting of brushless dc-servomotor, ball-nut and screw, elastic component, measuring elements, guide columns, ball bearings and bushes. The force output of the elastic mechanism is calculated by measuring the displacement of the spring with the linear potentiometer. An above-knee prosthesis consisting of this elastic mechanism was also designed and produced. General view of this above-knee prosthesis, and the principle of application on human body are shown in Fig. 6.

Although substantial effort has been made in the field of developing active prostheses, these devices have not adequately become widespread because of the necessity of carrying large and heavy batteries which must have been charged frequently. This system has to carry its power generating system consisting motor component and battery set, which is heavy and bulky. Our system consists of the 220 W servomotor and Li-ion battery set. Mobile energy requirement is the most crucial difficulties faced in the externally powered artificial orthopaedic devices. Therefore, in order to solve this difficulty, utilizing the studies mentioned above is proposed [a mechanism converting the mechanical energy from the vertical movement of carried suspended-load in backpack to electricity (see Rome, 2005); an energy harvester mounted at the knee joint which generates electricity (see Donelan, 2008); an energy recycling device (Collins, 2009)]. To meet a portion of the energy requirement for amputees wearing the active lower limb prosthesis, and for individuals having high electricity demands in rural areas are aimed with the harvested energy.

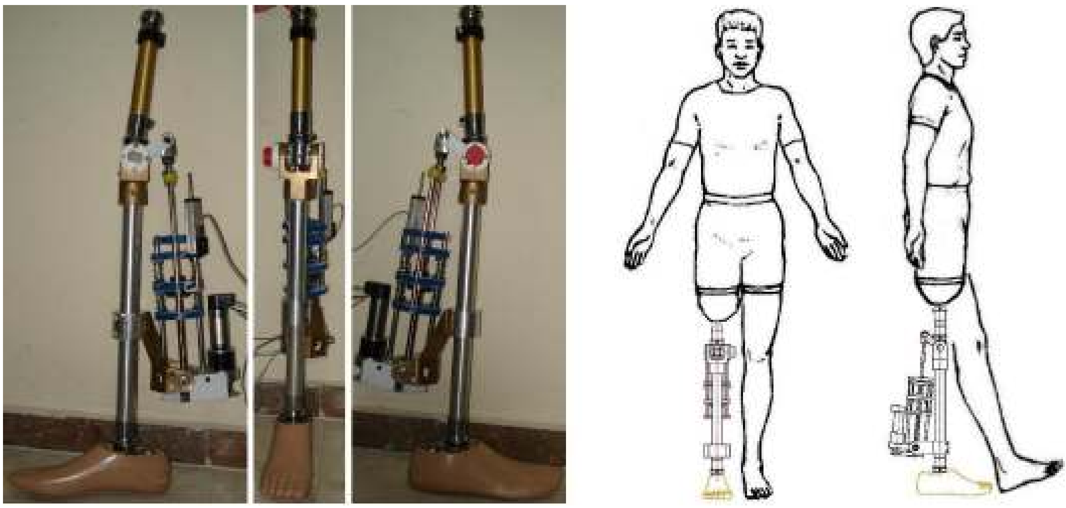

Figure 6. General view of the active above-knee prosthesis, and the principle of application on human body (Kapti, 2009).

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