A CONTEMPORARY SURVEY ON ENERGY HARVESTING TECHNIQUES FOR NEXT GENERATION IMPLANTABLE BIO-MEDICAL DEVICES

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ABSTRACT

Providing a constant and perpetual energy source is a key design challenge for implantable medical devices. Harvesting energy from the human body and the surrounding is one of the possible solutions. Delivering energy from outside the body through different wireless media is another feasible solution. In this paper, we review different state-of-the-art methods that process "in-body" energy harvesting as well as "out-of-body" wireless power delivery. Details of the energy sources, transmission media, energy harvesting, coupling techniques and the required energy transducers will also be discussed. The merits and disadvantages of each approach will be presented. Different types of mechanisms have very different characteristics on their output voltage, amount of harvested power and power transfer efficiency. Therefore different types of power conditioning circuits are required. Issues of designing the building blocks for the power conditioning circuits for different energy harvesting or coupling mechanisms will be compared.

KEYWORDS

Energy Harvesting, Impalntablr Bio-Medical Devices , Pacemaker, Power Delivery, Sensors

1. INTRODUCTION

Implantable Medical Devices (IMDs) have been used for more than 50 years. The early IMDs dated back to the implantable pacemaker in 1958 [1]. Since then numerous types of IMDs were introduced to tackle different health issues. Implantable cardioverter-defibrillators were developed for detecting cardiac arrhythmia and correcting through electric pulses [2]. Implantable insulin pumps were developed to deliver insulin into the body depending on the blood sugar level of the diabetic patients [3]. These traditional IMDs mainly function by monitoring the local signals and activating certain event for reaction. The required power level ranges from μW to mW. With the advancement in VLSI technology, more sophisticated implantable circuits and systems have been developed that have more sensing capability and stimulation functions. Low power wireless data transmissions are also possible. This creates a new class of IMDs which not only monitor and activate signals in the local region, but also collect data, send it back through wireless channel to a local host to do signal processing and receive commands wirelessly to execute massive stimulation and activation. Examples are the implantable retinal prosthesis devices [4, 5], of which the goal is to restore some vision to people who have degenerative eyesight.

Implants are either on the retina (epiretinal implants) or behind the retina (subretinal implants). Images are either captured by an external camera or the implanted micro-photodiodes. After processing, the signals are either generated locally or transmitted from a host processor through a wireless channel to generate electrical simulation signals to the retina cells. Power hungry circuits

DOI : 10.5121/vlsics.2018.9202

such as wireless receivers and electrical stimulators are required. Another example is cochlear implants, which generate electrical stimulation to the auditory system to recover some of the auditory function for the hearing-impaired [6]. Also neural implants are used to directly stimulate the neural cells at the areas of the brain that are dysfunctional due to diseases [7, 8]. Neural implants that have the capability of capturing the activity of the brain and using it for brain–computer or brain–machine interface are also becoming reality. These neural implants require circuits to do electrical stimulation, data capture and also wireless communication, and hence require significant amount of power. Table 1 gives a summary of the power requirement of different types of IMDs.

There are many design challenges for IMDs. Power consumption, size, durability, reliability and biocompatibility are some of the key ones. Among them, power consumption is probably the dominant issue as it also affects the other factors. Traditional IMDs such as pacemaker and defibrillator use battery to provide power to the device. Even though the current consumption of the device is in the range of μA, the battery only lasts for a certain period of time (15 years for pacemaker and 7–8 years for defibrillator). When the battery is gone, another operation is required to replace the old device with a new one. For other devices such as neural implants which consume significantly larger power, either a larger battery is used which leads to a larger volume or the frequency of replacement is increased.

Table I. Power Requirement Of Imds

IMDs	Neural Implants	Cardiac Pacemaker	Cochlear Implants	Retinal Implants	Insulin Pump
Power Required	10mW ~200mW	1 μW ~10 μW	10mW ~100mW	1mW ~100mW	10mW ~50mW

Both are not desirable as we want the IMDs to work perpetually and we want them small. To achieve this contradicting goal, energy harvesting and wireless power delivery methodologies were proposed as the power supply methods for IMDs. Energy harvesting has become popular recently as a strategy to provide power for low power sensors or microsystems used in areas with environmental hazard where it is difficult to recharge or replace the battery [9, 10]. Different types of devices were developed to scavenge energy from the environment. Sources of energy include solar, wind, vibration, radioactivity, ambient RF and heat. Human body, at the same time, is also a great source of energy.

Every day for an adult, the average daily diet provides about 10MJ of energy. Different amounts of power are generated for different daily normal activities, e.g., housekeeping generates 175W and 81W is produced during sleeping [11].Therefore it is tempted to use this as the energy source for IMDs. For the case that we cannot harvest energy from the body, if we can obtain the power wirelessly from outside to power up the implanted devices or recharge the battery, it will remove the requirement of an internal battery or help to reduce its size, and prolong the lifetime of the devices. In this monograph, we will review the recent trends in the research and development of the power provisioning methodologies for IMDs.

We categorize the methods of providing power to IMDs into two types, the "in-body" type and the "out-of-body" type. The in-body energy sources come from the human body. These include the kinetic, thermal, biochemical and direct electrical energy. The movements of the human body or even the internal organs [12] are good sources of kinetic energy. In [13] and [14] it is shown that several μW to mW of power can be extracted from the trunk and the head of the body during walking or running. The inner human body temperature is maintained at a relatively constant value of 37◦C and there is a temperature gradient between the inner body, the skin and the air ambience. Exploiting this existing thermal gradient, thermal electric generator (TEG) can be used to generate electric power.

Another abundant source of energy inside the body is glucose. Implantable biofuel cells using glucose as a reactant have been investigated and researched for a long time. It has been demonstrated recently that tens of μW/cm2 power density can be generated constantly for over a month using glucose biofuel cell. Some of our body part is itself a natural electrical battery, e.g., endocochlear potential. If the potential is large enough and the corresponding power condition circuits can be designed to match with the requirement, electrical energy can be harvested directly from this potential. In [15] it has been shown that nW of power can be extracted from the ear of a guinea pig for up to 5 hours. For out-of-body power delivery, external energy source is used to couple the energy into the IMDs or directly activate the energy harvester implanted inside the body. The external power sources come from magnetic energy, ultrasonic wave, optical wave and the most common one, electromagnetic induction.

The whole implantable energy harvester/receiver consists of three parts (Fig. 1), the primary energy source, the energy transducer and the power conditioning blocks. The primary energy source is either "in-body" or "out-of-body" source. The energy transducer collects the harvestable energy in a certain form and transforms it into electrical energy.

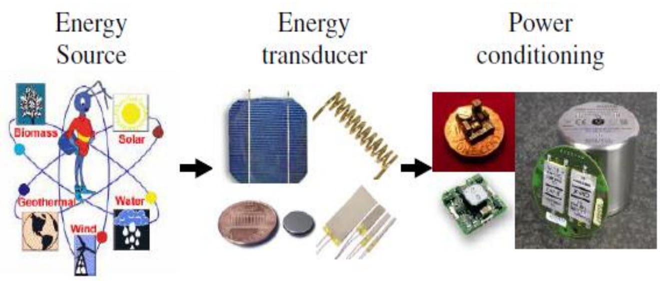

Fig. 1. Building blocks of the implantable energy harvester

The harvested electrical energy is time varying and usually the output voltage and power levels are low. Therefore power conditioning system is required to regulate the output voltage and deliver the output to the load in the required form. In some energy harvesters, the harvested power varies with the environment and there exist some operation points that the harvested power is maximized. In this situation, the power conditioning block is also required to track and operate the system in the maximum power point (MPP) in order to optimize the power transfer efficiency.

The rest of the paper is organized as follows. Section II will discuss different types of energy harvesting sources and wireless power delivery mechanisms. The corresponding energy transducer designs will also be presented and the optimum design strategies will be discussed. The building blocks of the power condition circuits will be presented in Section III. Detail system and circuit design will also be provided. Section IV concludes the review paper.

2. ENERGY SOURCES AND TRANSDUCERS

2.1 IN-BODY ENERGY HARVESTING

In-body energy harvesting converts the mechanical energy, chemical energy, heat energy and electrical potential available in the human body into useful electrical energy form to power up the IMDs. In the following subsections, we will present these energy sources inside the human body and discuss the transducers used to harvest these energies.

2.1.1 KINETIC ENERGY HARVESTING

In human body, kinetic energy is obtained from the voluntary movement of the limbs and the body, the involuntary movement of the organs and the muscles and the vibration induced from the actions of the body. They can be regular or random. Kinetic microgenerators are used to convert the kinetic energy into electrical energy. They are based on three different energy transduction methods, electromagnetic, piezoelectric and electrostatic.

(i) Electromagnetic

Electromagnetic transduction uses the change of magnetic flux to induce a voltage to drive current in a circuit loop. The change of the flux comes from either the movement of the circuit over a fixed field or a moving magnet. One classical body-based electromagnetic microgenerator is used in the automatic power-generating system (AGS) (Fig. 2) of Seiko Kinetic wristwatch where the wrist motion causes the rotation of an imbalanced rotor that drives a microgenerator through a train of speed up gears [16]. A few μW of power is extracted from this device. AGS was extended to use in pacemaker applications. In [17], it is estimated that with a heartbeat rate of 200 beats per minute (bpm), 80 mJ is generated in 30 minutes, i.e., 13 μJ per heartbeat. This amount of harvested energy is sufficient to pace a mongrel dog. More recently, *in vivo* experiment was done in [18]

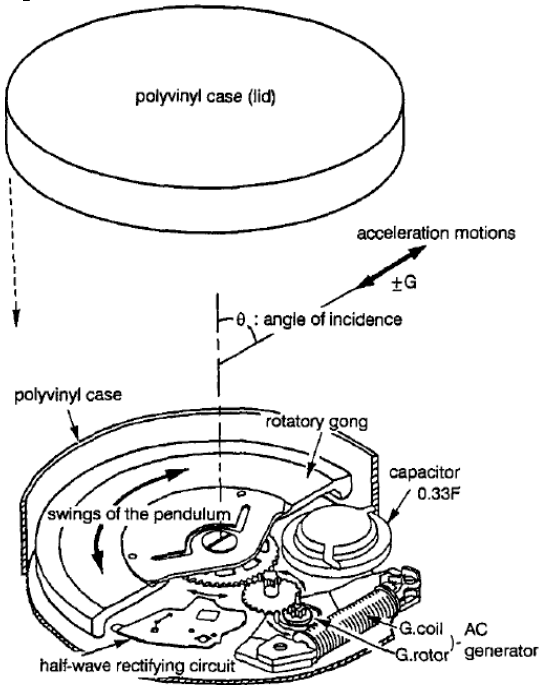

Fig. 2. Structure of AGS [27]

where the AGS was affixed onto a sheep heart for 1 hour. With a rate of 90 bpm, the average power collected by the AGS is 16.7 μW, i.e., 11.1 μJ per beat. With the power consumption of the latest pacemakers being reduced to 8 μW [19], it is enough to use AGS to harvest the heartbeat energy to power up the pacemaker. One of the drawbacks for AGS is that moving mechanical parts have friction. Wear and tear will happen and regular maintenance is required. This means extra operation to the patient and may not be desirable.

(*ii*) *Piezoelectric*

Piezoelectric transducer produces charges when a mechanical force, either stress or strain, is applied on it. The resonant frequency of the piezoelectric generator is inversely proportional to it size. Thus the smaller it is, the higher its resonant frequency. At the same time, the human motion is usually of low frequency and high amplitude. Thus most of the human piezoelectric generators obtain power either through direct straining or by impacting through direct force instead of periodic vibration. In [20], an integrated piezoelectric lead zirconate (PZT) dimorph was put in the shoe's insole for energy harvesting (Fig. 3).

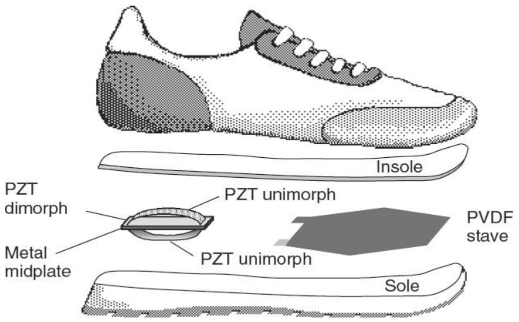

Fig. 3. Harvesting energy through PZT dimorph placed inside the shoe [20].

Low frequency direct force during walking and running creates mechanical stress on the material and generates charge accordingly. On average 8.4mW is produced for a 500-kOhm load at a 0.9-Hz walking pace.

However piezoelectric generation based on limb movement is not a good choice for power supply for IMDs since no energy is generated when the patient is not walking or running. As sensors based on MEMS technology become more mature and sophisticated, it is anticipated that some of the IMDs only need power in the range of μW. This triggers new investigations on using piezoelectric generator to provide very low amount of power. Ramsay and Clark [21] and Sohn et al. [22] investigated the feasibility of using piezoelectric membrane to extract energy from the change of blood pressure at each heart pulse. The change of blood pressure is typically in the range of 30 to 50mmHg. It was estimated that theoretically 23 μW can be extracted for a membrane size of 1cm2 and 9 μm thick. Nanodevices were also proposed for biomechanical energy harvesting [23].

Piezoelectric nano-wirebased nano-generator that consists of a flexible substrate with Zinc-Oxide (ZnO) nanowires affixed at two ends is used to harvest energy from muscle stretching. It was shown that when the device is put on a running hamster, the short-circuit output current and open circuit voltage reached 0.5 nA and ~50mV to 100mV, respectively. Nano-generator is interesting because it can be integrated with other electronic circuits in the same substrate or chip. However, currently the power and output voltage level generated are still too low to be practical.

(*iii*) *Electrostatic*

Electrostatic transducers use variable capacitors to transform the mechanical energy into electrical potential energy. The capacitance value is varied by changing the separation or the overlapping of the parallel plates of the capacitor using the harvested mechanical force. There are two types of electrostatic transducers, fixed-charge and fixed-voltage. For the first one, the capacitor is pre-charged and then the relative motion causes the plates to move. When the separation of the plates is increased, the capacitance is reduced. With a constant charge, the voltage of the capacitor increases and hence energy is stored in the capacitor. For the fixed-voltage transducers, if the voltage is fixed, the increase in the plate separation shrinks the electric field strength and hence causes charge to flow into an external circuit as a current. Meninger et al. [24] gives a detail description of the operation cycles for the two modes and the energy conversion cycles.

2.1.2 Biofuel Cell

Fuel cell is an electrochemical device that generates power by providing catalyzed complementary electrochemical reactions at the anode and cathode pairs. Reducing-agent fuel is on the anode while an oxidant flows across the cathode. In biofuel cell, glucose, which is of abundant quantity in the human body, is used as the fuel and oxidized at the anode while oxygen is reduced to water at the cathode. Fig. 4 shows the schematic of a simple membrane-based biofuel cell.

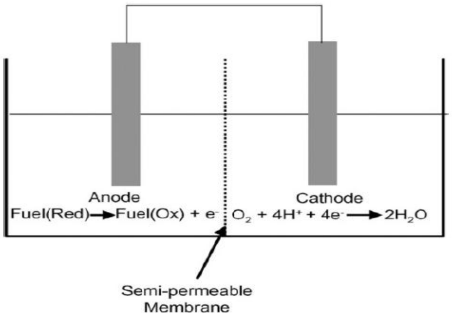

Fig. 4. Device schematic of a membrane-based biofuel cell.

There are three types of glucose fuel cells, depending on the type of catalyst used. They are microbial-catalyzed, abiotically-catalyzed and enzymatically-catalyzed. Using living microorganisms such as bacteria, microbial fuel cells can catalyze complete oxidation at the anode and generate 24 electrons per molecule of glucose consumed. Thus it has the highest efficiency and the power density is in the range of >1000 μW/cm2. However because of safety and biocompatibility, it is not suitable for IMDs.

Enzyme-based fuel cell is the next highest in catalytic efficiency. Many researches have been done in the recent years and the output power density has been improving. The operating voltage and power density was improved from 0.4V and 140 μW/cm2 [25] to 0.52V and 430 μW/cm2 [26].

2.1.3 THERMOELECTRIC ENERGY HARVESTING

Human body is a large heat energy source. Thermoelectric generator (TEG) uses the temperature difference between the body and the ambience to convert thermal energy into electric energy. In [27], it is advocated that using the temperature difference between the inner surface of the skin and the core body, enough energy can be generated to provide power for implants such as pacemaker. It is also theoretically demonstrated that using thin film super-lattice thermoelectrics, enough power can be obtained even when the temperature difference is only 0.8◦C. In [28], flexible TEG is fabricated and attached on the human body to demonstrate that about 1 μW can be obtained when the temperature difference between the skin and the ambience is 10ºC. TEG is based on the See beck effect. A voltage is created when there is a temperature difference at the two junctions where two metals or semiconductors are joined. The voltage is given by the following equation

$$V = \int_{T_1}^{T_2} (S_B(T) - S_A(T))dT$$

where $T1$ and $T2$ are the temperatures at the two junctions, S_A and S_B are the Seebeck coefficients of the two materials, respectively. The voltage and power generated by individual thermocouple may not be enough for applications. We can put them electrically in series and thermally in parallel to form a thermopile to increase both the output voltage and power. Fig. 5 shows an example structure of a thermopile. For a certain heat flow, the temperature difference depends on the thermal resistances of the thermal circuit.

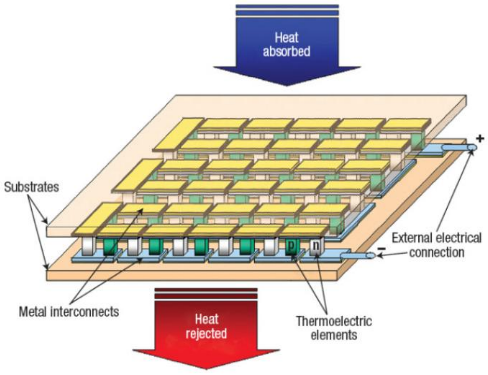

Fig. 5. A typical thermopile structure [29].

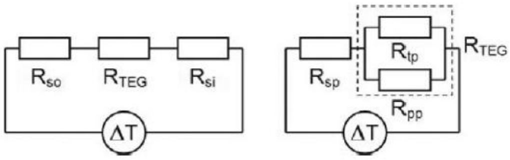

Fig. 6. Thermal circuit for TEG energy harvester [30].

Figure 6 shows the equivalent thermal circuit when the thermopile is placed between the body core or the skin (the hot plate) and the air (the cold plate). R_{si}, R_{so} and R_{teg} are the thermal resistances of the heat source, sink and the thermopile, respectively. To extract the maximum power from the TEG, it seems that we need to maximize the thermal resistance of the TEG (R_{teg}) so that the temperature gradient across the thermopile is the maximum. However, since the thermal and electrical impedances both affect the power extraction efficiency and are also related to the dimension of the thermopile, we need to consider their matching together [30].

A 5mm×5mm TEG is built with 540 × 540 wires, each has a diameter of 80 nm and 1 μm height. While these micro-TEGs are very small in size and can be integrated with other CMOS circuits, the extracted power and output voltage are too low to be practical under the current technology. In general the output voltage ranges from a few mV to 100mV and the effective power extracted ranges from a few nW to a few μW.

A complete thermal energy harvesting power supply for implantable pacemakers [31] wass presented. The designed power supply includes an internal startup and does not need any external reference voltage. The system is designed such that no failure occurs under overload conditions. Using this approach, a thermal energy harvesting power supply has been designed using 180-nm CMOS technology.According to HSPICE simulation results, the circuit operates from input voltages as low as 40 mV provided from a thermoelectric generator and generates output voltages up to 3 V. A maximum power of 130 µW can be obtained from the output of the boost converter. TE energy harvesting system architecture for the implantable biomedical devices has been shown in Fig. 7.

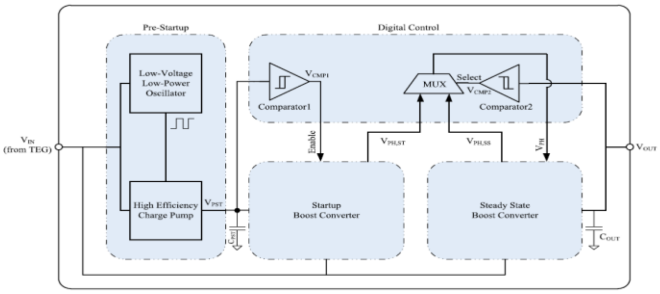

Fig. 7. TE energy harvesting system architecture [31].

2.1.4 HUMAN BODY BATTERY

Human body has many natural electrochemical potentials, for example, membrane potential across the cell, and it can be treated as a kind of biological battery. It has been known for a long time that for mammals the inner ear has the highest positive electrochemical potential, which is about 70–100mV [32]. It is called endocochlear potential (EP) and is actively maintained by the inner ear to facilitate the cochlear mechanotransduction of sound pressure vibration into electrical signal and excite the auditory nerve. A chip was designed, and fabricated to harvest this tiny electrical energy directly from the ear [32]. To demonstrate the usefulness of the extracted energy, it is integrated with a wireless sensor which monitors and reports the EP itself. *In vivo* demonstrations were done and a minimum of 1.12 nW was extracted from the EP of a guinea pig for up to 5 hours. The amount of power extracted is smaller than the theoretical value that can be generated from the EP of the guinea pig, which should be in the range of 14–28 μA [33].

It is because the pin-shaped electrode used has high impedance. Figure 8 shows the implantation of the energy harvesting chip in the cochlea. Since the extracted power level is so low, there are many challenges to use it to power practical circuits. Extremely low duty cycling is needed to reduce the active power consumption. Also the quiescent power of the circuit has to be much lower than the extracted power in order to allow the energy to accumulate during the sleep period. In addition, since the output voltage is very low, ~100mV, step-up boosting circuit is needed and hundreds of mV is required to start up the circuit. In [32], a wireless kick-start energy receiver and buffer are used to solve the start-up problem. At the same time extreme leakage reduction and ultra-low power design techniques are used to reduce the quiescent power. It was shown that the power extracted can enable a 2.4-GHz radio to transmit one EP measurement data every 40–360 s. This shows the potential of using biologic battery inside the body to drive some very low-power IMDs.

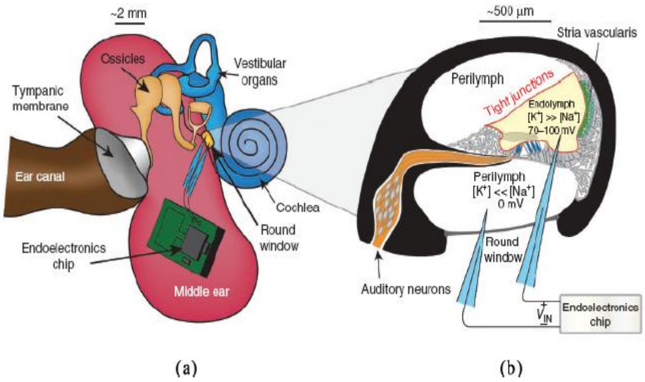

Fig. 8. Energy harvester for the EP harvesting [32].

2.2 OUT-OF-BODY WIRELESS POWER DELIVERY

Besides harvesting energy from the human body directly, we can also obtain energy from the outside sources through different wireless media. The difference from the environmental energy harvesting is that the energy source is actively applied near the human body and the transducers in the IDMs pick up and convert the coupled energy into electrical energy. The external energy source can be mechanical, optical or electromagnetic waves.

2.2.1MECHANICAL ENERGY COUPLING

Mechanical energy can be picked up inside the body through coupling using different wireless media. In the following, we will discuss two wireless mechanical coupling media, ultrasound and magnetic field.

(*i*) *Ultrasound*

Ultrasonic instruments have been used for human body diagnosis for a long time. At the same time, ultrasound can be used to transfer energy from one point to another point inside the human body. Ultrasound has advantages over electromagnetic coupling since it does not create any EM interference to other devices and also is immune to EM interference itself. Mechanical transducers, mainly piezoelectric elements, pick up the mechanical energy induced by the ultrasonic wave and vibrate. The induced vibrational energy is then transformed into electrical energy. Figure 9 shows how to use the ultrasonic wave to transfer energy into the body implants [34]. The idea of using ultrasonic wave to transfer energy for implantable devices can be dated back to 1988 when Cochran et al. [35] incorporated a piezoelectric element near the fracture bone to generate small current to stimulate the bone healing when it is excited mechanically by an external ultrasonic source. *In vivo* experiments showed that with a piezoelectric material of 5 × 5 × 0.9mm3 size, using a ultrasound wave with 2.25 MHz frequency and 10–20V input on the external ultrasonic transducer, a rectified current of 20 μA was generated.

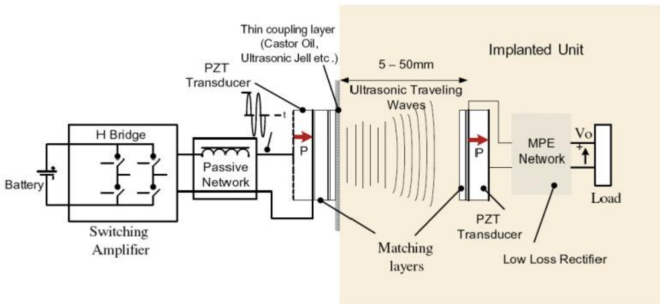

Fig. 9. An ultrasonic energy harvesting system [34].

Phillips et al. [36] used ultrasound to power up a nerve stimulator. *In vivo* experiments were done on a large American bullfrog. Using a 3.5-mm diameter receiver and a source from a 6mm diameter 2.25MHz ultrasound transducer, a 1.75–2.75 V rectified output voltage is obtained when an average 1–1.25mW/cm2 ultrasound intensity is applied and the two transducers are 6mm apart. It is also shown that the current generated is sufficient to produce muscle twitches at the bullfrog. It demonstrates that ultrasound can be a good way to couple energy for nerve stimulation. The efficiency of the ultrasonic energy coupling depends a lot on the acoustic impedance and hence the packaging of the devices. Packaging issue of the piezoelectric implant was tackled in [37]. Soft biocompatible cohesive gels are used as packaging and they are soft enough to absorb the incident wave from the subcutaneous tissues. Experiments were done using real pork as the separation medium and it was also shown that a spherical package is more suitable than the cubic one when the device is buried in the fatty layer.

(*ii*) *Magnetic coupling*

Suzuki et al. [38, 39] proposed a power supply system for implantable medical devices using magnetic coupling. The energy source comes from a rotor which has permanent magnets at the circumference. A similar rotor is placed inside the IMD. Figure 10 shows the principle of the magnetic coupling through rotor and the design of the implanted rotor. The magnets of the two rotors form the magnetic coupling. When the external rotor is rotated at a low frequency by an external mechanical force, the internal rotor rotates synchronously due to the varying magnetic field coupling.

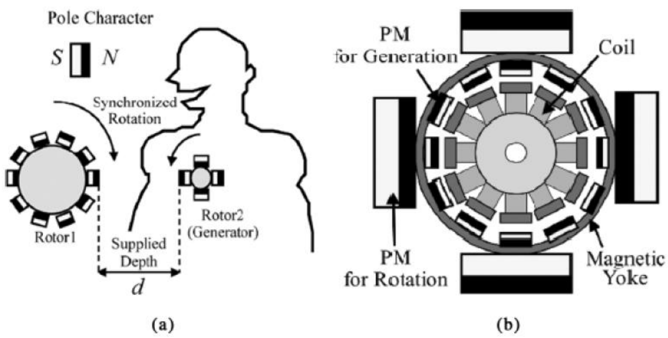

Fig. 10. (a) Power delivery using magnetic coupling; (b) design of rotor [40].

A micro generator is attached with the internal rotor and thus AC power is generated which is then rectified by the power conditioning circuit to generate the required DC voltage source. Output power is increased through the use of high-ratio gear, and by increasing the number of magnets on the rotors and using multi-polarized generator. The major advantage of this scheme is that it can provide large amount of power even the implant is located deep under the skin. From the experimental result in [39], the maximum power is 3.1W when the distance between the two rotors is 1.5 cm and the internal rotor is rotating at 87 rps. The power can keep at a level larger than 1W when the distance is below 2.5 cm. The power is dropped to zero when the distance is larger than 4 cm. In [40], more powerful generators are used to increase the power generation. The peak power is increased to 6W while the distance is extended to 3 cm if we want to keep the power larger than 1W. No *in vivo* experiment has been reported for this technology.

There are several disadvantages of using magnetic coupling and these restrict its deployment in IMDs. First the volume of the overall implanted generator is large, about 10 cm^3. Second, biocompatibility is another problem. Putting a rotating device inside a body may cause issues such as heat dissipation to the body. In addition, magnetic coupling device is not maintenance free. Since the device has rotational components, constant lubrication is required to reduce the frictions.

2.2.2 OPTICAL ENERGY COUPLING

Using solid-state light and free-space optical communication, we can transmit both power and signal through visible light. In [41], a sensor that can receive power, clock and data from optical sources at different wavelengths were presented. Clock data recovery (CDR) and delta-sigma-modulator (DSM) circuits were used to demonstrate the optical communication and data processing part. Integrated photodiodes are used to supply power and capture optical data.

Twelve photodiodes, each of 75 × 75 μm2, were used. When the photodiode is illuminated by a 10mm white LED and the transmission distance is 1 inch, the CDR functions correctly at a data rate of 50 kbps with a supply voltage of 300mV. Also the performance of the DSM for an output sampling rate of 256KHz with an input tone at 177Hz was shown for different input light intensities.

2.2.3 INDUCTIVE COUPLING

Electromagnetic inductive coupling is the most common ways of providing power to IMDs externally. Usually near-field electromagnetic coupling is preferred than far-field coupling due to its higher efficiency and less absorption by the human body. For near-field coupling an internal coil (secondary coil) picks up the power induced by an external coil (primary coil) as shown in Figure 11 [42]. An ideal model for the electromagnetic inductive coupling is shown in Figure 12. $L1$, $L2$ and M are the self-inductances of the two coils, and their mutual inductance, respectively.

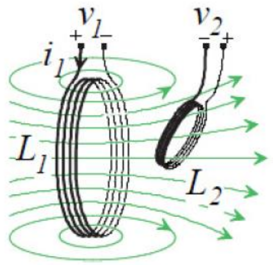

Fig. 11 Coil coupling [42].

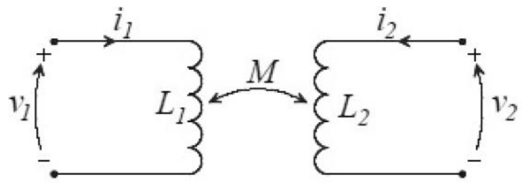

Fig. 12. Model of two coupled coils.

In [43] three-coil transmission system was proposed to achieve both high link efficiency and high power delivered to load (PDL). It was shown that for a link with 13.56 MHz carrier frequency, if the coupling distance is short, two-coil links is still the best choice when the driver has large output resistance or the load power is small. When the distance exceeds 10 cm, three-coil system gives the highest link efficiency and PDL. Link efficiency is 147% and 5% higher and PDL is 1.5 and 59 times higher than its equivalent two- and four-coil links, respectively.

2.3 COMPARISON OF DIFFERENT ENERGY PROVISIONING TECHNOLOGIES FOR IMPLANTABLE DEVICES

Table 2.1 shows a comparison of different energy provisioning technologies discussed in the previous subsections in terms of amount of power harvested or coupled, size, location of deployment and possible applications. The power numbers are obtained from measurement data of real prototypes. From the table, we can see that in general out-of-body wireless electromagnetic inductive coupling power delivery provides the highest power delivery than other methodologies. Also other out-of body power delivery methods show good power density number. However most of the out-of-body wireless power delivery devices work only when the device is not implanted deeply into the body. They are either implanted in the eyes or just underneath the skin surface or the skull. It also needs an active source of energy and cannot work without the attention of the patient. It can also be seen that although MEMS-based and nanotechnology based devices can be easily integrated with the electronics together, in general the energy outputs of these devices are quite small, in the range of nW. It is only useful if the power consumption of the application is very small.

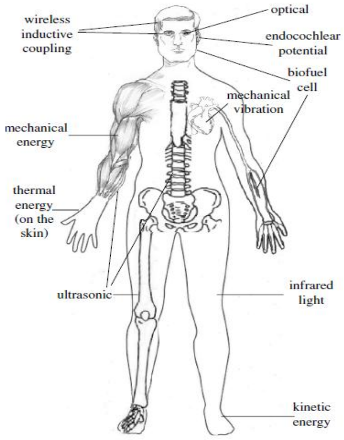

Fig. 13. Possible locations of deploying different energy provisioning technologies.

Figure 13 shows a diagram of a human body, which illustrates the possible locations of deploying the devices that use different types of energy provisioning techniques.

2.3.1 POWER CONVERTER

In the power conditioning system, the power converter is the most important component. It should have high power conversion efficiency and also cover a large range of input voltage as the operation environment varies significantly. Different power converters are used for different energy provisioning techniques depending on whether the received power is AC, DC or pulse type.

2.3.2 DC TO DC POWER CONVERTER

When the harvested power is DC, DC to DC power converter is needed. Either switching converter or charge pump can be used. Switching converter supports large input voltage range and has high efficiency. For applications that require the output to supply the load and recharge the secondary storage at the same time, a single inductor dual-input dual–output DC–DC converter as shown in Figure 14. was proposed [44] to reduce the number of external inductor to 1.

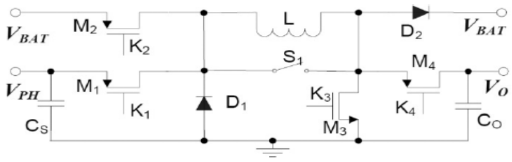

Fig. 14. A single inductor dual input dual output DC–DC converter [44].

If the harvested energy from V_{PH} is larger than the loading (*VO*), the extra energy flows into the battery V_{BAT} through *D*2. If the harvested energy from *V*PH is less than the loading, the battery V_{BAT} supply the extra energy required from *M*2 to the loading. Despite having high efficiency and supporting large input voltage range, switching converter requires a big off-chip inductor, which is disadvantageous for the miniaturization of the IMDs. Recent researches proposed to use bonding wire as the inductor so that the whole DC–DC converter can be integrated into a single chip [45]. However the inductance value is quite small, only around a few *nH*. Thus high switching frequency of around 100MHz is required to maintain the normal operation of the converter. This causes high switching power overhead, reduces the power conversion efficiency and hence is not suitable for IMD applications where the extracted power level is usually not very high.

To implement a fully on-chip integrated power converter, charge pump (QP) can be used as it only consists of switches and capacitors and both of them can be integrated on-chip. The power requirement of the IMDs is usually in the order of *mW* or even μW. On-chip QP is a good choice for IMDs. In [46], a solar cell is integrated and utilized to harvest the optical energy to charge a re-chargeable battery. Charge pump circuit is used to provide the load power from the harvested power. QP can be either step-up or step-down. For the step-up QP, there are linear QP [32, 47], Fibonacci QP [48] and exponential QP [49], of which the conversion ratio, i.e., the ratio of the output voltage to the input voltage, varies with the stage number accordingly. If the step-up QP is used reversely, it becomes a step-down QP. For IMD applications, the energy source varies with the operation environment and so does the input voltage. QP with a fixed number of stages cannot always guarantee to provide a high-enough output voltage to satisfy the load requirement. A reconfigurable charge pump was proposed in [50] (Figure 14) in which the conversion ratio is tunable depending on the input voltage values. By doing so, the allowable input voltage range is enhanced significantly.

For DC energy harvesting, when there is no initial energy, the switching converter and the charge pump cannot start properly. In [51], a small step-up charge pump with an oscillator working at a very low supply voltage is used as the start-up circuits. It is shown in Figure 15.

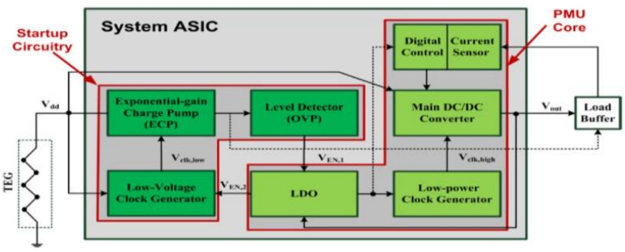

Fig. 15. A start-up circuit based on DC oscillator and charge pump for DC–DC converter [51].

As long as the output voltage of the energy generator can start up the oscillator, the step-up charge pump will generate a higher voltage to wake up the control circuits in the main DC–DC converter or the main charge pump. After that, the main DC–DC converter can work properly.

2.3.3 POWER CONDITIONING CIRCUITS

The basic building blocks of the power conditioning circuits are power converter, start-up circuit and the Maximum Power Point Tracking (MPPT). The power converter is the most important block in the power conditioning circuit. A DC-DC converter with rectifier which is shown in Fig. 16 has been implemented [52]. This type of power conditioning circuit enhances the overall efficiency of the power converter.

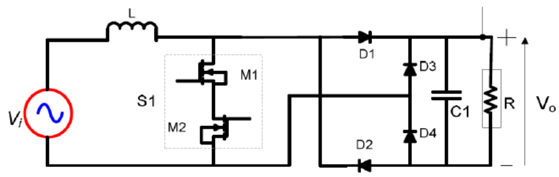

Fig. 16. DC-DC converter with rectifier [52]

In the power conditioning circuit of the energy harvesting [55, 56, 57] system, start-up is another very important issue. For the power converter circuit to work, we need to power up the control circuit first. If there is a rechargeable battery with initial energy in the system, it can provide power for the control circuit during the starting up of the system. A kinetic start-up mechanism [53] for DC–DC converter is shown in Fig. 17.

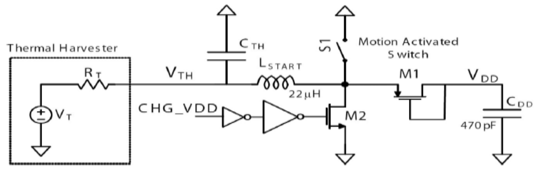

Fig. 17. A kinetic start-up mechanism for DC–DC converter [53].

For some energy generators such as piezoelectric material or PV cells, there exists a maximum power operation point. If the electrical model of the energy generator model is linear, e.g., piezoelectric generator, the maximum output power is achieved when the input impedance of the power converter is conjugate matched with the load impedance. A continuous MPP tracking method is shown in Fig. 18.

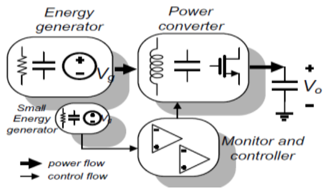

Fig. 18. A continuous MPP tracking method [54]

3. CONCLUSION

Various energy harvesting and wireless power delivery techniques for IMDs were reviewed. Different energy sources and the corresponding energy transducers and generators were presented and discussed. For different energy generators, different power conditioning circuits are required. The building blocks of the power conditioning circuits such as the power converters, the protection circuits and maximum power point tracking circuits were presented. Other design issues such as start-up issues and power control mechanisms were also discussed. From this review paper it is evident that the energy harvesting circuits are necessary and unavoidable for the next generation implantable biomedical devices such as cardiac pacemaker, implantable insulin pumps and neural implants etc.

TABLE I. COMPARISON OF DIFFERENT ENERGY HARVESTING TECHNOLOGIES FOR IMDS (IN BODY / OUT BODY)

Energy Harvesting Mechanism	Ref no.	Location of deployment	Applications	Amount of harvested power/power density	Device area/volume/size
Kinetic mechanical electromagnetic	[17]	Heart	Pacemaker	44 μ W or 13μJ/heartbeat	-
Kinetic mechanical–piezoelectric	[18]	Heart	Pacemaker	16.7 μ W (in vivo)	-
Kinetic mechanical–piezoelectric	[20]	Foot	Collecting energy during	1.3mW (0.8Hz walking pacing frequency)	5 cm $\times$5 cm $\times$0.015 in
Kinetic mechanical–piezoelectric	[22]	Heart	Collecting energy from heart	0.33 μ W (1Hz force of 5333N/m2)	1 cm2 $\times$ 9 μm
Kinetic mechanical–electrostatic	[27]	Heart	Pacemaker	58 μW (using simulated goat heartbeat magnitude)	3 cm $\times$10 cm (area), 1.2 kg (weight)
Endocochlear potential	[15]	Ear	Sensing	1.12nW (from EP of pig)	2.4 $\times$ 2.4 $\times$0.2mm^3
Mechanical– ultrasound	[35]	Bone	Stimulator	20 μ A (2.25MHz input ultrasonic wave with intensity of 10mW/cm^2)	5 $\times$ 5 $\times$ 0.9mm^3
Mechanical– ultrasound	[36]	Nerve	Stimulator	1.75–2.75V rectified output voltage (2.25MHz input ultrasonic wave with intensity of 1–1.25mW/cm2)	3.5 diameter receiving surface
mechanical– magnetic coupling	[39]	Underneath the skin	-	1W with primary rotor rotating at 87 rps and 2.5 cm away	10 cm^3
Wireless inductive coupling — two coil system	[5]	Eye	Retinal prosthesis	1.3mW @125KHz and 15mm distance	32mm $\times$13mm (area of the whole device)
Wireless inductive coupling — four-coil system	[43]	-	-	80–140mW output (80% power transfer efficiency) @ 700 kHz and 10 to 20mm distance	22mm diameter implanted coil

REFERENCES

[1] B. Larsson1, H. Elmqvist, L. Ryd´en, and H. Sch¨uller, "Lessons from the first patient with an implanted pacemaker," Pacing and Clinical Electrophysiology, vol. 26, no. 1, pp. 114–124, Janaury 2003.

[2] A. J. Moss, W. J. Hall, D. S. Cannom, J. P. Daubert, S. L. Higgins, H. Klein, J. H. Levine, S. Saksena, A. L. Waldo, D. Wilber, M.W. Brown, andM. Heo, "Improved survival with an implanted defibrillator in patients with coronary disease at high risk for ventricular arrhythmia," The New England Journal of Medicine, vol. 335, pp. 1933–1940, December 1996.

[3] W. J. Spencer, "A review of programmed insulin delivery systems," IEEE Transactions on Biomedical Engineering, vol. 28, no. 3, pp. 237–251, March 1981.

[4] W. Liu, M. S. Humayun, and J. D. Weiland, "A variable range bi-phasic current stimulus driver circuitry for an implantable retinal prosthetic device," IEEE Journal of Solid-State Circuits, vol. 40, no. 3, pp. 763–771, 2005.

[5] L. Theogarajan1, J. Wyatt, J. Rizzo, B. Drohan, M. Markova, S. Kelly, G. Swider, M. Raj, D. Shire, M. Gingerich, J. Lowenstein, and B. Yomtov, "Minimally invasive retinal prosthesis," in Proceedings of the IEEE International Solid-State Circuits Conference, pp. 99–108, February 2006.

[6] J. Georgiou and C. Toumazou, "A 126-uW cochlear chip for a totally implantable system," IEEE Journal of Solid-State Circuits, vol. 40, no. 2, pp. 430–443, February 2005.

[7] A. M. Sodagar, G. E. Perlin, Y. Yao, K. Najafi, and K. D. Wise, "An implantable 64-channel wireless microsystem for single-unit neural recording," IEEE Journal of Solid-State Circuits, vol. 44, no. 9, pp. 2591–2594, September 2009.

[8] R. R. Harrison, "The design of integrated circuits to observe brain activity," Proceedings of the IEEE, vol. 96, no. 7, pp. 1203–1216, July 2008.

[9] S. P. Beeby, M. J. Tudor, and N. M. White, "Energy harvesting vibration sources for microsystems applications," Measurement Science and Technology, vol. 17, no. 12, pp. 175–195, December 2006.

[10] J. A. Paradiso and T. Starner, "Energy scavenging for mobile and wireless electronics," IEEE Pervasive Computing, vol. 4, no. 1, pp. 18–27, 2005.

[11] T. Starner, "Human-powered wearable computing," IBM Systems Journal, vol. 35, no. 3&4, pp. 1–12, 1996.

[12] J.-L. Gonz´alez, A. Rubio, and F. Moll, "A prospect on the use of piezoelectric effect to supply power to wearable electronic devices," in Proceedings of the International Conference on Materials Engineering Resources (ICMR), pp. 202–206, October 2001.

[13] P. D. Mitcheson, "Energy harvesting for human wearable and implantable bio-sensors," in Proceedings of the IEEE Annual International Conference on Engineering in Medicine and Biology Society (EMBC), pp. 3432–3436, September 2010.

[14] E. Goll, H.-P. Zenner, and E. Dalhoff, "Upper bounds for energy harvesting in the region of the human head," IEEE Transactions on Biomedical Engineering, vol. 58, no. 11, pp. 3097–3104, November 2011.

[15] P. P. Mercier, A. C. Lysaght, S. Bandyopadhyay, A. P. Chandrakasan, and K. M. Stankovic, "Energy extraction from the biologic battery in the inner ear," Nature Biotechnology, vol. 30, no. 12, pp. 1240–1244, December 2012.

[16] M. Kayakawa, "Electronic wristwatch with generator," U.S. Patent 5001685, March 19 1991.

[17] H. Goto, T. Sugiura, Y. Harada, and T. Kazui, "Feasibility of using the automatic generating system for quartz watches as a leadless pacemaker power source," in Proceedings of the IEEE Annual International Conference on Engineering in Medicine and Biology Society (EMBC), vol. 1, pp. 417–419, October 1998.

[18] A. Zurbuchen, A. Pfenniger, A. Stahel, C. T. Stoeck, S. Vandenberghe, V. M. Koch, and R. Vogel, "Energy harvesting from the beating heart by a mass imbalance oscillation generator," Annals of Biomedical Engineering, vol. 41, no. 1, pp. 131–141, January 2012.

[19] L. S.Wong, S. Hossain, A. Ta, J. Edvinsson, D. H. Rivas, and H. Naas, "A very low-power CMOS mixed-signal IC for implantable pacemaker applications," IEEE Journal of Solid-State Circuits, vol. 39, no. 12, pp. 2446–2456, December 2004.

[20] N. S. Shenck and J. A. Paradiso, "Energy scavenging with shoe-mounted piezoelectrics," IEEE Micro, vol. 21, no. 3, pp. 30–42, 2001.

[21] M. J. Ramsay and W. W. Clark, "Piezoelectric energy harvesting for bio MEMS applications," in Proceedings of the SPIE's Annual International Symposium on Smart Structures and Materials, pp. 429–438, 2001.

[22] J. W. Sohn, S. B. Choi, and D. Y. Lee, "An investigation on piezoelectric energy harvesting for MEMS power sources," in Proceedings of the Instiute of Mechanical Engineers, Part C: Journal of Mechanical Engineering Science, vol. 219, pp. 429–436, 2005.

[23] R. Yang, Y. Qin, C. Li, G. Zhu, and Z. Wang, "Converting biomechanical energy into electricity by a muscle-movement driven nanogenerator," Nano Letters, vol. 9, no. 3, pp. 1201–1205, 2009.

[24] S. Meninger, J. O. Mur-Miranda, R. Amirtharajah, A. P. Chandrakasan, and J. H. Lang, "Vibration-to-electric energy conversion," IEEE Transactions on Very Large Scale (VLSI) Systems, vol. 9, no. 1, pp. 64–76, February 2001.

[25] C. Ting, B. S. Calabrese, B. Gary, G. Zhiqiang, Z. Yongchao, K. Hyug-Han, and H. Adam, "A miniature biofuel cell," Journal of the American Chemical Society, vol. 123, no. 35, pp. 8630–8631, September 2001.

[26] N. Mano, F. Mao, and A. Heller, "Characteristics of a miniature compartmentless glucose-O2 biofuel cell and its operation in a living plant," Journal of the American Chemical Society, vol. 125, no. 21, pp. 6588–6594, May 2003.

[27] C. Watkins, B. Shen, and R. Venkatasubramanian, "Low-grade-heat energy harvesting using superlattice thermoelectrics for applications in implantable medical devices and sensors," Proceedings of the International Conference on Thermoelectrics, pp. 250–252, 2005.

[28] S. E. Jo, M. K. Kim, M. S. Kim, and Y. J. Kim, "Flexible thermoelectric generator for human body heat energy harvesting," Electronic Letters, vol. 48, no. 16, August 2012.

[29] G. Jeffrey Snyder, J. R. Lim, Chen-Kuo Huang, and Jean-Pierre Fleurial, "Thermoelectric microdevice fabricated by a MEMS-like electrochemical process," Nature Materials, vol. 2, pp. 528–531, August 2003.

[30] V. Leonov and P. Fiorini, "Thermal matching of a thermoelectric energy scavenger with the ambience," in Proceedings of the European Conference on Thermoelectrics, pp. 129–133, 2007.

[31] Mohammadreza Ashraf and Nasser Masoumi, "A Thermal Energy Harvesting Power Supply with an Internal Startup Circuit for Pacemakers", IEEE Transactions on Very Large Scale (VLSI) Systems, vol. 24, no. 1, pp. 26-37, January 2016.

[32] J. Dickson, “On-chip high-voltage generation in MNOS integrated circuits using an improved voltage multiplier technique,” IEEE Journal on Solid-State Circuits, vol. 11, no. 3, pp. 374–378, June 1976.

[33] C. Fernandez, “Dimensions of the cochlea (guinea pig),” Journal of American Acoustic Society, vol. 24, pp. 519–523, 1952.

[34] S. Ozeri and D. Shmilovitz, “Ultrasonic transcutaneous energy transfer for powering implanted devices,” Ultrasonics, vol. 50, pp. 556–566, 2010.

[35] G. V. Cochran, M. P. Kadaba, and V. R. Palmieri, “External ultrasound can generate microampere direct currents in vivo from implanted piezoelectric materials,” Journal of Orthopaedic Research, vol. 6, no. 1, pp. 145–147, 1988.

[36] W. B. Phillips, B. C. Towe, and P. J. Larson, “An ultrasonically driven piezoelectric neural stimulator,” Engineering in Medicine and Biology Society, vol. 2, pp. 1983–1986, 2003.

[37] P. Shih, W. Weng, W. Shih, Y. Tsai, and P. Chang, “Acoustic polarization for optimized implantable power transimittion,” in Proceedings of the IEEE International Conference on Micro Electro Mechanical Systems (MEMS), pp. 879–882, 2007.

[38] S. Suzuki, T. Katane, H. Saotome, and O. Saito, “A proposal of electric power generating system for implanted medical devices,” IEEE Transactions on Magnetics, vol. 35, no. 5, pp. 3586–3588, September 1999.

[39] G S. Suzuki, T. Katane, H. Saotome, and O. Saito, “Electric power generating system using magnetic coupling for deeply implanted medical electronic devices,” IEEE Transactions on Magnetics, vol. 38, no. 5, pp. 3006–3008, September 2002.

[40] S. Suzuki, M. Ishihara, and Y. Kobayashi, “The improvement of the noninvasive power-supply system using magnetic coupling for medical implants,” IEEE Transactions on Magnetics, vol. 47, no. 10, pp. 2811–2814, October 2011.

[41] T. Kleeburg, J. Loo, N. J. Guilar, E. Fong, and R. Amirtharajah, “Ultralow-voltage circuits for sensor applications powered by free-space optics,” in Proceedings of the IEEE International Solid-State Circuits Conference, pp. 502–504, February 2010.

[42] K. Schuylenbergh and R. Puers, Inductive Powering: Basic Theory and Application to Biomedical Systems. Springer, June 2009.

[43] M. Kiani, U. M. Jow, and M. Ghovanloo, “Design and optimization of a 3-coil inductive link for efficient wireless power transmission,” IEEE Transactions on Biomedical Circuits and Systems, vol. 5, no. 6, December 2011.

[44] H. Shao, C. Y. Tsui, and W. H. Ki, “A single inductor dual input dual output DC-DC converter with hybrid supplies for solar energy harvesting applications,” in Proceedings of the ACM/IEEE International Symposium on Low Power Electronics and Design, pp. 69–74, 2009.

[45] Y. Ahn, H. Nam, and J. Roh, “A 50-mhz fully integrated low-swing buck converter using packaging inductors,” IEEE Transactions on Power Electronics, vol. 27, no. 10, October 2012.

[46] G. Chen, H. Ghaed, R. Haque, M. Wieckowski, Y. Kim, G. Kim, D. Fick, D. Kim, M. Seok, K. Wise, D. Blaauw, and D. Sylvester, “A cubic-millimeter energy-autonomous wireless intraocular pressure monitor,” in Proceedings of the IEEE International Solid-States Circuit Conference, pp. 310–312, February 2011.

[47] W. H. Ki, Y. Lu, F. Su, and C. Y. Tsui, “Design and analysis of on-chip charge pumps for micro-power energy harvesting applications,” in Proceedings of IEEE/IFIP International Conference on VLSI and System-on-Chip, pp. 374–379, 2011.

[48] A. Cabrini, L. Gobbi, and G. Torelli, “Voltage gain analysis of integrated fibonacci-like charge pumps for low power applications,” IEEE Transactions on Circuits and Systems II, vol. 54, no. 11, November 2007.

[49] L. Gobbi, A. Cabrini, and G. Torelli, “A discussion on exponential-gain charge pump,” in Proceedings of the European Conference on Circuit Theory and Design, pp. 615–618, 2007.

[50] X. Li, C. Y. Tsui, and W. H. Ki, “Solar energy harvesting system design using re-configurable charge pump,” in Proceedings of the IEEE Faible Tension Faible Consommation (FTFC), pp. 1–4, Paris, France, June 2012.

[51] M. AdbElFattah, A. N. Mohieldin, A. Emira, A. K. Hussien, and E. Sanchez-Sinencio, "Ultra-low-voltage power management unit for thermal energy harvesting applications," in Proceedings of 2012 IEEE International New Circuits and Systems Conference (NEWCAS), pp. 381–384, June 2012.
[52] S. Dwari, R. Dayal, L. Parsa, and K. N. Salama, "Efficient direct AC-to-DC converters for vibration-based low voltage energy harvesting," in Proceedings of IEEE Annual Conference of Industrial Electronics, pp. 2320–2325, 2008.
[53] Y. K. Ramadass and A. P. Chandrakasan, "A batteryless thermoelectric energy-harvesting interface circuit with 35mV startup voltage," in Proceedings of the IEEE International Solid-State and Circuit Conference, pp. 486–489, February 2010.
[54] T. Esram and P. L. Chapman, "Comparison of photovoltaic array maximum power point tracking techniques," IEEE Transactions on Energy Conversion, vol. 22, no. 2, pp. 439–449, June 2007.
[55] Tengku Norliza Tengku Mohamad, Jahariah Sampe and Dilla Duryha Berhanuddin, "Architecture of Micro Energy Harvesting Using Hybrid Input of RF, Thermal and Vibration for Semi-Active RFID Tag", ENGINEERING JOURNAL, Volume 21 Issue 2, pp. 184-197, Mar. 2017.
[56] Adrian Zurbuchen, Andreas Haeberlin, Alois Pfenniger, Lukas Bereuter, Jakob Schaerer, Frank Jutzi, Christoph Huber, Juerg Fuhrer, and Rolf Vogel, "Towards Batteryless Cardiac Implantable Electronic Devices—The Swiss Way", IEEE Transactions on Biomedical Circuits and Systems, vol. 11, no. 1, pp-78-86, Feb. 2017.
[57] Hao Liu, Qi Shao and Xuelin Fang, "Modeling and Optimization of Class-E Amplifier at Subnominal Condition in a Wireless Power Transfer System for Biomedical Implants", IEEE Transactions on Biomedical Circuits and Systems, vol. 11, no. 1, pp. 35 - 43, Feb. 2017.
[58] Ozgur B. Akan, Oktay Cetinkaya, Caglar Koca, Mustafa Ozger and Student Member, "Internet of Hybrid Energy Harvesting Things", IEEE Internet of Things Journal, vol. PP. no. 99, pp. 1-11, Aug. 2017.
[59] Caiwei Shen, Sixing Xu, Yingxi Xie, Mohan Sanghadasa, Xiaohong Wang and Liwei Lin, "A Review of On-Chip Micro Supercapacitors for Integrated Self-Powering Systems", Journal of Microelectromechanical Systems, vol. 26, no. 5, pp. 949-965, Oct. 2017.
[60] Indrani Dakua1 and Nitin Afzulpurkar, "Piezoelectric Energy Generation and Harvesting at the Nano-Scale: Materials and Devices", Nanomaterials and Nanotechnology, vol. 3, no. 21, pp. 1-16, Aug. 2013.