WIRELESS ENERGY TRANSFER AND WIRELESS COMMUNICATION FOR IN-BODY SENSORS

Department of Electronics and Telecommunications

Author: Sandra Yuste Muñoz

Supervisor: Kimmo Kansanen
Co-Supervisor: Ilanko Balasingham
Co-Supervisor: Narcís Cardona Marcet

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Abstract

Wireless communications in itself is a never-ending growing technology. One growing field of research relates to the implantable biomedical devices which have found applications in a wide range of areas. Some implants use traditional batteries to supply power for the electronic circuits within the sensor. However, any battery has a limited energy storage and life span, and percutaneous links are susceptible to infection and reliability problems. Wireless power transfer (WPT) offer the opportunity to provide power for longer periods without the risk of infection from a percutaneous lead. Inductive power transfer is the most common method of wireless power transfer to the implantable sensors which consist of a primary external power circuit and a secondary implantable power pick-up unit. A common characteristic associated with biomedical applications is loose coupling between the primary and secondary coils. Compensation for loose coupling can be achieved through the use of resonance circuits which enables the voltage or current at the secondary to boost up to useful levels even in the presence of low coupling coefficients. The ability to achieve power transfer is dependent on the match between the resonant frequency of the primary with the resonant frequency of the secondary.

Resonance-based wireless power delivery is investigated for improved energy transfer efficiency and reduced dependence on the distance between the primary and secondary coils. However, in practice the resonant frequency of the secondary pick-up circuit is often mismatched with the operating frequency of the primary because of the variations in load, coupling and other circuit parameters. When mismatching occurs, the voltage magnitude control approach can only respond by operating at a high magnitude to attempt to maintain the power flow to the load. One of the main constraints of the system is to achieve the minimum power required by the application by still keeping the implant size small enough for the living subject's body. This report also focuses on the design issues associated with the wireless exchange of data between the implant and the external world and also with the telemetry of power through the inductive link.

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List of Abbreviations and Symbols

WPT	Wireless Power Transfer
RFID	Radio Frequency Identification Device
SCMR	Strongly Coupled Magnetic Resonance
AET	Acoustic Energy Transfers
EM	Electromagnetic
WHO	World Health Organization
IEEE	Institute of Electrical and Electronic Engineers
ICNIRP	International Commission on Non-Ionizing Radiation Protection
BR	Basic Restriction
MPE	Maximum Permissible Exposure
RF	Radio Frequency
FCC	Federal Communications Commission
ANSI	American National Standards Institute
SAR	Specific Absorption Rate
DC	Direct Current
AC	Alternating Current
CMT	Coupled-Mode Theory
RLT	Reflected Load Theory
KVL	Kirchoff's Voltage Law
ISM	Industrial, Scientific and Medical
MICS	Medical Implant Communications Service
VAD	Ventricular Assist Device
FREE-D	Free-range Resonant Electrical Delivery
ASK	Amplitude Shift Keying
BER	Bit Error Rate
FSK	Frequency Shift Keying
PSK	Phase Shift Keying
PW	Pulse Width
SNR	Signal to Noise Ratio
UWB	Ultra Wide-Band
PL	Path Loss
FDTD	Finite Difference Time Domain
HP	Horizontal Polarization
VP	Vertical Polarization

Chapter 1

Introduction

Biomedical sensors and implantable devices are gaining momentum due to their variety of applications and are being utilized to perform important therapeutic, prosthetic and diagnostic functions. Some examples of these applications are automatic drug delivery systems, devices to stimulate specific organs and monitors to communicate internal vital signs to the outer world. Despite the fact all of these devices perform different tasks, one of their common issues is that of power requirements,which is a widely researched area over the past decade.

Providing required power to implanted devices in a reliable manner is of paramount importance. Some implants use batteries, however, their applications are limited due the longevity of the batteries and the size. Hence, wireless power-transfer schemes are often used in implantable devices not only to avoid transcutaneous wiring, but also to either recharge or replace the device battery.

This chapter introduces the topic of the present work. Section 1.1 provides a justification for using implantable sensors. Section 1.2 presents briefly the benefits of Wireless Power Transfer in contrast to typical batteries. Lastly, Section 1.3 describes the outline of this report.

1.1 Need for Implantable Sensors

Implantable sensors offer a unique opportunity for continuous monitoring of a patient's vital sign as well as an increased self-management of chronic conditions. This is a good approach to introduce remote healthcare solutions.

Nowadays the measurement of vital signs can be done without the physical presence of medical staff. Information can be sent directly from home, with a combination of wireless and wired communication links, to hospital or healthcare institution. People can receive divers benefits in this way such as self management and home as a care environment.

An increased awareness of patient's personal health by providing a continuous monitoring capability of own health parameters can provide a self management to the user. Furthermore, the possibility of staying at home longer can becomes home as a care environment where the patient can maintain a normal daily life instead of being hospitalized. This can have a positive influence on patient's entire healing process.

These benefits have also an important socio-economic impact. The quality of life is increased and the expenses for healthcare facilities are reduced, if not even suppressed by adopting new wireless technologies in healthcare and caring processes. The use of these wireless technologies can be seen as a fast and practical way for finding new ways in homecare related issues. Self-care, self management and cost effectiveness will be the key factor towards the development of new technological solutions and challenges. As a result, not only healthcare professionals will benefit from a reduction in hospitalizations and routine in-office follow-ups, but also patients will benefit from efficient management of their diseases.

1.2 Need for Wireless Power Transfer

The electronics inside the biosensor are used for signal processing and telemetry. These electronics blocks need power to function properly. Powering of implantable biomedical sensors is a major concern due to various constraints. Typically the leading source of powering involves batteries which can be used inside the human body if placed into a body cavity and are hermetically sealed. Conventional biomedical sensors, such as pacemakers, use batteries which have a typical lifetime of 5 to 7 years. These types of batteries are hermetically sealed and are placed inside human cavity via surgical procedures as well as surgical means are required in order to replace them after a discharge cycle.

It is also true that some sensors cannot use regular batteries as they directly come in contact with blood and thus it can cause injuries to the patients. The main risk is leakage which may lead to chemical burns, poisoning, and even death. This lead to decreasing number of applications for sensors using these types of batteries. In addition to this, another important feature is the dimensions of the sensors which must be sized according to the application for which it is required. A key limitation in the miniaturization of these biomedical implants relates to the size of the power supply required to drive an active device with even limited functionality. In some applications no available battery is sufficiently small for the space available to the implant.

In order to deal with these problems, researchers have been investigating over last decades in a new technique to power those implants to function correctly. Here is where the concept of Wireless Power Transfer (WPT) appears. Because of its working lifetime is the same of the electronics (15-20 years), it is much cheaper than traditional batteries.WPT is clean, controllable and can be available 24/7. Moreover,it can provide sufficient miniturization of the in-body sensor if implemented properly.

1.3 Thesis Outline

This research project studies the feasibility of a wireless communication link from an in-body sensor to the body surface in the upper torso. The energy required by the transmitter is provided by wireless energy transfer from the body surface. The main goals of this thesis are:

- To study on the energy transfer feasibility and to achieve the best procedure to supply the power required for an in-body sensor wirelessly placed in the heart.

- To investigate the requirements to support the communication through the human body as well as the main components in the telemetry system in order to characterize the channel model.

As a result of this research we will try to prove the existence of a valid energy transfer system for implanted sensors in human body which fulfils all the requirements for a wireless communication link.

The rest of the report is organized as follows:

In Chapter 2, diverse types of Wireless Power Transfer are described. Discussions on limitations and health related issues are also presented.

In Chapter 3, the resonance-based wireless power transfer circuit is exhaustively analyzed.

In Chapter 4, the data link requirements are studied and a communication channel is defined according to a briefly study of the State of the Arts in order to assess the wireless telemetry system.

The research work is summarized in Chapter 5. Conclusions and suggestions for future work are presented.

Chapter 2

Wireless Power Transfer

Wireless Power Transfer (WPT) is the propagation of electrical energy from a power source to an electrical load without the use of interconnecting wires. Wireless transmissions is useful in cases where interconnecting wires are difficult, dangerous, or non-exist. WPT is becoming popular for induction heating, charging of consumer electronics (electric toothbrush, Wii charger), radio frequency identification (RFID), contact-less smart cards, biomedical implants, and even for transmission of electrical energy from space to earth.

However, wireless power transfer is not strange to human beings, since the first demonstration of WPT dates back to 1889, performed by Nikola Tesla in Colorado Springs, Colorado. In his experiment, 200 incandescent lamps were lightened when powered by a base station 26 miles away, thereby inventing the famous Tesla coils which can transfer power wirelessly [1],[2]. Thereafter, researches on wireless power transfer began. In 1964, William C. Brown proposed a point-to-point wireless power transfer scheme on the basic of microwave beams [3]. In 1968, American engineer Peter Glaser presented a concept of a space solar power station, and further conceived that solar energy could be converted into electric energy first, and then transmitted to the Earth in the form of microwaves [4]. Huge progress took place in the solar power satellite (SPS) project during the 1970s [5], which indicated that human beings had realised how important was this spatial electric energy transfer technology. In 2007, wireless power transfer shocked the world again, the research team headed by Professor Marin Soljacic of MIT proposed strongly coupled magnetic resonance (SCMR), and they were able to transfer 60 watts wirelessly with approximately 45% efficiency over distances in excess of 2 meters [6]. Subsequently, Intel and Qualcomm also demonstrated their wireless power transfer systems, which indicated that this novel technology would soon appear in our daily life.

2.1 Types of Wireless Power Transfer

Wireless power transfer technology can be classified into three kinds in accordance with the working principles: electromagnetic induction, electromagnetic radiation and acoustic waves. Electromagnetic induction can be of two types: inductive coupling (electrodynamic) or capacitive coupling(electrostatic). Electromagnetic radiation can be divided into microwave power transfer (MPT) and laser. The types of WPT are shown in following Figure 2.1.

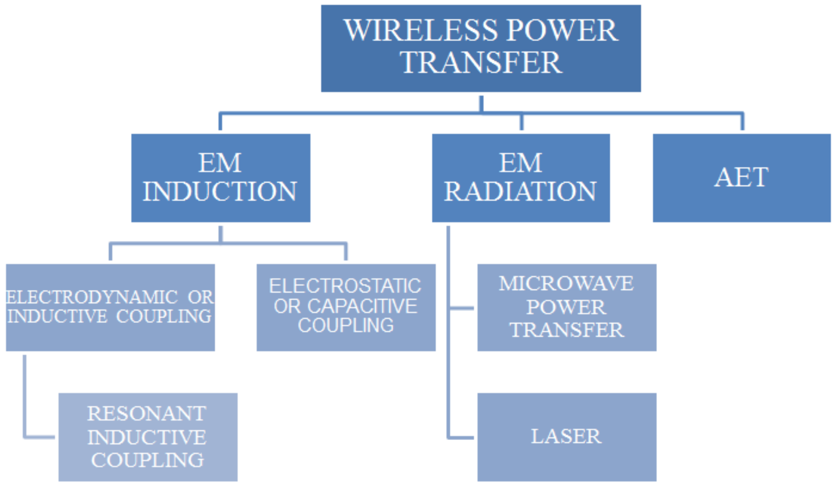

Figure 2.1: Types of wireless power transfer.

2.1.1 Electromagnetic Induction

It is the production of an induced voltage in a circuit which is excited by means of the magnetic flux. The condition for an induced current to flow in a closed circuit is that the conductors and the magnetic field must rotate relative to each other. It is a near-field technique which means that the area within is about 1 wavelength, λ, of the antenna. In this region the oscillating electric and magnetic fields are separate and power can be transferred via two different ways as detailed below. These fields are not radiative meaning the energy stays within a short distance of the transmitter.

- Electrodynamic or inductive coupling: It is the oldest and most widely used wireless power technology where power is transferred using near field radiation by magnetic fields.
  - Resonant inductive coupling: Also called electrodynamic coupling and strongly coupled magnetic resonance (SCMR), is a form of inductive coupling in which power is transferred by magnetic fields between two resonant circuits, one in the transmitter and one in the receiver. Inductive coupling and resonant inductive coupling are discussed in detail in Chapter 3.
- Electrostatic or capacitive coupling: In this context, the power is transferred via electric fields by electrostatic induction between metal electrodes. The transmitter and receiver electrodes form a capacitor, with the intervening space as the dielectric. Capacitive coupling has only been used practically in a few low power applications, because the very high voltages on the electrodes required to transmit significant power can be hazardous and can cause unpleasant side effects. In addition, in contrast to magnetic fields, electric fields interact strongly with most materials, including the human body, due to dielectric polarization [7].

2.1.2 Electromagnetic Radiation

Electromagnetic radiation is a far-field technique, beyond about 1 wavelength (λ) of the antenna, where the electric and magnetic fields are perpendicular to each other and propagate as an electromagnetic wave. It is a radiative technique meaning the energy leaves the antenna whether or not there is a receiver to absorb it. In general, visible light (from lasers) and microwaves (from purpose-designed antennas) are the forms of electromagnetic radiation best suited to energy transfer.

- Microwave power transmission: Microwave signal is used to transmit directional power to a large distance, usually in kilometers. Rectifying antennas are used to convert the energy back to electricity.
- Laser: Power can be transmitted by converting electricity into a laser beam which in then pointed at a solar cell receiver. That receiver can convert the laser beam into electricity.

Because of its high power density and good orientation features, electromagnetic radiation mode is usually suitable for the long distance transfer applications, especially for the space power generation or military applications. However, its transfer efficiency is severely affected by the meteorological or topographical conditions, and the impacts on creatures and ecological environment are unpredictable. Hence, wireless power transfer based on electromagnetic radiation mode is not appropriate for the civilian use.

2.1.3 Acoustic Energy Transfer

Summarizing preceding types of WPT, microwaves and lasers dominate in the long-range wireless power transfer applications, and electromagnetic induction dominates in the short-range wireless power transfer applications. Nevertheless, recent researches have been investigated in another approach of WPT based on sound waves called acoustic energy transfer (AET) [8]. AET is a new emerging method of transferring energy wirelessly which exploits vibration or ultrasound wave. Basically this technique is applied using ultrasonic transducer, where *ultrasonic* refers to any frequency that is beyond human hearing which is greater than approximately 20 kHz. This kind of systems offer certain advantages in the face of traditional WPT systems as explained next.

A typical acoustic energy transfer system basically consists of primary and secondary unit where both sides comprise of ultrasonic piezoelectric transducer and divided by a transmission medium as shown in Figure 2.2.

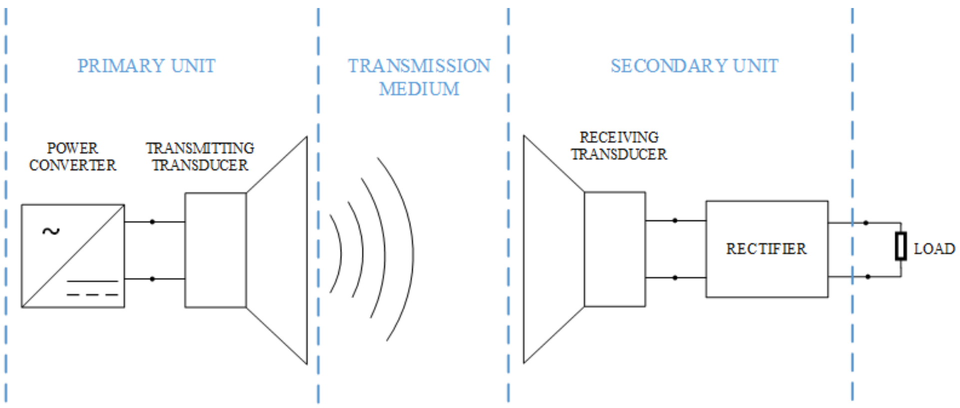

Figure 2.2: Acoustic Energy Transfer system scheme.

At the primary unit, the power converter used to drive the amount of power needed by the primary transducer. The primary transducer transforms electrical energy into a pressure or acoustic waves. It generates wave in the form of mechanical energy and propagates through a medium. The secondary transducer is placed at a point along the path of the sound wave for the inverse process of converting back into electrical energy which can be used for powering up an electrical load.

Although others WPT systems were established earlier years ago, AET has benefits in some traits. It is a good alternative to inductive energy transfer because of the absence of electromagnetic fields and the possibility of using a miniature receiver. One of the main advantages of AET in comparison to WPT based on electromagnetic fields lies in the much lower propagation speed of acoustic waves in air (c_0) with respect to the electromagnetic (EM) propagation velocity. Therefore, the sound waves have a smaller wavelength for a given frequency than their EM counterpart. This in turn means that the transmitter and receiver can be several orders of magnitude smaller for a given directionality of the transmitter. Alternatively, if the desired transmitter and receiver dimensions are given, then the frequency that is used in an AET system can be much lower than that of the EM system. Accordingly, losses in the driving power electronics will be much lower.

In [9] the authors compared the attainable energy transfer efficiencies of a biomedical AET system and an inductive coupling system. They concluded that AET outperforms inductive coupling for large distances, between source and receiver, and for smaller implants. Thus it is well suited for biomedical applications. However AET is still in its infancy and these results are not enough suitable for our application of interest, as can be seen in Table 2.1

Table 2.1: Efficiency vs distance for different energy transfer systems for a receiver of 10 mm diameter.

Distance	Efficiency	
	AET	WPT
d = 1 cm	$\eta = 39\%$	$\eta = 81\%$
d = 10 cm	$\eta = 0.2\%$	$\eta = 0.013\%$

The major contender for AET is of course inductively coupling, being the *de facto* standard, with good reason since it usually performs very well. Nevertheless, acoustic energy transfer is a technique with many capabilities to consider in the future of biomedical applications.

2.2 Limitations of Wireless Power Transfer

There are a number of limitations to the full implementation of wireless energy transfer:

- **Size**: The size of the transmitter or the receiver sometimes becomes too large to implement in a smaller systems.
- **Range**: The range of wireless energy transfer is just a few meters, which represents a major hurdle towards its practical implementation.
- **Frequency of operation**: Selection of the carrier frequency is another challenge which could be governed by a few considerations: compactness of the implant, data

transmission rate, absorption in human's tissue and radio frequency allocation by state authorities for Industrial, Scientific and Medical (ISM) application. To obtain reasonable small and flat induction coils of a high quality factor, frequencies above 5 MHz are to be preferred. On the other hand, the absorption by tissues increases with frequency. Thus, as a rule o thumb the carrier frequency should not be higher than about 50 MHz.

- **Efficiency**: Typical efficiency of wireless energy transfer ranges between 45% and 80% and is less efficient than conventional wire based energy transfer methods.

2.3 Health Related Issues

A common issue about wireless power transfer is human safety considerations. It is essential to consider the associated health risks in designing WPT system for biomedical applications. In this section, we will discuss what the human safety limits are, where they come from, and how it is established that wireless power systems conform to these safety limits.

The safety limits for human exposure to EM fields are determined by on-going reviews of scientific evidence of the impact of electromagnetic fields on human health. The World Health Organization (WHO) is expected to release a harmonized set of human exposure guidelines in the near future. In the meantime, most national regulations reference, and the WHO recommends, the human exposure guidelines determined by the Institute of Electrical and Electronic Engineers (IEEE) and by the International Commission on Non-Ionizing Radiation Protection (ICNIRP) [10]. The recommendations are expressed in terms of basic restrictions (BRs) and maximum permissible exposure (MPE) values. The BRs are limits on internal fields, specific absorption rare (SAR), and current density; the MPEs, which are derived from the BRs, are limits on external fields and induced and contact current. The safety factors are conservative so that exposures that exceed the BR or MPE are not necessarily harmful. The safety factors incorporated in the MPEs are general greater than the safety factors in the BRs. Thus, it is possible to exceed an MPE while still complying with the BRs.

In the most recent reviews of the accumulated scientific literature, both the IEEE and ICNIRP groups have concluded that there is no established evidence showing that human exposure to radio frequency (RF) electromagnetic fields causes cancer, but that there is established evidence showing that RF electromagnetic fields may increase a person's body temperature or may heat body tissues and may stimulate nerve and muscle tissues. This is referred as "thermal" effect [11] - [13]. It has been known for many years that the exposure to very high levels of RF radiation can be harmful due do the ability of RF energy to rapidly heat the biological tissues. Tissue damage in humans could occur during exposure to high RF levels because of the inability of the body to deal with or dissipate the excessive heat that could be generated. Federal Communications Commission (FCC) regulates the time and the amount of exposure of the electromagnetic waves to human tissues at various frequencies. American National Standards Institute (ANSI) standard C95.1-1982 sets the electromagnetic field strength limits for the general public for frequencies between 300 kHz and 100 GHz. This standard is superseded by the IEEE standard C95.1-1991, which sets the electric and the magnetic field strength limits for the general public for frequencies between 3 kHz and 300 GHz.

This last standard states the specific absorption rate (SAR) which is the quantity used to measure how much energy is actually absorbed in a body when exposed to RF EM field. It is defined as the power absorbed per mass of the tissue and has units of watts per kilogram (W/kg) or milliwatts per gram (mW/g). Mathematicaly, at any point in the human body, the SAR for a sinusoidal excitation can be calculated as:

$$SAR(x, y, z) = \frac{\sigma(x, y, z)E^2(x, y, z)}{2\rho(x, y, z)} \tag{2.1}$$

where σ is the conductivity (in S/m), E is the electric field amplitude (V/m) and ρ is the tissue density (kg/m^3).

In the case of whole-body exposure, a standing human adult can absorb RF energy at a maximum rate when the frequency of the RF radiation is in the range of about 80 MHz and 100 MHz, meaning that the whole-body SAR is at a maximum under resonance conditions. Because of this resonance phenomenon, RF safety standards are generally most restrictive for these frequencies. SAR should be within the tolerable acceptable range for biological tissue. A whole-body average SAR of 0.4 W/kg has been set as the restriction that provides adequate protection for workers in controlled environments (also called occupational exposure), and a SAR limit of 0.08 W/kg for the general public [11] while the FCC limit for public exposure from cellular telephone is an SAR level of 1.6 W/kg [14].

Note that the 0.08 W/kg limit is the whole body average, and corresponds to effects when a person's whole body is exposed to an electromagnetic field. However, under conditions of non-uniform or localized exposure, it is possible that the temperature of certain areas of the body may be raised by more than 1°C, even though the average field does not exceed the whole body SAR limit. To accommodate these circumstances, recommendations are also made for limiting the localized field exposure. Without question, the power dissipation characteristics of implanted electronic systems will have increasing importance for the design of future implantable devices.

Chapter 3

Inductive Coupling

Inductive link is a common method of wireless powering of implantable biomedical electronics and data communication with the outer world. Generally the inductive link for biomedical applications involving implantable devices consists of two coaxially aligned circular coils, of which one coil is meant to reside inside the human body, while the other one to be placed in an external unit located just outside of the body. The link provides a means for transferring electrical power from the external to the internal unit that can be used by the implantable sensor interface electronics and communications module via transformer action. The same link or a different one can be used for transmission of digital data from the implant to the external unit, which is known as backward or reverse telemetry.This will be explained in Chapter 4.

The use of this technique to wirelessly transfer energy across short distance is, however, expected to see an explosive growth over the next decade. Achieving high power energy transfer is necessary in high-power implantable microelectronic devices to not only reduce the size of the external energy source that should be carried around by the patient but also to limit the tissue exposure to the AC magnetic field, which can result in excessive heat dissipation if it surpasses safe limits, and to minimize interference with near by electronics.

3.1 Theory of the Inductively Power Coils

Tipically, a inductive link is formed by a loosely coupled transformer consisting of a pair of coils that are usually placed in a coaxial arrangement as shown in Figure 3.1. The external, also called primary coil, is excited by an alternating current (AC), and thus an EM field is produced with its magnitude dependent on the dimensions of the coil, the drive current and the frequency operation.

A portion of the alternating flux lines generated this way link to the internal, also called secondary coil, and the change in flux linkage produces a voltage in the secondary coil, which is proportional to the rate of change of the flux and the number of turns in the secondary coil (Faraday's Law). If the number of turns is n and the magnetic flux linking each turn is ψ_m, then the induced voltage for the circuit can be written as,

$$V = n\frac{d\psi_m}{dt} \tag{3.1}$$

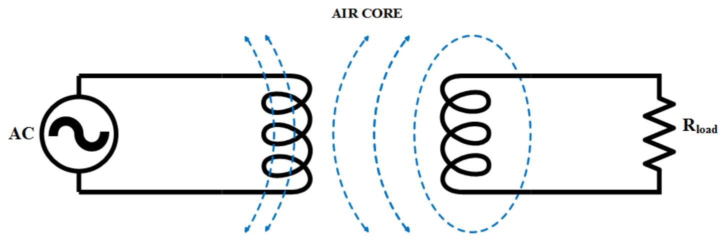

Figure 3.1: An inductive link produced by alternanting EM.

The fundamental concepts of designing an inductive link are presented in this section following the work reported by W. H. Ko *et al.* [15]. The theory and practical design equations were developed using the basic inductively coupled circuit (Figure 3.2) and its equivalent secondary circuit (Figure 3.3).

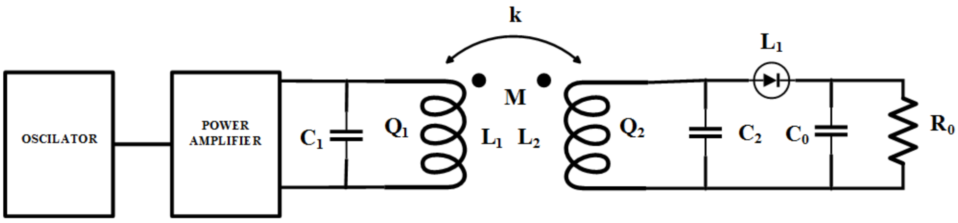

Figure 3.2: Basic inductively coupled circuit.

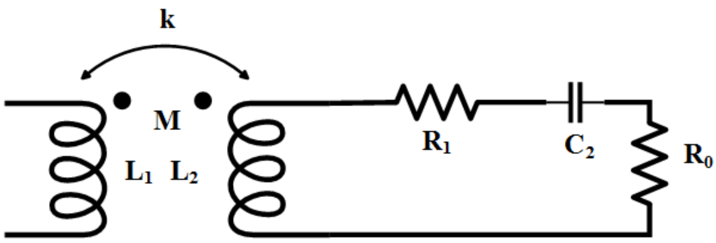

Figure 3.3: Equivalent secondary circuit for basic inductively coupled circuit.

As can be seen in Figure 3.2, L_1 and L_2 are the inductances of the primary and secondary coils, respectively. In the same figure, M is the mutual inductance of the coils, Q_1 and Q_2 are the unloaded quality factors of the primary and secondary coils, respectively and k is the mutual coupling which has value ranging from 0 to 1. This parameters are defined as:

$$M = k\sqrt{L_1 L_2} \tag{3.2}$$

$$Q_1 = \frac{\omega L_1}{R_1}; Q_2 = \frac{\omega L_2}{R_2} \tag{3.3}$$

The equivalent AC load resistance R which will dissipate an amount of AC power equivalent to the DC power in load resistance R_0 is $R = \frac{R_0}{2}$. The equivalent AC series resistance R_L due to the load R_0 is:

$$R_L = \frac{(\omega L_2)^2}{R} = \frac{2(\omega L_2)^2}{R_0} \tag{3.4}$$

The total equivalent series resistance in the secondary tank circuit is $R_2 + R_L$, where R_2 is the series resistance of the unloaded secondary tank circuit and R_L is the load resistance. The equivalent resistance R_e, reflected back into the primary coil is:

$$R_e = \frac{(\omega M)^2}{R_2 + R_L} = \frac{Rk^2 Q_1 Q_2}{R + {Q_2}^2 R_2} R_1 \tag{3.5}$$

Therefore, the equivalent circuit referred to the primary side is shown in Figure 3.4

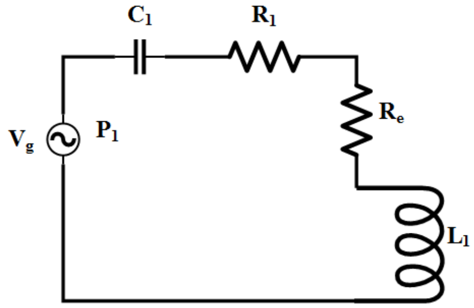

Figure 3.4: Primary referred equivalent circuit of basic inductively coupled circuit.

From the primary equivalent circuit shown in 3.4, the efficiency of the circuit at resonance can be derived as:

$$P_i = \frac{1}{2}\left(\frac{|V_g|^2}{R_e + R_1}\right) \tag{3.6}$$

$$P_i = P_o = \frac{R_L}{R_L + R_2} \frac{R_e}{R_1 + R_e} P_i \frac{1}{2}\left(\frac{|V_g|^2}{R_e + R_1}\right) \tag{3.7}$$

$$\eta = \frac{P_o}{P_i} = \frac{k^2 Q_1 {Q_2}^3 R_2 R}{(R + Q_2)^2 R_2((1 + k^2 Q_1 Q_2) R + {Q_2}^2 R_2)} \tag{3.8}$$

As can be seen in (3.8), the overall efficiency depends on coupling factor, k. However, value of k is dependent on size of the coil, coil spacing, and lateral and angular misalignment.

If the derivative of the efficiency expressed also in equation (3.8) is taken with respect to R_2 (for a given set of k, Q_S and R) and set to zero, then the optimum value of R_2 required for maximum efficiency is found to be:

$$R_{2opt} = \frac{R\sqrt{1 + k^2 Q_1 Q_2}}{{Q_2}^2} \tag{3.9}$$

Substituting this result into equations (3.5) and (3.8) yields the optimum efficiency of the circuit:

$$\eta_{opt} = \frac{k^2 Q_1 Q_2}{(1 + \sqrt{1 + k^2 Q_1 Q_2})^2} \tag{3.10}$$

This equation confirms that the optimum efficiency increases as $k^2 Q_1 Q_2$ increases, ant therefore the first and the foremost design consideration in an inductive link design is the attainment of the highest possible unloaded Q and k. These two essential parameters are functions of the shape, size and relative position of the coils. Putting (3.3) in the equation (3.10) results in:

$$\eta_{opt} = \frac{k^2\omega^2(\frac{L_1L_2}{R_1R_2})}{\left(1+\sqrt{1+k^2\omega^2(\frac{L_1L_2}{R_1R_2}}\right)^2} \tag{3.11}$$

The energy stored in magnetic field is $U_B = \frac{1}{2}LI^2$ while the energy stored in electric field can be expressed as $U_E = \frac{1}{2}CV^2$. Thus the total energy in a LC circuit can be expressed as:

$$U = U_B + U_E = \frac{1}{2}LI^2 + \frac{1}{2}CV^2 \tag{3.12}$$

At resonance condition, collapsing magnetic field of the inductor generates an electric current in its windings that charges the capacitor, and then the discharging capacitor provides an electric current that builds the magnetic field in the inductor. In resonance mode energy is transferred between inductor and capacitor and they are equal: $U_B = U_E$

$$Energy = \frac{1}{2}LI^2 = \frac{1}{2}CV^2 \tag{3.13}$$

From equations (3.11) and (3.13) it can be concluded that the inductance is a very important design parameter of the inductive link. Higher value of inductance is required for a proper performance with higher energy and better efficiency.

3.2 Resonant Inductive Coupling

In an inductively coupled power-transfer system consisting in two coils, power-transfer efficiency is a strong function of the quality factor (Q) of the coils as well as the coupling between the two coils as seen. Hence, the efficiency depends on the size, structure, physical spacing, relative location and the properties of the environment surrounding the coils. The coupling between the coils decreases sharply as the distance between the coils increases and causes the overall power-transfer efficiency to decrease monotonically.

Recently, the Massachusetts Institute of Technology (MIT) has proposed a new scheme based on strongly coupled magnetic resonances (SCMR)[6], thus presenting a potential breakthrough for midrange wireless energy transfer. A scheme of this system is shown in Figure 3.5.

The resonance–based method is based on the fact that two same–frequency resonant objects tend to couple, while interacting weakly with other off–resonant environmental objects, and even more strongly where the coupling mechanism is mediated through the overlap of the non–radiative near–field of the two objects. This resonant energy–exchange can be modeled by the appropriate analytical framework called coupled–mode theory (CMT) [16], but it can also be transformed into a simple circuit-based model know as reflected load theory (RLT) [17]. The autors in [17] have claimed that although CMT is a more physics-based approach and RLT is circuit-based, both the methods produce the same results for inductively-coupled power transfer. However, CMT produces relatively simplified equations but works only for very low coupling and high-Q coils. In the RLT

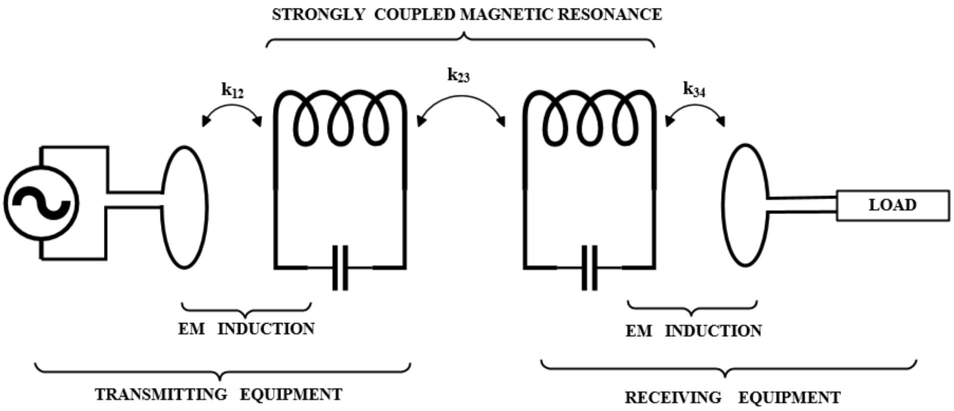

Figure 3.5: Strongly coupled magnetic resonance scheme.

method, the resistive load R_{load} is transformed into a reflected load onto the primary loop at resonant frequency. It has been shown that the highest power transfer energy across such inductive links can be achieved when all LC-tanks are tuned at the same resonance frequency.

This resonant-based power delivery is an alternative wireless power-transfer technique that typically uses four coils, namely: driver, primary, secondary and load coils.

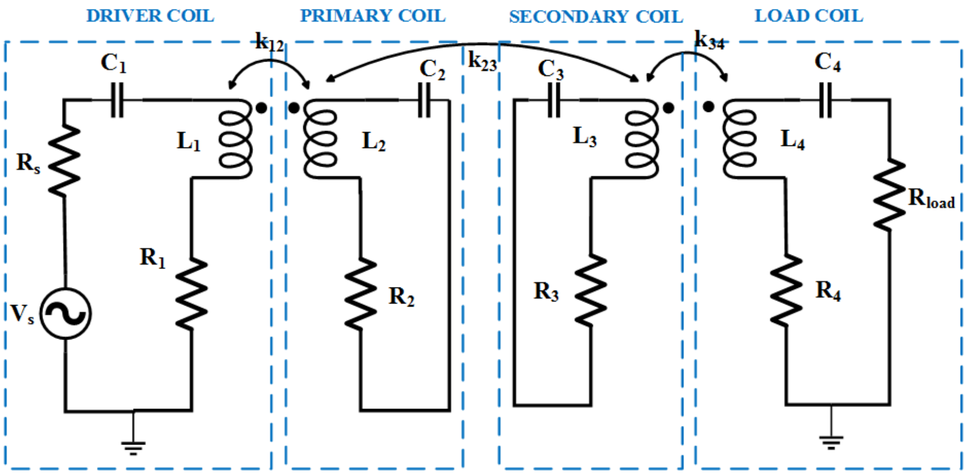

Figure 3.6: Electrical model of the resonance-based power-transfer circuit.

Figure 3.6 shows the equivalent circuit for the system in terms of the lumped circuit elements L, R and C. The transmitter drive loop and multiturn resonator are modeled as inductors L_1 and L_2, and the receiver multiturn resonator and drive loop are modeled as inductors L_3 and L_4, respectively. Capacitors $C_1 - C_4$ are selected such that each magnetically coupled resonator will operate at the same resonant frequency according to:

$$f_{res} = \frac{1}{2\pi\sqrt{L_i C_i}} \tag{3.14}$$

The resistors $R_1 - R_4$ represent the parasitic resistances of each resonator, and are typically less than 1Ω. Each resonant circuit is linked by the coupling coefficients k_{12}, k_{23}

and k_{34}. These coupling coefficients are typically an order of magnitude greater than the cross coupling terms (k_{13}, k_{14} and k_{24}). The relationship between the coupling coefficient and the mutual inductance between each resonator is given in:

$$k_{ij} = \frac{M_{ij}}{\sqrt{L_i L_j}} \tag{3.15}$$

When circuit theory in the form of Kirchoff's Voltage Law (KVL) is applied to the system, we achieve the following matrix that defines the relationship between voltage applied to the driver coil and current through each coil:

$$\begin{bmatrix} V_s \\ 0 \\ 0 \\ 0 \end{bmatrix} = \begin{bmatrix} Z_{11} & Z_{12} & Z_{13} & Z_{14} \\ Z_{21} & Z_{22} & Z_{23} & Z_{24} \\ Z_{31} & Z_{32} & Z_{33} & Z_{34} \\ Z_{41} & Z_{42} & Z_{43} & Z_{44} \end{bmatrix} \begin{bmatrix} I_1 \\ I_2 \\ I_3 \\ I_4 \end{bmatrix} \tag{3.16}$$

where

$$Z_{mn} = \begin{cases} R_n + j\omega L_n + \frac{1}{j\omega C_n}, & for \;\; m = n \\ j\omega M_{mn}, & for \;\; m \neq n \end{cases} \tag{3.17}$$

Therefore, using KVL and flux linkages the transfer function for the circuit model can be obtained as:

$$\frac{V_L}{V_S} = \frac{j\omega^3 k_{12} k_{23} k_{34} L_2 L_3 \sqrt{L_1 L_4} R_{load}}{({k_{12}}^2 {k_{34}}^2 L_1 L_2 L_3 L_4 \omega^4 + Z_1 Z_2 Z_3 Z_4 + \omega^2 ({k_{12}}^2 L_1 L_2 Z_3 Z_4 + {k_{23}}^2 L_2 L_3 Z_1 Z_4 + {k_{34}}^2 L_3 L_4 Z_1 Z_2))} \tag{3.18}$$

Being:

$$\begin{aligned} Z_1 &= R_1 + R_S + j\omega L_1 + \frac{1}{j\omega C_1} & Z_2 &= R_2 + j\omega L_2 + \frac{1}{j\omega C_2} \\ Z_3 &= R_3 + j\omega L_3 + \frac{1}{j\omega C_3} & Z_4 &= R_4 + R_{load} + j\omega L_4 + \frac{1}{j\omega C_4} \end{aligned} \tag{3.19}$$

The transfer function neglects the cross coupling terms due to the small size of the driver and load coils and relatively large distances between the respective coils.

In addition to this, another essential parameter to take into account is the ratio of the output power over the input power, which determines the power transfer efficiency. It is an important parameter in wirelessly-powered biological implants because of safety issues and standards regarding tissue exposure to RF electromagnetic radiation [11]. Therefore, maximizing the efficiency will guarantee a relatively high power output at the load even with a relatively low power wave that has to travel through the body.

However, the optimization of the efficiency does not automatically optimize the output power. In cases where we are safely below the exposure limit, and we require a high power delivered to the load, we can choose to maximize it, even if at the cost of a lower power transfer efficiency.

3.2.1 Impedance matching for maximum power transfer

The impedance matching method adopted in many wireless power transfer projects is based on the maximum power transfer theorem. In general, any wireless power transfer system, regardless of it being a two-coil or four-coil system can be represented as an equivalent circuit as shown in Figure 3.7.

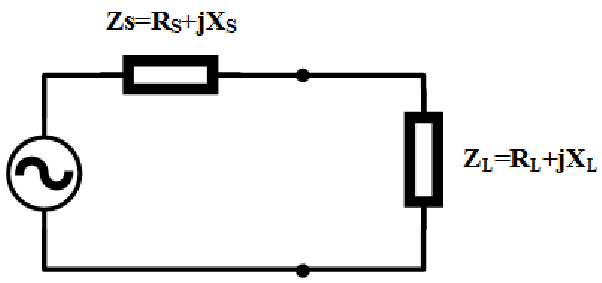

Figure 3.7: Equivalent circuit for an AC power source and an equivalent load.

The maximum power transfer principle requires impedance matching between the source and the load. If the source impedance is $Z_S = R_S + jX_S$ and the load impedance is $Z_L = R_L + jX_L$, then maximum power can be delivered to the load if $R_S = R_L$ and $X_S = -X_L$. Recent mid-range wireless power transfer research based on the four-coil systems adopts this approach. However, it should be noted that the maximum power transfer and maximum energy efficiency concepts are not identical. The maximum power theorem applies to a situation in which the source impedance is fixed. For a given R_S, the maximum power output (i.e. maximum power transfer) is achieved when R_L is equal to R_S. When R_L is larger than R_S, the larger R_L is, the higher the energy efficiency becomes.

For impedance matching, the system energy efficiency that includes the power loss in the power source is:

$$\eta_E = \frac{i^2 R_L}{i^2 R_S + i^2 R_L} = \frac{R_L}{R_S + R_L} = 0.5 \tag{3.20}$$

For this reason, when maximum power transfer occurs at impedance matching, the maximum system energy efficiency under the maximum power transfer approach cannot exceed 50% as shown in (3.20). Therefore, at least half of the power will be dissipated in the source resistance (R_S) if the maximum power theorem is adopted.

Researchers with an RF background are familiar with the use of the scattering matrix and two-port network approach as shown in Figure 3.8 for analyzing wireless transfer systems. It is important to differentiate the terms system energy efficiency η_E and transmission efficiency η_T. The system energy efficiency refers to the ratio of the output power P_3 and total input power P_1 from the power source. Its calculation includes the power loss in the power source. The transmission efficiency is the ratio of the output power P_3 and available power from the output of the power source for Port-1 P_2, and does not include the power loss in the power source. Therefore, high transmission efficiency does not necessarily imply high system energy efficiency because the source resistance can consume a significant amount of power if the impedance matching or maximum power transfer concept is adopted.

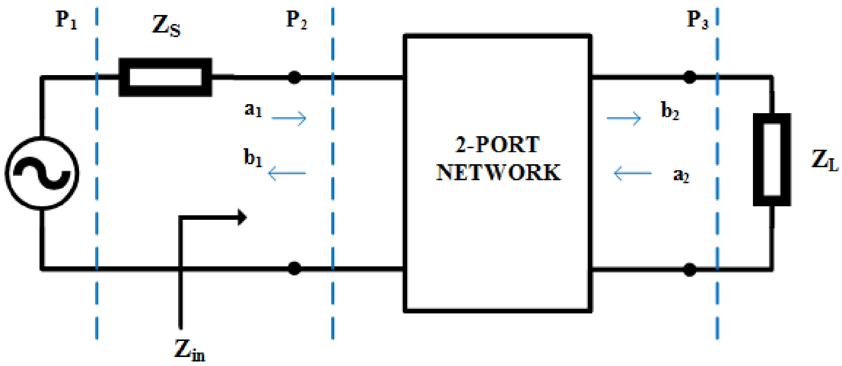

Figure 3.8: Schematic of a two-port network.

$$\eta_E = \frac{P_3}{P_1} \qquad \eta_T = \frac{P_3}{P_2} \tag{3.21}$$

The scattering parameters are used to analyze the forward gain of the mid range wireless systems. For a two-port system shown in Figure 3.8, the relationship of the incident and reflected waves can be represented by (3.22), where a_1 and a_2 are the incident power waves on Port-1 and Port-2, respectively, and b_1 and b_2 are the reflected power waves.

$$\begin{bmatrix} b1 \\ b2 \end{bmatrix} = \begin{bmatrix} S_{11} & S_{12} \\ S_{21} & S_{22} \end{bmatrix} \begin{bmatrix} a_1 \\ a_2 \end{bmatrix} \tag{3.22}$$

According to the definition of S-parameters, if Port-2 is terminated with a load identical to the system's source impedance, then by the maximum power transfer theorem, b_2 will be totally absorbed by the load making the reflected power a_2 equal to zero.

The forward voltage gain S_{21} (also known as the transmission coefficient) is defined as:

$$S_{21} = \frac{b_2}{a_1} = \frac{{V_2}^-}{{V_1}^+} \tag{3.23}$$

Therefore, the S_{21} parameter, which is the ratio of the output and input voltage values, has been used as an indicator for transmission performance of mid-range wireless power systems. The maximum power transfer condition can be met by maximizing this parameter.

3.2.2 Maximum Energy Efficiency

The maximum energy efficiency principle aims at maximizing the energy efficiency in the power transfer process. Previous researchs ([18],[19]) and (3.20) indicate that high efficiency can be achieved by using a power source with very small source impedance. If R_S is very small, the i^2R_S loss is very small and most of the power goes to the load (i^2R_L), resulting in high energy efficiency.

For a WPT system based on the use of coil resonators, since air-core resonators are usually used for mid-range wireless power transfer, there is no magnetic core loss. Assuming the capacitor's equivalent series resistance is negligible and nonradiative power transfer is employed, the only types of losses are the conduction loss due to the AC resistance of the coils and the power loss in the source resistance. Any loss from unwanted stray loads will decrease the energy efficiency. The control objective is therefore to maximize the system energy efficiency function (3.24). In order to achieve high energy efficiency, a power source with very low source resistance R_S will be employed. The value of R_S will not be matched with the equivalent load. Litz wire [1] or copper tube will be considered for reducing the AC winding resistance (R_1 ,..., R_N) under high-frequency operation. Investigations of using superconductors are also underway to further improve the system energy efficiency. In principle, system energy efficiency higher than 50% is possible if this approach is adopted. Therefore, this approach is suitable for relatively high-power applications where η_E is system energy efficiency, i_N and R_n are the current in and the AC winding resistance of the nth coil respectively; R_L is the load resistance.

$$\eta_E = \frac{{i_N}^2 R_L}{{i_N}^2(R_S + R_1) + {i_2}^2 R_2 + ... + {i_N}^2(R_N + R_L)} \tag{3.24}$$

The maximum energy efficiency operation relies on high-magnetic coupling coefficients between the coil resonators, which increase with the quality factor and decrease with the transmission distance.

3.3 Factors Affecting Inductive Link Perfomance

In this section some of the factors that mostly influence the performance of an inductive link are discussed. Most of them are considerably interdependent, and consequently, extensive trade-offs are associated with the design choices.

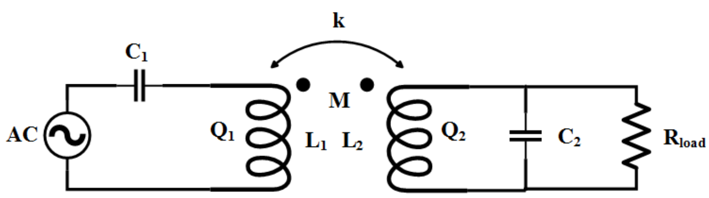

Figure 3.9: Schematic of a basic inductive link based on series-parallel resonance.

- **Mutual inductance (M):** It is a measure of the extent of magnetic linkage between current-carrying coils. The maximum value of M that can exist between two coils of inductance L_1 and L_2 is $\sqrt{L_1 L_2}$ and this occurs when all the flux of one coil links with all the turns of the other.

- **Self inductance:** The self inductance of a current-carrying coil is the amount of magnetic flux through the cross-sectional area that it encloses.

[1]It is a type of cable used in electronics to carry AC and is designed to reduce the skin effect and proximity effect losses in conductors used at high frequencies. It consists of many thin wire strands, individually insulated and twisted or woven together, following one of several carefully prescribed patterns often involving several levels [20].

- **Coupling coefficient (k):** Is the ratio of the mutual inductance present between two inductances to the maximum possible value. It is a dimensionless quantity which ranges from 0 to 1, as seen before.

- **Parasitic capacitance:** Parasitic or stray capacitance between turns is a common issue with inductors. It affects the inductor operation by causing self-resonance and limiting the operating frequency of the inductor.

- **AC Resistance:** At high frequencies, skin and proximity effects increase the effective series resistance, which decreases the quality factor of the inductor coils. In order to reduce its AC resistance, the coils are commonly made by using multistrand Litz wire.

- **Link efficiency (η):** It is defined as the ratio of the power delivered to the load to the power supplied to the primary coil. Its expression has been previously presented.

- **Diameter of the coils:** Since the mutual and the self-inductances of the coils vary proportionally with their diameters, the link efficiency increases with increasing diameters. The voltage gain on an inductive link also depends on the diameters of the receiver and transmitter coils. For implantable systems the limits on the receiver coil size are usually more rigid than those of the transmitter coil.

- **Spacing between coils (d):** This can significantly affect the coupling. The coil spacing can be varied by keeping one coil fixed while moving the other coil along the axis. The mutual inductance varies inversely proportional to d.

- **Misalignment:** Variation in coil alignment also changes the mutual inductance and the link gain. Two types of misalignment can be present in the system: lateral and angular misalignments.

  - <u>Lateral:</u> The centers of the coils are displaced in the horizontal direction. The planes of the coils are still parallel to each other. Authors in [21] concluded that as the lateral misalignment increases, the output voltage decreases due to the reduction in mutual inductance which is inversely proportional to the lateral misalignment.
  - <u>Angular:</u> In this case, the centers of the coils are kept along the same axis, but their planes are tilted to form an angle (φ). In [21] was observed that as φ is increased, output voltage drops due to the reduction in mutual inductance, which is inversely proportional to φ. When the planes of the coils become orthogonal to each other ($\varphi = 90^o$), there is no mutual inductance at all between the coils and output voltage drops to zero.

- **Quality factor (Q):** Quality factors of the primary and the secondary coils can change the link efficiency of a typical inductive link. Therefore, reasonably high Q values are desired at the frequency of operation in order to achieve satisfactory power transfer. Besides the output voltage becomes sensitive to load changes when Q is low. It is defined as:

$$Q(\omega) = \omega \cdot \frac{Maximum\ Energy\ Stored}{Power\ Loss} \tag{3.25}$$

- **Frequency of operation (f):** It is a vital design parameter since determines the maximum power through human tissue. The size of the coil, M, and the voltage transfer ratio are determined by the frequency. The efficiency and the bandwidth of the link are also impacted by the choice of the frequency because Q is a function of f. Wireless-based systems should not interfere with existing communication systems. Due to this stringent requirement, medical radios tend to use ISM frequency bands with low data rate operation. Moreover, in this frequency band, amount of allowable maximum electric field through human tissue is higher than high frequency bands which is strictly controlled by FCC.

- **Frequency splitting:** This phenomenon occurs when the conditions for the maximum power theorem cannot be met at the resonance frequency of the resonators within the over coupled region, and maximum S_{21} occurs at two different frequencies. To avoid the complication of frequency splitting, adaptive matching methods based on frequency tracking have been developed [22], [23].

3.4 Harvesting Unit

So far we have only considered the transmission of power from the external circuitry to the implant coil. But the power in this form is not suitable for direct consumption by the implant. Therefore, the power harvesting unit that rectifies and regulates the output power into a constant DC form is presented in this section.

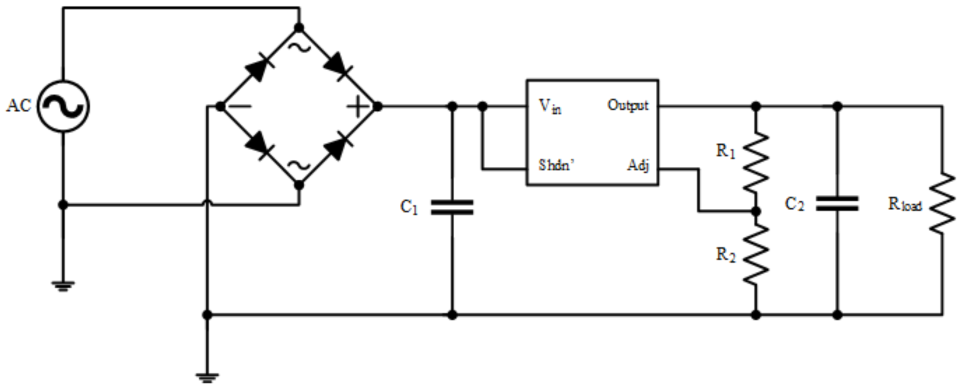

Figure 3.10: Power harvesting unit with rectifier and voltage regulator blocks.

Since the power requirements depends on the application of the implant Figure 3.10 shows an example of the harvester circuit. The power harvester is entirely made up of discrete components. The system of diodes is a full–bridge rectifier or AC–to–DC converter. Commonly used D1N914 diode have been used in the scheme above presented because it has a high switching speed and low forward voltage drop (≈ 0.6 V). Any diode with similar rating can be used for the rectifier section. The capacitor C_1 is for smoothing the output waveform, and can be calculated using the following relations:

$$V_{DC} = V_{AC} - 2V_D \qquad r \approx \frac{V_{DC}}{2fC_1R_{load}} \tag{3.26}$$

where V_{AC} is the AC input, V_{DC} is the DC output, V_D is the forward voltage of the diode and r is the height of the ripple in the output waveform.

Next we have a voltage regulator or DC-to-DC converter for forcing the output voltage to a certain constant value. Its four pins can be configured as per the requirements. In this example the *shutdown* pin is shorted with the V_{in}. The *output* and the *adjust* pin are arranged as a potential divider so that the output voltage at R_{load} can be adjusted by changing the value of R_1, thereby changing the R_1/R_2 ratio. Without the potential divider, the output voltage would be constant.

3.5 Review on previous inductive links for implanted sensors.

Wireless power transfer via strongly coupled magnetic resonance system have been presented and exhaustively analyzed. It has been demonstrated that such system is a really complex scheme. That becomes even more complicated whether talking about use this leading technique in order to power an in-body sensor in the upper torso. Several researches about WTP in the biomedical field have been reported in the literature ([24]-[30]). Table 3.1 shows different biomedical applications by using wireless power transfer, carrier frequency used and achieved efficiency of the overall system.

Table 3.1: Comparison among previous works based on WPT .

Applications	Carrier frequency	Efficiency
Blood-pressure monitoring [24]	4 MHz	N/A
Blood-glucose monitoring [25]	13.56 MHz	N/A
Capsule endoscopy [26]	8.2 MHz	26.14%
Retinal prosthesis [27]	1 MHz	67%
Neural recording [28]	4 MHz	73%
Implantable system[29]	13.56 MHz	21.9-29.32%
Neuroprosthetic implantable device [30]	13.56 MHz	56.56%

However, due to the novelty of this technique (SCMR) only a few studies have been done in our area of interest and specifically only one research about ventricular assist devices have been presented [19].

Ventricular assist devices (VAD) have traditionally been used as a stopgap measure for patients awaiting heart transplants, but more frequently are operating in patients for years. The awkward power cord that protrudes through the patient's abdomen becomes infected in many patients which implies that infection is the leading cause of rehospitalization and can be fatal. For this reason a wireless system is essential in order to deal with the risk of infection and improving the patient's quality of life.

The researchers in [19] have been applied the free-range resonant electrical delivery (FREE-D) wireless power system concept in order to use magnetically coupled resonators to efficiently transfer power across a distance to a VAD implanted in the human body, and to provide robustness to geometric changes. The (FREE-D) system allows devices to be charged in free space, without direct physical contact between the transmitter and the receiver. The key feature that distinguishes FREE-D from prior inductive schemes is the use of high-quality factor resonators combined with an automatic tuning scheme that keeps the system operating at maximum efficiency. The FREE-D system is able to adapt to variations in transmitter-receiver separation distances, orientations, and power requirements of the load.

The main idea for implementing the FREE-D system with a VAD consists of installing multiple transmitter resonators throughout the household that are hidden inside walls, floors, couches,tables and beds. Therefore, the patient is free to maneuver throughout their home while receiving wireless power from the nearest transmitter resonator. There will be two receivers resonators: one installed in an exterior vest that could be worn by the patient, and another smaller resonator implanted in the human body. The larger size of the receiver vest resonator will increase the working range of the wireless power transfer.

The researchers implemented the FREE-D system in their laboratory to test its performance with the commercially available axial pump and a VentrAsist centrifugal pump by two experimental studies. Table 3.2 shows the resonator sizes for the experimental configuration. The coils were tuned to resonate at 13.56 MHz. Even though 13.56 MHz is on the high end of the frequencies that have been traditionally adopted for inductive power transmission to implantable devices, it is still within the range in which the specific absorption rate (SAR) in the tissue is quite low compared to the thermal power dissipation in the coils.

Table 3.2: Resonator sizes for FREE-D experimental configuration.

Resonator Description	Diameter [cm]
Drive	31
Transmitter	59
Relay	59
Receiver vest	28
Implantable receiver	9.5

For their first experiment an 8-h continuous monitoring was conducted with resonators separated by a 1 m distance. The axial VAD required 8.1 W to be powered operating at a typical pump speed of 2400 r/min. They achieved a 56% system efficiency and an average efficiency of around 80% for the coils, the regulator and the rectifier of the system.

On the other hand, in their second experiment the investigators implemented the FREE-D system with the Ventracor VentrAssit centrifugal VAD while monitoring for a two-week time period. In this case, although the wireless power transfer efficiency is upwards of 90% for every pump speed, the rectifier and the regulator efficiencies significantly reduces the system efficiency due to an impedance mismatch between the rectifier and the impedance of the VAD.

As presented, wireless power transfer system for heart failures applications is still under test, thus it will require further study. Needless to say, it can be leveraged to simplify sensor systems, to power medical implants, to reduce electrical wiring in day-to-day care of the patients and improve their quality of life.

Chapter 4

Wireless Telemetry System

Central to a completely implantable system is the wireless power delivery which can sustainably power the implant for its entire lifetime but also the data telemetry to record data from and send control data to the implant. For this reason, another important feature of implantable devices is the wireless transmission of the sensor data for real-time monitoring and diagnosis. Data telemetry can be done in two ways as will be explained later; forward telemetry (data transmission from power transmitter to power receiver)and backward telemetry (data transmission from power receiver to power transmitter).

In this chapter the wireless data link is explained as well as its requirements to function properly. The communication channel is modeled and the propagation characterization of implantable antenna is presented based on the State of the Arts.

4.1 Wireless Power and Data Link

A schematic of the system is shown in Figure 2.1

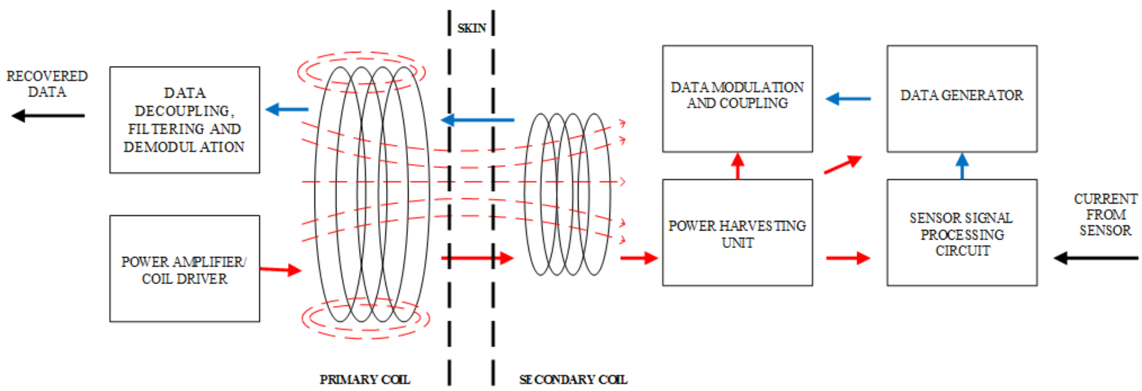

Figure 4.1: Block diagram of a WPT and backward data telemetry system .

A simple wireless power transfer and data transmission unit work as follows:

- ***Power amplifier/coil driver:*** It generates an alternating current which drives energy through the primary coil. Power is transferred from the primary coil to the secondary coil via the inductive link through the skin.

- ***Power harvesting unit:*** rectifies the received energy from AC to DC and then regulates the output voltage and supplies to the *sensor signal processing circuit*, *data generator* and *modulation and coupling* blocks.
- ***Sensor signal processing circuit:*** This block takes current from sensor as input, processes the current and sends signal to the *data generator*.
- ***Data generator:*** It converts the signal to digitally encoded data signal and sends it to the *data modulation and coupling* block.
- ***Data modulation and coupling:*** This unit takes the data as input, modulates it and couples it to the secondary coil. The data is transmitted through the inductive link back to the primary side.
- ***Data decoupling, filtering and demodulation:*** This unit decouples the data and filters ir out from the power signal and demodulates the data signal to get the information back.

4.1.1 Requirements of the link

A wireless data link for an application such as the present study must meet a number of requirements, several of which are unique to implanted systems:

- **Adequate Data Rate:** Depending on the application a certain data rate must be achieved in order to recover the information accordingly.
- **Adequate Power Transfer:** The implanted unit consumes on the order of tens of mW of power, which must be transferred over an inductive link. High power implantable devices such as VADs and artificial hearts require up to 10–30 W of power. As the inductive link is also used for data transfer, higher data rates require a wider bandwidth (and hence lower quality factor) inductive link, leading to a reduction in power transfer.
- **FCC Compliance:** Radio transmissions must not violate FCC regulations. Unlicensed operation is possible in any of the ISM bands. Specifically, the Medical Implant Communications Service (MICS) (402–405 MHz) band is the most commonly used for medical implant communications. The system is typically operating in the sleep mode at a higher ISM band and will not transmit data wirelessly at the MICS band until it receives a wakeup signal in the ISM band.
- **Size:** The implanted unit must have dimensions on the order of mm to be minimally invasive. This mandates the use of fully integrated transceivers with small, less efficient antennas.
- **Minimal Power Consumption:** The power consumed by the implanted unit must not exceed 80 mW/cm^2 to avoid tissue damage through heating effects [13]. Lower power consumption is important in terms of both the long-term performance of the device and the safety to the patient. For battery-less devices powered by an RF link, the low power restriction is also applied to ensure the electromagnetic energy radiated or backscattered by the device during wireless communication is in line with the IEEE human tissue exposure standards.

- **Fabrication Technology:** To keep fabrication cost to a minimum, it should be implemented in standard CMOS technology.

4.2 Communication Channel Modeling

As an implantable device, the sensor inside the human body must operate wirelessly. This means that the transfer of data to and from the implant must be done through wireless links. Two RF links are used for this purpose: the *forward telemetry link*,through which data is transferred toward the implanted sensor, and the *reverse or backward telemetry link*, which is used for transferring the recorded information back to the external host.

4.2.1 Forward Communication Channel

Due to its simple and easy implementation, amplitude shift keying (ASK) is the most common choice for this link. The advantage of this approach is the use of a simple circuit and its power dissipation is very low. However, this approach has a number of associated concerns.

- The RF signal received through the forward link not only carries data, it is also used for power transfer to the sensor and subsequently for on-board power generation. If amplitude modulation is chosen (with amplitude levels A_L and A_H assigned to 0's and 1's, respectively, and assuming that $A_H > A_L$), the lower amplitude level A_L must be high enough to provide sufficient voltage overhead on top of the regulated supply voltage. At the same time, for high-quality data transfer, i.e., a low bit error rate (BER), the modulation depth needs to be large enough for the data detector/de-modulator to distinguish between the two amplitude levels in the presence of different sources of amplitude noise (e.g., RF noise in the environment, motion artifacts, and misalignment of the transmitting and receiving coils). In other words, must be kept high enough for reliable power generation and has to be enough higher than that to meet data transfer requirements. This results in the transfer of needlessly high power through the tissue, which is undesirable.

- Moreover, if the amplitude is modulated with the data, variations in the amplitude make voltage regulation more challenging in terms of achieving an acceptable line regulation quality. This, in turn, translates into regulator circuit complexity and increased are for the required storage capacitance.

- Another issue is the maximum achievable bit rate, which for an ASK scheme will not be as high as for other digital modulation techniques such as frequency shift keying (FSK) and phase shift keying (PSK).

Many communication schemes for implantable devices have been demonstrated in recent literature spanning amplitude, frequency and phase modulation, and ranging from binary to higher-order encoding in complexity. One common justification for constant amplitude modulation is the need for constant power flow since on-chip energy storage may be infeasible. This is the reason why many designers choose phase or frequency shift keying modulation over amplitude modulation. The disadvantage, however, is the need for an on-chip clock and carrier synchronization in order to receive data. Integrating synchronization circuitry in the implant receiver significantly increases power consumption, which can easily offset its advantages. Therefore, a reasonable compromise is provided by an asynchronous data link as in the case of amplitude shift keying (ASK) with data encoded in the pulse width (PW) that avoids the need for clock and carrier synchronization.

4.2.2 Backward Communication Channel

Reverse data link from the implantable device to the external reader can be implemented in many different ways ranging from very complex to low-power solutions. Firstly, a dedicated transmitter with a local oscillator can be used to transmit data to the external reader. This approach allows for full-duplex communication at the cost of high complexity and power consumption. Secondly, load modulation can be also used. For low-power devices load and backscatter modulation for the reverse link are preferable. Load can be modulated in several different ways: resistive, reactive, or a combination of the two. Depending on the link transfer function and how the load is varied, the phase and/or amplitude of the carrier will be modulated. In [31], the authors provide analysis for selecting optimum load impedance that maximizes the difference between the transmitted encoded bits and thus maximizes the received signal to noise ratio (SNR) at the external reader. It is not always optimal to reflect the maximum amount of energy back to the external reader since it interferes with the power delivered to the sensor. Therefore, phase modulation is a better approach for the reverse data link. Additionally, load modulation can be easily combined with continuous-time sensing and processing that was recently described by researchers in [32]. They demonstrated a system that converts an analog waveform into a digital representation without a clock or sampling. Not only does this eliminate the need for clock generation, but also saves power during idle time intervals because the signal is not sampled when there is no activity. This is perfectly suitable for a variety of biological waveforms as they tend to have long periods of low or no activity. The modulator that is driven by the event-driven sampling will only transmit data to the external reader during physiological activity and will save energy during the idle state. Since the forward link is asynchronous and does not require a clock, the event-driven sampling with load modulation minimizes power for the reverse data link by only enabling circuitry when necessary.

To conclude this section some prior works on data telemetry for biomedical applications using wireless power transfer are summarized in Table 4.1.

Table 4.1: Prior works on data telemetry for biomedical applications with WPT.

Application	Forward Telemetry		Backward Telemetry	
	Modulation type	Data rate, Carrier	Modulation type	Data rate, Carrier
Cortical stimulator[33]	ASK-BPSK	1.507 Mbps, 13.56 MHz	LSK	1.13 Mbps, 13.56 MHz
Visual neuro-prosthesis[34]	FSK	100kbps, 1 MHz	DBPSK	15.625 kbps, 500 kHz
Peripheral neural recording[35]	ASK	N/A, 4 MHz	PWM-ASK	125kbps, 4 MHz

4.3 Propagation Characterization of Implantable Antenna: State of the Art

To the best of my knowledge only a few models that describe the radio propagation inside the human body have been published. Measuring the radio channel for in-body communications is challenging because a number of ethical and technical issues prevent the realization of measurements with human subjects. Therefore, in order to approximately characterize the propagation of radio signals through the human body a chemical solution specially formulated to emulate the dielectric properties of biological tissues can be used. This solution is generally referred to as a phantom. Two ultra wideband (UWB) channel models for implanted sensor communications have been reported in this way in [36], [37] and [38], [39], respectively. The UWB system is a promising candidate for use in biomedical applications because of its efficiency with respect to multi-path fading, high transmission speed, and simple structure.

Nevertheless, the current technology of implant devices was stated to have a standard frequency of 402-405 MHz allotted by the FCC [40]. This frequency range is said to have appropriate propagation characteristics in human tissues. MICS band of 402-405 MHz are used currently due to its features such as ultra-low power, unlicensed, mobile radio service for transmitting data of the diagnosis from the implant to the external receiver. On the other hand, it is found that UWB frequency range offers an exceedingly wide transmission bandwidth that helps in giving more accurate real-time data.

4.3.1 Path Loss

Since human body is made up of various organs consisting of different types of tissues, the electrical characteristics of the entire body show vast heterogeneity and anisotropy, for instance the conductivity, power absorption, path loss and relative permittivity. Path loss, which is sometimes known as path attenuation, is a reduction in power density of electromagnetic waves as it propagates through space. It is also a measure of the average attenuation of a signal that propagates from a transmitter to a receiver. It can be presented as:

$$PL_{dB} = -|S_{21}(f)|_{dB} \tag{4.1}$$

Around the average value, there will be a variation called scattering that is caused by the difference of material dielectric properties in terms of conductivity and permittivity along the propagation path. This path loss (PL) cannot be directly measured in vivo for practical in-body implant applications due to legal restrictions, thus it is carried out by simulations. Researches on the scattering parameter have been made using a technique of finite difference time domain (FDTD). In order to model a path loss between the transmitting and the receiving antenna as a function of distance,it can be expressed as the following as shown in [39]:

$$PL_{dB}(d) = PL_{o,dB} + a(d/d_o)^n + \aleph(0, \sigma) \tag{4.2}$$

where d the depth of the implant from skin surface, in mm, a is fitting constant, d_o is the reference depth that will remain constant, n is the path loss exponent that is commonly subjected to the environment where the radio signal is propagating through. Whereas, the $\aleph$ is the random variable that is distributed by Gaussian in which it approximately shows the scattering in decibels.

4.3.2 Influence on propagation characteristics

There are many influences that are able to effect the results of propagation characteristics including antenna separation for different tissues, antenna misalignment, antenna rotation, line-of-sight as well as size and shape of subject. The authors in [41] investigated the subject-specificity of the on-body and in-body radio channels by considering nine body models of different sizes and genders. Their concluded that parameters such as the height of the subject and the curvature radius at the trunk affect the path loss in on-body radio channels. However, in the inter-body and off-body cases, the signal propagates away from the body and reaches the receiver through free space propagation and reflections from surrounding scatterers, and hence the impact of the subject on the radio link is found to be minimal.

4.3.3 Recent researchs

UWB communication system is an interesting topic that has recently gained more attention instead of the MICS band that is being currently used in medical implant systems. As said, only two studies have been found on this UWB characterization. Both studied on UWB characterization propagation from inside the human body using an adult male's chest as test area.

The first model characterizes an UWB channel in the 3.4-4.8 GHz frequency band for implanted devices inside the chest within 6-18 mm depth [36]. They achieved a data rate of 10 Mbps. The second model is more general and covers the 1-6 GHz frequency range for depths up to 120 mm inside the chest [38]. In this experiment, both electric and magnetic fields probes were set in a rectangular cube space with dimensions of 140 mm x 160 mm x 80 mm as well as resolutions of 20 mm x 10 mm x 20 mm. These probes were inserted in various tissues such as blood, lung, hear, fat, bones and muscles. Considering the probes as isotropic radiating antennas, they operated at a frequency range of 0.1-1 GHz and 1-6 GHz for spectral bandwidth of -15 dB. The characterizations that were investigated include power density, spectral analysis and delay spread. As can be noticed, it has been revealed that significant power loss befell at the superficial area near to the skin. This may be caused by impedance mismatching between free-space and the multilayer body tissues. Moreover, delay spread was examined in order to evaluate different multipath channels for wireless system. It was concluded that the delay spread increased with penetration depth where the maximum delay spread was 1 ns.

In the same experiment, both 0.1-1 GHz and 1-6 GHz bandwidths were compared in the power field. The range 0.1-1 GHz had nearly the same power density at both horizontal polarized (HP) and vertically polarized (VP) at depths of 60 mm to the surface. However, the power density with HP was higher than VP at a depth of 120 mm. Thus, it proved that UWB signal with HP was able to penetrate deeper inside the human tissues. On the other hand, for 1-6 GHz, the power density was lower with higher power loss. As for delay spread, RMS delay spread was more with HP rather than VP at 1-6 GHz signal. It could reach a maximum value of 1.7 ns delay spread at 160 mm depth. Similarly, at 0.1-1 GHz, it indicated the same pattern of delay spread yet with different maximum value of 1.2 ns at a depth of 100 mm.

Chapter 5

Conclusions

Wireless technology has opened up new expectations in the fabrication of tiny devices implanted in the human body for healthcare applications. Nowadays much emphasis is placed on taking pro-active measure by health care professionals before the condition of the patient becomes acute. It is only possible if patients are equipped with wearable sensors. Certain sensors are place inside the human body to obtain information on vital physiological phenomena. These implantable sensors have associated circuits for signal processing and data transmission. Powering the circuit is always a crucial design factor. Battery cannot be placed inside the human body due to serious health risks such as poisoning and chemical burn. An alternative approach is to supply power using inductive link where power is transferred via two loosely couple inductors.

The aims of this thesis were to find a procedure to supply the power required for an in-body sensor wirelessly placed in the heart and characterize the human body as the channel model for recover data from the in-body sensor. Different wireless power transfer system have been exhaustively studied. The method of resonance-based inductive coupling has been presented as a leading technique to power implant sensors wirelessly. Due to biomedical constraints, the physical dimension of the implanted sensor cannot be large. In addition, the frequency cannot be increased due to FCC regulations. From the review of state of the art it can be observed that inductive coupling operating in HF (13.56 MHz and below) frequency range is now one of the most common methods to wirelessly send power and data from off-body interrogator to the implanted device inside human body. However, only two researches have been made for the propagation characterization of the channel model for implantable antennas in the human chest within ultra-wideband frequency range.

A radio channel in such applications is affected by several phenomena, such as human or limb position, age, weight and sex. Only the change of the posture of an arm obstructing the line-of-sigh in a link causes significant difference to the received signal power. In addition, the environment has impact on the radio wave propagation and interference level. On the other hand, age, weight and sex have impact to fluid concentration and tissue structure inside a body, and therefore to the dielectric behaviour. This means that the relative permittivity of a body is changing, and for that reason the radio propagation differs.

Although much work has been done in the field of wireless communications and wireless powering for biomedical implants, there is still room for improvement and optimization of the entire system.

5.1 Future directions

Based on the achievements of this thesis, there is some scope for future developments to be made on the work done, which will increase its better understanding. They are listed as follows:

- Manufacturing of a prototype with inductors to implement the proof of concept of WPT in biomedical applications.
- Realization of the wireless power transfer system with on-chip inductors and associated electronics. This can be used to miniaturize the wireless power transfer system which could be a good step towards implementation of complete on-chip WPT system.

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