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New and Emerging Energy Sources for Implantable Wireless Microdevices

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***Abstract*— In this paper, we review new and emerging energy sources for wireless implantable microdevices. After a brief historical background, we review the developments in power sources in the decades following the pioneering works of Zworykin and Mackay in the late 1950s. These include deployment of lithium batteries and inductive powering in the 1970s, which resulted in significant growth and commercialization of implantable medical devices such as cardiac pacemakers and cochlear implants. Recent research in nano-scale materials for energy generation has created intriguing possibilities for next generation implantable power sources in the form of flexible and biodegradable batteries and super-capacitors. In addition, energy harvesting/remote powering from various environmental physical and chemical sources within the body utilizing nano-scale materials can also offer unique possibilities for autonomous implantable micro and nano-scale devices.**

***Index Terms*— Implantable microdevices, nanomaterials, biodegradable electronics, flexible bioelectronics, flexible batteries, flexible super-capacitors, energy scavenging.**

I. Introduction

BIOMEDICAL telemetry has an old and celebrated history dating back to the late 1950's. Wireless transmission of information from inside the body did not become a reality before the invention of the transistor in the late 1940's and its commercialization in 1954 by Texas Instrument. The first use of miniature components to remotely measure a physiological parameter, in this case gastrointestinal motility, was reported almost simultaneously by two groups in 1957 (one group included V. K. Zworykin, the television pioneer, and the other R. S. MacKay who subsequently made significant contributions to the field of biomedical telemetry) [1], [2]. Both devices used a battery-powered single transistor oscillator, made pressure-sensitive by the incorporation of a ferrite core coupled to a flexible membrane, Figure 1-a. It is interesting to note that after several decades of being a laboratory curiosity, a smart capsule which measures pressure, temperature, and pH was finally commercialized and is currently in clinical use (SmartPill®by Given Imaging Ltd.), Figure 1-b. In the decades following the initial pioneering work by Zworykin and Mackay, the field experienced a considerable growth as various investigators employed discrete transistors and integrated circuits to create custom-made systems for remotely measuring a variety of physiological parameters (pressure, flow, temperature, bio-potential, etc.). A comprehensive review of the field up to 1970 can be found in the classical book by MacKay [3]. Even today, this book is a pleasure to browse, describing many ingenious techniques employed in the early years to collect physiological information from hard to access areas (e.g., deep body temperature from a 170-kg Galapagos island tortoise). Several review articles in a special issue of the *IEEE Engineering in Medicine and Biology Magazine* published in the March of 1983 summarized the state of the technology up to that point. The readers interested in a more up-to-date and in-depth discussion of various technical issues can consult the most recent edited reference title, the Handbook of Bio-medical Telemetry [4].

A fundamental requirement for successful operation of implantable systems is an adequate and reliable power supply. Two significant advances in the energy source area in this period were the appearance of lithium batteries and inductive powering [5], [6]. Wilson Greatbatch, the pioneer entrepreneur and inventor of the cardiac pacemaker, was the first to develop and commercialize lithium batteries for implantable medical devices [7], [8]. This resulted in a significant increase in the device lifetime, with current systems lasting up to 10 years without the need for battery replacement. Inductive powering offers an alternative approach for devices that cannot utilize primary batteries, mainly due to a large power consumption (cochlear implants) or limited anatomical space (intra-ocular implants). These two methods have reached a mature status and have dominated wireless implant energy source technology for the past four decades. Recent interest in flexible and biodegradable devices for implantable and ingestible electronics has created new challenges as related to on-board or remote power sources that cannot be easily met with existing technologies. These systems require flexible (or sometimes stretchable) batteries and super-capacitors that can be integrated with biosensors and other electronic components in a biocompatible low-profile package, Figure 2a. In addition, energy harvesting/remote powering from various physical and chemical sources can also offer unique possibilities for such systems, Figure 2b. In the following sections, we will discuss some of the more important recent developments in this area, in particular as related to flexible/biodegradable batteries, super-capacitors, energy harvesting, and remote powering.

II. Flexible and Biodegradable Batteries and Super-capacitors

Due to their high energy density, convenience, portability, and reliability (long shelf life), batteries are the *de facto* standard for powering today's portable electronics and implantable

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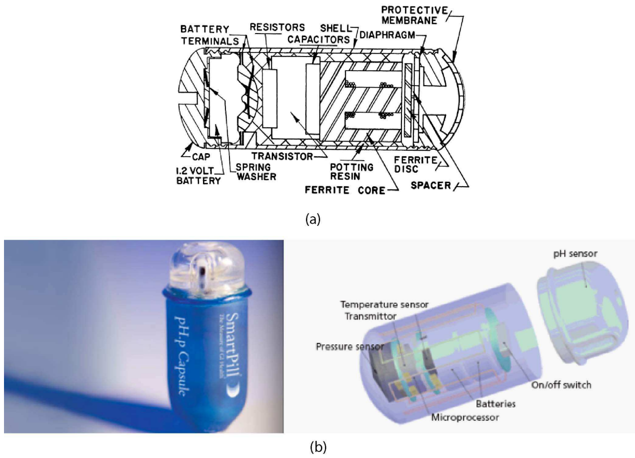

Fig. 1. (a) One of the first wireless pressure transmitters in the shape of a swallowable capsule, showing various components including a variable inductance pressure sensing element [1], (b) SmartPill® by Given Imaging Ltd. measuring pressure, pH and temperature in the GI tract

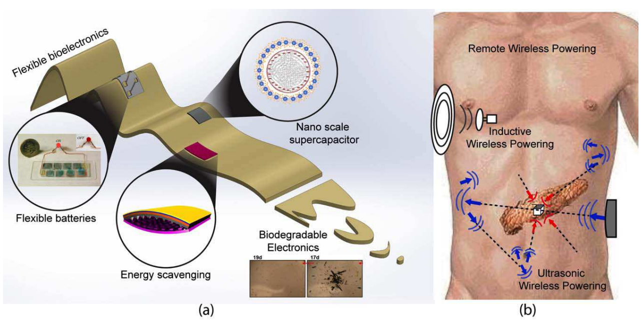

Fig. 2. Next generation of energy sources for flexible biomedical systems; (a) flexible batteries, energy scavenging, super-capacitors, and biodegradable batteries; (b) remote wireless powering (e.g. ultrasonic, inductive)

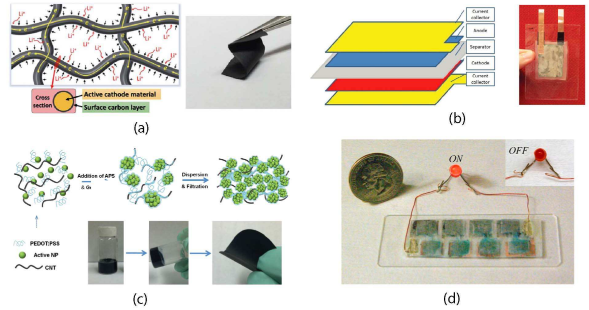

Fig. 3. Flexible batteries, (a) $LiMnPO_4$ lithium-ion [9], (b) Zinc-MWCNT alkaline [10], (c) TiO_2 (a high-rate capability) or SiNP (high cycling stability) with CNT and PEDOT:PSS [11], (d) flexible paper based battery [12]

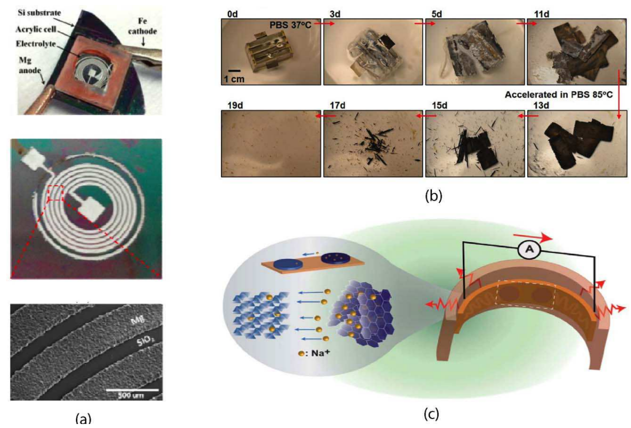

Fig. 4. Biodegradable batteries: (a) MEMS-enabled Mg-$MgCl_2$-Fe battery [19], (b) dissolution of Mg-Mo battery in PBS solution [15], (c) biodegradable sodium-ion battery: active carbon and MnO_2 electrode on PGScin/AgNW [20]

devices. Advances in battery technology over the past few decades have been consistently increasing the energy density of batteries while reducing their size. Their established reliability guarantees their importance among future implantable systems. However, with the advent of modern light-weight, flexible, stretchable, and biodegradable microsystems, it has become necessary to adapt battery technologies in similar form factors. Over the past few years, researchers have been developing a variety of flexible and biodegradable batteries while investigating techniques for increasing their energy density. In this section, we describe some of the latest battery technologies which can power next-generation implantable systems.

A. Flexible batteries

Aside from the primary goal of improving battery performance, there has been a strong push to impart flexibility, biocompatibility, and biodegradability to batteries used in biomedical microsystems. Applications for such batteries include ingestible electronics, key-hole deployed bioelectronics, and many cases in which the system has to conform to the curvature of an organ. An obvious implementation is to incorporate traditional battery chemistries onto flexible biocompatible substrates such as parylene and polyimide.

Recently, Lu et al. reported on a flexible lithium-ion battery created by incorporating magnesium (Mg-) into $LiMnPO_4$ to form a flexible nanofibrous cathode [9], Figure 3-a. The battery exhibited high flexibility as well as 50 % increased energy density (135 mAh/g). A similar approach used a multi-walled carbon nanotube composite (10 %) electrode with a polyvinyl alcohol (PVA)-ploy (acrylic acid) (PAA) copolymer separator to achieve flexibility with up to 236 mAh/g of energy density, Figure 3-b [10]. There have also been some recent efforts in utilizing conductive polymers as the electrodes; these include hybrid carbon nanotube (CNT) hydrogel composites as well as combinations of either TiO_2 or Si nanoparticle with CNTs and PEDOT:PSS, Figure 3-c [11].

As an alternative to these batteries which require expensive/complex CNT processing technology, Chitnis et al. developed a low-cost flexible paper-based battery using a laser ablation technique, Figure 3-d [12]. Here, Cu and Zn were used as electrodes, with a water-triggered polyacrylamide hydrogel as the salt bridge. The materials were patterned on laser-machined wax paper, which became hydrophilic at selected regions by the laser treatment. This battery was capable of generating 3 V from a four-cell arrangement, with up to 625 μW for 30 minutes after activation. Overall,

TABLE I
SELECTED RECENT FLEXIBLE AND BIODEGRADABLE BATTERIES.

Ref.	Type	Energy density	Electrodes
[9]	Flexible Li-ion	135 mAh/g	LiMnPO4
[10]	Flexible Alkaline	236 mAh/g	CNT/PVA-PAA
[11]	Flexible	227 mAh/g	CNT/PEDOT:PSS
[12]	Flexible paper-based	625 mW/m^2	Copper/Hydrogel/Zinc
[15]	Biodegradable	276 mAh/g	Mg/PDMS/Mo, W, or Fe
[19]	Biodegradable	0.7 mAh	Mg/MgCl2/Fe
[20]	Biodegradable	9.58 mAh/g	Ac/(PGScin/AgNW)/MnO2

these recent flexible batteries demonstrate their potential for powering implantable microsystems; however, future flexible batteries should reduce cost and extend lifetime for long-term applications.

B. Biodegradable batteries

Biodegradability (i.e., the ability to benignly degrade at the end of the life cycle without toxic byproducts/residues) is another important property of emerging battery technologies, allowing for transient implantable devices [13]–[15]. Well-known biodegradable metals such as magnesium (Mg), iron (Fe), and zinc (Zn), which are routinely used in biodegradable stents and bone screws [14], have played a major role in the recent developments of biodegradable batteries [15]–[19]. Tsang et al., for example, reported a MEMS-enabled biodegradable battery which uses Mg as the anode, Fe as the cathodes, and $MgCl_2$ as the electrolyte [19]. Its fabrication involved traditional microfabrication techniques to allow for miniaturization, Figure 4-a. To overcome Mg's natural degradability in aqueous solutions, the Mg was coated with other biodegradable passivation layers such as polycaprolactone (PCL) and poly(glycerol-sebacate) (PGS). This enabled a storage capacity of up to 0.7 mAh with 26 μW of power [19]. Similarly, Yin et al. developed batteries with Mg as the anode and Fe, W, or Mo as the cathode. Among the Mg–X batteries the Mg-Mo combination produced 2.4 mAh from 1 cm^2 active area (50 μm thick Mg, 8 μm thick Mo). Accelerated tests in an 85 °C phosphate buffered solution demonstrated reliable performance of a stacked cell Mg-Mo battery over the course of 19 days, Figure 4-b [15]. Other researchers have focused on combining flexibility and biodegradability; in one example, a composition of poly(glycerol-co-sebacate)-cinnamate and silver nanowires was used as the electrolyte to achieve flexibility with a sodium ion electrochemical cell, Figure 4-c [20]. Here, the anode and cathode were fabricated using active carbon and λ-manganese oxide, both of which are edible (active carbon is used in metal detoxification therapies, and manganese oxide is a recommended daily health supplement). This battery could generate 0.6 V and 5–20 μA.

Table 1 summarizes some recent flexible and biodegradable batteries. As these works demonstrate, biodegradable batteries show a tangible power output that can be utilized in future implantable devices. However, controlling and limiting Mg's natural degradation in aqueous solution remains a challenge.

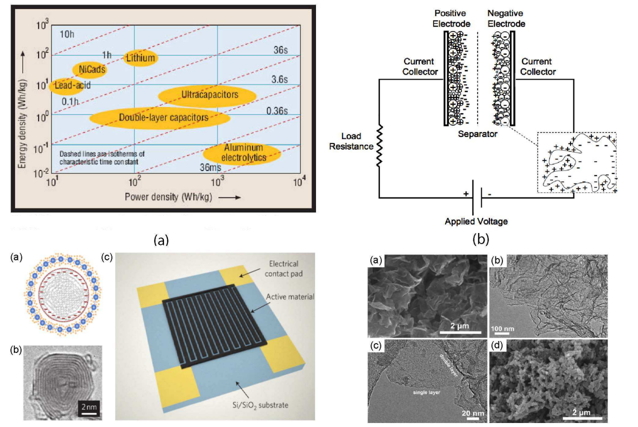

Fig. 5. (a) Power vs. energy characteristics of batteries and super capacitors [22]; (b) principle of a single-cell EDLS [23]; (c) design of the interdigitated micro-super-capacitor with OLC (onion-like-carbon) electrode; (a) cross-section of OLC capacitor, (b) TEM image of OLC, and (c) schematic of the micro device [35]; (d) SEM image of G-mPANI (a-c) and pristine PANI [36]

C. Flexible super-capacitors

Super-capacitors offer an alternative electrochemical energy source that can be used for storing a large amount of charge in a small volume. Although the concept of electrochemical capacitors is not new, the use of nanomaterials for electrodes has created unique opportunities for considerable miniaturization and enhanced power density. Typically, super-capacitors have higher power densities as compared to batteries but cannot store energy for a long time as the batteries do, Figure 5-a [21] [22]. However, the recent incorporation of high-surface-area carbon-based materials has enabled the development of high-performance super-capacitors which can sustain higher energy densities and allow multi-layer designs. This new technology opens the doors for super-capacitors as energy sources for future and emerging biomedical microsystems [22].

Super-capacitors can be categorized into electrostatic double-layer super-capacitors (EDLS) and Faradaic super-capacitors (FS) (i.e., pseudo-capacitors). EDLSs are associated with accumulated electrostatic charges at the interface of an electrode and an electrolyte, Figure 5-b [23]. While charging, the electrons travel from the cathode to the anode through the load; meanwhile within the electrolyte, cations move towards cathode and anions towards the anode. The reverse process takes place during discharge. In FSs, however, charging/discharging is based on rapid faradaic reactions (i.e., redox reactions), similar to those in batteries. Compared to EDLSs, FSs have higher capacitances and energy densities but lower power densities due to the slower nature of faradaic processes. Table 2 summarizes various recent flexible super-capacitors.

1) Electrostatic double layer super-capacitors (EDLS): Figure 5-b depicts the operation principle of electrostatic double layer super-capacitors. The electrostatic charge created at the interface of the electrodes and its potential depends on

TABLE II

SELECTED RECENT FLEXIBLE SUPER-CAPACITORS.

Ref.	Type	Capacitance or Energy density	Electrodes
[26]	ELDC	100 F/g (@ 22.5 mA/cm^2)	Activated carbon aerogel
[28]	ELDC	100 F/g (@ 1 A/g)	Crumpled graphene balls
[29]	ELDC	50~54 F/g	SW CNT
[30]	ELDC	115.83 F/g, 48.86 Wh/kg	Paper / SW CNT
[31]	ELDC	33 F/g, 250 kW/kg	Paper / SW CNT
[32]	ELDC	0.48 F/cm^2	Fabric / SW CNT ink
[33]	ELDC	11 - 19.2 F/g	Elastic fiber / CNT sheet
[35]	FS	1.3F/cm^3, 1kW/cm^3	Nanostructured carbon onions
[36]	FS	749 F/g (@ 0.5 A/g)	G-mPANI

TABLE III

SELECTED RECENT PIEZO- AND TRIBO-ELECTRIC ENERGY SCAVENGERS

.Ref.	Type	Power	Materials
[40]	Piezoelectric	1.189 μW	PMN-PT
[41]	Piezoelectric	1.2 μW/cm2	PZT with Ti/Pt and Cr/Au
[42]	Triboelectric	1.5W	EFP with Au
[43]	Triboelectric	128 mW/cm3	Kaptopn/PDMS/PTFE
[44]	Triboelectric	3.68 mW	Kaptopn/Al/PTFE
[45]	Triboelectric	3.2 μA/cm2	PTFE/ TiO2 nanowire
[46]	Triboelectric	100 μW	PTFE/Al
[47]	Triboelectric	0.13 μA/cm2	(PET/ITO)/PDMS

the surface area of the material. Therefore, electrodes featuring a rough nanoscale surface morphology provide a large surface area, allowing for high power densities. Recent investigations have focused on improving such electrodes by using various forms of carbon nanostructures, utilizing their remarkable electrical and mechanical properties [24]; these include activated carbon, carbon aerogels, carbon nanotubes, porous carbons, and carbon nanofibers [23], [25]–[27]. These materials can be engineered to optimize the surface area for higher capacitances and easier diffusion of electrolyte ions. To further increase the spatial efficiency of super-capacitors, various researchers are exploring 2.5- or 3-dimensional super-capacitor structures. One particular architecture is reported by Luo et al., which consists of super-capacitor sheets that are crumpled into balls [28]. These structures are capable of delivering much higher specific capacitance and exhibit an improved charge/discharge performance since the reduced density of the crumpled ball (compared to plain sheets) provides increased free volume, which favors ionic flow and electron transport. In a similar approach, Yu et al. reported the fabrication of stretchable ELDC super-capacitors by laminating a CNT macrofilm to a pre-strained PDMS substrate; this led to the spontaneous formation of a periodically-buckled pattern of CNT films, which served as both stretchable electrical connection and high surface area for the ELDC super-capacitor [29].

In addition to polymers, paper and fabric are also being investigated as super-capacitor substrates due to their low cost, high porosity, and stretchability (in the case of certain fabrics). Processing these materials typically entails loading the porous matrix of paper or fabric with CNTs using techniques such as solution immersion [30] and inkjet printing [31]. For instance, Hu et al. developed a highly conductive and stretchable super-capacitor by impregnating a stretchable fabric with SWCNT [32]. The resulting conductive textile featured resistance as low as 1 Ω/□ with capacitances of up to 0.48 F/cm^2 and excellent performance for strains as high as 230 %. Taking the fabric concept a step further, Yang et al. created a super-capacitor on a fiber by sequentially wrapping aligned CNT sheets and H_3PO_4–poly(vinyl alcohol) (PVA) gel electrolyte on an elastic fiber. The super-capacitor maintained specific capacitance of 18 F/g with minimal degradation after stretch by 75 % for 100 cycles [33]. The remaining challenge for future EDLSs is to further increase their energy density; this can best be achieved by implementing electrodes with: 1) high specific surface area; 2) suitable pore size, distribution, and length for diffusing electrolyte ions at higher rates; and 3) low internal resistance for efficient charge transfer.

2) Faradaic super-capacitors (FS): In contrast to EDLSs, FSs mange charge via the transfer of charge between electrode and electrolyte. Typical electrodes for FSs include conductive polymers, metal oxides, and composite polymers of metal and carbon-based materials to increase reduction-oxidation reactions. Conductive polymers feature relatively high capacitance and conductivity as well as low cost, compared to carbon-based materials; however, their subpar charge-transfer efficiency and instability under mechanical stress have limited their development [34]. Instead, researchers have focused on developing composite electrodes by integrating carbon-based materials with conductive polymers to enable both physical and chemical charge storage mechanisms [35], [36]. Pech et al., for example, reported micro-super-capacitors with power density comparable to that of electrolytic capacitors, three orders of magnitude higher than conventional super-capacitors. The devices were fabricated by electrophoretic deposition of a layer (several microns) of nanostructured carbon onion-like structures [35], Figure 5-c; they report very high scan rates (on par with electrolytic capacitors) while sustaining charges four orders of magnitude higher than that of the electrolytic capacitors [35]. Another recent example is that of Wang et al., which consists of a hybrid mesoporous polyaniline (PANI, a common conductive polymer) film on an ultra-thin graphene nano-sheet (G-mPANI). This structure was able to achieve a higher specific capacitance (749 F/g vs. 315 F/g), Figure 5-d, while remaining electrochemically stable for up to 1000 cycles, which is not possible with pristine PANI [36]. These examples show a promising future for FSs utilizing nanomaterials; thus, their further development will depend heavily on discovering/composing new nanostructured materials such as nano-aerogels, nano-plates, and nano-spheres.

TABLE IV
SELECTED RECENT REMOTE WIRELESS POWERING

Ref.	Type	Efficiency	Power	Operating frequency	Receiver size	Transmitter power/size	Distance
[53]	Multi-coil Inductive	82%	-	0.7 MHz	22 mm Ø	64 mm Ø	3.2 cm
[54]	Mid-field inductive	-	0.2 mW	1.6 GHz	2 mm Ø	500 mW	5 cm
[58]	Ultrasonic	-	0.816 mW	2.15 MHz	1 x 5 x 1 mm^3	50 x 50 x 1 mm^3	10 cm
[59]	Ultrasonic	-	0.11 mW	1 MHz	1.33 mm Ø	26 mm Ø	12cm
[60]	Acoustic	-	1.098 μW	350 Hz	20 x 2 x 0.38 mm^3	11.665 W	15 cm

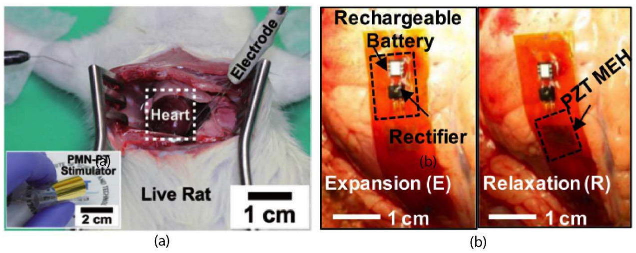

Fig. 6. (a) Piezoelectric energy scavenger based on PMN-PT tested on live rat [40], (b) *in vivo* demonstration of PZT MEH paired with rechargeable battery on heart [45]

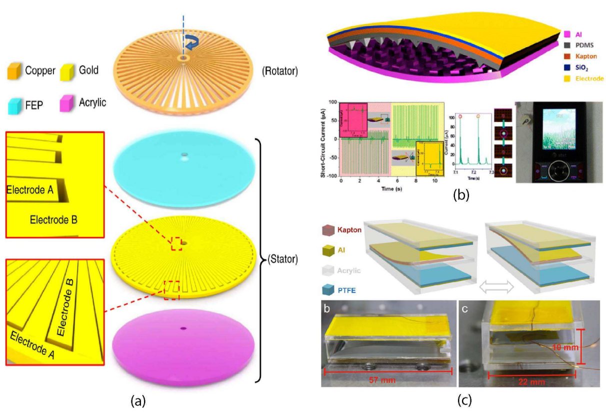

Fig. 7. Triboelectric generator designs: (a) radial-driven using gold and FEP [42], (b) nanomaterials using PDMS and Kapton® [43], and (c) flow-driven using Kapton® and PTFE [44]

III. Energy Scavenging

Energy scavenging or energy harvesting is a very attractive method for powering implantable microdevices since its operating lifetime is theoretically unlimited (limited only by device failure or biological complications). The term "energy scavenging" often refers to methods of energy transduction from various sources which are not primarily intended to power the device in the first place. It is important to differentiate between scavenging and remote powering, since they pose different limitations. As mentioned above, energy scavenging refers to techniques which collect background energy which already exists in a typical operating environment and which would otherwise remain unused. As such, the burden of receiving sufficient power for operating lies solely on the design of the device itself, which must take into account any possible fluctuations in the environment. This differs fundamentally from remote powering, which requires, or at least permits, a dedicated remote energy source; here, the power source can be tuned to improve reception at the device or to compensate for transmission/ reception inefficiencies on the device side. Both of these schemes have their own merits and are useful among implantable devices. In this paper, we will focus on energy scavenging techniques which rely on kinetic energy from the within and without the body. Table 3 summarizes some of the most recent developments in each type of scavenger.

A. Piezoelectric energy scavengers

The human body is a great source of kinetic energy in the forms of voluntary (skeletal) and autonomic (cardiovascular, respiratory, and gastrointestinal) muscular movements, which can be scavenged by appropriate techniques. One of these is piezoelectricity, by which mechanical strain imposed on structures of certain electrically-polled materials generates an electrical potential. The concept has been extensively used with various materials ever since its discovery by Jacque and Pierre Curie in 1880 [37]. Some of the most common materials today which exhibit this phenomenon include lead zirconate titanate (PZT), barium titanate ($BaTiO_3$), polyvinylidenefluoride (PVDF), and zinc oxide (ZnO). Their differences in mechanical properties, in particular brittle PZT vs. flexible PVDF, provide a broad range of options for creating piezoelectric energy scavenging devices.

Early examples of piezoelectric energy scavenging from body motion include the shoe energy harvesters of the MIT Media Lab, which produced up to 8.3 mW from human walking [38]. Since then, piezoelectric powered devices have been developed with applications ranging from blood pressure monitoring to orthopedic implants and knee replacements [39]. More recently, the technique has been adopted to scavenge energy from the motion of internal organs, including heart [13], [40], [41]. This is accomplished by using thin, flexible piezoelectric films of high-efficiency materials such as $Pb(Mg_{1/3}Nb_{2/3})O_3$-$xPbTiO_3$ (PMN-PT) [40]; these can generate up to 145 μA and 8.2 V, sufficient for stimulation of rat cardiac muscle, which normally requires 2.7 μJ, Figure 6-a. Similarly, Dagdeviren et al. developed devices made with custom PZT ribbons coupled with Ti/Pt and Cr/Au electrodes for use in bovine *in vivo* experiments, Figure 6-b; the devices were able to generate up to 1.2 μW/cm^2 [13], [41].

As these examples show, piezoelectric energy scavenging can be successfully applied to implantable microsystems, but its applications are currently limited by their overall low energy density (compared to batteries). Hence, as Hwang et al. suggest [40] this technology still needs further development, in particular with respect to higher electromechanical coefficient piezoelectric materials that are biocompatible.

B. Triboelectric energy scavengers

Triboelectricity is another phenomenon which can scavenge background motion from the body. The effect occurs when two different dielectrics come into frictional contact, thus generating charges that can be separated via backside contacts to provide an electrical voltage. Wang. et al. introduced the first triboelectric generator in several publications [42]–[47]. One design featured a radial-arrayed rotary design with a gold rotator and fluorinated ethylene propylene (FEP) as an electrification material, Figure 7-a. Using 3 kHz rotational motion, the scavengers were able to induce AC power with open circuit voltage of 850 V and short circuit current of $\sim$3 mA or 1.5 W with a 0.8 MΩ load [42]. Subsequent miniaturized designs used Kapton tape with PDMS or with polytetrafluoroethylene (PTFE) film as the electrification materials to yielded 230–400 V open circuit voltage with efficiencies of 10–39 % (for the PDMS version), Figure 7-b,c [43], [44]. All of these designs offer very high voltages which can be used for powering devices and systems that are primarily capacitive (the energy densities are sufficient to power many small devices which may require as little as a few mW power). Such triboelectric generation schemes are, therefore, a very promising technology that may one day replace current batteries. However, for implantable systems, triboelectric generators will need to be further optimized for lower frequency and amplitude mechanical motions.

IV. Remote Wireless Powering

Remote wireless powering techniques offer a convenient compromise between batteries and energy-scavenging devices, achieving an effective balance between power density and component autonomy. Such techniques eliminate the need for transcutaneous wiring or invasive medical procedures (e.g., for battery replacement), which are common among battery-operated systems; this is typically at the expense of diminished power density. Nevertheless, their available power density remains sufficiently large for powering implantable medical devices and is superior to that of emerging energy scavenging systems. Hence, despite their reduced autonomy (compared to energy scavenging systems) and decreased power density (compared to batteries), remote wireless powering techniques are gaining much popularity due to their reliable and convenient operation. Their various implementations can be classified into one of two categories: inductive and acoustic/ultrasonic powering (although some limited work have also been performed on using near infrared to power implantable devices, their limited penetration depth constrains them to subcutaneous and shallow implants [48]). The following subsections describe the typical operation of each type as well as their merits and drawback; the techniques are also summarized in Table 4.

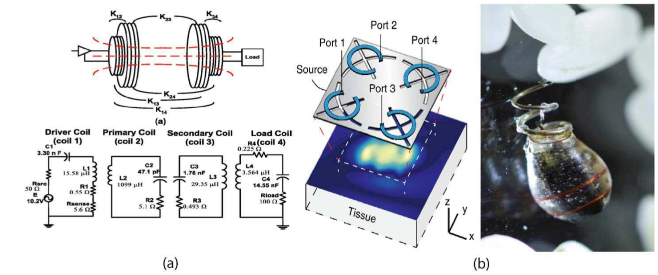

Fig. 8. (a) Schematic of a multi coil inductive powering and electrical model of power transfer circuit [53], (b) primary coil and implant receiver ($\sim$2mm) of mid-field inductive powering [54]

A. Inductive powering

As discussed in Section 1, the most common technique for remote wireless powering of implantable medical devices is inductive coupling, i.e., near-field magnetic power transmission. Its basic principle is based on Faraday's law, by which a varying magnetic field created by an alternating current in a primary coil concatenates with a nearby secondary coil to induce an alternating current in the secondary coil. The overall performance of this technique is dictated by three component parameters: 1) self-inductance of primary and secondary coils, 2) their mutual inductance (i.e., coupling coefficient), and 3) their quality factor (Q). The compact form factor for emerging micro and nano-scale devices significantly limits these parameters. This in turn significantly reduces the overall energy transmission efficiency. For instance, self-inductance and quality factor, which positively correlates with coil size, are restricted due to the small size of the implanted coil; at the same time, the coupling coefficient between the transmitter and receiver coils drops exponentially with their separation and dimensional mismatch. The ability to tune the resonance for wideband receiving (usually by incorporating additional capacitive components) may also be limited by the packaging and spatial confinement. Despite such volumetric constraints, wireless inductive powering remains a very attractive method for remote powering among those implantable devices which can support sufficiently large coils (e.g., > 1 cm in diameter) and those which are positioned near to outside of body.

The fundamentals of induction powering enabled the development of various early wireless devices, including cochlear implants [49] and implantable neuromuscular stimulators [50]–[52]. More recently, researchers have enhanced the capabilities of inductive powering by increasing the power density and operating range [53], [54]. This has been accomplished by the incorporation of multi-layer coils, coil geometry optimization [53], and focused/adaptive electromagnetic energy transport via propagating modes in tissue [54]. RamRakhyani et al., for instance developed a high-Q four-coil configuration (two coils for primary and another two for receiver, 22 mm-diameter, 2.5 mm-thickness) to achieve power efficiency up to

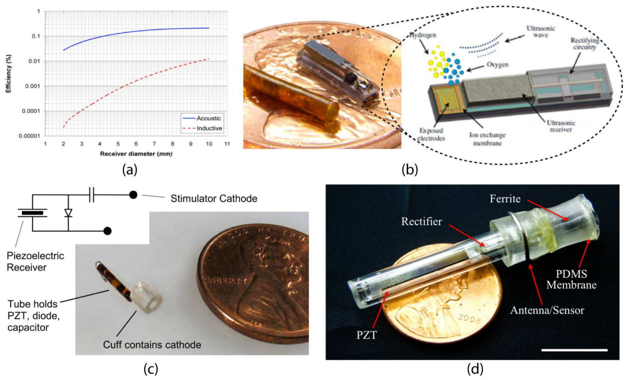

Fig. 9. (a) Efficiency of inductive and ultrasonic powering as a function of receiver diameter at 10 cm [55], (b) ultrasonically powered micro oxygen generator for treating tumor hypoxia [58], (c) miniature ultrasonically powered nerve-cuff stimulator [59], and (d) music powered implantable pressure sensing transponder [60]

72% at a distance of 3.2 cm (resonant frequency at 0.7 MHz), Figure 8-a [53]. Meanwhile, Ho et al. developed a two-coil system with a modified primary coil [54]; a non-typical metal pattern allowing mid-field operation of time-varying current through the coil, which increased operative distance to more than 5 cm (suitable for cardiac applications). As a result, mid-field power transmission enabled the miniaturization of implant receiver down to 2 mm, Figure 8-b [54]. As these examples show, inductive powering is still a promising solution for future mm-scale implantable medical devices. There are, however, still some practical issues that need to be addressed; these include coil misalignment, tissue heating, and difficulties in fabricating sub-mm coils with adequate inductance and quality factor.

B. Acoustic/ultrasonic powering

Acoustic/ultrasonic powering is another potential energy source for emerging micro and nano-scale implantable devices that can overcome certain limitations of inductive powering (i.e., directionality and penetration depth). Unlike electromagnetic signals, acoustic waves are not sensitive to alignment; rather, they can be omni-directional and can be effective and efficient at transferring power at longer distances (up to 10 cm) for [55]. The omni-directionality of ultrasonic powering arises from the reflection of waves at the tissue/air interfaces due to the large impedance mismatch (reflection coefficient of 0.99 between soft tissue and air [56]), Figure 2-b. In addition, since it operates at frequencies below 10 MHz, tissue absorption is not excessive, resulting in penetration depths of 10-20 cm; however, one must be cognizant of the fact that at too low of a frequency (low kHz range) the receiver dimensions become too large (trade-off between implant size and penetration depth). It is interesting to note that at diagnostic ultrasound frequencies (1-20 MHz) one can achieve a reasonable penetration depth and receiver dimensions; this is advantageous since ultrasound instrumentation in the market are usually designed for this frequency band, making them easily accessible in most clinical settings. Another important consideration in ultrasonic powering is whether the implant is in the ultrasonic transmitter's near or far field region (demarcation border for this is given by $D^2/4\lambda$, where D is the diameter of the transmitter and λ is the wavelength, for example, near field region of 3 cm-diameter ultrasonic transducer operating at 1 MHz is within 15 cm). In the near-field region, ultrasonic wave intensity exhibits a series of maxima and minima along the axial direction, whereas in the far-field region the ultrasonic wave intensity gradually decreases (typical $1/R^2$ dependence). Due to its larger energy density, it is advantageous to operate the device within the near-field region. The tissue-air boundary reflections to a large extent alleviate the power drop in minima regions. Denisov and Yeatman recently compared the efficiency of inductive and ultrasonic powering methods for different receiver sizes. According to their study, as the receiver size becomes smaller, the efficiency of inductive powering significantly decreases with smaller receiver sizes, whereas ultrasonic powering demonstrates a much better performance (0.02×10^{-3}% vs. 0.02% for a 2 mm receiver at 10 cm), Figure 9-a [55].

The first remote wireless power transmission using acoustic waves was that of Cochran et al. in 1985 [57], in which an external ultrasonic generator excited an internal fixed plate with an attached piezoelectric element to produce electric current (used for treating bone fractures). Since then, researchers have developed systems with miniaturized components suitable for implantation using minimally invasive procedures. One recent example is an ultrasonically powered micro oxygen generator (IMOG) developed by Maleki et al. for *in-situ* generation of oxygen (to enhance radio-sensitivity of hypoxic tumors prior to radiation therapy), Figure 9-b [58]. The IMOG consisted of a lead zirconate titanate (PZT) receiver, a rectifier, electrodes, and an ion exchange membrane. An external ultrasonic transducer was used to energize the implant (2x2x8 mm^3), which in turn created a DC voltage for *in-situ* water electrolysis (received power of 0.816 mW). Another example of using ultrasound to power implantable devices is that of Larson and Towe who made a miniature cuff stimulator powered by a PZT receiver (received power of 0.11 mW) [59], Figure 9-c. A hybrid device (acoustic signal to power the implant and RF to send the measured signal back to the outside) operating at much lower audio acoustic frequencies was reported by Kim et al. [60]. The device relied on lower harmonics of a music signal to excite a PZT cantilever into resonant vibrations and power a passive LC tank in which the inductor was also a pressure sensor. At intervals in which no harmonics was present the filter capacitor was discharged through the inductor resulting in a pulsed RF signal frequency of which was a measure of pressure, Figure 9-d.

Due to its higher efficiency and omni-directionality at powering mm and sub-mm scale devices at larger distances, acoustic/ultrasonic method is a strong candidate for remote powering of next generation implantable microsystems. Its use in biomedical area can be expected to increase rapidly in the coming years, especially given the availability of ultrasonic transducers in medical settings. However, in order to ensure safety, thermal and cavitational effects of ultrasound must be avoided (staying within FDA limits, i.e., exposure intensity limit of 720 mW/cm^2 should alleviate these concerns [61]). Also, among the remaining challenges is applying such techniques for lung and brain implants, both of which present a barrier to acoustic transmission.

V. Conclusions

Next generation implantable microsystems require power sources that can be fundamentally different from the ones currently used in pacemakers and cochlear implants. Such power sources are expected to be increasingly smaller, flexible (for intimate contact with human tissue), and in some cases even biodegradable. These requirements have stimulated the research on novel flexible and biodegradable batteries and super-capacitor. Additionally, energy scavenging using nanoscale materials and composites promise systems that can last for many years and possibly decades using small environmental vibrations. Many of these technologies are still in their infancy and require further development as related to increased power densities, smaller form factors, and packaging.

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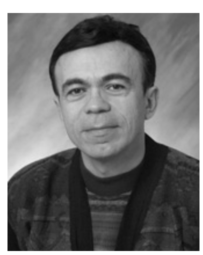

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