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Topical Review

Kinetic and thermal energy harvesters for implantable medical devices and biomedical autonomous sensors

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Abstract
Implantable medical devices usually require a battery to operate and this can represent a severe restriction. In most cases, the implantable medical devices must be surgically replaced because of the dead batteries; therefore, the longevity of the whole implantable medical device is determined by the battery lifespan. For this reason, researchers have been studying energy harvesting techniques from the human body in order to obtain batteryless implantable medical devices. The human body is a rich source of energy and this energy can be harvested from body heat, breathing, arm motion, leg motion or the motion of other body parts produced during walking or any other activity. In particular, the main human-body energy sources are kinetic energy and thermal energy. This paper reviews the state-of-art in kinetic and thermoelectric energy harvesters for powering implantable medical devices. Kinetic energy harvesters are based on electromagnetic, electrostatic and piezoelectric conversion. The different energy harvesters are analyzed highlighting their sizes, energy or power they produce and their relative applications. As they must be implanted, energy harvesting devices must be limited in size, typically about 1 cm^3. The available energy depends on human-body positions; therefore, some positions are more advantageous than others. For example, favorable positions for piezoelectric harvesters are hip, knee and ankle where forces are significant. The energy harvesters here reported produce a power between 6 nW and 7.2 mW; these values are comparable with the supply requirements of the most common implantable medical devices; this demonstrates that energy harvesting techniques is a valid solution to design batteryless implantable medical devices.

Keywords: energy harvesting, batteryless devices, micropower generator, implantable medical devices, self-powered systems, electromagnetic generators, electrostatic generators, piezoelectric generators, thermoelectric generator

(Some figures may appear in colour only in the online journal)

1. Introduction

In recent years, development in the bioengineering field has considerably increased the employment of implantable medical devices, such as pacemakers, defibrillators, drug pumps, cochlear implants, muscle stimulators and neurological stimulators. Several types of implantable medical devices perform real-time monitoring and/or control different human-body physiological conditions, some of them communicate wirelessly with an external unit avoiding the use of transcutaneous wires and the consequent risk of infections and opening the possibility to perform at-home monitoring [1].

The wide diffusion of these devices is shown in the data presented in [2]; about three million heart patients in the world have implanted pacemakers, and each year 600 000 new pacemakers are implanted. The implanted pacemakers or defibrillators aid in treating different heart diseases or conditions, such as bradycardia, ventricular and atrial tachyarrhythmia, or fibrillation [3]. Other examples of implantable medical devices, such as blood pressure, implanted microelectrodes to monitor cerebral or intra-muscular electromyographic (EMG) signals, are described in [4–7].

The power supply of an implantable medical device is usually provided by a battery that determines the longevity of the whole implantable system. In most cases, the implantable medical devices must be replaced because of dead batteries, even if the battery box of some heart pacemakers can be replaced separately under a local anesthetic without disturbing the implanted electrode. Otherwise, for example, the lifespan of an implanted defibrillator is about 10 years, but it must be replaced after about 4.7 years [8]. Therefore, the patient must undergo a surgical operation; this procedure causes physical and psychic pain in the patient and, furthermore, it is an economic burden for the patient itself and for the national health system. In the literature, research activities are in progress for powering pacemakers in an alternative way, such as power harvesting systems. This is just one example of where research is focusing.

In general power source techniques alternative to batteries are: telemetric and energy harvesting techniques. In the first case, a coil inside the implant harvests energy from a magnetic field, externally generated by another coil placed outside the body, operating at a frequency in the 1–10 MHz range in agreement with the low tissue absorption [9]. This technique can be uncomfortable for the patient due to presence of the external device that has to be close to the implant. Otherwise, energy harvesting techniques convert the human-body energy sources into electrical energy in order to power implantable devices without an external power source; hence, with respect to telemetry, an energy harvesting technique is more comfortable for the patient. This paper focuses on the second technique; in particular implantable devices supplied by means of kinetic or thermal principles have been reviewed.

Energy harvesting from the human body is typically more difficult than energy harvesting in the industrial environment [10]. First of all, since it must be implanted in the human body, the harvester must have limited volume (typically less than 1 cm^3). Furthermore, for kinematic harvesters, the matching between the resonant frequency of the generator and the human-body frequency, in order to maximize the power efficiency, is obstructed by the size requirements; in fact, the size of a resonant generator is typically inversely proportional to its resonant frequency. For thermoelectric harvesters, the thermal gradients in the human body are low making it difficult to achieve significant power values.

This paper surveys energy harvesting systems for powering implantable medical devices. Section 2 describes the human-body energy sources. Section 3 presents and discusses the state-of-the-art of implantable energy harvesters.

Table 1. Energy consumed by the human body for different daily activities [11].

Activity	kcal/h	watts
Sleeping	70	81
Sitting	100	116
Strolling	140	163
Hiking, 4 mph	350	407
Swimming	500	582
Mountain climbing	600	698
Long-distance running	900	1048
Sprinting	1400	1630

Table 2. Calculated work in different joints during walking with a speed of 1 m s^{-1}, a leg length of 0.98 m, and one range motion of 30° [12].

Joint/Motion	Work (J/step)
Ankle	34.9
Knee	24.7
Hip	19.6
Elbow	1.07
Shoulder	1.1

Section 4 summarizes the main features of the analyzed energy harvesters through tables. Finally, section 5 concludes the paper.

2. Human-body energy sources

The energies consumed for different daily activities, such as sleeping, eating and walking, are reported in table 1 [11]. The evaluated available power is 1 W from breath, 0.93 W from blood pressure, 35 W from upper limb motion, and 67 W from walking at two steps per second. Table 2 [12] shows the amount of kinetic energy used, while table 3 [13] shows the thermal gradient for typical daily activities. Table 4 reports average values of power required to supply some implantable medical devices. Comparing the four tables, it is evident that typical daily activities consume relatively large power so that the extraction of a very small amount for powering an implanted medical device, even in the presence of low efficiency, is a true possibility.

The most common energy harvesters from human-body motion exploiting kinetic energy are electromagnetic, electrostatic or piezoelectric generators. In particular, a generic kinetic harvester is composed of a resonance mechanism [14, 15] that generates the swing of a proof mass and is designed according to the input vibration frequency in order to operate in resonance condition. This is generally due to the fact that the harvester generally has low efficiency; therefore, to increase the value of the electrical power produced, the harvester is led to work near or in correspondence to the mechanical resonance condition. Typically, the mechanical resonant frequency depends on the system mechanical features and is equal to

$$f_r = \frac{1}{2\pi}\sqrt{\frac{k}{m}}, \tag{1}$$

Table 3. Human temperature gradients at ambient temperature of 25 °C [13].

Site	Muscle thickness (mm)	Fat thickness (mm)	Resting ($v = 0.2$ m s^{-1}) ΔT(K)	Walking ($v = 1.56$ m s^{-1}) ΔT(K)	Running ($v = 4.25$ m s^{-1}) ΔT(K)
Abdomen	16.34	14.8	1.73	3.8	4.75
Biceps	34.60	3.33	0.45	1.22	1.70
Calf-posterior	65.36	4.93	0.65	1.74	2.40
Chest	33.45	7.26	0.94	2.37	3.18
Forearm	26.04	3.24	0.44	1.16	1.63
Hamstring	69.29	6.97	0.91	2.32	3.14
Lumbar	37.00	6.54	0.85	2.18	2.96
Quadriceps	54.54	6.42	0.82	2.12	2.89
Subscapular	23.74	8.40	1.06	2.60	3.44
Suprapatellar	29.42	6.23	0.81	2.08	2.81
Triceps	41.84	5.92	0.78	2.02	2.75

Table 4. Power consumptions for different implantable medical devices.

Device	Power consumption
Cochlear	145 μW [21]
	5.16 mW [22]
Drug pump	400 μW [23]
Neurostimulator	50 μW [24]
Muscle stimulator	1.3 mW [25]
Pacemaker	8 μW [26]

where k is the stiffness of the spring and m is the seismic mass. The human-body motion frequency typically belongs to the range 0.5–10 Hz, for example measured frequencies for knee, ankle and hip during normal walking are 1.8, 1.4 and 1.8 Hz, respectively [16]. Kinetic energy harvesters, working as described by the Williams–Yates model, have generally higher conversion efficiency at the mechanical resonance frequency. Therefore, to obtain a mechanical resonant structure that uses the low frequencies of the human-body motion as excitation source, generally requires large dimensions (not compatible with human-body applications) and/or materials with a low Young modulus (difficult to obtain). In other words, lowering the resonant frequency generally increases the generator size, making the system too invasive for human-body applications. Different approaches are reported in the literature: for examples, in [17], a material with nonlinear elastic characteristics or MEMS frequency up-converter operating at very low frequency (1–100 Hz) [18]. An energy harvester based on this technique is composed of two vibrating systems: a low-frequency vibrating element excites a resonant element through mechanical impact or magnet interaction; the second element vibrates at higher frequency than the input vibration frequency. In this way, it is possible to operate in resonance condition while maintaining limited size.

Thermal energy is another exploitable energy reservoir. Thanks both to the non-uniform temperature distribution and to the Seebeck effect, the thermal energy produced by the thermoelectric generator (TEG), composed by several thermocouples, is proportional to the thermal gradient applied to the TEG itself. The highest temperature gradient occurs near the skin surface of the human body [19]; therefore, the TEG should be implanted as close as possible to the skin surface, where the temperature gradient between the two junctions of the TEG is maximum. This position guarantees good output voltage and consequently good output power. In order to compare the different implementation areas, table 3 shows the temperature gradients calculated in different parts of the human body, with a room temperature of 25 °C, supposing a simulated physical structure composed of muscle and fat for different human activities.

This paper is focused on different energy harvesters according to the transduction principle, output power, operating frequency and size. An important issue, not specifically treated, of an energy harvester is its biocompatibility: materials used for the manufacture of an energy harvester should be carefully and appropriately chosen [20]; otherwise its different parts must be housed in a biocompatible packaging.

3. Overview of energy harvesters for implantable devices

3.1. Kinetic energy harvesters using electromagnetic conversion

An electromagnetic harvester is typically composed of a permanent magnet and a coil; according to the Faraday–Neumann–Lenz law, relative motion between the coil and the permanent magnet produces a time-variable magnetic flux and consequently generates a voltage. A generic electromagnetic harvester is composed of a resonance mechanism, a permanent magnet and a coil. Typically, the electromagnetic harvester operates in resonance conditions in order to maximize the amplitude of swing and the magnetic flux variation. Mathematical expressions relating to the design of the electromagnetic harvesters can be found extensively in the literature, for example in [27]; the general expressions for the calculation of the generated power as a function of different parameters such as the geometrical dimensions are shown. In the literature, some electromagnetic harvesters have been designed to harvest energy from joint movements, such as knee and hip, or from internal human body parts, such as heart and diaphragm muscle. Here a series of devices of increasing size are reported and analyzed. The first are for specific locations in the human body: heart, diaphragm muscle, hip and knee. The remaining presented electromagnetic harvesters have not been designed for a particular part of the human body, but they

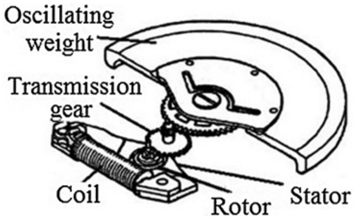


Figure 1. Schematic of the pacemaker system patented by Tran *et al* [28].

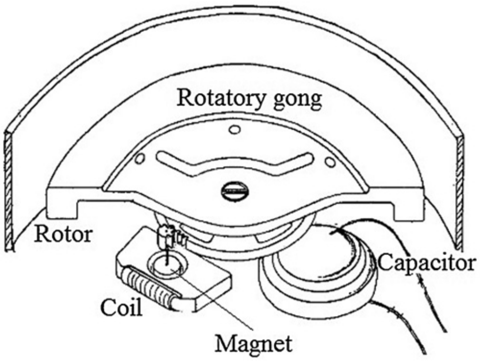


Figure 2. Structure of the pacemaker power source described in [29]. The generator is composed of an energy harvester also used in a quartz watch commercialized by Seiko.

are generically suitable for implantable medical devices, for instance thanks to their low resonance frequency.

A heart-powered pacemaker has been patented by Tran *et al* in [28]. In the designed device, contraction of the heart muscle produces relative motion between a magnet, placed on a rotor, and a fixed coil (figure 1). According to the authors, movements produced by the heart have a frequency between 0.5 and 2 Hz and the system could generate a power between 40 and 200 μW, values adequate to operate the pacemaker.

Another power generating mechanism that has been tested also *in vivo*, as a possible pacemaker power source, is proposed by Goto *et al* in [29]. The generator is also used in a quartz watch commercialized by Seiko. As shown in figure 2, it is composed of a rotator gong (the proof mass), a magnet (rotor), a coil (stator), a half-wave rectifier and a capacitor (0.33 F). The heart contraction moves the rotatory gong and consequently the magnet producing a voltage on the fixed coil whose energy is stored in the capacitor. *In vitro*, the capacitor stored energy of 0.66 J after 30 min; when the system was excited with a sinusoidal input with amplitude of 40 mm at 2 Hz. *In vivo* tests were carried out by placing the generator system on the wall of the right ventricle of a dog heart. During 30 min at about 200 bpm, 80 mJ were stored in the capacitor (13 μJ per heart beat).

A linear permanent electromagnetic generator for diaphragm muscle movement is described and analyzed by Nasiri *et al* in [30]. As reported in figure 3, this device is composed of two layers of permanent magnets and a layer of coil fixed on a spring; its volume is about 16 cm^3, much greater than previously reported and can be hosted, as suggested by the authors, in the abdominal wall. The device was tested in the laboratory by using an audio speaker at a resonance frequency

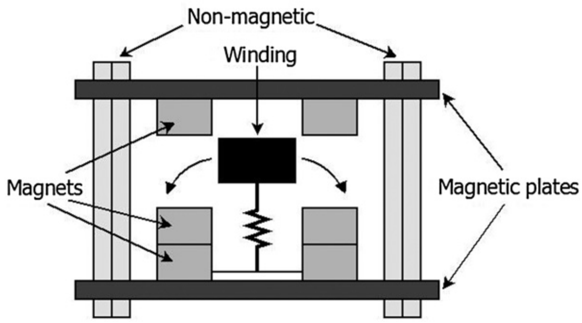


Figure 3. Schematic of a permanent electromagnetic generator to be implanted in the diaphragm muscle [30].

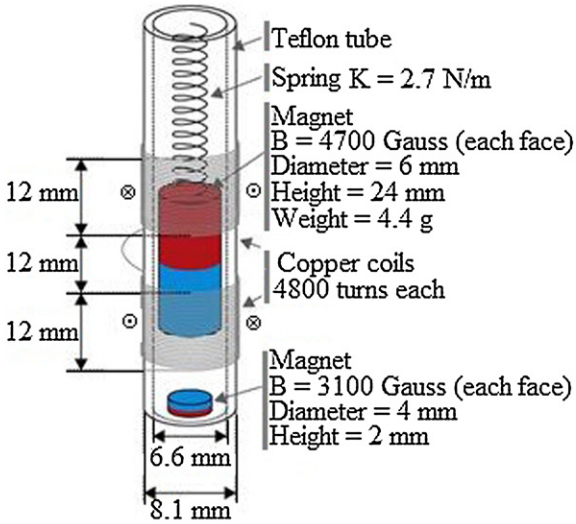


Figure 4. 3D representation of the hip electromagnetic harvester reported in [31].

of about 0.3 Hz to simulate the abdominal body motion; in this condition, the generator produced a power of about 1.1 mW. Afterwards, Nasiri explains that the device is also able to produce energy from the human gait. After 22.5 s of walking, the generator, fixed to an individual's waistline, was able to harvest 9 mJ. However, the end use is not indicated. The authors have not specified the electronic circuits and sensors that will be powered using the power harvesting device.

A nonlinear electromagnetic generator implantable in a hip prosthesis is described by Morais *et al* in [31]. This device is able to harvest energy from the human gait by using a velocity-damped resonant generator to power the electronics embedded in the hip prosthesis. As shown in figure 4, the main structure of the prototype is made up of one or two external coils and a Teflon tube in which a neodymium magnet is attached to a spring. At the bottom of the tube there is a magnetic brake in order to avoid collisions of the magnet against the bottom of the Teflon tube itself. The generator stores a total energy of 1912.5 μJ in the two capacitors under normal walking conditions. For a subject moving at a rate of 1.3 Hz and a 100 kΩ load, the generator was not able to supply the embedded electronics. The target of this work is to energize a telemetric system inserted in a smart hip prosthesis implant for early detection of loosening. The device is connected to an

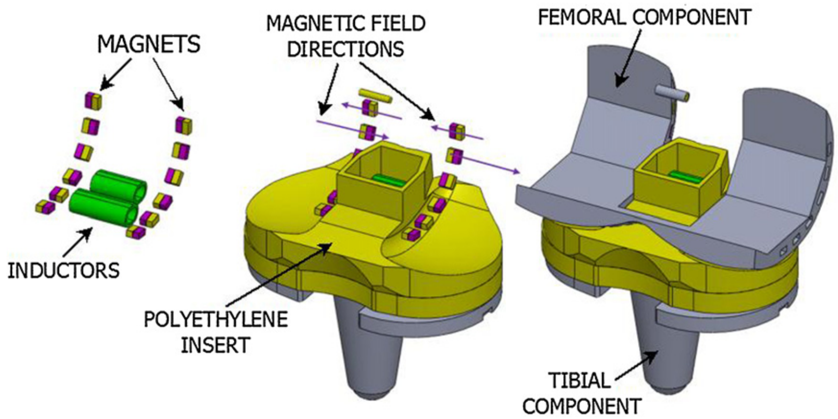


Figure 5. Structure of the generator for knee prosthesis [32].

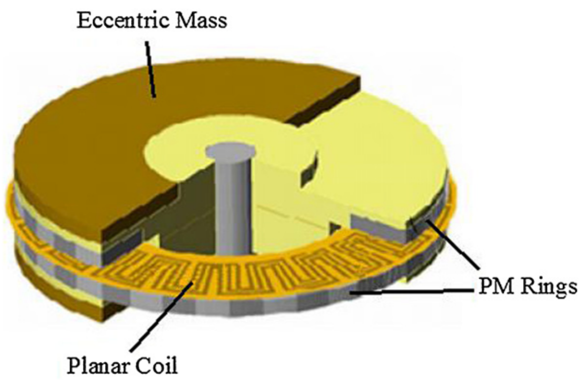


Figure 6. 3D schematic of the rotational EM generator proposed by Romero *et al* from [16].

energy management circuit that for a walking speed of 1.3 Hz is able to turn on the measuring circuit for 9.2 s after a charging phase of 34.8 s.

A non-resonant electromagnetic generator is proposed by Luciano *et al* in [32]. It is an electromechanical generator implantable in human total knee prosthesis in order to convert the mechanical energy of the knee movement into electrical energy directly. As shown in figure 5, the energy harvester is composed of six prismatic magnets placed in each condyle of the knee prosthesis and two copper coils disposed between the condyles, in a prominence of the tibia plate. The relative movement between tibia and femur produces a relative movement between coils and magnets and hence a voltage on the coils is generated. The proposed system was tested in laboratory by using a motor to simulate the human gait and an open circuit output voltage of about 1.6 V peak–peak was obtained. Luciano proposes a conditioning circuit composed of a voltage multiplier and a pump charge. With this conditioning circuit and a gait frequency of 1.02 Hz, the whole system produced output energy of about 22.1 μJ able to turn on the measuring circuit for 13 ms after a charging phase of 7 s.

An EM generator based on a rotational mechanism is presented in [16] by Romero *et al.* The rotor is made up of two rings with multiple pole-pairs of permanent magnets (PM) and an eccentric mass; the stator is composed of a few stacked layers of planar coils (figure 6). During human gait, body motion causes the rotation of the eccentric mass and therefore, an ac output voltage is produced on the coils. The authors consider different possible locations from which energy can be harvested during human walking. *In vivo* tests were carried out by placing the fabricated device externally on the limbs of a subject. Ankle and knee gave higher values of harvested power. With two layers of coils, the device generated an average power of about 3.9 and 3 μW by placing the device itself on the ankle and the knee, respectively. Romero estimates that the generator can have a long lifespan and it could be suitable for implantable devices, however the end use is not indicated.

3.2. Kinetic energy harvesters using electrostatic conversion

An electrostatic generator is based on a variable capacitor, whose plates are moved by human-body motion. Electret material produces an initial charge distribution and when the capacitor is moved by the external human-body motion, a charge movement is generated. The voltage variation between the capacitor plates is described by the following equation:

$$\Delta V = \sqrt{2F\frac{\Delta z}{\Delta C}}, \tag{2}$$

where ΔV is the voltage variation between the capacitor plates, F is the electrostatic force, z is the distance between the plates, and C in the capacitance value.

Electrostatic power harvesting systems can be built by MEMS technology and produce energy whose values are, in some cases, independent of movement frequency. These two characteristics make electrostatic techniques suitable candidates for applications inside the human body, with no great problem of space, but the value of energy produced is very low, being at least one order of magnitude less than the electromagnetic harvesters.

In [33], Tashiro *et al* propose an electrostatic generator for a cardiac pacemaker which exploits ventricular motion. The device was created with a honeycomb structure, by using aluminum-evaporated polyester films (figure 7). The variable capacitor was inserted into a resonant structure, whose resonant frequency is 6 Hz, comprising a mass suspended on springs. As the prototype was too large to put it on the left ventricle, the device was tested by employing a vibration mode simulator, driven by the signals produced by a 3-axis accelerometer module placed on the left ventricle of a dog.

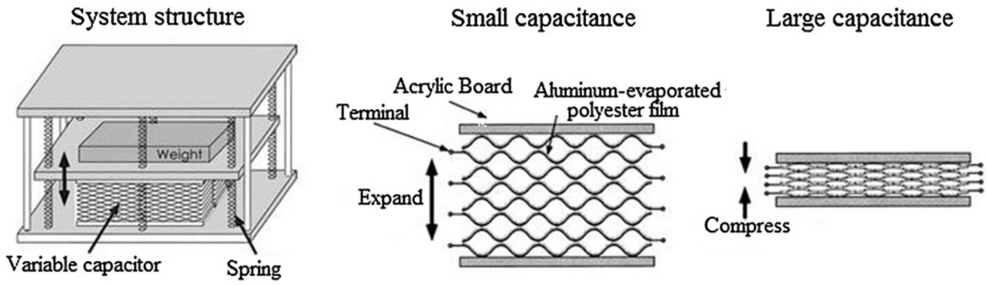


Figure 7. Honeycomb structure for an electrostatic harvester, from [33]. The resonant structure vibration expands and compresses the variable capacitor, corresponding to small and large capacitance, respectively.

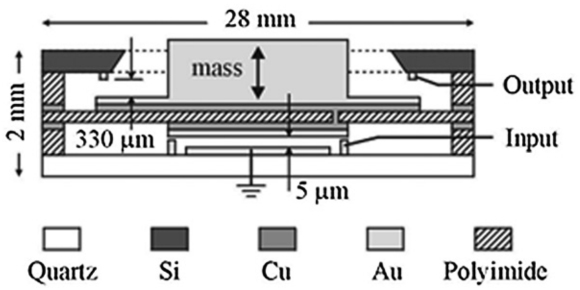


Figure 8. Cross section of the electrostatic harvester described in [34].

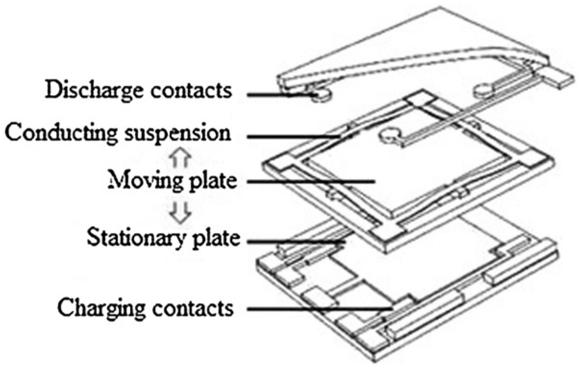


Figure 9. Exploded view of a MEMS electrostatic generator, from [35].

The authors report a mean power of about 36 μW produced by the VCES with 180 bpm. According to Tashiro, this value is enough to supply the cardiac pacemaker.

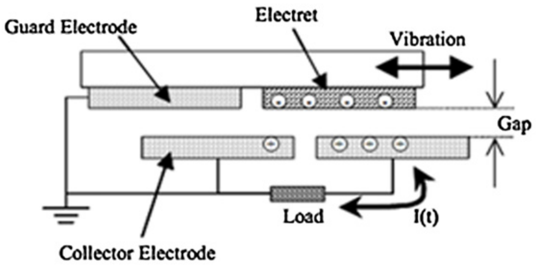
In [34, 35], Miao *et al* propose a MEMS electrostatic power generator. Figure 8 shows the device structure; it is made up of a gold proof mass, placed on a flexible polyimide membrane and suspended between a silicon top plate and a quartz base plate. Experimental results demonstrate the power produced is 24 μW at a vibration frequency of 10 Hz, frequency quite high for human-body motion.

To maximize the movable plate displacement, the two previous systems are composed of a resonant structure even if the resonant frequencies are not the optimized value for the human body. Two other papers, reported in the following, propose a non-resonant electrostatic microgenerator. The device in [35] was obtained by MEMS fabrication techniques and, as shown in figure 9, it consists of a variable capacitor (5.5 pF up to 150 pF) having one moving plate on which a silicon proof mass is attached. The device was tested on a low frequency shaker platform, for frequencies in the range 10–100 Hz. Miao report a generated power of 120 nJ/cycle. This is a significant value in comparison with other MEMS electrostatic generators [36], but the power obtained remains significantly below theoretically achievable values [37]. The authors attribute this variance to the viscous air damping and on the unwanted degrees of freedom of the proof mass. In any event, with an acceleration of about 1 g, the device should theoretically generate about 80 μW or 2.6 μJ/cycle for operation at 30 Hz.

In [38], Naruse *et al* present an electrostatic power generator, whose particular structure allows to match the operating frequency range with the human-body motion frequency. The electrostatic generator is based on the employment of micro ball bearings, and a particular electret structure. The micro ball bearing permits a long-range movement at low frequency although a spring with a low spring constant is used (figure 10). An electret is a dielectric material with a quasi-permanent electrical charge and it is the electrostatic equivalent of the permanent magnet. The authors adopted the employment of the new electret structure with high surface potential and narrow electrode width; the proposed electrostatic micro power generator can increase the number of times that the power generation (electrostatic induction) is done with high surface potential. The electrostatic power generator was tested with a shaker and experimental results show that the device has a resonance frequency of about 6 Hz. The authors state that the proposed device generates an output power of 40 μW with a mechanical input of 2 Hz and 0.4 G.

3.3. Kinetic energy harvesters using piezoelectric conversion

Piezoelectricity is a property that allows the generation of a potential difference when the material is subjected to a mechanical deformation (direct piezoelectric effect). The piezoelectric transducers are kinetic harvesters that exploit the piezoelectric effect to collect energy from human motions; for example, voltage is produced by the forces applied on a human joint or muscle twitches. The coupled distributed parameter solution of a piezoelectric energy harvester under base excitation was given for unimorph [39] and bimorph [40] cantilevers (i.e., for cantilevers with one or two piezoceramic layers). As reported in [27], for the unimorph cantilever configuration, the coupled beam equation can be expressed based on the Euler–Bernoulli beam theory. The piezoelectric



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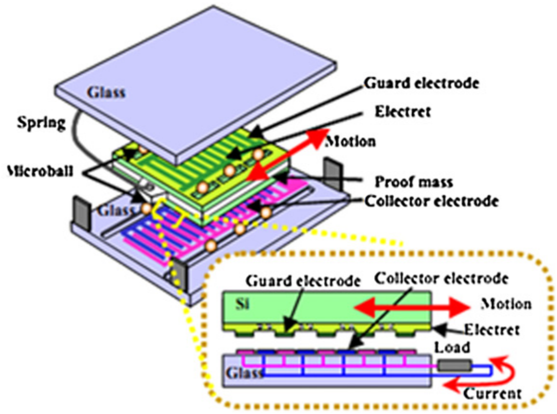

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Figure 10. Schematic of electrostatic induction power generation using electret principle (*a*) and 3D structure of an electrostatic harvester supported on micro ball bearings (*b*), from [38].

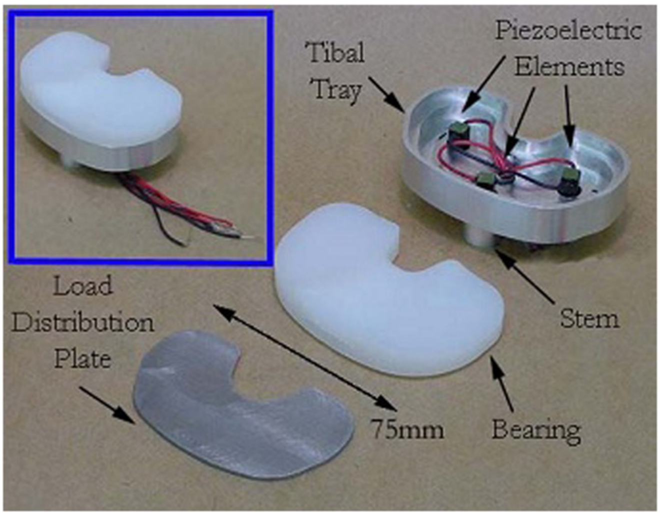

Figure 11. Components of the implant model proposed in [41]. Three piezoelectric elements are placed into the tibial tray.

energy harvester is capable of producing relatively high output voltages but only at low currents and the output impedance of piezoelectric generators is typically very high ($>$100 kΩ) showing a capacitive characteristic. For these reasons, these devices must be used with an appropriate energy management circuit that generally consists of a circuit for matching the capacitive impedance with up-conversion, an ac–dc rectifier, a storage supercapacitor, a control unit and a dc–dc converter.

In this section, different ways to collect electrical energy through piezoelectric elements are reviewed. The first reported regards the use of piezoelectric harvesters embedded in the lower limbs. These devices are composed of 'hard' materials, such as quartz or PZT, which is specific for application with high force but small displacement, which is the situation for the lower limbs.

Platt *et al* in [41, 42] use piezoelectric ceramics implantable in orthopedic prosthesis in order to exploit the human knee activity. Figure 11 shows a tibial tray instrumented with three commercial piezoelectric elements. These rectangular elements, whose dimensions are 10 mm $\times$ 10 mm $\times$ 20 mm, are composed of multiple wafers of lead zirconate titanate (PZT) ceramic. The authors use a self-powered total knee replacement (TKR) implant, and the movements of the knee are simulated by a single-axis Mini-Bionix MTS 858 test machine. The forces are applied by the femoral component to the bearing surface according to the International Standard ISO 14243-1: 2002 and ISO 14243-3: 2004 [43, 44]. Each output of the three piezoelectric elements was rectified and the unipolar output was stored in a 10 μF capacitor. The whole implant generated 4.8 mW of raw electrical power. The overall electrical efficiency was about 19%.

In [45], Chen *et al* draw on the work done by Platt: the authors used four piezoceramic elements, embedded into the implant, whose output power (about 1.2 mW) is adequate to supply the circuits of the knee implant to monitor the working condition of the TKR and to transmit the data to medical staff by RF signal.

Drawing inspiration from Platt's work, Almouahed *et al* [46–48] also designed and implemented a monitoring system for knee motion using piezoceramic elements both as electrical energy harvesters and as sensors. With respect to Platt, there are four piezoelectric elements between the femoral component and a custom-designed tibial component and also in this case the generated electric power supply circuits that implement several functions: acquisition, processing and transmission. SCMAP09 piezoceramic elements (Noliac, Inc.) with dimensions of 10 mm $\times$ 10 mm $\times$ 4 mm were used. Applying a load to each piezoceramic, the available power was estimated to be at most 1.8 mW for a single piezoceramic, so the total generated power was about 7.2 mW. With this power value, in [49] Lahuec *et al* show that a telemetry system, designed for a 0.35 μm CMOS device, is able to transmit its measurements outside the knee with a data transmit unit consuming 0.2 mW.

Lewandowski *et al* present in [50] a totally implantable piezoelectric solution that is able to employ power from an activated muscle. The authors carried out some experimental tests by applying several forces to a non-optimized prototype system with a material testing system machine (MTS Systems

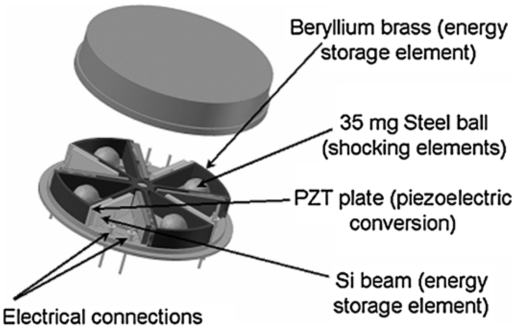

Figure 12. Schematic of the shock-based energy harvester proposed in [51]. The steel balls collide with the PZT plates and, through the piezoelectric effect, electrical energy is generated.

Corporation, Eden Prairie, MN). A PZT piezoelectric stack generator (T18-H5-104, Piezo System) with dimensions of 5 mm × 5 mm × 18 mm was used in the experimental trials. Lewandowski estimated that a stimulator needs an average power of approximately 40 μW to operate, which is possible to obtain with a force greater than 150 N; this requirement is satisfied by the gastrocnemius muscle.

In [51] Cavallier *et al* propose a device based on a vibrating structure for energy harvesting. The authors realized two basic structures both inserted in a HC45 package; the first one is made with a PZT plate, while the second one is a stack realized with a PZT layer, a silicon beam and another PZT layer. As shown in figure 12, the structure has an external diameter of 14 mm and a height of 2 mm and a 35 mg steel ball used to shock the PZT elements. According to the authors, this device is able to recover energy from human movements thanks to its size and its operating frequency. The experimental tests were made by exciting the whole structure with a shaker at 1.4 g acceleration and at a frequency of 6 Hz. Each element produces an output power of 62 nW, therefore the whole structure, with eight internal elements, generates about 0.5 μW of power.

With respect to the previously analyzed devices, which require high forces, now is presented an energy harvester that exploits flexible piezoelectric strips producing energy with weak forces.

Potkay *et al* in [52] propose an arterial cuff for blood pressure monitoring (figure 13). The cuff is realized with a silicone elastomer in which a thin piezoelectric film in polyvinylidene fluoride (PVDF) is embedded. Its function is to convert the expansion/contraction of the artery into electrical energy. The cuff was rolled around a latex tubing used like an artery. A peak voltage of 1.2 V, a maximum instantaneous power of 16 nW and an average power of 6 nW were measured. According to the authors, the microfabricated version of this implantable device for *in vivo* tests would generate more than 1.0 μW.

3.4. Thermoelectric energy harvesters

Energy scavengers exploit the thermoelectrical effect to collect electrical energy thanks to the temperature difference in the human body. When a temperature difference ΔT is applied

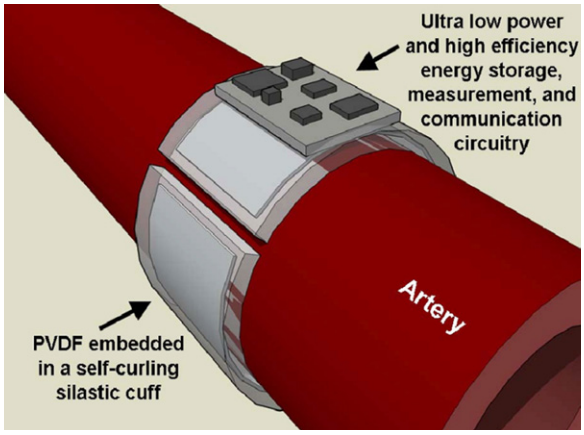

Figure 13. Schematic of an implanted arterial cuff power source integrated into a self-powered blood pressure sensing system, from [52].

between the TEG faces, an open-circuit output voltage V_G is generated according to the following:

$$V_G = N\alpha\Delta T, \tag{3}$$

where α is the Seebeck coefficient of the TEG materials, N is the number of thermocouples and ΔT is the temperature difference applied. The output power P_L delivered by the TEG to the load R_L is given by the following relationship:

$$P_L = N^2\alpha^2\Delta T^2 \frac{R_L}{(R_L + R_{\text{in}})^2}, \tag{4}$$

where R_{in} is the internal electrical resistance of the TEG and R_L is the load resistance.

The voltage values are usually too low and unregulated for directly powering electronic circuits, although the available current has values up to 100 mA. In order to power autonomous systems, the converted energy must be properly handled, and in particular, the thermopile output voltage must be increased and regulated.

The thermoelectric harvesters could be used in the neurostimulator systems thanks to their small dimension and coherent energy value. Furthermore, the position of the thermoelectric energy harvesters, usually under the skin, allows easy connections with neurostimulator systems with respect to other devices.

In the literature, several examples of thermo-power generators are reported. In [53] Stark *et al* present a device (Thermo Life) that is able to convert thermo energy into electrical energy. It is a small and compact energy source. Thermo Life is composed of over 5000 thermocouples in series and made of a highly efficient thermoelectric material (Bi_2Te_3) deposed on thin Kapton films. The working principle is based on a plate in contact with a warm source and another one with a cold source; when a heat flux flows through the thermopile into the device, a voltage is generated. The latest developed prototype achieves a performance of 120 μW at 3 V with a temperature difference of about 5 K. This is a nominal value, but it is also possible to use the harvester with lower gradients, obviously decreasing the output power. According to the authors, the produced power of Thermo Life

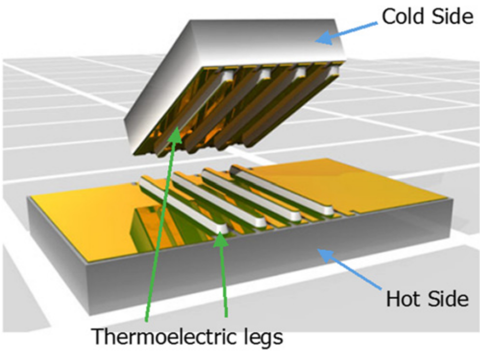

Figure 14. Micropelt thermoelectric microgenerator structure, from [57].

Table 5. Experimental data produced with TE modules at RTI with temperature differentials of 1 to 3 °C [56].

TEG fabrication	Number of couples	ΔT (K)	V_{oc} (mV)	P_{max} (μW)	Area (cm^2)
Pick and place	$4 \times 4 = 16$	0.8	5.5	140	0.09
Wafer-scale	$3 \times 10 = 30$	2.7	41.8	980	0.16

can vary from a few 10 to 100 μW according to the temperature difference.

Watkins *et al* [54, 55] describe a low-grade-heat energy harvester, consisting of an advanced thin superlattice thermoelectric film. Experimental data produced with two TEG modules are shown in table 5 [56]: ΔT is the temperature difference between the two surfaces of the devices, V_{OC} is open circuit voltage, and P_{max} is the generated output power. These tests are realized *in vitro* but the authors aim to package and test the modules in a simulated bio-environment.

An energy harvesting device based on the thermoelectric effect, called Micropelt, is reported by Böttner *et al* [57–60]. Micropelt technology with thin film thermoelectric layers in common vertical architecture on silicon substrates have an area of less than 1 mm^2 and a total thickness of less than 500 μm. The Micropelt technology is based on two wafer technologies separately for n- and p-type material. In figure 14, a schematic drawing of a Micropelt device is shown, the two wafer concept with the thermoelectric legs on the individual and n- and p-type parts before the soldering process is represented. Power factors of 30 μW cm^{-1} K^{-2} for p-material and of 40 μW cm^{-1} K^{-2} for n-material have been achieved. For example, the type MPG-D901, one of the Micropelt thermoelectric microgenerators, is a device with 1800 leg pairs integrated on an area of 5 mm^2 and produces 0.6 mW (load matched) at 2.5 K temperature difference. Micropelt devices for their dimensions and possible temperature difference ΔT could be used in medical implants inside the human body.

Finally, Yang *et al* [19] present an evaluation of TEG varying physiological or environmental thermal conditions. The authors performed *in vivo* and *in vitro* tests on a commercial TEG (TEC1-01706T125, Beijing Xinyu Kaimeng Electronic Technology, 15 mm × 15 mm × 3.9 mm). *In vitro* tests were made by using one piece of pork meat. The TEG was embedded just below the thickness of the skin layer and the fat layer (figure 15). A plate was positioned under the tissue layer and connected with a thermal water bath at a temperature of 37 °C in order to simulate the thermal human body; the skin surface was exposed to the room environment, 18 °C. Before starting the cooling of the skin, the thermal gradient was stabilized at 0.5 K and the output voltage was about 3.3 mV. After having cooled the surface through ice, the thermal gradient was nearly 1.1 K and the output voltage was about 6 mV.

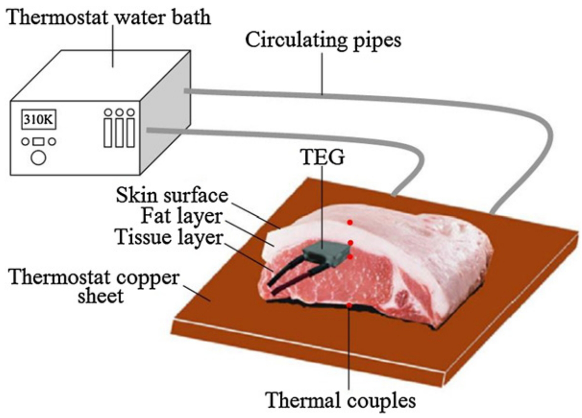

Figure 15. Setup for *in vitro* experiment on a TEG, from [19].

In vivo experiments were performed with a rabbit weighing 2 kg. One piece of TEG was implanted into a narrow opening in the abdomen. After 260 s, the temperature difference stabilized at around 1.3 K and the output voltage was about 5 mV. After the rabbit skin surface was covered with an ice water bag, the temperature difference and the output voltage increased up to 5.5 K and 25 mV, respectively.

4. Comparison and discussion of energy harvesters

Tables 6–9 summarize the main features of the electromagnetic, electrostatic, piezoelectric and thermoelectric harvesters, respectively. For each harvester, the relative table shows bibliography references, material characteristics for biocompatibility, suggested location of the application, source properties, harvester sizes, energy and/or power outputs and eventually additional notes. As can be seen, the harvesters are very heterogeneous; if one observes the column of publication years, there is a large time interval which denotes constant interest of research into the study of implantable devices. Furthermore, the same observations can be made for harvester sizes and energy and/or power outputs. The reported energy harvesters produce a power between 6 nW and 7.2 mW. In particular, piezoelectric harvesters produce a power between 6 nW and 7.2 mW, electromagnetic harvesters generate a power between 3 μW and 1.1 mW, thermoelectric harvesters produce a power between 24 and 150 μW, and electrostatic harvesters scavenge a power between 24 and 85 μW.

The piezoelectric harvesters seem the most promising since the value of power produced is the greatest. In particular, the harvesters [46–48] generate a power greater than 1 mW. These devices, anyway, need a significant force since they are suggested for the knee where the tibio-femoral forces

Table 6. Comparison of electromagnetic harvesters.

Article		Material characteristics	Source		Harvester	Output energy	
Author	Year/Ref	for biocompatibility	Location	Test condition	volume	and power	Notes
Goto	1998/[29]	Embedded into a biocompatible system	Heart	Laboratory sinusoidal input 40 mm, 2 Hz	~12 cm^3	0.66 J/30 min 91.7 μJ/each stroke and 367 μW	Same structure as the wristwatch generator—encapsulated in polyvinyl case
Goto	1998/[29]	Embedded into a biocompatible system	Heart	*In vivo* dog heart rate (200 bpm MAX)	~12 cm^3	8 mJ/30 min 1.3 μJ/heart beat and 4.4 μW	Same structure as the wristwatch generator—encapsulated in polyvinyl case
Nasiri	2011/[30]	Embedded into a biocompatible system	Diaphragm	Laboratory audio speaker	16 cm^3	1.1 mW	Audio speaker models abdominal motion
Nasiri	2011/[30]	Embedded into a biocompatible system	Pocket	Laboratory normal human gait	16 cm^3	9 mJ (during 22.5 s of gait) and 400 μW	Fixed to a belt
Morais	2011/[31]	NdFeB—copper	Hip	Laboratory normal human gait (controlled using treadmill)	3.76 cm^3	1001.88 μJ and 108.9 μW	Output energy/power of the energy management circuit
Luciano	2012/[32]	NdFeB—copper -ferrite	Knee	Laboratory robotic human gait device	3.9 cm^3	22.1 μJ (each step) and 56 μW (each step)	Non resonant kinetic generator; gait frequency of 1,02 Hz
Romero	2009/[16]	NdFeB—copper	Ankle	Laboratory normal human gait	1.5 cm^3	Not available and 3.9 μW	Mechanical resonant frequencies at 2.8 Hz and subharmonics at 1.5 and 1 Hz
Romero	2009/[16]	NdFeB—copper	Knee	Laboratory normal human gait	1.5 cm^3	Not available and 3 μW	Mechanical resonant frequencies at 2.8 Hz and subharmonics at 1.5 and 1 Hz

Table 7. Comparison of electrostatic harvesters.

Article		Material characteristics	Source		Harvester	Output energy	
Author	Year/ref	for biocompatibility	Location	Test condition	volume	and power	Notes
Tashiro	2002/[33]	Embedded into a biocompatible system	Left ventricle	Laboratory/*In vivo* dog heartbeat (180 bpm)	150 dm^3	Not available and 36 μW	Stimulated by an accelerometer module placed in a canine heart
Miao	2006/[35]	Silicon (proof mass)	Generic human movements	Laboratory sinusoidal input (10 Hz)	1 cm^3	120 nJ/Cycle and 24 μW	Shaker tests—unpackaged
Miao	2006/[35]	Gold (proof mass)	Generic human movements	Laboratory sinusoidal input (10 Hz)	Not available	Not available and 24 μW	Shaker tests
Naruse	2008/[38]	Embedded into a biocompatible system	Generic low frequency human movements	Laboratory sinusoidal input (2 Hz, 0.4 G)	20 × 45 × 6 mm^3	Not available and 40 μW	Test with a shaker
Naruse	2008/[38]	Embedded into a biocompatible system	Generic low frequency human movements	Laboratory sinusoidal input (6 Hz, 0.4 G)	20 × 45 × 6 mm^3	Not available and ~85 μW	Test with a shaker

have relatively higher values than in other parts of the body. The electromagnetic harvesters produce lower power than the abovementioned piezoelectric devices; however, the electromagnetic harvesters do not require significant forces, being driven just by movements, and they are typically suitable to be applied in any body part. The electrostatic harvesters need a movement as well, but, as they produce lower power than electromagnetic harvester, they can only be suitable to power very low power circuits. In contrast, the thermoelectric harvesters are designed to be implanted in areas with a thermal gradient, such as under the skin, where thermoelectric harvesters should be implanted as close as possible to the skin surface, where the temperature gradient is maximum. The different values reported in the power and/or energy output column depend on the different sizes and thermal gradients used in the test conditions. As they must be implanted, energy harvesting devices must be limited in size, typically about 1 cm^3. The electromagnetic harvesters are typically larger

Table 8. Comparison of piezoelectric harvesters.

Article		Material characteristics for biocompatibility	Source		Harvester volume	Output energy power	Notes
Author	Year/ref		Location	Test condition			
Platt	2005/[41, 42]	PZT ceramic	Knee	Laboratory International Standard (ISO 14243-3)	1.2 cm^3 (1 piezo)	Not available and 4.8 mW (3 piezos)	Unpackaged (3 piezos)
Chen	2010/[45]	PZT ceramic	Knee	Laboratory International Standard (ISO 14243-1)	0.45 cm^3	Not available and 1.2 mW	Unpackaged (4 piezos)
Almouahed	2010/[46] 2011/[47, 48]	PZT ceramic	Knee	Laboratory international standard (ISO 14243-1) international standard (ISO 14243-3)	0.4 cm^3 (1 piezo)	Not available and 7.2 mW(4 piezos)	Unpackaged (4 piezos)
Lewandowski	2007/[50]	PZT ceramic	Gastrocnemius	Laboratory 250 ms triangle force F = 250 N	0.5 cm^3	Not available and 68.9 μW	Unpackaged (1 piezo)
Cavallier	2005/[51]	PZT	Generic human movements	Laboratory shock by using a tin ball (1,4 g, 6 Hz) vibration of the beam	87 mm^3	Not available and 0.5 μW	Non-resonant structure—requires high forces
Potkay	2008/[52]	PVDF (polyvinylidene fluoride film)	Artery	Laboratory blood pressure of 135/60 mmHg heart rate of 90 bpm	0.25 cm^3	Not available and 6 nW	*In vitro* test mock artery

Table 9. Comparison of thermoelectric harvesters.

Article		Material characteristics for biocompatibility	Source		Harvester size, volume or area	Output voltage and power	Notes
Author	Year/ref		ΔT	Test condition			
Stark	2006/[53]	Bi_2Te_3	Up to 5 K	Laboratory	95 mm^3	Not available and from 10 μW to 100 μW	Thermo life commercial product
Watkins	2005/[56]	n-$Bi_2Te_3/Bi_2Te_{2.7}Se_{0.3}$ p-Bi_2Te_3/Sb_2Te_3	0.8 K	Laboratory	0.09 cm^2	5.5 mV and 140 μW	Pick and place
Watkins	2005/[56]	n-$Bi_2Te_3/Bi_2Te_{2.7}Se_{0.3}$ p-Bi_2Te_3/Sb_2Te_3	2.7 K	Laboratory	0.16 cm^2	41.8 mV and 980 μW	Wafer-scale
Böttner	2008/ [57–60]	n-Bi_2Te_3 p-$(Bi, Sb)_2Te_3$	2.5 K	Laboratory	7.8 mm^3	Not available and 0.6 mW	Micropelt technology (commercial product)
Yang	2007/[19]	$Bi_2Te_3/Bi_2Te_{2.7}Se_{0.2}$	0.5 K	Laboratory exposed in the environment	390 mm^3	3.3 mV and Not available	*In vitro* test (piece of pork meat)
Yang	2007/[19]	$Bi_2Te_3/Bi_2Te_{2.7}Se_{0.2}$	1.3 K	*In vivo* exposed in the environment	390 mm^3	5 mV and Not available	*In vivo* test (rabbit)
Yang	2007/[19]	$Bi_2Te_3/Bi_2Te_{2.7}Se_{0.2}$	1.2 K	Laboratory cooled by the ice water	390 mm^3	6 mV and Not available	*In vitro* test (piece of pork meat)
Yang	2007/[19]	$Bi_2Te_3/Bi_2Te_{2.7}Se_{0.2}$	5.5 K	*In vivo* cooled by the ice water	390 mm^3	25 mV and Not available	*In vivo* test (rabbit)

than the other devices (more than 1 cm^3). In particular, the harvester proposed by Nasiri [30] exceeds the limit of 1 cm^3 abundantly; nevertheless it is compatible with human-body applications. The other electromagnetic devices differ little from this value. The electrostatic harvesters are smaller than electromagnetic and they can be realized in MEMS technology. While the size of the thermoelectric harvester is smallest, it must be considered that the calculation has been done on the thermoelectric module; to this should be added the remaining parts of a power harvesting device such as the case, the connections of the wires, etc. This aspect introduces the biocompatibility problem. The piezoelectric devices may present biocompatibility problems since they must have a means of transmission for the force directly on the piezoelectric and a contact wire with the measuring device. The electrostatic and electromagnetic devices have no particular biocompatibility problems. In fact, they are applicable within the device that requires their power. They may be proposed for pacemakers, neurostimulators or force gauges (knee). While for the thermoelectric harvesters, special precautions are not specified in the articles with regard to biocompatibility.

Another aspect, treated only briefly in the articles, regards energy management systems. In fact, the energy derived from a power harvester has to be managed before being ready to supply a circuit. Electrostatic harvesters require complex conditioning electronics for managing the charge–discharge cycle of electromechanical conversion and for interfacing the transducer with the load. The conditioning electronics must take into account the variation of the environmental parameters (typically, frequency and amplitude of the external vibration source). This requires intelligent power management and adaptive control of system operation. However, few studies have addressed these points. For this reason, a reliable model for the design of the system is necessary; this can afford to study the system behavior under various conditions of frequency and amplitude of the external vibration source. Piezoelectrics produce high voltage values but low currents and they have high capacitive component, so an impedance matching circuit is required to maximize the generated power. An inductor can be added to take advantage of the capacitive component contribution, but cannot be adaptable to the environmental changes, and the inductance value that must be added for applications that require low frequencies is too big. To overcome this drawback, charging circuits, such as switching type, have been proposed and commonly used in recent years. In switching circuits, the switches operate synchronously with the structure vibration; this is to optimize the power flow. Although the techniques of synchronous switching can increase the energy or the charging speed of the accumulation stage, these circuits require electricity to operate the active components as well. Therefore, in recent years, circuits for totally self-powered devices are under study. However, thermopiles, compared to piezoelectric harvesters, when exposed to low-temperature gradients can deliver high currents at low voltages. Therefore dc–dc converters are required to increase the voltage at the expense of the current value. Usually charge pumps or dc/dc converters based on inductive components are widely used. The great effort to improve the performance of these converters has produced several architectures. Since an electromagnetic power device generates an ac voltage while a power storage element usually requires a dc voltage input, an energy harvesting circuit needs an ac–dc rectifier connected to the two ends of the power generator coil in the first stage. If the output voltage of the rectifier is high enough to be used, the electric energy can be collected directly. However in many cases, the output voltage of the rectifier has too low a value, then it must be intensified to be used or to be collected in the energy storage capacitor and generally low voltage charge pumps can be used. Therefore, in order to develop autonomous systems, the voltage should be increased by a dc–dc converter and the energy should be accumulated. Integrated dc–dc converters are widely used in modern integrated circuits in order to boost the internal voltage where required. Unfortunately, those circuits are not suitable when the input voltage is very low (100–200 mV), or when the voltage is lower than the threshold voltage of the transistors themselves; and innovative converters, based on SOI or natural threshold devices, must be investigated. The dc/dc conversion is usually carried out by the use of charge pumps, but, unfortunately, the classical architectures are unsuitable for ultra low-voltage power harvesting applications. Therefore, for the arduous goal of boosting such low voltages, attention should be focused on novel architectures, based on both inductive and capacitive effects.

An important step in the power management circuit is the design of a charge storing circuit, which can collect the excess charge on a high density capacitor. Then, autonomous implantable systems with the power management circuit cannot measure continuously, but firstly the energy produced by the harvester is stored in a capacitor; in this phase the energy on output load is null, and then the energy previously stored is used to supply the system. These aspects affect the frequency and the technique of the measurement. The measurement phase must, in some way, take into account two main aspects. The first concerns the time between one measurement and the following one; this time cannot be too short because, as pointed out, a certain charging time is necessary to achieve the amount of energy sufficient to perform a measurement phase. The second aspect is related to the time in which a measurement is performed; if a specific event has to be monitored, the measurement itself must be performed when the event occurs. This aspect of synchronism is not easy to obtain, because it is necessary to awaken the overall system and use the stored energy to perform the measurement phase.

5. Conclusions

In an implantable medical device powered by batteries, the longevity of the whole device is typically determined by the battery lifespan. Battery replacement requires a surgical operation, which causes physical and psychic pain to the patient and is an economic burden for the patient itself and for the national health system.

Electronic circuits can be supplied by means of telemetric or energy harvester techniques in order to avoid the use of batteries. However, telemetric techniques require the employment of an external device that can be uncomfortable for the patient or can limit her/his movement.

In this paper, different energy harvesting systems for powering implantable medical devices have been described, analyzed and compared.

Referring to table 4, some of the analyzed implantable medical devices can be supplied by most of the reviewed energy harvesters.

Other implantable medical devices for which the value of the power harvested by the environment is too low to be supplied continuously may require to store the energy up to a level for which the system can operate. Then the implantable system cannot measure continuously but their operation is divided into (1) a ‘charging phase’ in which the energy produced by the harvester is stored in a capacitor and a (2) ‘measuring phase’ in which the energy previously stored is used to supply the system. These two phases introduce a control level to manage the energy, in order to measure the energy level, to awake or turn off the system, to synchronize the measurement, elaboration and communication functions of the system. These aspects affect the frequency and the technique of the measurement. The purpose of the research activity on power harvesting system is therefore not limited to identifying power harvesting devices compatible in size and characteristics with the energy associated to human movement, but also to study and develop search strategies and power management circuits that allow optimizing the energy harvested.

This paper demonstrates that in many cases the most implantable medical devices can be powered by energy harvesters. Hence, energy harvesting represents a promising technique to be used in the design of batteryless implantable medical devices. Anyway, research efforts should be oriented not only to the optimization of the produced power but also to other important features such as biocompatibility and reliability. Biocompatible materials are required for the harvester fabrication; otherwise the different parts of the energy harvester must be fully sealed in a biocompatible material packaging. Encapsulation seems a viable solution for electromagnetic and electrostatic harvesters, since a magnetic field can pass through material. Reliability is an additional main characteristic that has not be tested in the different analyzed prototypes but it stands to reason that, in order to be effective, the lifespan of the implanted energy harvester must be greater than the lifespan of the implanted medical device or at least similar.

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