KINETIC ENERGY HARVESTER FOR BODY MOTION

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Abstract: This paper reports the design, fabrication, and testing of a permanent-magnet energy harvester generator for body motion. The described generator was capable to extract up to 117µW of power from walking, employing a non-resonant rotational mechanism. The planar generator proposed is composed of a rotor having multiple NdFeB permanent magnet pole-pairs with an eccentric proof mass. The stator has a multi-stacked gear-shaped planar coil. A change on the magnetic flux induces a voltage in the coil when the rotor moves as a result of body movement. A 2cm^3 prototype was tested on several body locations while walking.

Keywords: Kinetic Energy Harvesting, Body Motion

INTRODUCTION

Portable electronic devices are a common denominator, but they need to be connected to the electric grid or powered by batteries. Although battery technology has advanced in the last decades, finite battery lifetime is still a limitation for some applications; such as embedded or implantable devices. Standalone systems that need to be operational for several years are mostly dominated by battery size and limited by battery power constraints. For instance, batteries for cardiac pacemakers occupy at least half the device's volume [1].

Energy harvesting is a technology that has progressed at large scale from wind mills to wind turbines. Lately, energy scavenging research has been focusing on smaller applications for portable electronic devices or standalone systems. For example, body motion for energy scavenging has been used for decades to power self-winding wristwatches. Actual power requirements for standard electronic applications tend to exceed the power outputs produced by energy harvesters [2]. As a result of the technology progress, new low-voltage and low-power circuitry can close the gap to make energy scavenging a feasible option for powering more devices.

Therefore, if kinetic generators can be developed for scavenging energy from human movements, the power produced can be employed to energize portable electronic devices or biomedical applications. As a consequence, wearable, embedded, or surgically implantable devices can have longer lifespan, reduced sizes, or increased functionality.

The design of kinetic energy harvesters for body motion is a challenging area because traditional generator architectures follow the cantilever beam design, which is well suited for high-frequencies (>100Hz) and small displacements (<1mm) applications. In addition, these types of generators rely on large quality (Q) factors that need to be tuned at a fixed or narrow frequency range. On the other hand, human activities are characterized by large amplitudes (>1cm), and low frequency content (<10Hz). This kind of motion source makes it difficult to design a small generator that matches a person's broad frequency content while having large Q-factors.

The non-resonant planar rotational design presented here is found to be satisfactory for body motion. The proposed generator is composed of an eccentric weight rotor having multiple NdFeB permanent magnet (PM) pole-pairs and a stacked planar coil stator, as shown in Fig. 1. The eccentric rotor induces a voltage on the coil when it turns due to an external movement, a behavior akin to self-winding wristwatches. Although research has been pursued for energy harvesting; little is reported for rotational generators that convert energy from human-based activities [3-5].

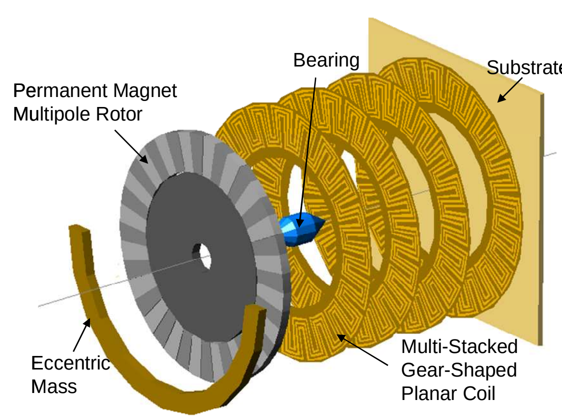

Fig. 1: Simplified schematic of the proposed design.

GENERATOR DESIGN AND TESTING

Previous research [6] has shown that body motion is an untapped source for energy harvesting with available power levels comparable to those obtainable from machines. For example, it is described that activities such as walking have available power densities on the order of $1mW/cm^3$ or even higher for more energetic movements. This is immediately related to the available power for a system oscillating at resonance with mass *m*, acceleration *a*, frequency ω, and quality factor *Q* ($P_{available} = ½\, mQa^2/\omega$). From the previous expression, the acceleration-squared-to-frequency (*ASTF*) term is the only one related to how energetic the source is. Therefore, the available power will depend on the mass of the vibrating generator, the *Q* factor and the *ASTF* value. Higher values of these terms are desirable for higher power output. Narrow frequency bandwidth is inherently associated with large *Q* factors for machine designs while presenting small *ASTF* values. The broad frequency bandwidth of body motion is translated into low *Q* factors, but with high *ASTF* values as presented by [6]. In summary, lower body locations present *ASTF* values as high as 20-120, while upper body locations show *ASTF* numbers from 1 to 8. Hence, the placement of energy harvesters on the lower body can produce up to 100 times more power output than on upper body locations. A preliminary test measuring the peak acceleration from several body locations was performed on a treadmill, Fig. 2. Lower body locations exhibit larger acceleration peaks than upper body targets.

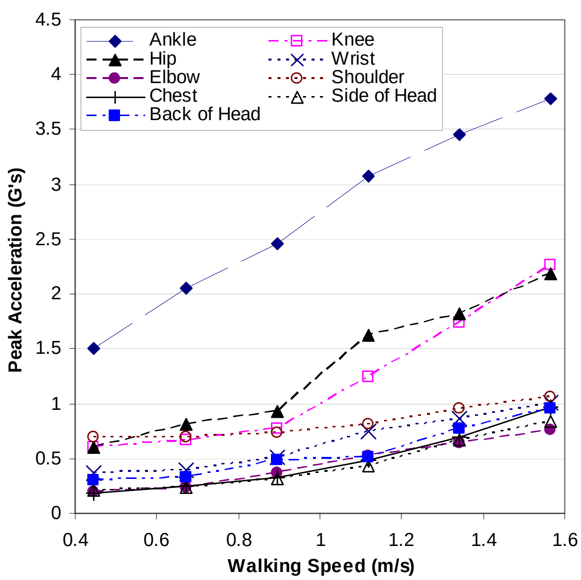

Fig. 2: Peak acceleration on several body locations while walking at different speeds on a treadmill ($1G=9.8m/s^2$).

Given that human body movements are not constrained exclusively to individual translations or rotations, the planar prototype design presented is found to be sensitive to translation, rotation, or position changes. The generator is more susceptible to motion if the device is held vertically rather than lying flat since it will oscillate as a pendulum does. Two different prototype designs were considered: the first having two rotors with one stator in between, and the second having two stators with one rotor in between. The first design maximizes the magnetic flux to induce more voltage, while the second one maximizes the number of coil layers to induce a larger voltage. The first design has been tested for a meso-scale prototype, while the second architecture is planned to be implemented at MEMS-scale.

A meso-scale prototype was built with a volume of $2cm^3$ (without the casing) as shown in Fig. 3. 20 PM pole-pairs for the rotor assembly were made of discrete NdFeB pieces (1.1mm x 1.1mm x 5.1mm) inserted radially in a 25mm slotted PMMA disk. Two eccentric 1.6g bronze masses were added to the rotor assembly. The stator was made by stacking layers of a gear-shaped planar coil (25mm diameter) made of commercial copper-clad polyimide film using photolithography techniques, as shown in Fig. 4. Coil linewidth dimensions are limited by the photoplotter resolution on the actual prototypes. Two PMMA casing structures were machined with a recess to contain the generator parts. The prototype was assembled by placing the stacked planar coil in between the two rotors (1mm gap separation) using an alignment fixture. Jewel bearings were used to provide low-friction rotor support. Coil resistance for each 200μm linewidth layer was calculated as 4.4Ω, but measured as 4.9Ω. This difference is explained by copper undercut while etching as evidenced by micrograph analysis.

Fig. 3: Two-rotor meso-scale prototype. Rotor with 20 pole-pairs of NdFeB PM (1.1x1.1x5.1 mm) and a jewel bearing design (bottom rotor not shown).

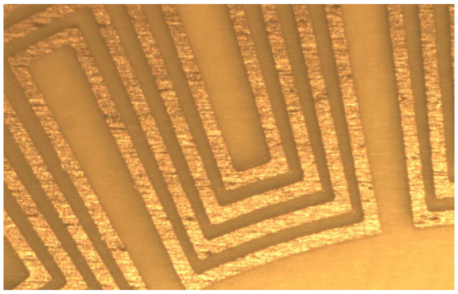

Fig. 4: Detail of a 200μm linewidth and 4 turns copper planar coil on polyimide substrate.

Testing was done by placing a 2-layer and a 4-layer coil prototype on different body locations while walking. The test with the 2-layer coil was executed a total of seven times (1min recording time). The results for this test (average, maximum, and minimum values) are shown in Fig. 5. The 4-layer coil test was done on a treadmill at several walking speeds. A PC-based data acquisition system was used to record the voltage output with matching resistive loads. Analysis of the recorded data was performed on MATLAB.

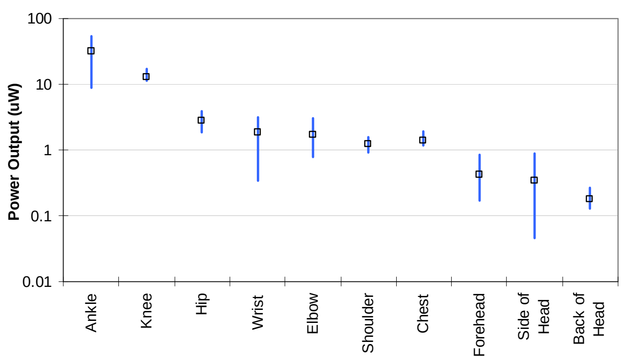

Fig. 5: Average, minimum and maximum power output for a 2-layer coil generator placed on different body locations while walking (seven tests).

Fig. 6 shows an example of the recorded voltage output. A peak voltage of 326mV (70mV$_{rms}$) and an average power output of 54.7μW were obtained on one of the trials when the generator was placed on the ankle while walking (power density of 27.3μW/cm^3). Fig. 7 presents the treadmill test results for 30s recordings. Power output levels from 10μW to 100μW were obtained for the prototype placed on the lower limbs. The generator produced power higher than 10μW at the lower limb locations even for slower walking speeds (0.45m/s). In contrast, power output levels less than 10μW were recorded for most of the upper body locations. Power output as high as 117 μW (137mV$_{rms}$) was obtained for one of the trials when walking at a faster pace (1.34m/s), for a peak power density of 58.5μW/cm^3.

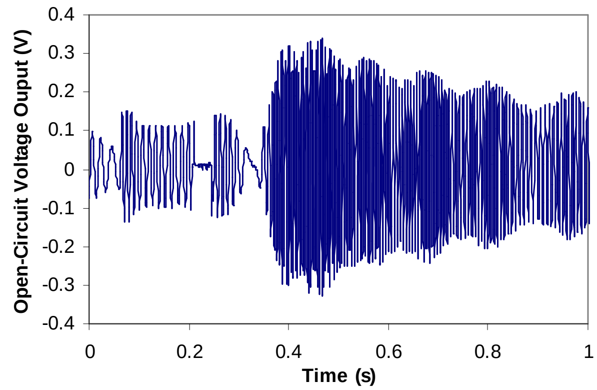

Fig. 6: Open-circuit voltage while walking with the generator laterally placed on the ankle for the 2-layer coil test (200 μm line width and 4 turns) and 20 discrete NdFeB PM pole-pairs (1.1x1.1x5.1mm). A peak voltage of 326mV was reported (70mV$_{rms}$).

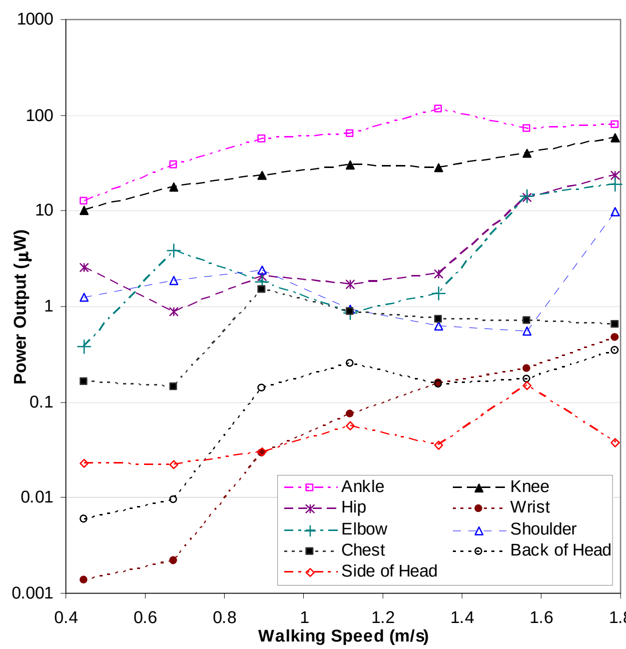

Fig. 7: Power output for a 4-layer coil generator placed on several body locations while walking at different speeds on a treadmill.

DISCUSSION

Test results showed that prototype energy harvester generators situated on the lower limbs are capable of producing up to 100 times more power output than upper body located generators, as appeared in Fig. 7. *ASTF* values indicate that generators on lower limb locations can produce between 10 to 100 times more power than those devices at upper body locations, which correlates well with experimental results.

Generators placed on the ankle and knee produced an increasing power output with increasing walking speed as evidenced by the semi-log plot in Fig. 6. Other body locations followed a similar pattern, although not as uniform as the one from ankle and knee placed generators. The hip-placed generator was expected to produce energy at levels comparable to the knee location, as indicated by Fig. 2. The larger angular momentum of the limbs can explain this difference. More testing is needed to describe the nature of this behavior.

Only 2-layer and 4-layer coil designs were tested for the generator prototype because of the ease of assembly. Further prototypes with an increased number of layers or with smaller coil linewidth (for a higher number of turns per coil) are expected to produce higher induced voltages and therefore larger power outputs. Larger output voltages are also an advantage for using solid state diode rectification. Generated energy can be stored in super capacitors or rechargeable batteries for posterior use when there is no body motion activity.

Sasaki et al [7] described a similar mechanism based on an electronic self-winding wristwatch. They reported the energy generation as being produced by posture change (as a transient rotation), by oscillatory or swinging motion, or by self-excited rotation. It was suggested that the primary mode for energy generation in wristwatches is due to postural changes rather than by the inertial forces of the arm's motion. The self-excited rotation mode happens when the rotor continues to rotate because of an initial momentum. This mode has the added advantage of increasing up to 10 times the power generated in comparison to swinging motion or postural change. The conditions to maintain these self-excited rotations were reported as well by Sasaki et al. From the tests performed it was observed that the generator tended to rotate more often at high-energetic locations (lower limbs) rather than at less-energetic locations (upper body). This agrees with observations from wristwatch generators by Sasaki et al. Although, this also could be explained by higher *ASTF* factors found on lower limb joints which translates into a higher available power.

CONCLUSION

The non-resonant rotational planar prototype design reported here was shown to harvest kinetic energy from human walking. Generators placed on body locations with high *ASTF* values presented larger power outputs. In contrast, devices situated on body locations with smaller *ASTF* values produced smaller power outputs. A maximum of 117µW was produced while walking at 1.34 m/s with the generator placed on the ankle (power density of 58.5µW/cm^3). The low profile design makes it a practical choice to be embedded or incorporated into portable electronics or biomedical implantable devices. Such hybrid designs, consisting of rechargeable batteries and energy scavengers, can enhance the operation of actual devices for longer operation without replacement.

ACKNOWLEDGEMENTS

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