Chapter 6

Recent Advances on Implantable Wireless Sensor Networks

Hugo Dinis and Paulo M. Mendes

Additional information is available at the end of the chapter

http://dx.doi.org/10.5772/intechopen.70180

Abstract

Implantable electronic devices are undergoing a miniaturization age, becoming more efficient and yet more powerful as well. Biomedical sensors are used to monitor a multitude of physiological parameters, such as glucose levels, blood pressure and neural activity. A group of sensors working together in the human body is the main component of a body area network, which is a wireless sensor network applied to the human body. In this chapter, applications of wireless biomedical sensors are presented, along with state-of-the-art communication and powering mechanisms of these devices. Furthermore, recent integration methods that allow the sensors to become smaller and more suitable for implantation are summarized. For individual sensors to become a body area network (BAN), they must form a network and work together. Issues that must be addressed when developing these networks are detailed and, finally, mobility methods for implanted sensors are presented.

Keywords: implantable medical devices, sensors, communication, powering, mobility

1. Introduction

Implantable electronic devices are becoming ever smaller and more efficient, which drives their suitability for many new applications to levels never seen before. Examples of such devices are implantable chemical sensors [1], glucose and oxygen sensors for diabetics [2], neural implants [3, 4] and cochlear implants [5]. The constant evolution of these devices is paving the way for their large-scale use in the human body. It is not hard to imagine a cluster of sensors gathering data from several different locations in the human body, giving birth to what is referred to as a body area network (BAN). A BAN is a wireless sensor network (WSN) that consists of devices operating in, on or close to the human body [6]. It is composed of a small number of devices, equipped with biomedical sensors and wireless communications [7].

BANs and WSNs share many of the same challenges, but BANs pose a particular set of problems to be addressed:

- Size constraints imposed by the limited available space inside the body.
- Lossy materials surrounding the implant that heavily attenuate electromagnetic signals used for communication, degrading the quality of the link.
- Biocompatibility concerns.
- High power efficiency requirement, due to the limited available energy, whether from batteries or other powering methods [8].
- Communications must be reliable, as they can be conveying urgent information about life-threatening conditions of the individual.
- Data safety must be guaranteed during communication, as personal and confidential medical information will be transmitted.

In this chapter, implantable sensors and sensor networks will be studied, starting with examples of their applications in the biomedical field and state-of-the-art sensors. Methods that allow the devices to communicate with the outside world will be reviewed and discussed, as the sensors must transmit data to an external reader, so that it can be accessed by the individual or medical personnel. Alternative powering methods that allow the device to have smaller form factors than those possible with batteries will be presented. Advances in material science and fabrication techniques lead to the integration of electronics with smart materials, thus birthing a new generation of devices that are more suitable than ever for implantation. These integration efforts will be presented in this chapter. As the main interest of this publication is wireless sensor networks, networking issues faced by in-body sensors will be presented. Finally, there have been reports of self-propelled devices, and the possibility of having sensors capable of moving to different places in the human body inspired the authors to present some of these propulsion methods.

2. Applications

The biomedical field has a vast range of devices and techniques capable of aiding medical staff to diagnose, manage and treat diseases. This section focuses on sensors, which are responsible for gathering data on a given biomedical signal and relaying them to physicians. Since the scope of this book is wireless sensor networks, only sensors with wireless capabilities will be considered and presented. Examples of applications of sensors in the medical field will be presented, along with proposed devices.

2.1. Intraocular pressure

Intraocular pressure (IOP) monitoring is an important tool for medical staff to diagnose and control glaucoma. This disease is the second most common cause of blindness, and it is predicted to affect around 80 million people by 2020 [9]. Different approaches for measuring IOP are possible and range from non-invasive devices, such as contact lenses [10, 11] that measure the deformation of the cornea curvature due to the extra pressure, to invasive, implantable sensors [12–16] that directly measure the IOP inside the eye. The device presented in Ref. [12] is considered by the authors of this chapter to be the state of the art of IOPs. It is a 1.5 mm^2 sensor with wireless communication capabilities and a power requirement of only 7 μW, which is satisfied by solar energy harvesting.

2.2. Neural activity

Neural activity provides useful data for a number of different applications. It can be used, for example, to diagnose neural dysfunctions, such as epilepsy [17], to control prosthetic limbs through what is called a brain-machine interface [18], and behavioural studies [19]. Neuron action potentials can be measured from deep brain tissues through implantable needles [20], or from the surface of the cortex [17, 21–23], which reduces cortical scarring and allows for chronical and stable measurements [21]. There are even examples of devices used to record the electrical activity of neuron in the peripheral nervous system [18]. The state of the art in implantable neural sensors is considered to be the device presented in Ref. [21]. It is a radio frequency (RF) wirelessly powered, 42.25 mm^2, 64-channel sensor with a 1 Mbps data rate, consuming only 225 μW. A smaller device consuming 120 μW is available in Ref. [18], but its single-channel topology puts it at a disadvantage.

2.3. Bladder pressure

Bladder pressure monitoring is an important tool for the diagnosis of bladder dysfunctions. As some symptoms may only be induced in normal daily activities, such as walking, they cannot be registered in an acute measurement at the hospital. Implantable, chronic reading is necessary, preferably with no discomfort to the patient. Examples of such devices are presented in Ref. [24–27]. In Ref. [27], a 40 mm^2 sensor consuming 16 μW, with sound wave power transfer capabilities and LC resonance-based communication, is presented and considered to be the state of the art in this field.

2.4. Glucose

Glucose monitoring is traditionally done by the patient himself, usually by pricking the fingertip and drawing a small blood sample. Unfortunately, this method is not comfortable for the patient and is only capable of getting a measurement in given points of time. Implantable alternatives are being researched and have already been presented [28, 29]. These allow for continuous glucose-level monitoring and can be used to trigger alarms or even to automatically control implantable insulin pumps, thus improving the patient's quality of life. In Ref. [28], a needle implantable $0.5 \times 0.5 \times 5$ mm^3 wireless sensor with light powering and communication is presented.

2.5. Blood pressure

High blood pressure is the main cause for morbidity and mortality worldwide [30]. It is responsible for a higher risk of cardiovascular diseases, heart problems, strokes and aneurisms. Being such a critical vital parameter, continuous monitoring can prove important to the medical staff when it comes to diagnosing conditions. Implantable wireless blood pressure sensors have been proposed in Ref. [30–33]. Blood pressure can also be useful to control vascular graft degradation through blood flow measurements, and a sensor capable of performing this task is presented in Ref. [34]. The state of the art is considered to be the vascular graft blood pressure sensor presented in Ref. [34]. This sensor has a 2.67 mm^2 chip with two coils that hold it in place inside a vascular graft. Pressure is digitized and backscattered, with the device consuming only 21.6 μW and with a sensitivity of 0.176 mmHg.

2.6. pH

The pH of a solution plays an important role in chemical processes that it undergoes, therefore affecting several physiological parameters and functions. pH can be used to identify microbial presence in tumours and monitor wound healing [35]. In Refs. [36, 37], a sensor is used to monitor gastroesophageal reflux disease (GERD) by measuring pH in the oesophagus. In Ref. [38], oral pH is measured to control the pathogenesis of dental caries. The device presented in Ref. [35] is the state of the art of implantable pH sensors. It integrates carbon nanotube-based sensors, which do not require a reference electrode, with an RFID tag that modulates data into an externally provided carrier. It is capable of accurately detecting pH levels between 2 and 12 during 120 days.

2.7. Intracranial pressure

Intracranial pressure is a vital biomedical parameter when it comes to the management of traumatic brain injuries. Current methods require catheters inserted into the cranial cavity, which cause patient discomfort and carry a risk of infection and haemorrhage [19, 39]. Minimally, invasive techniques based on wireless sensors have been presented in Refs. [39–42]. Chen et al. presented, in Ref. [39], passive sensors with volumes down to $1 \times 1 \times 0.5\ mm^3$. Pressure ranges from 0 to 100 mmHg were registered, with wireless and batteryless operation.

2.8. Electromyography

Electromyography (EMG) measures the electrical potentials present in muscle, and this data can be useful for the diagnosis of illnesses and injuries, functional electrical stimulation, and to control prosthetic limbs. EMG sensors with wireless capabilities have been presented in Refs. [43, 44]. The sensor presented in Ref. [44] is an EMG and electrocardiogram (ECG) monitor with four analog channels, a chip that consumes 19 μW (when sampling from one channel) and communicates at a data rate of 200 kbps with a power consumption of 160 μW. It includes RF power transfer and thermoelectric energy harvesting powering modules, giving the device versatility.

2.9. Electrocardiogram

Electrocardiogram (ECG) measurements allow physicians to have a closer look at the patients' heart, and it can be used to detect arrhythmias and heart attacks (myocardial infarctions), for example. Wireless ECG monitors have been proposed in Refs. [44, 45]. The device presented in Ref. [45] is notable for its extremely low power consumption of 64 nW, which raises the bar in terms of power budgets. Nevertheless, it does not allow for continuous monitoring, as it only stores abnormal events into the memory for posterior wireless relaying. For continuous monitoring, the previously discussed device of Ref. [44] is considered the state of the art.

3. Communications

Promising and viable communication strategies have been reported, such as intra-body communication (IBC) [46, 47] and ultrasound (US) [48]. The first consists on using biological tissue of the system's host as a conductive medium for electrical signals conveying data. The second is based on ultrasounds, a mechanical wave of frequencies above 20 kHz, which suffers low tissue absorption. Radio frequency (RF) is the most widely implemented communication method; therefore, this section will focus on sensors with RF wireless communications. Passive and active RF communication methods will be presented, with examples of devices resorting to them.

3.1. Passive RF communication (PRFC)

This communication method relies on the resonant frequency of a pair of coupled coils, one in the wireless implant and the other in an external device. The sensor is attached to the implant's coil, and a change in the parameter to which the sensor is sensitive to translates into a varying impedance of the coil. Consequently, the resonant frequency of the coupled coils will shift as the parameter of interest, for example, IOP, varies. Generally, this approach requires no power from the implant [13, 14, 32, 33, 39, 40], as the external reader is responsible to detect the impedance change in the implant's coil and, from it, calculate the sensed parameter's value. This allows for smaller implants, as power budget is reduced and no processing electronics are required. In a BAN perspective, this communication method can be applied in situations where on-body readers are a possibility (e.g. intraocular pressure monitoring where the external reader is placed in a pair of glasses worn by the patient).

3.2. Active RF communication (ARFC)

Implantable sensors described in this subsection communicate with the outside world resorting to an on-board antenna and an RF signal, at the expense of power. In Ref. [30], the authors resorted to capacitive coupling, in contrast to the more common inductive coupling. This method consists of using the host's biological tissue as a dielectric between two sets of metallic plates, one on the sensor and another on a reader, which can be body-worn or implantable. Operation frequencies must be kept as low as possible, since the tissue becomes more conductive as frequency increases.

Inductive coupling communication is performed between two coils and has the advantage of being more efficient than far-field communication. On the other hand, a precise alignment between coils is necessary, under the penalty of drastically losing efficiency. Additionally, the distance between coils must be kept as small as possible, unlike far field, which can be used at long ranges. The choice between one of these two powering methods must be made considering the available space, power budget and radiation safety guidelines. From the examples provided, no connection can be made between the choice of inductive coupling or far-field communication and the application of the sensor.

The choice between passive and active communication technologies is one that cannot be taken lightly when designing and developing an implantable sensor. Considering BANs, if wearable or large implantable relays are available near the implanted sensors, and the latter have severe volume limitations, passive communication can be a viable option, as the relays can support the bulky batteries or wireless power transfer (WPT) components, and the implanted sensor can use an oscillator to modulate the data into the relay's RF signal and backscatter it. In applications where the sensor has available space for computing capabilities, inductive links can be employed. With this, the sensor can process larger amounts of data, such as multiple channels. When long-distance operation is desirable, e.g. when no on-body or implantable relays are desirable, far-field communication is the best option, as it removes those constraints.

4. Powering

Sensor miniaturization is a desired goal; therefore, a compromise must be made between battery size, and consequently the size of the device itself, and its autonomy, bearing in mind that battery replacement may require an invasive surgical procedure, which could potentially lead to health complications [49, 50]. The urge to research for new and reliable powering solutions for implantable devices to increase their lifespan and reduce their volume is evident, and the interest in this field is proven by the amount of publications made available over the previous years. **Figure 1** contains a diagram representation of the different types of device powering that will be discussed in this section.

4.1. Energy harvesting

Energy harvesting techniques consist of harvesting useful amounts of energy from the ambient environment in order to power a device or charge a battery, having potential to provide power to biomedical devices since they could yield unlimited energy, drastically increasing the devices' lifespan. However, harvesting useful amounts of energy from the environment can be proven challenging, as the amount of available energy is volatile and often very limited, which imposes the need of special power management circuitry [49, 51]. Despite of the aforementioned limitations, research in the field of energy harvesting is of high interest due to the constant reduction of the power demands of electronic circuits [52].

Several energy harvesting techniques have been proposed by researchers, and special attention is given to thermoelectric generators, biomechanical energy, solar power, biofuel and RF energy harvesters.

http://dx.doi.org/10.5772/intechopen.70180

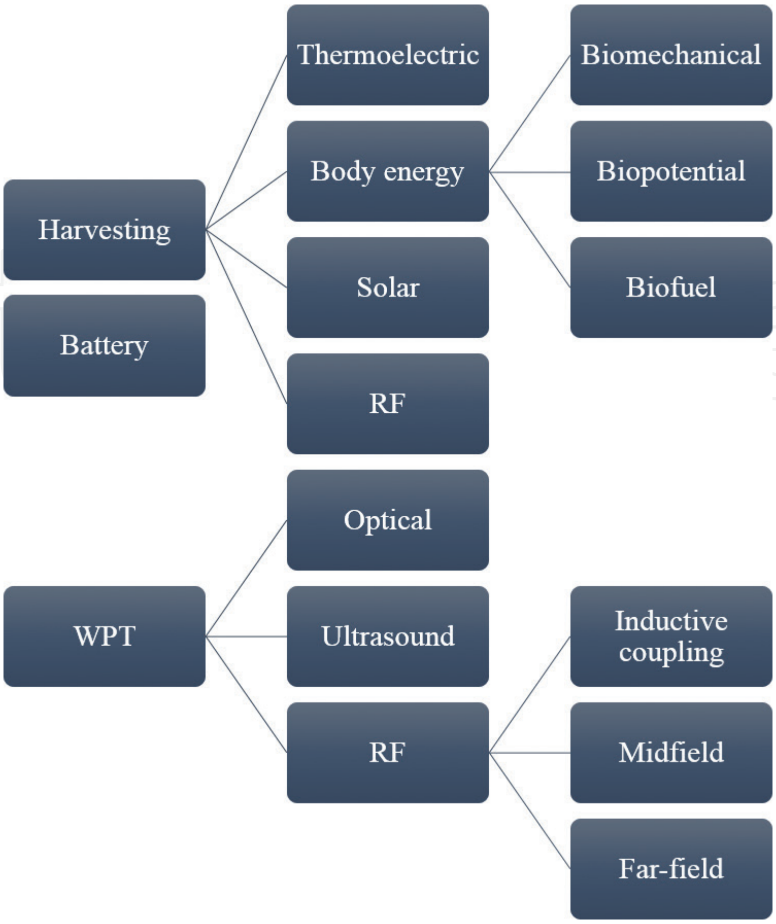

Figure 1. Implantable device powering methods.

4.1.1. Thermoelectric generators

Thermoelectric generators are solid-state devices that convert the thermal energy from temperature gradients into electrical energy [53]. These are appealing power sources for implantable devices, as they possess high reliability, are compact and do not require moving parts [54]. Such generators are based on the Seebeck effect, which states that an electrical voltage is generated across a metal or semiconductor when it is exposed to a temperature gradient [55].

4.1.2. Biomechanical energy harvesters

Biomechanical energy is generally abundant in the human body. It is generated by breathing, muscle stretching, body weight during motion and heart beats. The conversion between energy types is achieved resorting to a transduction mechanism, with electromagnetic and piezoelectric mechanisms being the most common. Biomechanical energy harvesters usually fall into two categories: vibrational or force-driven. Vibrational harvesters use inertial energy of a given mass, while force-driven ones rely on direct application of mechanical force [49].

This generator's feasibility for implantable medical devices was also studied. It was implanted on the right ventricular wall of a dog's heart and produced 80 mJ of energy after 30 minutes of operation [56].

4.1.3. Solar power

Solar cells were found to be capable of powering implantable devices. Even when implanted below a skin layer, these cells can harvest some power, as a small amount of light is able to penetrate the skin, in particular near-infrared light [57]. An absorption of around 10% of the incident power per millimetre of the skin occurs for a wavelength of 632.8 nm and 11.5% for 904 nm [58]. Nevertheless, large size, low efficiency and tissue heating are the major drawbacks of these systems [59]. Solar power harvesting cells have been developed in Refs. [12, 60], and they are capable of generating 1.1 $\mu W/mm^2$ in the eye and 34 $\mu W/mm^2$ below the skin, respectively.

4.1.4. Biofuel

Biofuel cells transform biochemical energy into electric energy by making use of electrochemical reactions. Oxidation and reduction reactions occur in the anode and cathode of the biofuel cell, generating a flow of electrons that generates power that a device can be used to power itself. Advantage of this technology is, for example, the biocompatibility between the fuel cell and the human body. On the other hand, low harvested power levels can pose a limitation as well as the anode and cathode degradation over time. Examples of reviews of implantable biofuel cells in living animals are available in Refs. [61, 62].

4.1.5. RF energy harvesting

RF energy harvesting consists on harnessing electromagnetic waves that exist in the environment, generated by communication towers, for example. These waves have the potential to provide power for electronic devices. The quantity of available radiation, the efficiency of the power conversion system and the size constraints of the device will dictate whether this method suffices in powering a given application. Even though technological advances are constantly being made, the size constraint of implantable medical devices and the typical ambient RF power densities cause some uncertainty about the suitability of this device powering method, as power levels below 1 μW can be recovered [51].

4.2. Wireless power transfer

The previously studied energy harvesting techniques suitable for implantation generate small power outputs. Consequently, the use of a dedicated power emitter for charging the devices has to be considered. Pertinent technologies such as the use of optical energy, ultrasounds or RF waves emerge as alternatives.

4.2.1. Optical link

Optical waves have been suggested to power medical implants, as they do not interfere with nearby communication systems like RF waves do. In Ref. [63], an array of silicon diodes with an area of 2.1 cm^2, implanted 1–3 mm under the skin, was used for transcutaneous power transmission. Using near-infrared irradiation at 810 nm with a power density of 22 mW/cm^2, the charging of a lithium battery capable of powering a commercial pacemaker for 24 hours was reported, while the temperature rise on the skin during light irradiation was 1.4°C.

4.2.2. Ultrasonic link

Ultrasonic waves, akin to optical waves, do not interfere with nearby electromagnetic fields and communication devices. They induce a vibration in the tissue, and the resulting kinetic energy is converted to electrical energy through a transducer, e.g. a piezoelectric transducer [49]. Ultrasonic power transmission has some disadvantages that limit its application to implantable medical devices. This transmission is very sensitive to the contact between the transmitter and the tissue, as an impedance mismatch between these elements or a misalignment between transmitter and receiver can severely reduce transmission efficiency [64].

4.2.3. Radio frequency link

Electromagnetic radiation, more specifically RF, is adequate to transport energy over long distances and presents one of the highest miniaturization potentials [65]. Additionally, its absorption by biological tissues does not induce damage, as long as the specific absorption rate (SAR) is not exceeded.

One of the most common methods of power transmission to medical devices bases itself on inductive coupling, as it has the lowest absorption rate by body tissue at lower frequencies. This method has been previously used to power cochlear implants, total artificial hearts and neural implants, among others [66–69]. Despite its popularity, this method has some drawbacks, such as coil decoupling due to misalignment, since it requires rigorous positioning of transmitter and receiver coils [70–72]. Moreover, the range of inductive coupling complies with exponential decay, a near-field phenomenon, meaning that the external coil must be close to the implant. These limitations can be overcome by establishing links in the middle (see Ref. [73]) or far field, resorting to antennas. Although energy transportation is less efficient, it allows for greater distances between the power source and the target than the previous inductive methods [74].

5. Integration

Most of today's implantable electronic devices, such as the ones so far reviewed in this chapter, rely on silicon microelectronics. The evolution of fabrication techniques and microelectronics has translated into a reduced size of implantable electronics. Nowadays, there is an urge to further miniaturize them to make them easier to implant and less traumatic for the patient. Efforts made towards this goal over the past few years will be reported through examples of success cases.

Chen et al. [39] have designed a wireless pressure monitoring sensor with dimensions down to $1 \times 1 \times 0.1$ mm^3. A $2.5 \times 2.5 \times 0.1$ mm^3 device was used to validate the design in vivo. A resonant circuit composed of an inductive antenna and a pressure-sensitive capacitor is the heart of the sensor. An applied pressure changes the resonant frequency of the LC circuit, and the frequency shift is detected by an external reader, which then converts it into pressure values.

Mostafalu et al. [75] created threads with different properties to act as sensors, microfluidics and electronics. Hydrophobic threads were used as microfluidic channels, while threads infused with materials such as carbon nanotubes were used as electrodes for sensing pH, glucose and so on. Conventional electronics were present in a different layer, and these established communication links and processed the electrodes' signal. Fabric devices were tested in pH and strain sensing, in vivo, having been successful. This research has the potential to lead to the creation of smart sutures and bandages.

The examples above serve to demonstrate how fabrication technology enables devices to become smaller than ever, while still packing enough features to perform their given tasks.

5.1. Biodegradable and stretchable sensors

In the past few years, new materials for implantable sensors have been proposed, studied and validated, namely, stretchable and biodegradable materials [76–80]. Biodegradable materials allow for transient sensors that can, for example, be implanted after a surgery to monitor wound healing and bacterial activity, and after a predefined period, the device would start to degrade inside the human body. The by-products of this process would then be eliminated naturally by the organism. This would mitigate the need for implant retrieval surgeries, along with all associated negative aspects, e.g. patient discomfort, risk of infection, surgery room scheduling and so on.

Kang et al. [41] demonstrated an ICP sensor in a rat, fabricated with a polymer (PLGA) and either a magnesium or a silicon foil. Continuous monitoring of ICP was achieved during 3 days, after which the materials composing the sensor were reabsorbed into the body.

Luo et al. [81] fabricated a pressure sensor based on a variable capacitor and a coil. The biodegradation of this device was documented by the authors. During the first 21 hours of immersion in a saline solution, the resonant frequency of the sensor changed, as if it was stabilizing itself in the system. In the following 86 hours, the resonant frequency stayed constant, showing stability of the device, thus being the optimal operation period of the sensor. After this, the quality of the sensor starts to degrade until it is unusable.

A transient device capable of managing bacteria growth in a region of the body, possibly a surgery or implant site, has been proposed in Ref. [80]. Using magnesium for an inductive coil, a silicon resistor and silk encapsulation, a heater was produced. An external RF field would power the resistor, which would heat up by 5°C and prevent bacteria proliferation in that location. The longevity of the device is controlled by the silk's crystallinity.

Biodegradable batteries have also been achieved (see Ref. [78]). These were capable of powering a LED and a 58 MHz wireless signal generator.

http://dx.doi.org/10.5772/intechopen.70180

Finally, stretchable electronics are also becoming a reality [79]. A sensor that can be bent and twisted without losing its properties is an important step in implantable devices, as patient discomfort would be greatly reduced. Devices, such as electronic eyeball cameras and coplanar waveguides, have been demonstrated.

The evolution of integration techniques and material science is of paramount importance for the medical sensor area. Smaller devices with the same powerful capabilities are in high demand, allowing for new applications and the improvement of current ones. Bendable and stretchable sensors can be a positive step in patient comfort and device reliability, reducing the negative response from the human body. Finally, biodegradable sensors have an enormous potential, as they can be used to monitor a parameter for a limited period of time, after which it is simply absorbed by the human body without any harm, eliminating the need for a retrieval surgery.

6. Networking issues

The increase of implantable sensor solutions for the medical field brings the necessity of such sensors to work together to collect and relay measurements of biomedical parameters. The most used communication method for implants is based on electromagnetic radiation. Due to the conductive nature of biological tissue, it suffers great attenuation, as tissues absorb energy and dissipate it as heat. Experimental path loss models were presented in Ref. [82] for in-body to in-body, in-body to on-body and in-body to off-body communications between 2.36 and 2.5 GHz. This work is a proof of the challenges that lossy biological tissue presents to sensor development and networking.

Networking solutions of implanted sensors must consider SAR limits and temperature increase of tissues to guarantee patient safety. According to Ref. [83], the high quality of service (QoS) required for biomedical systems can only be achieved in such a propagation medium if performance-enhancing techniques, such as adaptive coding and modulation and link diversity, are adapted from miniature wireless electronics to implantable sensors.

In Ref. [7], three networking methods for on- and in-body sensors are presented. Of these, the use of on-body beacons shows good promise. The beacons would be responsible for forwarding data between sensors and relaying it to base stations, thus reducing the power dissipation inside the human body. Since the beacons can be larger than implantable sensors, these can also be used as power sources or controllers for the sensors, as their power budget can be significantly higher. In Ref. [84], the authors agree with the previous statement, and they present a study of QoS and power consumption variation of BAN nodes with different on-body beacon placements.

BANs pose yet another challenge for engineers. As the human body is a flexible, moving environment, the relative position of the network's nodes can change frequently, thus altering signal attenuation in communication links. For example, a wearable, on-body relaying node, such as a smartwatch, changes its position relatively to the in-body sensors all the time during normal day-to-day activity of the wearer. The same concept applies to in-body sensors placed in moving organs and members or even in the bloodstream. Ramachandran et al. [85] proposed a medium access control (MAC) protocol based on human activity, which they named HAMAC, that aims to work around the previously mentioned problems by adjusting the timing of communication between nodes and from nodes to relays.

In 2012, the IEEE has published standard IEEE 802.15.6. It addresses the communication protocol for BANs for medical applications (see Ref. [6]). Works around this standard have been found, such as the one found in Ref. [86], which aims to improve it by adding an ultrawideband channel model.

Alternatives to RF networking have also been studied. Santagati et al. [48, 87] proposed a MAC protocol for US communications, ultrasonic wideband (UsWB). It aims to establish intra-body communication between BAN nodes without the previously mentioned setbacks of RF radiation. UsWB was reported to be resistant to the multipath caused by the multitude and inhomogeneity of tissues in the propagation medium, i.e. the human body, thus making it a viable alternative to RF-based communications.

6.1. Security concerns

When sensors transmit data to one another or to the outside world, sensitive medical information is therein contained. The theft of such data by a third party is a serious danger and must be prevented. In the case of sensors or actuators within the body network that receive instructions from a controller, the possibility of having an attacker sends commands to these devices must be completely eliminated to guarantee the safety of the patient. Furthermore, an attacker must not be able to modify the content of the data being exchanged in the network without the receiver noticing the change, thus guaranteeing the integrity of the communication.

Steps towards the protection and encryption of transmitted data have been taken and reported. In Ref. [88], the authors proposed a method to share secret data inside a network by using ECG as a decryption key. Only an external reader with access to real-time ECG of the patient would be able to read the data, and given the random nature of the ECG wave, this safety method presents great potential. Nevertheless, it must not be forgotten that implantable sensors have limited power budgets; therefore, this encryption must be lightweight. In Ref. [89], the same authors have improved upon this method by using characteristic parameters of ECG signals, the P, Q, R, S and T peaks, and generate random binary sequences with the time intervals between these peaks. This approach was reported to have low latency and to benefit of the same randomness of ECG signals as the previously reported one.

7. Node mobility

In recent years, propulsion methods for small implantable devices, or robots, have been proposed. Having sensors capable of moving in body fluids has medical interest, as it can allow one device to perform measurements and diagnostic in an area wider than ever before.

Minimally invasive surgery or targeted drug delivery could also be performed with steerable devices.

In Ref. [90], a 3 × 4 mm^2 wireless implantable device is presented. Its propulsion method is based on magnetohydrodynamics (MHD). It requires a constant magnetic field of around 0.1 T, which can be achieved with a permanent magnet. The device applies currents in the mA order of magnitude through the medium's conductive fluid, and a force is generated in the magnetic field. The device then experiences an equal and opposite force that propels it. Power is provided by a 1.86 GHz WPT system, which also carries movement commands modulated into the power carrier. The device can controllably move at a speed of 5.3 mm/sec in salt water.

Hsieh et al. [91] developed a remote-controlled device with a propulsion mechanism based on gas pressure from electrolytic bubbles generated on the surrounding fluid. It can move at a rate of 0.3 mm/s, at around 200 μW power consumption. Electrolysis electrodes are present all around the device, so it is possible to define where the electrolysis will occur and, consequently, steer the device. It is powered by a 10 MHz inductive coupling link which also carries commands to control movement direction and speed. The receiving coil and electrodes are integrated in the locomotive chip, which has a total area of 21.2 mm^2. Despite the slower speed of this device, especially comparatively to the one reported in Ref. [90], this approach does not require external components such as permanent magnets.

This section presented propulsion mechanisms for wireless devices operating in a liquid medium. The reported WPT, communication and steering capabilities of these devices are an important stepping stone towards fully autonomous or remote-controlled sensors and actuators integrating a BAN that are capable of navigating, for example, through the bloodstream, digestive tract or bladder. Simultaneously, they would be performing measurements, relaying them to the outside world and performing microsurgery or drug delivery at the required locations.

8. Conclusion

BANs comprised of implantable sensors are becoming closer and closer to being a common tool in the medical field. This would mean a significant improvement on healthcare for patients, as close monitoring of critical parameters can be done full-time and without constraints. Several powering methods that allow these devices to be as small as possible and to operate indefinitely are available and maturing. The same applies for communication methods, which tend to be less power consuming, and there were even reported completely passive methods that can be used in situations where extremely small devices are required. Still in the topic of communications, security issues and networking difficulties have been raised, with efforts to mitigate them being presented. Integration techniques that allow the fabrication of sensors with more host-friendly materials have been detailed, with biodegradable and stretchable materials being a topic of high interest in the past few years. Finally, mobility mechanisms that allow for controllable exploratory sensors have also been shown, and these pave the way for large area monitoring by a single sensor, adding to their versatility and capabilities. In conclusion, the evolution of biomedical sensors is leading the way to completely functional and tailored BANs that in the near future will prove to be indispensable tools for health monitoring in both the hospital environment and daily life of patients.

Acknowledgements

This work is supported by FCT with the reference project PTDC/EEI-TEL/5250/2014, by FEDER funds through Projecto 3599 – Promover a Produção Científica e Desenvolvimento Tecnológico e a Constituição de Redes Temáticas (3599-PPCDT) and by grant SFRH/BD/116554/2016.

Author details

Hugo Dinis* and Paulo M. Mendes

*Address all correspondence to: hugodcdinis@gmail.com

Department of Industrial Electronics, University of Minho, Portugal

References

[1] Frost MC, Meyerhoff ME. Implantable chemical sensors for real-time clinical monitoring: Progress and challenges. Current Opinion in Chemical Biology. 2002;**6**(5):633-641

[2] McKean BD, Gough DA. A telemetry-instrumentation system for chronically implanted glucose and oxygen sensors. IEEE Transactions on Biomedical Engineering. 1988; **35**(7):526-532

[3] Wise KD, et al. Wireless implantable microsystems: High-density electronic interfaces to the nervous system. Proceedings of the IEEE. 2004;**92**(1):76-97

[4] Olsson R, Wise K. A three-dimensional neural recording microsystem with implantable data compression circuitry. IEEE International Solid-State Circuits Conference. 2005;558-559. DOI: 10.1109/ISSCC.2005.1494117

[5] Bazaka K, Jacob M. Implantable devices: Issues and challenges. Electronics 2012;**2**:1-34 p

[6] IEEE Standards Association. IEEE Standard for Local and Metropolitan Area Networks—Part 15.6: Wireless Body Area Networks. IEEE Std. 2012. 271 p. DOI: 10.1109/IEEESTD.2012.6161600

[7] Honeine P, et al. Wireless sensor networks in biomedical: Body area networks. Systems, Signal Processing and their Applications (WOSSPA), International Workshop. 2011; (1):388-391. DOI: 10.1109/WOSSPA.2011.5931518

[8] Hannan MA, et al. Energy harvesting for the implantable biomedical devices: Issues and challenges. Biomedical Engineering Online. 2014;**13**(1):79

[9] Quigley HA. Number of people with glaucoma worldwide. The British Journal of Ophthalmology. 1996;**80**(5):389-393

[10] Chiou JC, et al. Toward a wirelessly powered on-lens intraocular pressure monitoring system. IEEE Journal of Biomedical and Health Informatics. 2016;**20**(5):1216-1224

[11] Leonardi M, et al. Wireless contact lens sensor for intraocular pressure monitoring: Assessment on enucleated pig eyes. Acta Ophthalmologica. 2009;**87**(4):433-437

[12] Ghaed MH, et al. Circuits for a cubic-millimeter energy-autonomous wireless intraocular pressure monitor. IEEE Transactions on Circuits and Systems I Regular Papers. 2013;**60**(12):3152-3162

[13] Chitnis G, et al. A minimally invasive implantable wireless pressure sensor for continuous IOP monitoring. IEEE Transactions on Biomedical Engineering. 2013;**60**(1):250-256

[14] Kouhani MHM, Weber A, Li W. Wireless intraocular pressure sensor using stretchable variable inductor. Proceedings on IEEE International Conference on Micro Electro Mechanical Systems. 2017. pp. 557-560. DOI: 10.1109/MEMSYS.2017.7863467

[15] Donida A, et al. A 0.036 mbar circadian and cardiac intraocular pressure sensor for smart implantable lens. Dig Tech Pap – IEEE International Solid-State Circuits Conference. 2015;**58**:392-393

[16] Shih Y, Shen T, Otis B. A 2.3 μW wireless intraocular pressure/temperature monitor. IEEE Journal of Solid-State Circuits. 2011;**46**(11):2592-2601

[17] Chen W, et al. A fully integrated 8-channel closed-loop epileptic seizure control. IEEE Journal of Solid-State Circuits. 2014;**49**(1):232-247

[18] Seo D, et al. Wireless recording in the peripheral nervous system with ultrasonic neural dust. Neuron. 2016;**91**(3):529-539

[19] Fan D, et al. A wireless multi-channel recording system for freely behaving mice and rats. PLoS One. 2011;**6**(7):1-9. DOI: 10.1371/journal.pone.0022033

[20] Rhew HG, et al. A fully self-contained logarithmic closed-loop deep brain stimulation SoC with wireless telemetry and wireless power management. IEEE Journal of Solid-State Circuits. 2014;**49**(10):2213-2227

[21] Muller R, et al. A minimally invasive 64-channel wireless μECoG implant. IEEE Journal of Solid-State Circuits. 2015;**50**(1):344-359

[22] Mestais CS, et al. WIMAGINE: Wireless 64-channel ECoG recording implant for long term clinical applications. IEEE Transactions on Neural Systems and Rehabilitation Engineering. 2015;**23**(1):10-21

[23] Gao H, et al. HermesE: A 96-channel full data rate direct neural interface in 0.13 μm CMOS. IEEE Journal of Solid-State Circuits. 2012;**47**(4):1043-1055

[24] Majerus SJA, et al. Wireless, ultra-low-power implantable sensor for chronic bladder pressure monitoring. ACM Journal on Emerging Technologies in Computing Systems. 2012;**8**(2):1-13

[25] Majerus S, et al. Wireless implantable pressure monitor for conditional bladder neuromodulation. IEEE Biomedical Circuits and Systems Conference: Engineering for Healthy Minds and Able Bodies, BioCAS 2015 – Proceedings. 2015. pp. 2-5. DOI: 10.1109/BioCAS.2015.7348337

[26] Lee WS, et al. UP-link: An ultra-low power implantable wireless system for long-term ambulatory urodynamics. 2014 IEEE Biomedical Circuits and Systems Conference (BioCAS) – Proceedings. 2014. pp. 384-387. DOI: 10.1109/BioCAS.2014.6981743

[27] Kim A, Powell CR, Ziaie B. An implantable pressure sensing system with electromechanical interrogation scheme. IEEE Transactions on Biomedical Engineering. 2014; **61**(7):2209-2217

[28] Vaddiraju S, et al. Needle-implantable, wireless biosensor for continuous glucose monitoring. 2015 IEEE 12th International Conference Wearable Implant Body Sensor Networks, BSN 2015. 2015. pp. 2-6. DOI: 10.1109/BSN.2015.7299421

[29] Dehennis A, Mailand M, Grice D, Getzlaff S, Colvin A. A near-field-communication (NFC) enabled wireless fluorimeter for fully implantable biosensing applications. IEEE International Solid-State Circuits Conference. 2013. pp. 298-299. DOI: 10.1109/ISSCC.2013.6487743

[30] Aldaoud A, Laurenson C, Rivet F, Yuce MR, Redouté J. Design of an inductively powered implantable wireless blood pressure sensing interface using capacitive coupling. IEEE/ASME Transactions on Mechatronics. 2015;**20**(1):487-491

[31] Cong P, et al. A wireless and batteryless 130 milligram 300 μW 10-bit implantable blood pressure sensing microsystem for real-time genetically engineered mice monitoring. IEEE International Solid-State Circuits Conference.. 2009;**44**(12):428-429

[32] Park J, et al. A wireless pressure sensor integrated with a biodegradable polymer stent for biomedical applications. Sensors (Switzerland). 2016;**16**(6):1-10. DOI: 10.3390/s16060809

[33] Murphy OH, Bahmanyar MR, Borghi A, McLeod CN, Navaratnarajah M, Yacoub MH, et al. Continuous *in vivo* blood pressure measurements using a fully implantable wireless SAW sensor. Biomedical Microdevices. 2013;**15**(5):737-749

[34] Cheong JH, et al. An inductively powered implantable blood flow sensor microsystem for vascular grafts. IEEE Transactions on Biomedical Engineering. 2012;**59**(9):2466-2475

[35] Gou P, Kraut ND, Feigel IM, Bai H, Morgan GJ, Chen Y, et al. Carbon nanotube chemiresistor for wireless pH sensing. Scientific Reports. 2014;**4**:1-6

[36] Cao H, et al. Batteryless implantable dual-sensor capsule for esophageal reflux monitoring. Gastrointestinal Endoscopy. 2013;**77**(4):649-653

[37] Ativanichayaphong T, Tang SJ, Hsu LC, Huang WD, Seo YS, Tibbals HF, et al. An implantable batteryless wireless impedance sensor for gastroesophageal reflux diagnosis. IEEE MTT-S International Microwave Symposium Digest. 2010. pp. 608-611. DOI: 10.1109/MWSYM.2010.5516775

[38] Farella M, et al. Simultaneous wireless assessment of intra-oral pH and temperature. Journal of Dentistry. 2016;**51**:49-55

[39] Chen LY, et al. Continuous wireless pressure monitoring and mapping with ultra-small passive sensors for health monitoring and critical care. Nature Communications. 2014;**5**:5028

[40] Behfar MH, et al. Biotelemetric wireless intracranial pressure monitoring: An *in vitro* study. International Journal of Antennas and Propagation. 2015 Apr;**2015**:1-10

[41]] Kang S, et al. Bioresorbable silicon electronic sensors for the brain. Nature. 2016;**530** (7588):71-6

[42] Meng X, et al. Dynamic study of wireless intracranial pressure monitoring of rotational head injury in swine model. Electronics Letters. 2012;**48**(7):363

[43] Kneisz L, Unger E, Lanmuller H, Mayr W. *In vitro* testing of an implantable wireless telemetry system for long-term electromyography recordings in large animals. Artificial Organs. 2015;**39**(10):897-902

[44] Zhang Y, et al. A batteryless 19 μW MICS/ISM-band energy harvesting body sensor node SoC for ExG applications. IEEE Journal of Solid-State Circuits. 2013;**48**(1):199-213

[45] Jeon D, et al. An implantable 64nW ECG-monitoring mixed-signal SoC for arrhythmia diagnosis. Dig Tech Papers – IEEE International Solid-State Circuits Conference. 2014;**57**:416-417

[46] Hayami H, et al. Wireless image-data transmission from an implanted image sensor through a living mouse brain by intra body communication. Japanese Journal of Applied Physics. 2016;**55**(4): 04EM03-1 - 04EM03-5. DOI: 10.7567/JJAP.55.04EM03

[47] Anderson GS, Sodini CG. Body coupled communication: The channel and implantable sensors. 2013 IEEE International Conference of Body Sensor Networks, BSN 2013. 2013. pp. 3-7. DOI: 10.1109/BSN.2013.6575490

[48] Enrico Santagati G, Melodia T. Experimental evaluation of impulsive ultrasonic intra-body communications for implantable biomedical devices. IEEE Transactions on Mobile Computing. 2017;**16**(2):367-380

[49] Rasouli M, Phee LSJ. Energy sources and their development for application in medical devices. Expert Review of Medical Devices. 2010;**7**(5):693-709

[50] Gould PA. Complications associated with implantable cardioverter-defibrillator replacement in response to device advisories. Journal of the American Medical Association. 2006 Apr 26;**295**(16):1907

[51] Hudak NS, Amatucci GG. Small-scale energy harvesting through thermoelectric, vibration, and radiofrequency power conversion. Journal of Applied Physics. 2008;**103**(10): 101301-1 - 101301-24. DOI: 10.1063/1.2918987

[52] Mitcheson PD, Yeatman EM, Rao GK, Holmes AS, Green TC. Energy harvesting from human and machine motion for wireless electronic devices. Proceedings of the IEEE. 2008 Sep;**96**(9):1457-1486

[53] Snyder J. Small thermoelectric generators. Electrochemical Society Interface. 2008;**17** (3):54-56

[54] Leonov V, Torfs T, Fiorini P, Van Hoof C. Thermoelectric converters of human warmth for self-powered wireless sensor nodes. IEEE Sensors Journal. 2007;**7**(5):650-656

[55] Bell LE. Cooling, heating, generating power, and recovering waste heat with thermoelectric systems. Science. 2008;**321**(5895):1457-1461

[56] Goto H. Feasibility of using the automatic generating system for quartz watches as a leadless pacemaker power source: A preliminary report. Medical & Biological Engineering & Computing. 1998;**20**(I):9-11

[57] The International Commission on Non-Ionizing Radiation Protection. ICNIRP statement on far infrared radiation exposure. Health Physics. 2006;**91**(6):630-645. link: http://www.icnirp.org/cms/upload/publications/ICNIRPinfrared.pdf

[58] Enwemeka CS. Attenuation and penetration of visible 632.8 nm and invisible infra-red 904 nm light in soft tissues. Official Journal of World Association for Laser Therapy. 2001;**13**:95-101

[59] Amar A Ben, Kouki AB, Cao H. Power approaches for implantable medical devices. Sensors (Switzerland). 2015;**15**(11):28889-28914

[60] Haeberlin A, et al. Successful pacing using a batteryless sunlight-powered pacemaker. Europace. 2014;**16**(10):1534-1539

[61] Katz E, MacVittie K. Implanted biofuel cells operating *in vivo* – methods, applications and perspectives – feature article. Energy & Environmental Science. 2013;**6**(10):2791

[62] Katz E. Implantable biofuel cells operating *in vivo*: Providing sustainable power for bioelectronic devices: From biofuel cells to cyborgs. In: 2015 6th International Workshop on Advances in Sensors and Interfaces (IWASI). IEEE; 2015. pp. 2-13. DOI: 10.1109/IWASI.2015.7184958

[63] Goto K, et al. An implantable power supply with an optically rechargeable lithium battery. IEEE Transactions on Biomedical Engineering. 2001;**48**(7):830-833

[64] Arra S, Leskinen J, Heikkila J, Vanhala J. Ultrasonic power and data link for wireless implantable applications. 2007 2nd International Symposium on Wireless Pervasive Computing. 2007;567-571. DOI: 10.1109/ISWPC.2007.342668

[65] Katz E. Implantable Bioelectronics. 2014. DOI: 10.1002/9783527673148

[66] Kurs A, et al. Wireless power transfer via strongly coupled magnetic resonances. Science. 2007;**317**(5834):83-86

[67] Kim S, Ho JS, Chen LY, Poon ASY. Wireless power transfer to a cardiac implant. Applied Physics Letters. 2012;**101**(7):1-5

[68] Ho JS, Yeh AJ, Neofytou E, Kim S, Tanabe Y, Patlolla B, et al. Wireless power transfer to deep-tissue microimplants. Proceedings of the National Academy of Sciences. 2014;**111**(22):7974-7979

[69] Ho JS, Kim S, Poon ASY. Midfield wireless powering for implantable systems. Proceedings of the IEEE. 2013 Jun 3;**101**(6):1369-1378

[70] Flynn BW, Fotopoulou K. Wireless power transfer in loosely coupled links. Power. 2011;**47**(2):416-430

[71] Flynn BW, Fotopoulou K. Rectifying loose coils: Wireless power transfer in loosely coupled inductive links with lateral and angular misalignment. IEEE Microwave Magazine. 2013;**14**(2):48-54

[72] Aldhaher S, Luk PCK, Whidborne JF. Electronic tuning of misaligned coils in wireless power transfer systems. IEEE Transactions on Power Electronics. 2014;**29**(11):5975-5982

[73] Agrawal DR, et al. Conformal phased surfaces for wireless powering of bioelectronic microdevices. Nature Biomedical Engineering. 2017;**1**(3):43

[74] Visser H. Far-field RF energy transport. IEEE Radio and Wireless Symposium. 2013. pp. 34-6. DOI: 10.1109/RWS.2013.6486632

[75] Mostafalu P, et al. A toolkit of thread-based microfluidics, sensors, and electronics for 3D tissue embedding for medical diagnostics. Microsystems Nanoengineering. 2016;**2**(April):16039

[76] Rogers JA, Someya T, Huang Y. Materials and mechanics for stretchable electronics. Science. 2010;**327**(5973):1603-1607

[77] Yin L, Cheng H, Mao S, Haasch R, Liu Y, Xie X, et al. Dissolvable metals for transient electronics. Advanced Functional Materials. 2014;**24**(5):645-658

[78] Yin L, Huang X, Xu H, Zhang Y, Lam J, Cheng J, et al. Materials, designs, and operational characteristics for fully biodegradable primary batteries. Advanced Materials. 2014;**26**(23):3879-3884

[79] Someya T, Bao Z, Malliaras GG. The rise of plastic bioelectronics. Nature. 2016;**540**(7633):379-385

[80] Hwang S-W, et al. A physically transient form of silicon electronics. Science. 2012; **337**(6102):1640-1644

[81] Luo M, et al. A microfabricated wireless RF pressure sensor made completely of biodegradable materials. Journal of Microelectromechanical Systems. 2014 Feb;**23**(1):4-13

[82] Chavez-Santiago R, et al. Experimental path loss models for in-body communications within 2.36-2.5 GHz. IEEE Journal of Biomedical and Health Informatics. 2015;**19**(3):1-1

[83] Cheffena M. Performance evaluation of wireless body sensors in the presence of slow and fast fading effects. IEEE Sensors Journal. 2015;**15**(10):5518-5526

[84] Ntouni GD, Lioumpas AS, Nikita KS. Reliable and energy-efficient communications for wireless biomedical implant systems. IEEE Journal of Biomedical and Health Informatics. 2014;**18**(6):1848-1856

[85] Ramachandran VRK, Havinga PJM, Meratnia N. HACMAC: A reliable human activity-based medium access control for implantable body sensor networks. BSN 2016 – 13th Annual International Body Sensor Networks Conference. 2016. pp. 383-389. DOI: 10.1109/BSN.2016.7516292

[86] Chavez-Santiago R. Propagation models for IEEE 802.15.6 standardization of implant communication in body area networks. IEEE Communications Magazine. 2013;(August): 80-87. DOI: 10.1109/MCOM.2013.6576343

[87] Santagati GE, Melodia T, Galluccio L, Palazzo S. Medium access control and rate adaptation for ultrasonic intrabody sensor networks. IEEE/ACM Transactions on Networking. 2015;**23**(4):1121-1134

[88] Zheng G, Fang G, Orgun MA, Shankaran R, Dutkiewicz E. Securing wireless medical implants using an ECG-based secret data sharing scheme. 14th International Symposium on Information and Communication Technology – ISC 2014. 2015. pp. 373-377. DOI: 10.1109/ISCIT.2014.7011935

[89] Zheng G, Member S, Fang G, Shankaran R, Orgun MA, Member S, et al. Multiple ECG fiducial points-based random binary sequence generation for securing. IEEE Journal of Biomedical and Health Informatics. 2017;**21**(3):655-663

[90] Pivonka D, Yakovlev A, Poon ASY, Meng T. A mm-sized wirelessly powered and remotely controlled locomotive implant. IEEE Transactions on Biomedical Circuits and Systems. 2012;**6**(6):523-532

[91] Hsieh JY, et al. A remotely-controlled locomotive IC driven by electrolytic bubbles and wireless powering. IEEE Transactions on Biomedical Circuits and Systems. 2014; **8**(6):787-798