The Pizzicato knee-joint energy harvester: characterisation with biomechanical data and effect of backpack load

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Abstract

The reduced power requirements of miniaturised electronics offer the opportunity to create devices which rely on energy harvesters for their power supply. In the case of wearable devices, human-based piezoelectric energy harvesting is particularly difficult due to the mismatch between the low-frequency of human activities and the high-frequency requirements of piezoelectric transducers. We propose a piezoelectric energy harvester, to be worn on the knee-joint, that relies on the plucking technique to achieve frequency up-conversion. During a plucking action, a piezoelectric bimorph is deflected by a plectrum; when released due to loss of contact, the bimorph is free to vibrate at its resonant frequency, generating electrical energy with the highest efficiency. A prototype, featuring four PZT-5H bimorphs, was built and is here studied in a knee simulator which reproduces the gait of a human subject. Biomechanical data were collected with a marker-based motion capture system while the subject was carrying a selection of backpack loads. The paper focuses on the energy generation of the harvester and how this is affected by the backpack load. By altering the gait, the backpack load has a measurable effect on performance: at the highest load of 24 kg, a minor reduction in energy generation (7%) was observed and the output power is reduced by 10%. Both are so moderate to be practically unimportant. The average power output of the prototype is 2.06±0.3 mW, which can increase significantly with further optimisation.

Keywords

Plucked piezoelectric bimorph, frequency up-conversion, energy harvesting, human gait, wearable energy harvester, knee-joint harvester.

Introduction

The continuous progress made in electronic miniaturization is delivering sophisticated systems with modest power consumptions. This has driven the development of wearable devices which rely on batteries as their power supply, but has also opened up the possibility of developing battery-free devices by exploiting a variety of techniques to harvest energy around the human body. The replacement of batteries with the renewable source of energy afforded by energy harvesters (EHs) has the two-fold advantage of reducing maintenance (battery replacement or recharging) and of increasing the reliability of the devices by removing the risk of being left without the power necessary for operation. This is of particular importance in medical applications, where any downtime might be life-threatening and where battery replacement may require surgery [1].

Among the areas on the human body where harvestable energy is available, the most interesting ones are the foot and the knee-joint [2]. The biomechanical advantage of a knee-mounted brace is that during normal walking the angular displacement of the knee-joint is large and it delivers significant angular velocities at typical walking speeds. In addition, the attachment of such braces is simple, stable and repeatable. An electromagnetic device for harvesting energy from the knee-joint was presented in [3]; the power generation was very high at 4.8±0.8W during walking, however, due to the nature of the electromagnetic generator, the prototype included gears and other mechanisms, which negatively impacted on its complexity, size and mass. Piezoelectric materials are smart materials with the ability to couple the mechanical and electrical physical domains; when used as energy generators, piezoelectric materials such as the ceramic PZT are efficient and compact. On the other hand, piezoelectric generators are at their most efficient when operated at high frequencies, which are not always available. To bridge this frequency gap, frequency up-conversion techniques have been proposed which rely on impact [4,5] or on slow deflection and quick release of bimorphs [6,7]. We refer to the latter technique as plucking or *pizzicato* for its similarity with the technique used for corded musical instruments In a previous work [8], two of the present authors, Pozzi and Zhu, have used finite element techniques to investigate the response of piezoelectric bimorphs to plucking excitation. Modelling and experimental results showed that during both the loading phase and the free vibrations following release, the direct piezoelectric effect converts a significant proportion of mechanical energy into electrical energy. It was also found that the energy produced in each plucking action increases with the speed of deflection in the loading phase. In a subsequent paper [9], Pozzi and Zhu presented a rotary piezoelectric harvester based on plucking excitation. The prototype featured one PZT-5H bimorph and was tested at a selection of constant speeds over a complete revolution, so as to highlight the contribution of all the plectra to the overall energy generation. Among other results, it was confirmed that higher speeds enhance energy generation and it was shown that the manufacturing quality of the plectra has a very important effect on the energy they cause a bimorph to produce when they pluck it. The controlled conditions of the tests permitted the extraction of statistically relevant information, which are useful for a range of applications, even beyond the rotary configuration.

In this paper, we present a knee-joint wearable energy harvester prototyped by the Piezoelectric Energy Harvesting Research Group at Cranfield University; the harvester features four bimorphs fixed to a central hub and plectra embedded in an external ring. The electrical outputs from the four piezoelectric devices are individually rectified and measured. For this work, the harvester was mounted on a knee motion simulator which reproduced the gait pattern of a human subject carrying a selection of backpack loads. The knee-joint kinematics data used were acquired with a camera-based motion capture system.

The principal aim of this work is to investigate the energy generation performance of the prototype in simulated real world conditions; we also show that additional weight carried by the subject affects his gait, which then impacts the overall energy and power output of the harvester. The paper contributes to the growing corpus of research on wearable energy harvesting by offering a yet unexplored approach to human-based piezoelectric energy harvesting as no other research group has worked on plucked-piezoelectric generators for human-based energy harvesting, nor are there other examples of piezoelectric knee- joint harvesters. By producing an average of around 2 mW in the present form, the harvester presented here could power sensors or monitoring systems for a wide range of applications, both medical and of general interest.

Experimental methods

Knee-joint piezoelectric energy harvester

The knee-joint piezoelectric energy harvester is fixed to the outside of the knee by braces (Figure 1). As the wearer walks, the inner hub and the outer ring rotate relatively to each other with reciprocating motion, so that the four bimorphs (mounted on the hub) are forced to pass in front of the plectra (embedded in the outer ring). In this way, the plectra pluck the four bimorphs. The prototype of the EH (Figure 2) was realised based on this design. For testing, the prototype was mounted on a knee-joint simulator (Figure 2), which uses a stepper motor to reproduce the kinematics of the knee-joint of a human subject. The main advantage of testing the device on a simulator rather than directly on a human subject is the reproducibility of the tests as the motor will accurately reproduce the recorded gait at every run whereas the real gait cycle changes slightly from one step to the next. This reproducibility is important as it allowed us to carry out statistical analysis and isolate the contribution of the harvester alone. The electrical outputs produced by the bimorphs were rectified, dissipated across resistors and the corresponding voltage drops were recorded. Data reported in this paper were taken during 18 experimental runs, where each load-specific gait cycle (0, 12 and 24 kg, as specified later) was run three times for each direction of motion. Reversing direction of motion means mounting the device on the other leg or, alternatively, swapping the links to thigh and shank. This is relevant because the objective of the plectra's design was to increase energy production (and resistance to motion) during the swing extension phase of gait and minimize them during the stance phase, making it asymmetric. The asymmetry is beneficial as the muscles of the leg perform negative work during the swing extension phase [3]: the device would harvest energy which otherwise must be dissipated by the muscles.

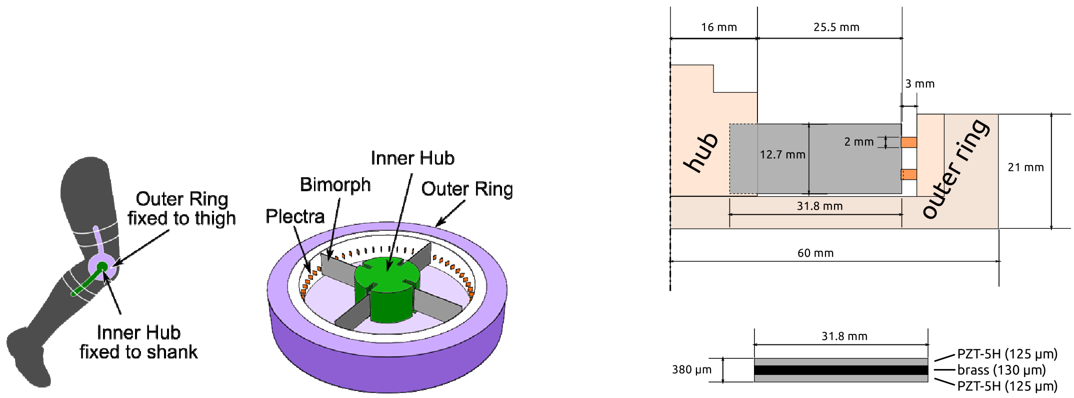

Figure 1: a. Knee-joint piezoelectric harvester. It is worn on the external side of the knee and fixed by braces. Inside, a hub carries a number of bimorphs which are plucked by the ring-mounted plectra as the joint rotates during walking. b. geometrical details of the harvester showing side view (above) and top view of mounted bimorph (below).

The EH features four bimorphs mounted on the inner hub and 74 active plectra mounted on the outer ring. The piezoelectric bimorphs, referred to as B1 to B4 in this paper, are of type T215-H4-303X produced by Piezo Systems Co.; more details are found in Figure 1b. The main selection criteria for the bimorph were commercial availability and appropriate dimensions to give a combination of compact size and good energy output.The bimorphs were mounted within copper-clad mechanical fixtures doubling up as pick-off electrodes. The plectra were cut out of a 125 μm-thick Kapton® polyimide film. The typical distance between plectra is 3.5±0.5 mm, although there are two gaps of about 8-9 mm along the circumference, due to manufacturing issues. The prototype harvester occupies a volume of 226 cm^3 and has an approximate mass of 235 g.

During testing, the hub was held steady by a bracket, while the ring was rotated by a stepper motor (M60STH88 from Motion Control Products Ltd.) controlled by a microstepper driver (MSD880, same supplier), which received low level control signals from a computer via a D/A card and current amplifiers. The controller requires two TTL-level signals: pulse and direction. A micro-step is generated by the controller at every rising edge of the pulse signal, whereas the high or low level of the direction signal controls the clockwise or counter-clockwise rotation. The controller was set for 20,000 micro-steps per revolution and the pulse signal was generated with a sampling period of 8μs, so as to ensure the necessary frequency of low-to-high transitions even at the highest speeds. Pulse and direction control files were prepared in MATLAB by an algorithm which calculated the time interval between two rising edges based on the instantaneous speed given by the biomechanical data. As all gait cycles were periodic, at the start of each experiment the relative position of hub and ring was the same, as verified by direct visual observation.

To measure the power generation performance, four electrical loads (55.9 kΩ resistors) were connected to the bimorphs via individual full bridge rectifiers (MULTICOMP – DBLS103G). The voltage-drops across the four loads were simultaneously sampled with a NI-9229 A/D card. The instantaneous power was then calculated as $P(t)=V^2(t)/R$, where $V(t)$ is the measured voltage and R is the equivalent resistor from the parallel between the electrical load and the internal impedance of the acquisition card, i.e. R = 55.9 kΩ // 1 MΩ = 52.9 kΩ. The vibration of the bimorph was measured with a Laser Doppler Vibrometer (LDV, Polytec CLV-2534); as it was not possible to shine the laser beam parallel to the direction of vibration, it was placed at a small angle (approximately 15°) and the resulting systematic error accounted for in the processing of the data. During the experiments reported here, the laser was shone on the tip of bimorph B4 (lower left in the photograph). The sampled velocity was numerically time-integrated to yield displacement and a base line was removed to account for the jitter of the hub caused by variable friction with the outer ring.

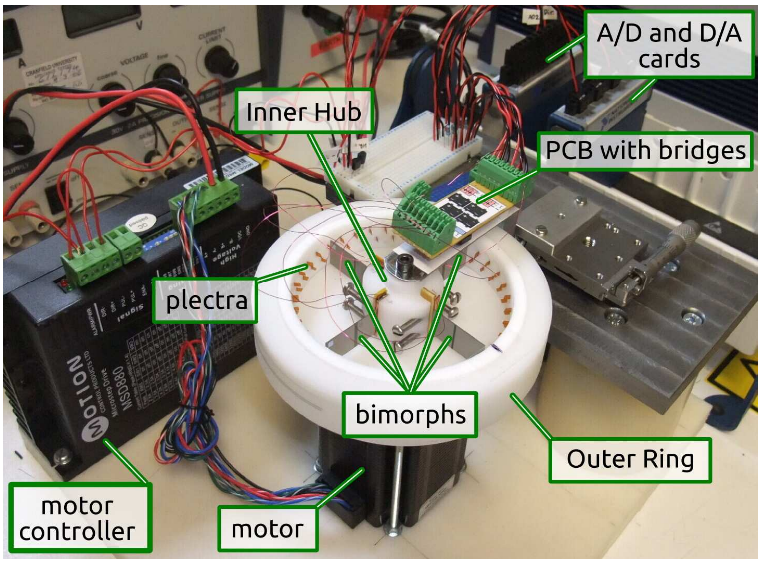

Figure 2. The energy harvester mounted on the knee simulator. The rectifying bridges are surface mount devices soldered to the underside of the circuit board (PCB), which is mounted above the bracket holding the hub steady. The motor and its controller are also visible.

Biomechanical data collection

Kinematic data derived from marker-based motion capture systems is the gold standard method for the accurate characterisation of human biomechanics. For this investigation, one healthy male subject was recruited. The trials were conducted at an unforced manner, at a speed self determined by the subject. A six camera Qualisys Proreflex MCU240 motion analysis system capturing at 100 Hz was used. Reflective markers were attached to the subject's lower limbs at the anterior superior iliac spine, posterior superior iliac spine, iliac crest, greater trochanter, fibula head, tibial tubercle, medial condyles, lateral condyles, calcaneus, lateral malleolus, medial malleolus. The markers form the basis of anatomical reference frames and centres of rotations of the joints. Five rigid plates, each consisting of four non-collinear markers, were also secured on the antero-frontal aspect of the leg, thigh and around the pelvis (Figure 3). The calibrated anatomical systems technique (CAST) [10] was employed to determine the movement of these segments during the walking trials. The subject undertook five repeated trials with three backpack load conditions of 0, 12 and 24 kg; the backpack loads were evenly distributed to avoid bias and all of the markers remained attached to minimise positional inconsistencies in re-attachment. The selection of backpack loads was dictated by the initial motivations of the project, sponsored within the Battery-free Soldier initiative of the MoD of the UK.

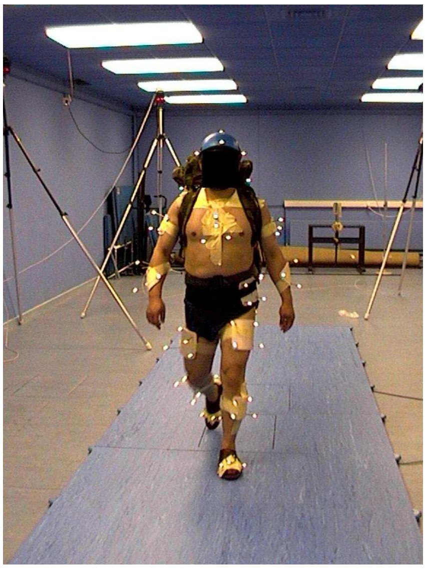

Figure 3. The marker configuration and motion analysis system used for the CAST.

The knee joint angle displacement was extracted from the main kinematics dataset. The joint kinematics was calculated using an X-Y-Z Euler rotation sequence in which the centre of the knee-joint is defined as the midpoint between the medial and lateral condyles markers [11]. The angle between the thigh and shank in the sagittal plane is used in the knee-joint simulator, where a naturally standing extension is calibrated to 0° and all higher angular displacements represent flexion (Figure 4), whereas further extension is possible beyond the natural standing position, giving negative angles. The angular displacement during each time interval of 0.01 s was averaged over all trials, giving a mean angular displacement sampled at 100 Hz. No temporal normalisation into one gait cycle was done in order to preserve the time values in seconds that are needed to measure the average power generation.

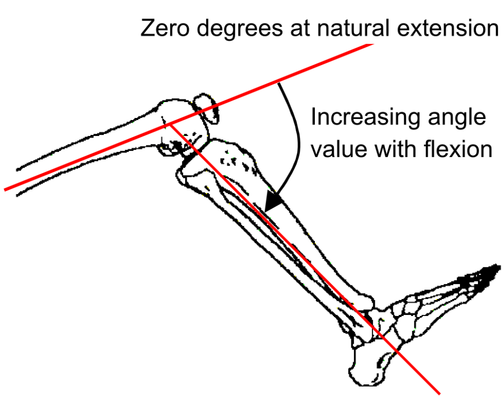

Figure 4. The joint angle values at the knee are calibrated to zero at a natural extension stance. Positive values represent flexion and negative values represent further extension.

Results and discussion

First, we show the effect of backpack load on gait as evidenced by kinematic measurements. Then, some typical measurements of tip displacement, voltage and instantaneous power are given as a function of time to clarify the operation of the harvester and the behaviour of the bimorphs in response to the plucking form of excitation. In a subsequent section, the energy generated during the gait cycle is presented, showing that the greatest contribution comes from the second half of the gait cycle and that there is a noticeable difference in the energy produced by the four bimorphs, with the best harvesting 26% more energy than the worst, in this instance. How this difference is affected by the backpack load and direction of rotation is discussed in the next subsection and its origin is elucidated. Finally, a statistical analysis of the total energy and average power produced in each experiment proves that the load carried by the subject has a measurable, albeit of limited practical importance, effect on the performance of the harvester. In particular, the average output power is lower with the highest backpack load of 24 kg, as less energy is harvested over a longer time, due to the increase in gait duration.

Biomechanical data, bimorph displacement, voltage and instantaneous power

Biomechanical data collected from a human subject carrying a selection of backpack loads confirm that the load influences the gait (Figure 5). The first peak in all three curves, reaching approximately 20°, is the flexion of the knee-joint immediately following heel-strike, when the leg is loaded with the body weight; the second peak, over 50°, is associated with lift-off, when the leg has lost contact with the ground and is carried forward in preparation for the following heel-strike. Whereas the gait cycles for 0 kg and 12 kg loads are very similar, the highest load of 24 kg forces the subject into a significantly different gait pattern, which also lasts approximately 0.1 s more, lowering the step frequency from an average of 0.95 Hz to 0.88 Hz.

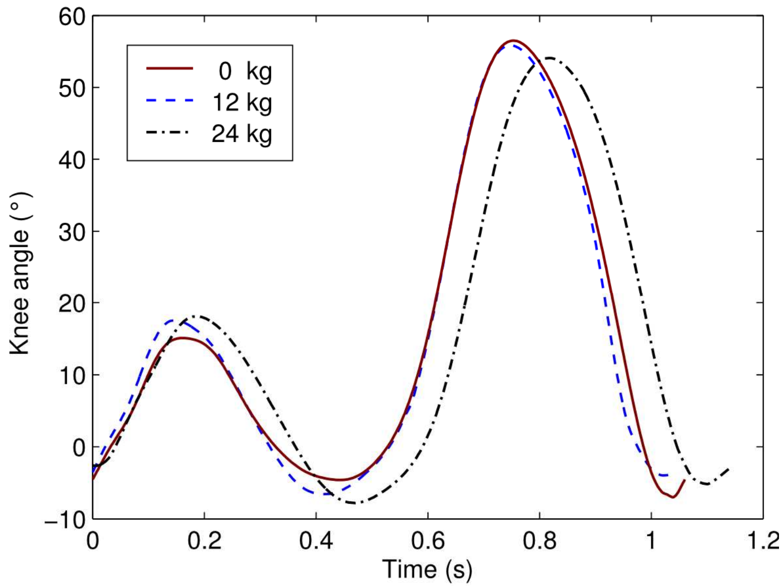

Figure 5. Mean angular displacement covered by the knee-joint of the subject when carrying a selection of backpack loads.

We now focus on time-domain results from bimorph B4 tested with the kinematic data for a 0 kg backpack load to present the typical behaviour of a bimorph in the harvester. In the first 0.5 s of motion (Figure 6) the speed is low and the bimorph experiences only a few well-spaced plucking actions, so that the ring-down following release can be clearly seen (although the 300 Hz resonance vibration of the bimorph cannot be discerned well at this scale). The peak centred at about 0.4 s is very wide, suggesting that the contact plectrum-bimorph was held for a long time; in fact in this case the plectrum didn't effectively pluck the bimorph, rather it deflected it and then re-accompanied to the rest position as the direction of rotation was reversed. There are two bursts of peaks in the second half of the cycle, where large angles are covered in a short time. Here the peaks are so close together that the bimorph vibrates continuously. More details on the different response of the bimorph at high plucking frequency vs. low plucking frequency can be found in [9].

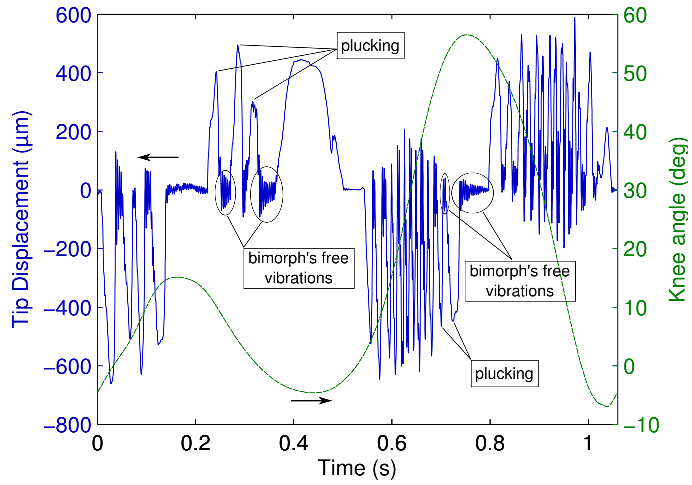

Figure 6. Displacement of the tip of bimorph B4 during the gait cycle with 0 kg load. The angle covered by the knee-joint is also plotted (dashed line, right ordinate axis)

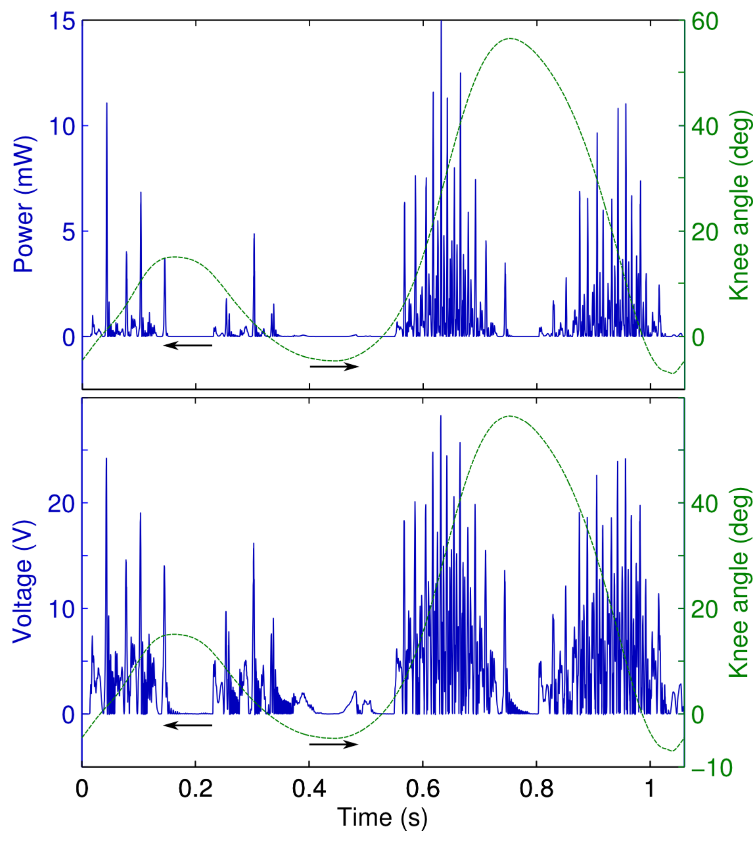

Figure 7. Voltage and instantaneous power detected across a 52.9 kΩ equivalent resistor connected to bimorph B4 as a function of time during the gait cycle with a backpack load of 0 kg. The angle covered by the knee-joint is also plotted (dashed line, right ordinate axes).

Similar features are seen also in the voltage and power signals (Figure 7): well spaced and not so high peaks in the first half of the gait cycle and two quick sequences of higher peaks in the second half of it. The wide displacement peak highlighted above (near 0.4 s) yields only very small voltage peaks at its beginning and end, and negligible power, because the deflection is very slow. This further clarifies that plucking can produce much more power than a quasi-static deflection of the bimorph: while the charges produced in open circuit conditions may be the same in the two cases, the corresponding current is higher if the deflection is faster. The response of the harvester in the second half of the gait cycle exemplifies the high power and voltages that can be achieved with plucking: in this area the voltage exceeds 20 V and the instantaneous power reaches 15 mW.

Energy generation during the gait cycle

The build-up of energy vs. time, *E(t)*, during the gait cycle is calculated from the instantaneous power, *P(t)*, by numerical integration via the trapezoidal method, approximating:

$$E(t) = \int_0^t P(t)dt$$

The resulting energy vs. time data are plotted in Figure 8 for all four bimorphs for one of the runs. The energy curves confirm that a small fraction (less than 20%) of the energy is produced in the first half cycle, with the second half contributing substantially more to the overall energy generation.

There are clearly some differences in the energy generated by the four bimorphs (Figure 8), which in the main can be ascribed to the variability in the plectra they encounter.

As the plectra were cut by hand, their tips do not have exactly the same shape and they do not protrude from the outer ring by the same distance. As a result, some plectra produce more modest final deflections of the bimorphs than others; more importantly, most of the plectra do not release the bimorphs sharply enough, leading to an "unclean" release. These issues are described in great detail in a previous paper [9], where it was also concluded, from statistical analyses, that the performance would increase by about 60-90% if all plectra were made like the seven best-performing plectra in the current harvester.

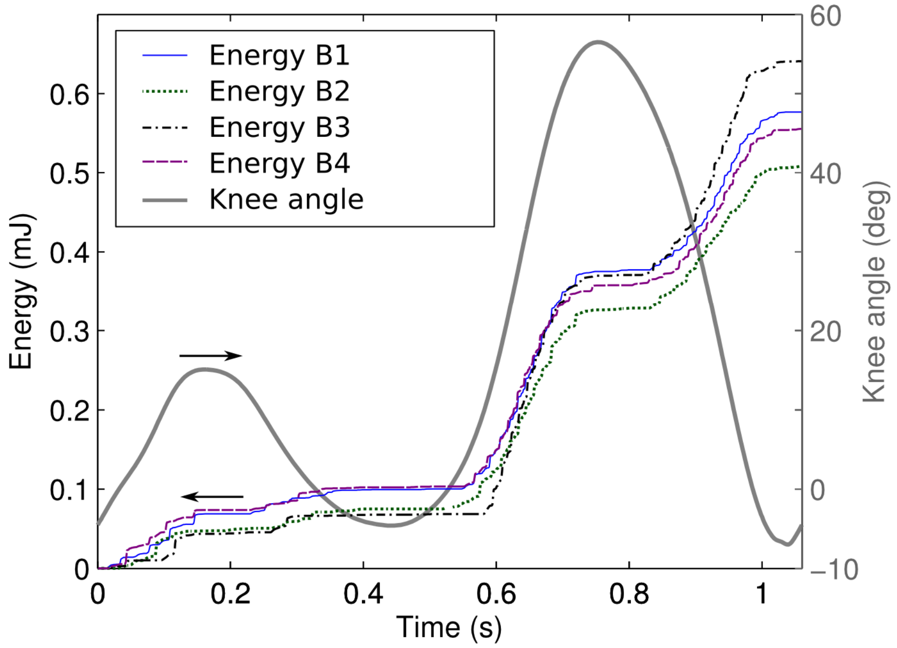

Figure 8. Build-up of total energy from each bimorph as a function of time during the gait cycle with a load of 0 kg. The angle covered by the knee-joint is also plotted (right ordinate axis).

End-of-gait energy for each bimorph

The final energy produced by the end of the gait cycle by each bimorph is plotted in Figure 9 for all the experimental runs discussed in this paper. First, three successive runs were performed for each of the three gait cycles, i.e. for the subject carrying 0, 12 and 24 kg. After these nine measurements, the direction of the motion was reversed, as if the device was mounted on the other leg or connections to inner hub-shank and outer ring-thigh were reversed, and a similar set of nine measurements performed. The four traces in the lower part of the plot, one for each of the individual bimorphs making the prototype, show that there is a significant difference in the energy produced by each bimorph, with bimorph B3 producing on average 30% more energy than bimorph B2 in the first nine runs, whereas both B1 and B4 generate an amount of energy almost equal (within 3%) to the average of B3 and B2 in the same experiments. Upon direction reversal, bimorph B4 produces 33% more energy than B3, whereas B1 and B2 produce the same energy (within 3%), and again close to the average of the other two (within 4%). The fact that the energy generation for the same bimorph can change drastically with direction, suggests that the cause of the variability is not the bimorph itself, rather the exact shape of the plectra and the angle with which they emerge from their mounting. In fact, most plectra are not normal to the outer ring, i.e. they are not exactly radially oriented. If this was correctly controlled, it would lead to higher energy production during the swing extension phase of gait and reduced resistance to joint rotation during the stance phase, as originally intended at the design stage. Coincidentally, the effects on the four bimorphs almost exactly balance themselves, so that the overall average energy production is a statistically-not-significant 0.7% higher in the first nine runs than in those following direction reversal. In other words, the energy generated by the harvester presented here is not affected by what leg it is mounted on or, alternatively, whether the hub is fixed to the shank or the thigh.

Total energy and average power

For each experimental run, the end-of-gait energies produced by the four bimorphs were added together to represent the total energy produced by the harvester in one walking step (upper trace with open circles in Figure 9):

$$E_{tot} = E_{B1} + E_{B2} + E_{B3} + E_{B4} \tag{1}$$

The horizontal line at 2.21 mJ represents the mean of the total energy in the 18 experiments.

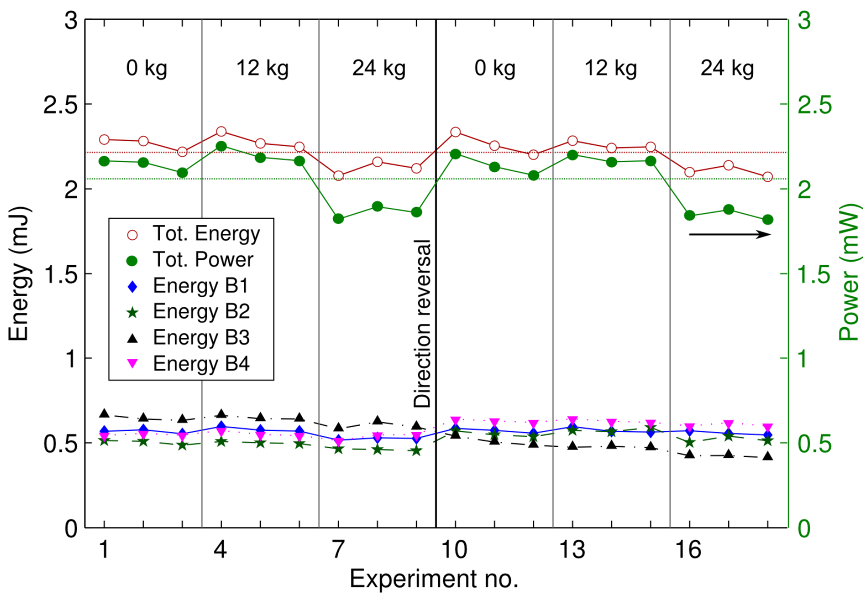

Figure 9. Consolidated data for end-of-gait energy production and power. The lower traces are energy for the individual bimorphs, according to the included legend. The upper traces are the total energy (open circles, Eq. 1), and average power (closed circles). The horizontal dotted lines indicate the means of energy and power. Vertical lines separate the different gait cycles, whose loads are indicated above. The thicker vertical line indicates the boundary between runs in one direction and runs in the opposite direction ("Direction reversal").

The total energy produced by the harvester (Figure 9) was statistically analysed (Table 1). The three experimental runs available for each combination of gait/direction were analysed together. Data in the last row were calculated from the the original dataset of 18 runs. The uncertainties were estimated as the 90% confidence level of the t-Student distribution. The total energy produced by the harvester is independent of the direction (Table 1). It is possible to notice a moderate dependence on the specific gait, with the 24 kg-load cycle yielding less energy than the others. As the total angle travelled is 166°, 168° and 170° for loads of 0, 12 and 24 kg, respectively, the gait with 24 kg load may actually trigger more plucking actions than the others, and could therefore produce more energy. The reason for the slightly lower energy production in the 24-kg gait is the lower average speed observed in this case (149°/s for 24 kg, 157°/s for 12 kg and 161°/s for 0 kg): it was shown earlier [9] that the energy generation of a plucked harvester increases as the plucking actions become quicker. At any rate, the difference is very close to the experimental uncertainty, and hence of moderate statistical significance.

The average power produced during each gait cycle was simply calculated as the ratio between average energy and corresponding gait duration. Its mean and uncertainty were then calculated within the three runs of equal conditions and within all the 18 runs (last two columns of Table 1), as described above for the energy. The seven mean values are in the region of 2 mW, with the overall mean at 2.06±0.3 mW. The data in the table show that the power is significantly affected, in a statistical sense, by the specific gait cycle. The power output when the subject carries 24 kg is 10% below the mean of all loads. This is a combined effect of the lower energy production just discussed and a longer time during which this energy is generated, as the gait is slower with the highest load. However, in a practical sense, the reduced power generation observed when the subject is carrying a very large load is likely to be of limited importance.

Table 1. Total output energy and power averaged over the gait cycle. A total of six conditions are considered, for the three gaits and the two directions. The three experiments performed at each condition are the samples for the statistical analyses. The uncertainties correspond to the 90% confidence level of the t-Student distribution.

Direction	Load (kg)	Gait duration (s)	Mean Energy (mJ)	Uncertainty (mJ)	Mean Power (mW)	Uncertainty (mW)
Plus	0	1.06	2.26	0.12	2.14	0.11
	12	1.04	2.28	0.14	2.20	0.13
	24	1.14	2.12	0.12	1.86	0.1
Minus	0	1.06	2.26	0.2	2.13	0.2
	12	1.04	2.26	0.07	2.17	0.07
	24	1.14	2.10	0.10	1.84	0.09
Overall	all	1.08	2.21	0.15	2.06	0.3

Conclusions and future work

Gait cycle data were collected with a marker-based motion capture system and processed to control a custom-made knee simulator designed to test a piezoelectric energy harvester based on the plucking technique of frequency up-conversion. The performance of the harvester, assessed using gait data collected with the human subject carrying three level of backpack loads, is in good agreement with previous predictions [9] and satisfies the design requirement of producing a few mW of continuous power during normal walking. At an average of 2.06±0.3 mW, the harvester's power is sufficient to supply a host of useful potential applications. For example, a wireless module sensing temperature, light intensity and acceleration, and transmitting the data via ZigBee, has been powered by the device described here, achieving a transmission rate of one data packet every 1.1 s.

The backpack load carried by the wearer has a measurable impact on the power generated; however, as the performance penalty is limited to 10% of output power, this is unlikely to have serious practical effects. It is possible to state that the harvester is not influenced by the direction of rotation, in the sense that the same performance is achieved if it is connected to the left or to the right knee or, alternatively, if the connections to thigh and shank are reversed.

Significant scope for increased performance may come from better selection of materials, different geometry of bimorphs and general optimizations, which have not yet been carried out on this harvester. Numerical optimisation techniques can be used to fine tune some design parameters so that the bimorphs are subject for a longer time to the highest plucking rates investigated here. This may be achieved with an improved distribution of plectra along the ring and/or an improved angular distance between the bimorphs. Finally, an industrially manufactured device would likely have a 60-90% power boost [9] from better construction quality and at least a further fourfold power increase due to the use of 16 or more bimorphs instead of the current four. Overall, an optimised plucked-piezoelectric harvester, with 16-20 bimorphs and good quality plectra could exceed 30 mW of output power after voltage rectification.

Field testing with a wearable prototype attached with braces would be useful to study the effects on gait patterns. In this case, portable monitoring systems based on accelerometers and gyroscopes should be used for a more natural and unconstrained experience. Whereas the prototype presented here is not suitable to be worn by a human subject, this is one of the objectives of our future research.

Acknowledgements

This research was sponsored by the Engineering and Physical Sciences Research Council (EPSRC), grant no. EP/H020764/1.

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