WIRELESS POWERING OF RFID DEVICES

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STATEMENT OF CANDIDATE

I, Chenggong Qu, declare that this report, submitted as part of the requirement for the award of Bachelor of Engineering in the Department of Electronic Engineering, Macquarie University, is entirely my own work unless otherwise referenced or acknowledged. This document has not been submitted for qualification or assessment an any academic institution.

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ABSTRACT

In recent years, there is a significant increasing in the demand for biomedical implants. Research shows that a larger number of Australians require implantable biomedical devices every year such as pacemakers, Cochlear implants and cardiovascular implants [9]. Figures indicate that the number of pacemakers sold in Australia was 12523 in 2009. However, the number reached 15203 in 2013 with an annual increasing rate of 4.28% [27].However, most of the implantable devices are still remain restricted due to certain constraints. The power supply is one of the dominating factors that limits the use of biomedical implants because such devices need to function at a stable, efficient and reliable rate for a period of time(normally for years). Currently, batteries are mostly used for powering implantable devices; however, it cannot be a desirable solution due to its shortcomings such as size, the risk of infection and need of replacement. Consequently, this project aims to develop a wireless charging system that can be implanted into the human body with no need of replacing the battery. This project will define and quantify the tradeoffs between power requirement, lifetime and functionality for such system. This document presents the outcomes of this project highlighting the problems encountered and solutions devised.

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Chapter 1

Introduction

The demand for biomedical implants has been significantly growing during the past few years as biomedical implants are becoming increasingly popular for monitoring, organ replacement and many other medical applications. Data shows that more than a million cardiac pacemaker was sold and installed in 2009 globally [28]. Today, biomedical implants are saving millions of lives and providing alternatives for both doctors and patients through the innovation of cochlear implants, cardiac pacemakers, neuroprostheses, implantable imaging devices, deep organ pressure sensors and implantable contraceptives which can be implanted into a woman's body to adjust estrogen and hormone level for more than 16 years and can be controlled and charged wirelessly [6]. Figure 1.1 indicates the most common applications across the breadth of different body systems.

The power requirement for these devices is very high as implant devices are becoming more highly integrated, robust, smaller in size, more reliable and much efficient. However, current battery technology cannot provide a stable, clean, safe and effective way of powering biomedical implants as batteries are exhaustible, and it can cause infection to body tissue [20]. In addition, most of the biomedical implants contain no built-in batteries due to particular concerns such as safety, lifespan, size and power requirement. For example, cochlear implants and radio frequency identification devices (RFID) do not have batteries. Consequently, wireless charging/powering technology has been introduced to the public. Wireless power transmission (WPT) is an approach that allows implantable devices to be charged wirelessly over a certain distance without taking the battery out of human body.

Therefore, this project will develop a wireless charging system for an RFID-style device that is compatible with the human body. The primary objective of this paper is to define and quantify the tradeoffs between power requirement, lifespan and functionality of such a system.

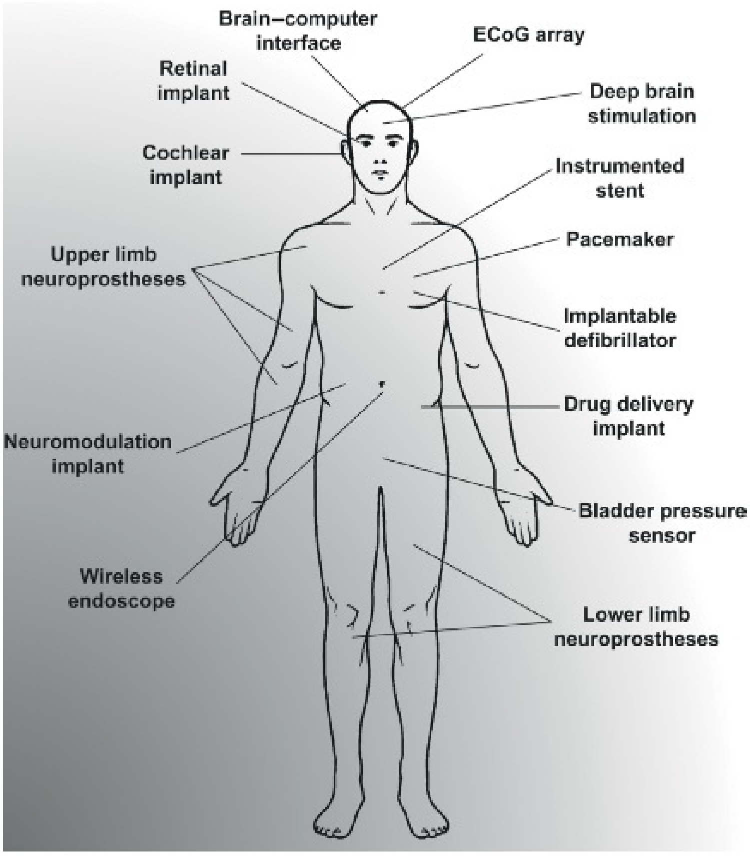


Figure 1.1: Biomedical applications for different body systems [41]

1.1 Biomedical Implants

Biomedical implants are usually divided into two types: sensing and therapy delivery (also can be called as stimulation). Sensing is a function that allows a device to sense different data or parameters inside the human body such as blood pressure, glucose, heart rhythm and blood oxygen content. Batteries are not necessary for this kind of implants because sensing devices do not require a constant power supply since it will only be activated when sensing parameters and monitoring events; otherwise, it stays in a low power mode or even off mode when inactivated. Based on the data collected from sensing devices, stimulation devices are used to response to the condition. Stimulation devices are used to deliver therapy to the patients. Pacemakers and Cochlear implants are excellent examples because they provide stimulation to a particular organ by generating current or voltage pulse [25]. The development of modern medical technologies regarding biomedical implants has provided an alternative way of treating diseases. As shown in figure 1.1, biomedical implants have been designed for different organs and systems with specific functionalities such as drug delivery, blood pressure monitoring, improving hearing abilities and organ stimulations.

1.1.1 Wireless Energy

Due to the rapid development of wireless power transfer technology, scientists and designers are considering wireless energy as a promising solution to provide reliable, robust and efficient power supplies for systems where wire connections are not convenient. Wireless power transfer was first introduced by Nikola Tesla in 1890s. He developed a wireless transmission of electrical energy by using radio frequency resonant transformer which is known as Tesla coil [37]. It has been more than 100 years since the principle of wireless transmission has been introduced to the public. However, the market for wireless power transmission still has tremendous potential due to massive demands on personal wireless connectivity, charging battery-enable devices and providing efficient power supplies. According to the new market research report, the market is expected to reach 13.11 billion dollars by 2020 [33]. Nowadays, wireless energy can be found in a variety of applications such as wireless charging for electronic devices, data transmission, imaging, heating patients with hypothermia and cancer treatment. Wireless charging is a technique that allows power to be transferred from a transmitting source to a receiving device wirelessly by using electromagnetic fields. Theoretically, wireless charging is based on the principles of inductive power transfer or resonant technology. Different wireless power technologies have different frequency range and charging distance. For example, charging electric toothbrush or A4 battery is often operating in the range of Hz - MHz with inductive coupling technology. According to Qi standard, resonant inductive coupling technology allows charging portable devices in the range of KHz - GHz [30]. Referring to Appendix C for a full radio wave spectrum. Power losses across a medium, which is the human body in the case of biomedical implants, is minimal below 10 MHz frequency range. Therefore, inductive coupling transmission in the near field region is often used in powering biomedical implants. Wireless communication (Data/Voice transmission) is another system function which enabled by RF-based wireless technology, and it has highly interacted with modern society. Wireless communications services such WiFi, Bluetooth and Zigbee, operate in the 2.4GHz band (Zigbee also operate in the 800 MHz and 900 MHz band). Those wireless communication services are designed based on IEEE 802.11 standards which are a standard for short-distance wireless networking [13](Referring to Appendix D for a full table of short distance wireless interfaces). Table 1.1 shows the standards for short-range wireless networks.

	Zigbee	**Bluetooth**	**UWB**
Frequency(GHz)	$2.4 - 2.4835$	$2.4 - 2.4835$	$3.1 - 10.6$
Bandwidth(MHz)	83.5	83.5	7500
Max. Data Rate(Mbps)	0.25	1	100
Power Consumption(mW)	$5 - 20$	$40 - 100$	$80 - 150$

Table 1.1: Short-Range Wireless Network Standards

Wireless energy has been used as an alternative to diagnose and cure a variety of diseases such as cancer in the 1920s [55]. Scientists and researchers have found electromagnetic fields (EMF) such as X-Rays and Gamma rays can be used to treat cancer based on the assumption that energy fields inside the human body or disease-causing altered electromagnetic frequencies can be corrected by electromagnetic energy. The frequency range for EMFs in diagnosing and treating cancer is between 300 PHz to 300 EHz [8]. Radiofrequency ablation is another alternative frequently used to treat breast cancer, colorectal cancer and hepatocellular carcinoma [55] which operates in the range of $460 - 550$ kHz. The human body and living tissue can absorb EMF power to cause heating. This has been applied to the medical practice to heat patients with hypothermia. However, too much EMF power can cause harmful effects to the living tissue which called thermal effects. Thermal effects occur in the RF frequency range of 30 MHz to 300 GHz [8]. Table 1.2 indicates different uses of wireless energies along with their frequency range.

Functions	**Frequency Range**	**Typical Energies**
Charging	kHz − MHz	RF−based
Data/Voice transmission	2.305−2.32 GHz, 2.345−2.36 GHz	Radio Frequency
Cancer treatment	300 PHz−300 EHz, 460−550 kHz	EMF
Heating people with hypothermia	30 MHz−300 GHz	Radio Frequency

Table 1.2: Different Uses of Wireless Energies and its Frequency

1.1.2 Wireless Power Transmission

Wireless powering technologies provide us with a solution to reduce the size, power consumption and circuit complexity of biomedical implants. Figure 1.2 shows a general block diagram of a wireless power transmission system which consists of an RFID-type transceiver, a power amplifier, a back telemetry system that sends information to the transmitter and a rectifier that converts alternating current (AC) to direct current (DC). From figure 1.2 we can see that the system is divided into three parts which are an external unit, skin and internal unit. The details of each block will be covered later in the report.

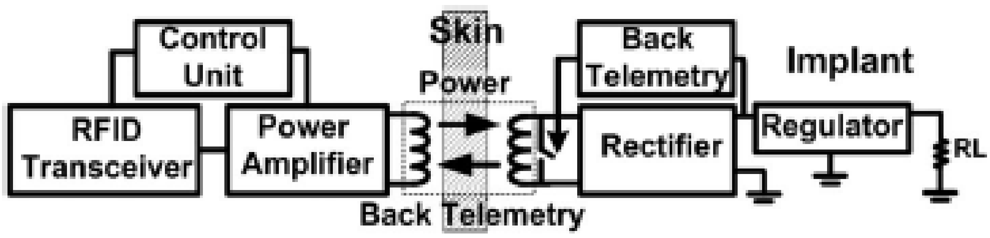


Figure 1.2: Block diagram of a wireless power transmission system [26]

1.2 Motivation

As the demands for implantable biomedical devices is raising and the current technologies are still under some restrictions such as battery size, power efficiency, lifespan and safety, thus, a desirable solution for powering such devices is required in the industrial to break through the constraints. The topic was given by Professor Michael Heimlich from Macquarie University, and the idea was based on the work and findings conducted by a research team at Stanford Universtiy. Assistant Professor Ada Poon has first reported in the Proceedings of National Academy of Sciences that her team has invented a way wirelessly to power implantable biomedical devices(IMDs) inside human body [53]. The technology could eliminate some constraints that prevent IMDs from being more widely used such as larger batteries, batteries lifetime and complex and clumsy circuitry. This is interesting because it allows us to transmit power to different organ implants safely and it also opens new opportunities in medicine as well as challenges. That is why the thesis is conducted.

Chapter 2

Thesis Overview

In this section, a general description of the thesis will be given along with project plans and deliverables which can be found in the appendix at the end of the report. This project is supervised by Professor Michael Heimlich from Department of Engineering, Macquarie University. To ensure the project is conducting successfully and accordantly with the project requirements, set of goals and deliverables, regular weekly meetings with the academic supervisor is needed. Consultation attendance form is attached to the Appendix as required by the Department.

The thesis is organised as follows. To begin with, Chapter 3 gives literature review about the topic and discusses some related work such as wireless charging technologies and methodologies, wireless charging standards, power supply options and applications of inductive power transfer(RFIDs). Moreover, according to the functionalities, a wireless charging system can be further divided into three parts: wireless power transfer, telemetry system and sensor system. Those will also be discussed in literature review.

Chapter 4 gives a detailed discussion on the main problem faced which is to develop a wireless charging system for an RFID-style device that is compatible with the human body. The primary objective of this paper is to define and quantify the tradeoffs between power requirement, lifespan and functionality of such a system. As a result, different possible approaches to solving the problem will be addressed and discussed after illustrating the main problem. Moreover, Chapter 4 will discuss the analytical solutions that could be used to solve the problem. Finally, Chapter 4 provides the system simulations developed by AWR VSS for different modulation techniques based on the objective. This thesis does not focus on circuit implementation; therefore, Chapter 4 will illustrate the tradeoffs by doing several system simulations in AWR rather than performing circuit simulations.

Chapter 5 will address some of the possible improvements and interesting fields for future work.

Chapter 6 will conclude and summarise the whole document.

2.1 Project Objectives

- 1. Review of wireless charging technologies(mainly on inductive coupling).

- 2. Familiarise with wireless power transmission and telemetry system.
- 3. Background on wireless power transmission system and telemetry system in detail and understand how they work in a wireless charging system. Most importantly how they function in terms of power consumption, frequency and datarate.
- 4. Analytical solutions to the problem.
- 5. Theoretical and simulation analysis.

2.2 Project Baseline Review

This project was planned to be executed from 1 March, 2016 through 6 June, 2016. Baseline plan was generated with the intention of utilising only the weekdays, inculding mide-semester break, for project activities.

2.2.1 Time Budget Review

Table 2.1 summarises the time budget on the project as well as the completion of the project in percentage. No amendments to the time schedule was required at the moment. Although the allotted time for each activity was followed closely, the order of accomplishment varied. A more detailed time plan can be found in Table A.1.

Estimated Work	85 days
Realised Work	85 days
Percentage Completion	100%

Table 2.1: Time Budget Summary

Task	**Start**	**End**	**Days**	**% Done**
Literature Revirew	1/03/2016	4/04/2016	34	100
Project specification and plan	5/03/2016	15/03/2016	10	100
Understanding the scope of the project	1/03/2016	4/04/2016	34	100
System level design	25/03/2016	1/05/2016	35	100
Analytical solutions	25/03/2016	10/05/2016	44	100
Simulations through AWR	10/04/2016	15/05/2016	33	100
Comparison between the results	15/05/2016	17/05/2016	2	100
Thesis report draft	18/05/2016	25/05/2016	8	100
Thesis report final	25/05/2016	5/06/2016	11	100

Table 2.2: Planned start day and end day for each of the activities

2.2.2 Financial Budget Review

The allocated financial budget from the Department of Engineering, Macquarie University was $300. However, this project did not require any circuit design or anything to buy expect a computer aided software AWR which can be obtained from online without any cost. Therefore, no further purchasing is intended, hence the budget surplus may not be utilised.

2.2.3 Scope Review

Given that the current activities are in line with the baseline plan, no amendments to the scope of the project is needed at this stage.

Chapter 3

Literature Review and Theory

3.1 Wireless Charging Technologies

The very first thoughts of wireless power transfer can be traced back to the early research of Heinrich Hertz [16]. Then Nikola Tesla became interested and conducted pioneering work on wireless power transfer topic which he has applied the concept of resonance to the transmission of electrical power without any wires. In the year of 1899, Tesla has built a set of coils that resonated at a frequency of 150KHz and with an input power of 300Kw [5]. Although Telsa couldn't further his research with limited financial support, he was still managed to provide us with a large number of massive findings and patents.

In the past few years, there has been a massive development of the wireless powering technologies. Currently, there are two types of wireless technologies which are inductive powering for the near-field region and high power density directive powering in the far-field region. Specifically, the near-field region is defined as the area within about one wavelength of the antenna [38]. On the contrast, far-field is anything beyond one wavelength of the antenna. For the purpose of this project, near-field will be the focus. As we mentioned in the introduction, larger batteries size and exhaustive characterises limit the use of implantable biomedical devices. Therefore, wireless powering provides us with two alternatives to (1) provide power directly to the system or (2) to recharge the built-in batteries within an implantable device. To achieve this, we need to introduce electromagnetic powering of the system(EMP). Inductive radio frequency (RF) coupling of energy using a carrier frequency in the range of few KHz to few MHz depending on the applications is one of the main options for EMP. The other primary option is the infrared powering which the process of infrared powering is considered to be inefficient and non-practical for most of the implantable biomedical devices [17]. Thus, the most practical and popular solution is near-field coupling because it can provide sensor operation in the range of one meter and it can significantly reduce the system dimension, at the mean time, also extend lifespan. Moreover, inductive coupling is widely used for RFID and biomedical applications. Figure 3.1 shows a variety of wireless power transfer techniques.

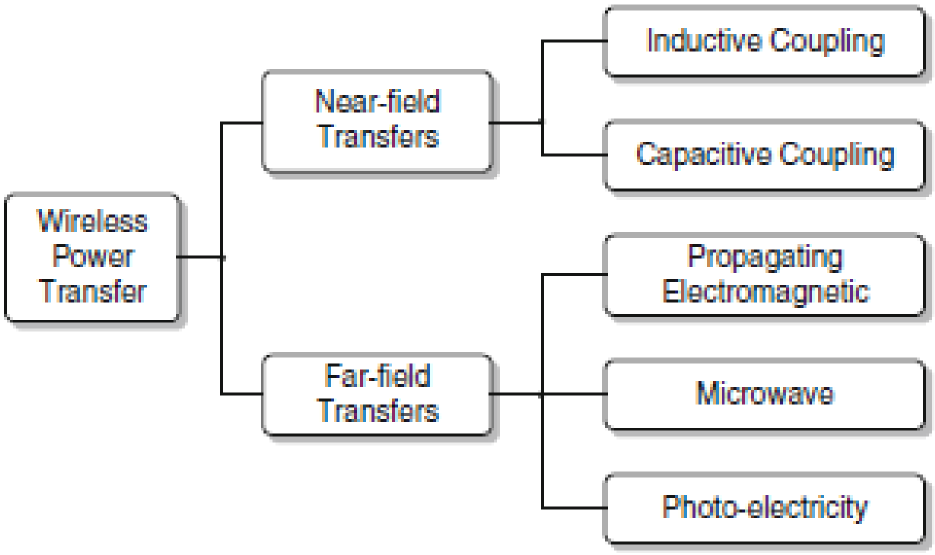


Figure 3.1: The category of the wireless power transfer methods [39]

3.1.1 Inductive coupling

In this section, a detailed review on inductive coupling will be addressed. In inductive coupling, power is transferred between two coils of wire by a magnetic field [35]. Two coupled resonant circuits with transmitter(TX) and receiver(RX) coils together acting as a transformer. An AC in the TX generates a magnetic field which induces a voltage in the RX. Then the induced voltage can be used as a power source.

In inductive coupling, the amount of power that can be delivered to the IMD can be determined by the coupling coefficient k. The variation in k can cause a change in the received power. Moreover, the distance variations between two coils are the primary causes of change in k. Changes in the received power can cause large voltage differences across the RX coils. Load changes as a result of stimulation, for example, can also cause variation in the receiver coil voltage. Such variations are highly undesired in implanted medical devices. Because too little power can cause a malfunction, and extra power can increase heat dissipation within the implant and damage the surrounding tissue [21].

Most inductive coupling studies found in IEEE were designed with frequencies lower than 20 MHz to avoid harmful effects to the living tissue caused by heating [34]. Figure 3.2 is a block diagram of an inductive coupling link system. The system contains two coils. One is implemented into the human body, and the other one is on the receiver side. Additionally, there are four topologies that can be used to form a resonance circuit in an inductive coupling system which is shown in figure 3.3. Both sides of the coil are tuned to the same frequency f_0 to maximise the power transfer efficiency of the inductive coupling link. Practically, f_0 can be calculated by the equation (3.1).

$$f_0 = \frac{1}{2\pi\sqrt{LC}} \tag{3.1}$$

There are many other parameters need to be considered when designing an inductive coupling which is mutual inductance (M) and coupling coefficient (K). M can be calculated by using equation (3.2) as proposed by [19].

$$k = \frac{M}{\sqrt{L_T L_R}} \tag{3.2}$$

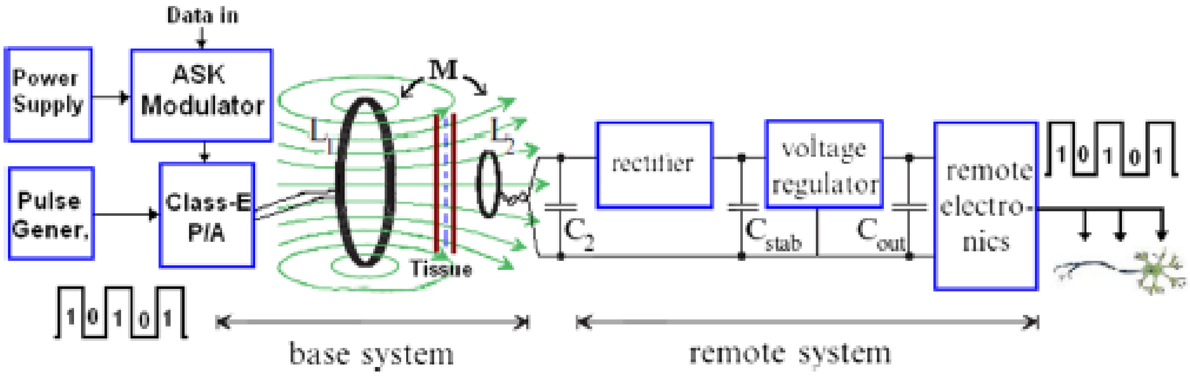


Figure 3.2: Block diagram of an inductive coupling link system [34]

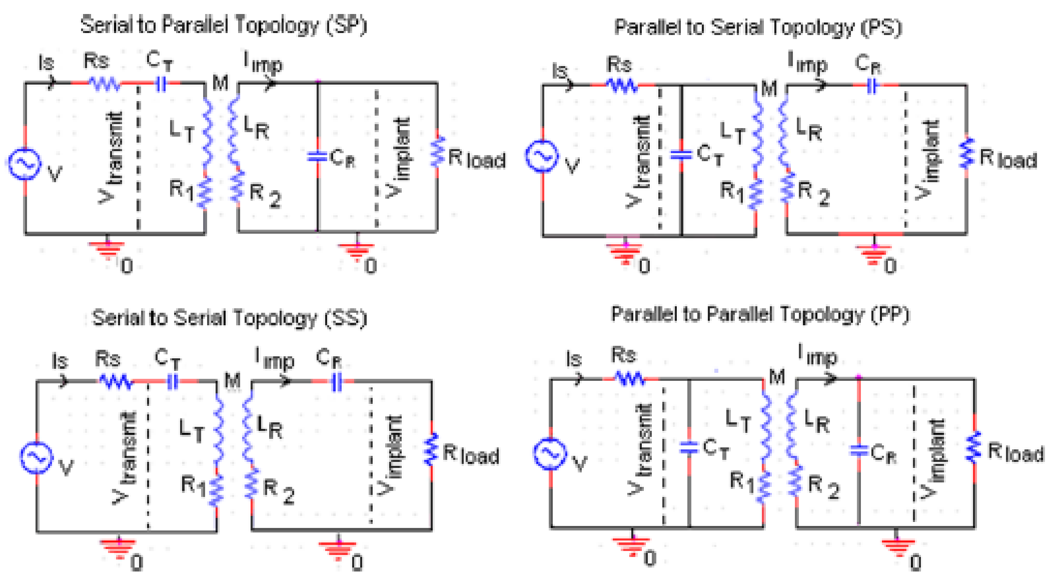


Figure 3.3: Topologies to form resonance circuits in inductive coupling [34]

As can be seen from figure 3.3, with serial to parallel topology R_1 is the combination of L_T, whereas R_2 is the effective series resistance of L_R [48]. The resonances can be created by using capacitors C_T and C_R. Therefore, the resonance frequency ω_o can be calculated by equation (3.3).

$$\omega_0 = \frac{1}{\sqrt{L_T C_T}} = \frac{1}{\sqrt{L_R C_R}} \tag{3.3}$$

Then the quality factor Q can be calculated separatly for both coils.

$$Q_1 = \frac{L_T \omega}{R_1} and Q_2 = \frac{L_R \omega}{R_2} \tag{3.4}$$

The efficiency for both side of link should be maximized for high frequencies [15].

$$1 \ll K^2 Q_1 Q_2 = \frac{K^2 L_T}{L_R} * \frac{1}{R_2 C_T} \tag{3.5}$$

Therefore, the maximum efficiency can be calculated by equation (3.6). The relationship between efficiency, coupling coefficient and the quality factor is shown in figure 3.4. Apparently, the increase in coupling coefficient and quality factor will result in an increase in the maximum efficiency.

$$\eta_{max} = \frac{K^2 Q_1 Q_2}{(1 + \sqrt{1 + K^2 Q_1 Q_2})^2} \tag{3.6}$$

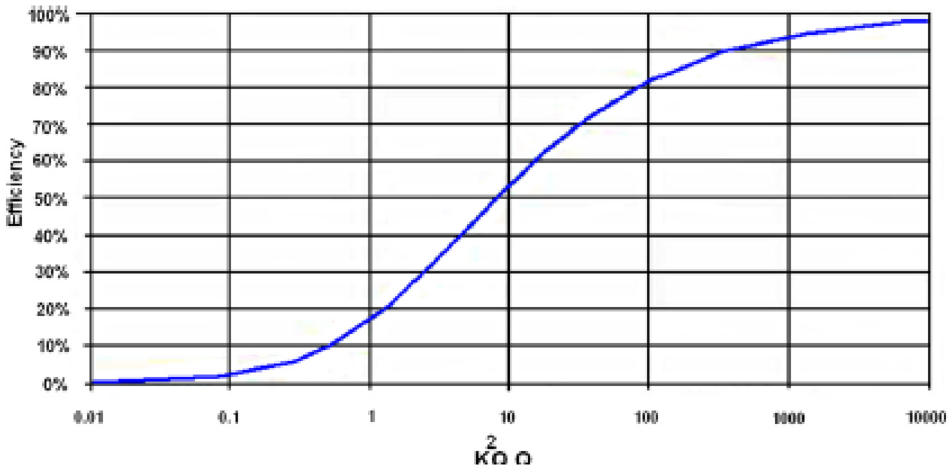


Figure 3.4: Maximum link efficiency as a function of coupling coefficient and quality factor [15]

The total power efficiency can be calculated by equation (3.7) by considering load resistance. Moreover, the total power efficiency is proportional to load resistance which is the resistance of implantable devices. As stated in [52], the total power efficiency for inductive coupling can achieve between 74 percent to 80 percent depending on the resistance of the implants.

$$\eta_{total} = \eta_T \eta_R = \frac{K^2 Q_1 Q_2^3 R_{LR} R_{Load}}{K^2 Q_1 Q_2^3 R_{LR} R_{Load} + K^2 Q_1 Q_2 R_{Load}^2 + Q_2^4 R_{LR}^2 + 2 Q_2^2 R_{LR} R_{Load} + R_{Load}^2} \tag{3.7}$$

3.1.2 Capacitive coupling

Capacitive coupling is the transfer of energy within an electrical network to different devices by using capacitors [36]. It is also an approach to transfer power and data to biomedical implants with a short region. Although the capacitive coupling is relatively easy to make, it is often unintended because when two signals coupled together, there appears to be noise. Figure 3.5 is a simplified capacitive coupling link. The major problem with capacitive coupling is that it causes harmful effects by generating heat to the living tissue and human body. Inductive coupling transfers energy through the alternating magnetic field while capacitive coupling transfers energy through alternating electric field [39]. Figure 3.6 shows the comparison between the capacitive coupling and inductive coupling. Compare to a magnetic field; electric field causes much more heating and harmful effects on human tissue because human body absorbs electric field.

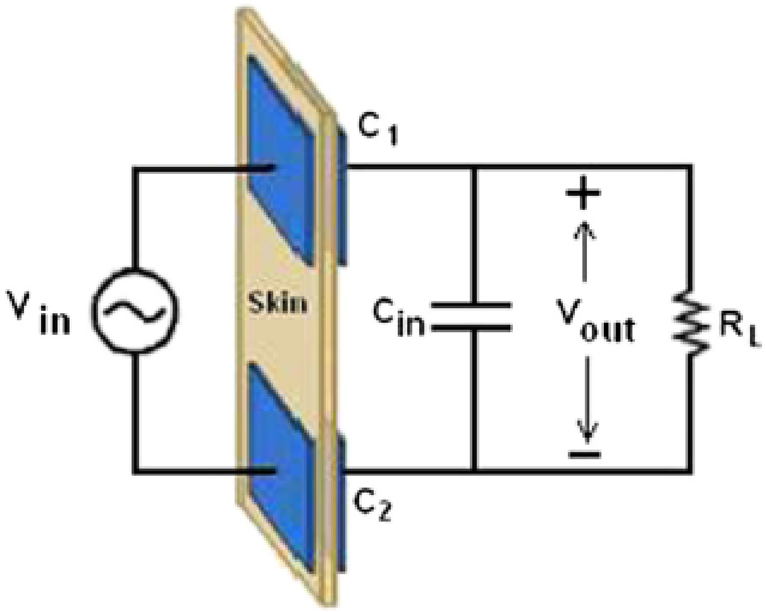


Figure 3.5: Example of capacitive coupling link [15]

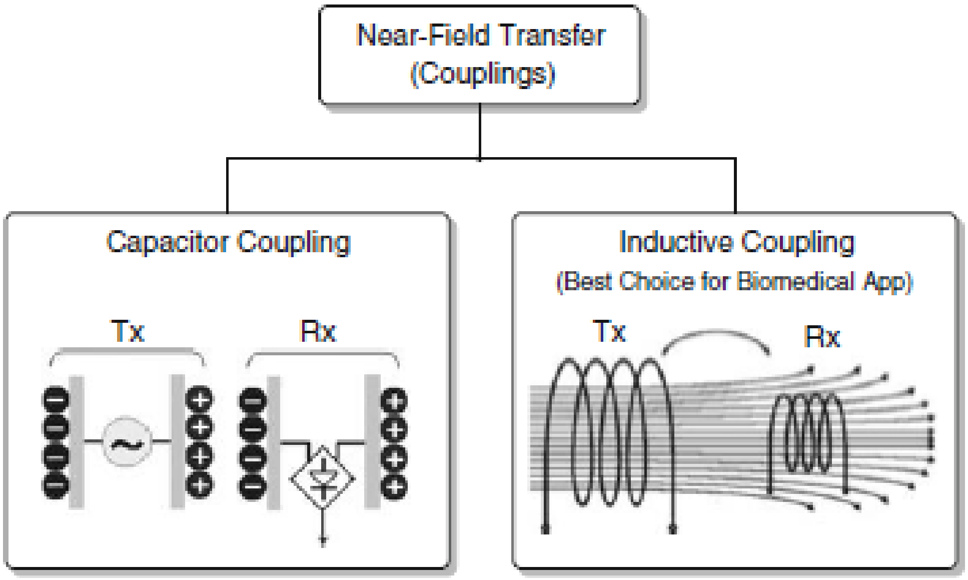


Figure 3.6: Example of capacitive coupling link [39]

3.1.3 Resonant energy transfer

Resonant coupled coils are often used in some embedded devices and RFID applications to enhance the efficiency of power transmission. It is the transmission of energy between two coils that are part of resonant circuits with the same frequency [54]. High power transfer efficiency can be achieved by using this method as two resonant coils are strongly coupled. Figure **??** is a general resonant inductive wireless power system. The quality factors for resonant magnetic coupling are relatively high because it often operates in MHz range. The charging efficiency can be further increased by decreasing the charging distance. However, it requires a large size of WiTricity receiver. Therefore, the implementation of magnetic resonant coupling link onto portable devices can be very difficult.

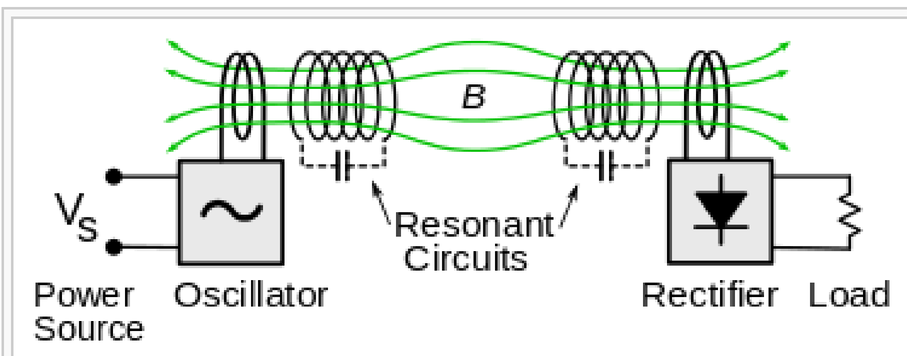


Figure 3.7: Resonant circuit [7]

3.1.4 Comparison

A comparison can be drawn from the review of different wireless charging methods. Inductive transmission is likely to be the most appropriate method to wirelessly power a biomedical implant by considering their size, frequency, distance, penetrability and efficiency. There are several reasons which support the conclusion. Firstly, parameters such as penetrability and efficiency are extremely important for biomedical applications. Secondly, the harmful effects caused by heating to the human body for inductive coupling is much less compare to conductive or capacitive links because the electric current is not significant in the induction coils and not in the living tissue. In an inductive coupling link, energy is transferred through a magnetic field rather than an electric field. The facts that human body only absorbs electric field makes inductive coupling the preferred choice in short-range wireless powering. Thus, inductive coupling method will be used and further illustrated in next section. Table 3.1 is the comparison among several wireless power transfers.

Wireless Power Transfer Methods	**Frequency**	**Range**	**Penetrability**	**Efficiency**
Inductive Coupling	Low Hz to MHz	Short	Strong	High
Capacitive Coupling	Low Hz to MHz	Short	Strong	High
Propagating Electromagnetic	Medium MHz to GHz	Medium	Medium	Medium
Microwave	High GHz to THz	Long	Weak	Low

Table 3.1: A comparison among the wireless power transfer technologies

3.2 Wireless Charging Regulation and Standards

3.2.1 International Charging Standards

1) Qi: Wireless Power Consortium(WPC) has developed Qi−Inductive Power Standard in 2008 [51]. It is designed for inductive electrical power transfer and data communication between the power transmitter and receiver with a maximum allowable distance up to 4 centimetres. Figure 3.8 is a typical wireless charging model with Qi standard. The technology behind Qi is resonant inductive coupling which allows up to 40 mm's charging distance. Normally, a Qi charging system contains one power transmitter and one power receiver [51]. Qi standard requires precise coil alignment between the power transmitter and power receiver which is a bit difficult for implementations. The design of power transmitter for Qi standardised system has to follow some requirements which are [51]:

- Qi standard states that a power charger should operates on 110 to 205 KHz frequency range and transfer power within 5 Watts for Low power design.
- Qi standard states that a power charger should operates on 80 to 300 KHz frequency range and transfer power up to 12 Watts for medium power design.

The Qi-compliant wireless power transfer system also supports data transmission which is operating within the same frequency range as charging.

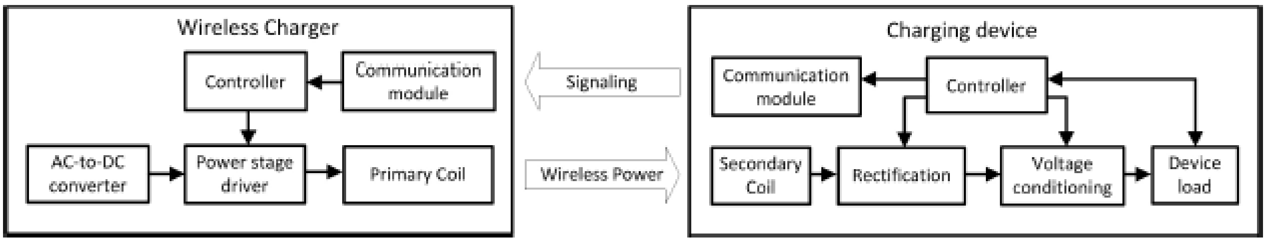
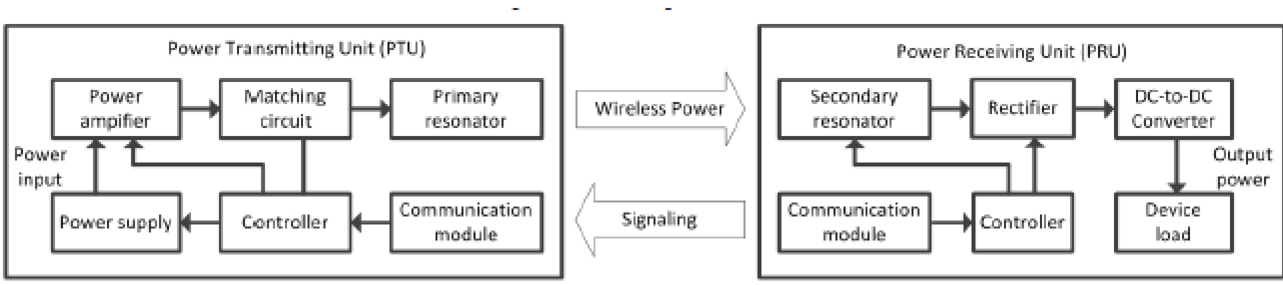


Figure 3.8: Wireless charging model with Qi standard [51]

2) A4WP: Compare to Qi, A4WP can charge multiple devices at the same time with just one power transmitter by using larger magnetic field [43]. Figure 3.9 is a general model of wireless power transfer system with A4WP standard. A4WP aims to produce a much larger electromagnetic field than Qi standard by utilising magnetic resonance coupling. Different to Qi, A4WP does not require precise coil alignment when charging and it can supports multiple chargings at the same time with distance up to several meters. The implementation and deployment of A4WP standardised system are flexible comparing to Qi because the effects caused by foreign objects and separation between transmitter and receiver is much lesser than that to the Qi model. Data transmission can also be done with A4WP. There are two blocks in figure 3.9; Power Transmitter Unit(PTU) which consists of matching circuit, power conversion circuit and signalling circuit, Power Receiver Unit(PRU) which consists of energy reception and conversion circuit, control and communication circuit [51]. In wireless charging system with A4WP standard, the transmission of power is controlled by a charging management protocol [51]. Data is then being decoded, and feedback signalling is being performed after the power has been transferred to the PRU. Wireless power is transferred at 6.78 MHz Industrial Scientific Medical (ISM) frequency band for A4WP which allows the signal to be transferred in a wider bandwidth compare to 2.4 GHz ISM band for Qi standard [24].



PTU Class	Power Input	PRU Support Requirement
1	On roadmap	PRU Category 1
2	10 W	Multiple PRU Category 1, 2, 3 concurrently or a single PRU Category 4
3	16 W	
4	22 W	
5	On roadmap	On roadmap

PRU Category	Power Output	Applications
1	On roadmap	Low-power applications (e.g., on order of 1 mW)
2	3.5 W	Feature phone
3	6.5 W	Smart phone
4, 5	On roadmap	Higher-power applications (e.g., Tablets and laptops)

Figure 3.9: Wireless charging model with A4WP standard [43]

3.2.2 Safety Standards

Safety has to be considered when it comes to the design of medical applications. Here two standards will be discussed since there are no direct standards for charging biomedical implants internationally.

- IEEE C 95.1 2005 Standard: Standard C95.1 2005 is also called "Standard for Safety Levels with Respect to Human Exposure to Radio Frequency Electromagnetic Fields, 3 kHz to 300 GHz". This standard states that the safety operating frequency range is between 3 KHz to 300 KHz which is where the majority of wireless powering system operates [40].
- ICNIRP Guideline: ICRIRP is a short term for International Commission on Non-Ionizing Radiation Protection. They have published guidelines of maximum allowable exposure to EMF which is in the frequency range of 0 Hz to 300 GHz in 1998 and 0 Hz to 100 KHz in 2010 [44].

Both standards provided the maximum level of allowable exposure for human and both of them are covered the frequency range of 0.1MHz to 100 MHz which is where the majority of the wireless powering system operates.

3.2.3 Inductive Impants

According to Federal Communications Commission(FCC), IMD used inductive coupling to power and communicate information between transmitter and receiver should have frequencies below 200 KHz. For inductive devices that meet the requirement of Section 15.209 of the FCC's rules are allowed for the use of small antennas [14].

3.2.4 Medical Device Radiocommunication Service

The FCC allocated spectrum in the 401-406 MHz band for radio communication between IMDs and external controllers in 2009. This particular frequency band is well suited for tissue penetration at the relatively low power which helps extend batteries lifetime [14]. Moreover, higher frequencies mean that transmission can be made at high data rates.

3.2.5 Emerging Standards

Recently, a new wireless medical standard of IEEE 802.15.6 has been developed for a body area network, which can be used to help real-time health monitoring and diseases diagnoses, to be operated at low frequencies over a short distance, with long battery life and high data rates [14]. IEEE 802.15.6 is a task group of IEEE 802 and it is used for the standardisation of Wireless Body Area Networks. Figure 3.14 shows the frequency bands for Wireless Body Area Networks. Figure 3.14 indicates that the Medical Implant Communications Service (MICS) band is applied worldwide for data communication within biomedical implants. MICS band is in 402 to 405 MHz range. However, MICS can not be used for applications with high data rate [22]. Wireless Medical Telemetry Services (WMTS) band is established for telemetry system in the range of 420 to 1429.5 MHz. Similarly, WMTS band does not support high data rate applications [22]. Ultra-wideband (3100 to 10600 MHz) allows high data transmission over a much wider bandwidth.

A detailed current medical telemetry bands can be found in the appendix. Several things we need to consider when choosing a wireless frequency band for IMDs and its applications [14].

- International availability and band allocation for medical devices.
- Whether the devices need to have primary or secondary radio service classification.
- How incumbent users of the selected and adjacent bands can impact a medical device's operation.
- Interference mitigation techniques for shared RF wireless frequency bands.
- For implantable devices, tissue propagation characteristics and specific absorption rates.

3.3 Frequency requirements

Also, approximate frequencies for different types of biological tissue is also important. In the last 15 years, the design of implantable medical devices, link optimisation, circuit simulation and implementation were usually done within the frequency range of lower than 10MHz. In the contrast, the design of telemetry system is always between 10 MHz to a few GHz [31]. As stated in the review, EMF that is used to heating patients with hyperthermia operates in both low-frequency (less than 10 MHz) and high-frequency (greater than 1 GHz) carries. Therefore, Prof Ada Poon from Standford University has done some researches on the topic of optimal frequency for the power transmission into the human body that allows the maximum ratio of received power to total tissue absorption [31]. Figure 3.10 summarizes the optimal frequencies for 17 different types of biological tissue.

Several design techniques have been introduced in the previous part of the thesis including high-frequency electromagnetic wave transmission, inductive power transmission, ultrasound power transmission and resonance-based coupling power transmission. The high-frequency electromagnetic wave transmission can provide near-field power links. However, the received power is often limited to mW which is not enough for most of the biomedical implants [47]. Inductive power transmission seems to be the best solution among all those options.

3.4 Power supply options

Research shows that there are two main options for power supply in low power wireless communication system, which are batteries and ambient power scavenging [4]. Battery

Type of tissue	Approximate f_{opt} (GHz)
Blood	3.54
Bone (cancellous)	3.80
Bone (cortical)	4.50
Brain (grey matter)	3.85
Brain (white matter)	4.23
Fat (infiltrated)	6.00
Fat (not infiltrated)	8.64
Heart	3.75
Kidney	3.81
Lens cortex	3.93
Liver	3.80
Lung	4.90
Muscle	3.93
Skin (dry)	4.44
Skin (wet)	4.01
Spleen	3.79
Tendon	3.17

Figure 3.10: Approximate optimal frequency for different biological tissue

technology is well developed, and it's been widely used in the industry. However, a battery has a large size and exhaustive characteristics which make it need to replace as soon as the power is exhausted. Clearly, this is not sustainable for the majority of implantable biomedical devices. For example, a pacemaker implanted inside human body has multiple functions, sensors and an integrated circuit which highly interacting with each other. This will require high power consumption and a replacement of batteries every few months. Obviously, this is not the desired solution for most of the biomedical implants. Currently, inductive RF coupling of energy, ambient-power scavenging, and energy harvesting are developed quickly. Ambient-power scavenging is the process of obtaining energy from sources like thermal and kinetic energy from human body [50]. Another possible solution to power IMD wirelessly is energy harvesting which harvests energy from the external environment. Nowadays, the near-field coupling is the most popular one in the industry which allows the transmission of power and data for short range.

3.5 Applications of Inductive Power Transfer(RFID type of devices)

Having explained the concept and principle of inductive coupling the primary focus in this thesis should be now introduced. This thesis will look at the applications involving inductively coupled RFID and biomedical embedded sensor system. Radio Frequency Identification (RFID) technology is widely in many applications. RFID can be categorised as passive or active depending on the power source. Active RFIDs are those with built-in batteries, and it can achieve a larger read range of (20-100m) as well as higher data rates. Active RFIDs operate at 455 MHz, 2.45 GHz or 5.8 GHz depending on the applications [49]. Passive RFIDs is suitable for small range and low data rates. Passive RFIDs is read by intercepting the magnetic field of the receiver. Figure 3.11 shows the general frequency bands for RFID systems. Implantable RFID can be used to send information

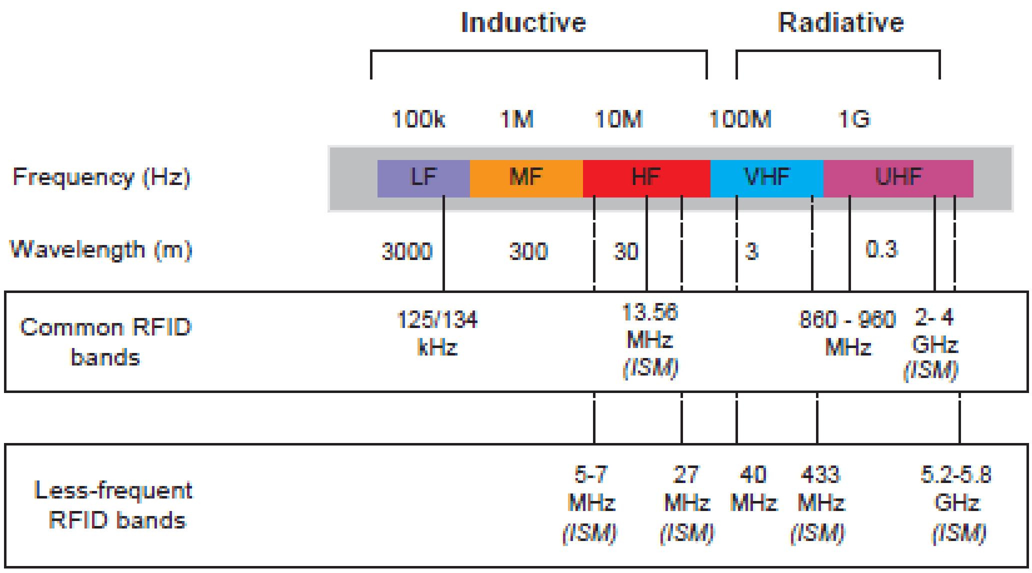


Figure 3.11: Frequency bands for RFID systems [11]

about patient's identity, physiological characteristics and health condition. This kind of application can be important when an emergency occurs. With the rapid development in radio-frequency technology, the use of embedded RFID devices for biomedical implants is expected to grow in the next few years.

3.6 Rectifier circuit - AC to DC

The use of rectifier circuit is essential in wireless power transmission system. The energy obtained from the electromagnetic field is high-frequency AC. However, this AC power can not be used directly. We need to make it DC with a magnitude of 1 to 3 V. Figure 3.12 is an example of rectifier circuit which contains diodes and capacitors. Diodes are commonly used in the design of rectifiers because of its characteristics. Diodes can only pass current in one direction which makes it suitable to obtain DC voltage from AC voltage. Based on the characteristics of the diode, a sine wave input should give us sinusoidal pulses at the output. Figure 3.13 indicates the result of a sine wave passing through a diode.

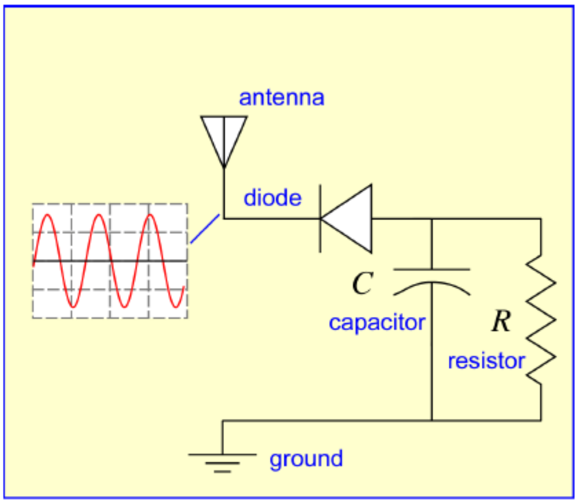


Figure 3.12: Rectifier circuit [7]

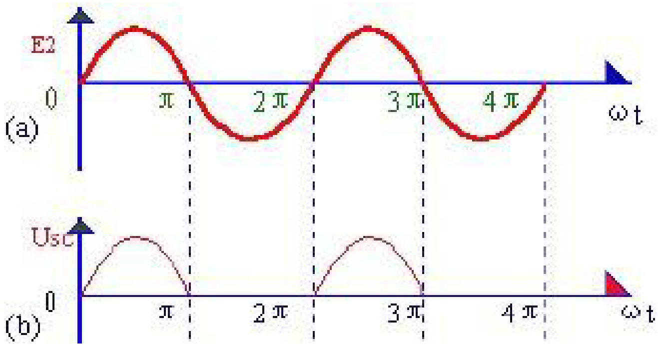


Figure 3.13: The result of a sine wave passing through a diode

The design and implementation of rectifier circuits are not difficult. This thesis will provide a review on three basic types of AC to DC circuits.

- Half Wave Rectifier

- Full Wave Rectifier
- Bridge Rectifier

3.6.1 Half Wave Rectifier

Half Wave Rectifier that formed with an ideal diode and a load could be the easiest design of a rectifier circuit. Figure 3.14 is a simple half wave rectifier circuit diagram. In half

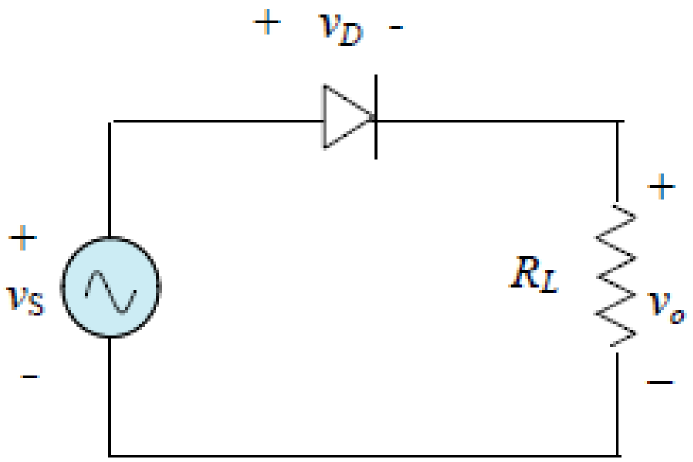


Figure 3.14: Example of half wave rectifier [32]

wave rectifier, current only flows in the posistive cycle of the Vin. The supply voltage Vs can be calculated by equation 3.8:

$$v_s = V_m sin\omega t \tag{3.8}$$

where ω is 2πf The diode acts as a closed switch when the source voltage is in the positive cycle, and it acts as an open switch when it's in the negative half cycle. During the negative half cycle, the Vs is disconnected from the load which result in Vo to be zero as no current flow to the load. Figure 3.15 shows the waveforms for Vo and Vs.

DC Output Voltage can be calculated by equation 3.9. DC output voltage is also the average load voltage.

$$V_{ave} = \frac{V_m}{\pi} \tag{3.9}$$

Average Load Current which is average load current can be calculated by equation 3.10.

$$I_L = \frac{V_{ave}}{R_L} \tag{3.10}$$

Peak Inverse Voltage is the maximum voltage that a rectifier can block. This can be calculated by equation 3.11.

$$PIV = V_m \tag{3.11}$$

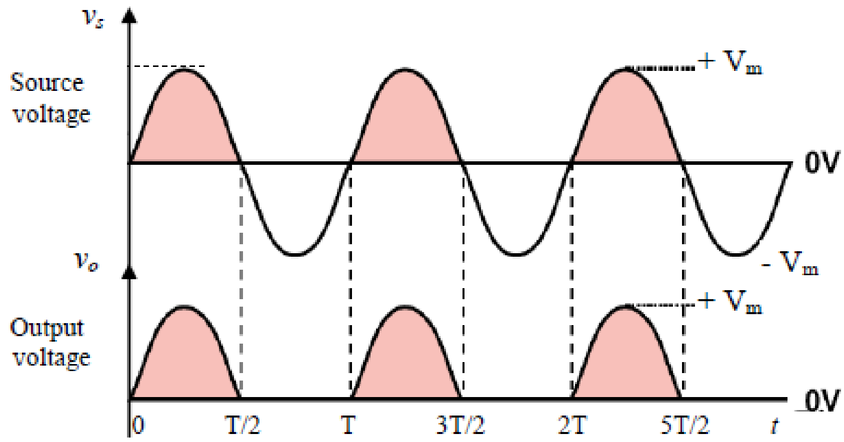


Figure 3.15: Waveforms for Vo and Vs [32]

3.6.2 Full Wave Rectifier

Figure 3.16 is a simple full wave rectifier which consists a resister R_L and two diodes. As can be seen from the figure, current flows clockwise through the circuit which means that diode 1 is forward biassed and diode 2 is reverse biassed. On the other hand, current flows counterclockwise when full wave rectifier is operating at the negative half cycle which results in diode 1 to be reverse biassed and diode 2 to be forward biassed [32]. Unlike half

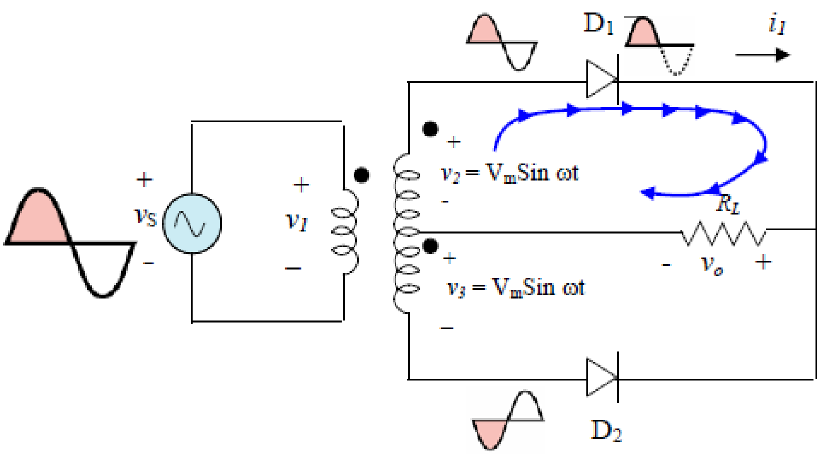


Figure 3.16: Full Wave Rectifier [32]

wave rectifier, a full wave rectifier has twice the output frequency and hence can produce twice the output. Therefore, the DC output voltage (V_{ave}) can be calculated as:

$$V_{ave} = \frac{2V_m}{\pi} \tag{3.12}$$

Based on the full wave rectifier operation, V_{total} is approximately twice as much as V_m.

$$PIV = 2V_m \tag{3.13}$$

3.6.3 Full Wave Bridge Rectifier

Apart from the half wave and full wave rectifier, there are another technology called full wave bridge rectifier. This rectifier can provide twice the output voltage as a full wave rectifier. Figure 3.17 is an equivalent bridge rectifier circuit. As figure 3.17 shows diode 1 and diode 3 are acting like closed switches and diode 2 and diode 4 are reverse biassed hence acting like open switches.

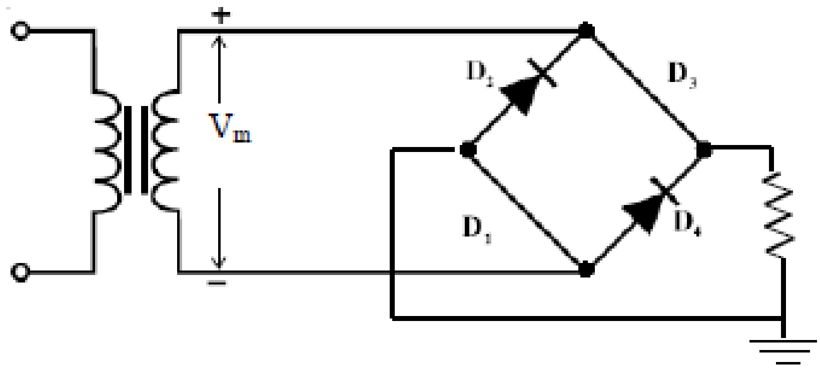


Figure 3.17: Equivalent circuit of a bridge rectifier [32]

The circuit can be further simplified as figure 3.18. As shown, diode 1 and diode 4 are both reverse biased and connecting in parallel. Therefore PIV is:

$$PIV = V_m \tag{3.14}$$

3.6.4 Use of Capacitors

A rectifier circuit can be further developed by adding capacitors as a filter. A capacitor is used to keep the voltage constant and make it into DC. A capacitor can store electric charges and its charging speed is much faster than its discharging speed. Therefore, a capacitor will start charge at the positive half of sine wave and discharge at the negative half. Since the discharging speed is slow, the voltage only decreases a bit. Thus, a constant voltage can be obtained by using a capacitor. Hence, a capacitor filter can be used to smoothen the ripple which is the output variations.

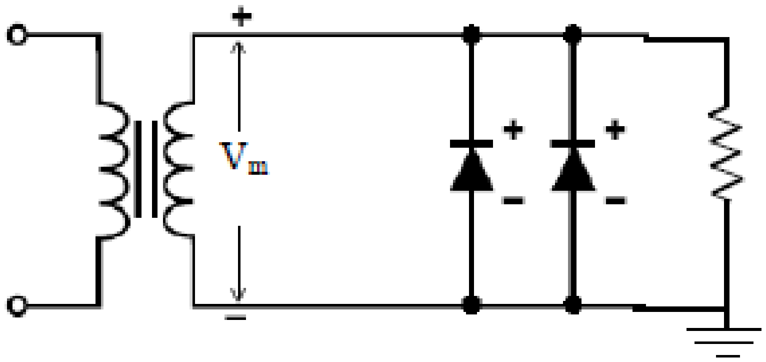


Figure 3.18: Simplified version of bridge rectifier circuit [32]

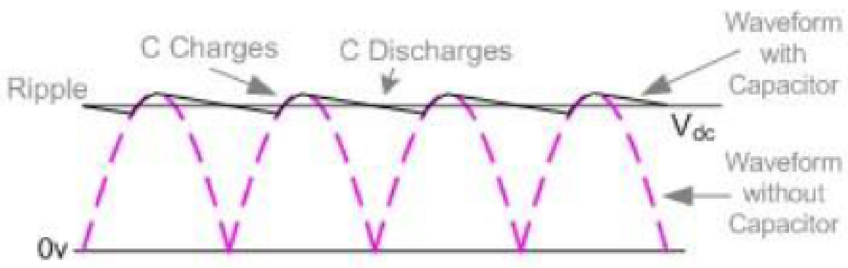


Figure 3.19: The result of sinusoidal pulses passing through a capacitor

3.7 Telemetry System

A telemetry system is essential in a wireless powering system. A telemetry system is a system that receives data and information from an external monitoring system to control the operation of biomedical implants. It can also receive external power and use it to recharge the onboard battery. Apart from that, a telemetry system is responsible for wirelessly transmitting data to an external receiver. There are two major techniques to build a telemetry system, namely RF and ultrasound. For the purpose of this thesis, RF approach will be chosen. The telemetry system needs to be miniaturised due to certain constraints such as size and weight of the implants. Several requirements need to be considered when designing a telemetry system such as data rate, power consumption and transmission distance [41]. Figure 3.20 is an architecture for inductively coupled RF power transfer. The transmission of command data from external unit to the biomedical implants and the telemetry of monitored physiological data from the implants back to the receiver can be achieved by applying the architecture in Figure 3.20. From figure 3.20, the voltage gain can be calculated by:

$$V_{out}/V_{in} = \frac{\omega^2 L_2 M}{R_1 R_2 + (\omega M)^2 + R_1(\omega L_2)^2/R_{Load}} \tag{3.15}$$

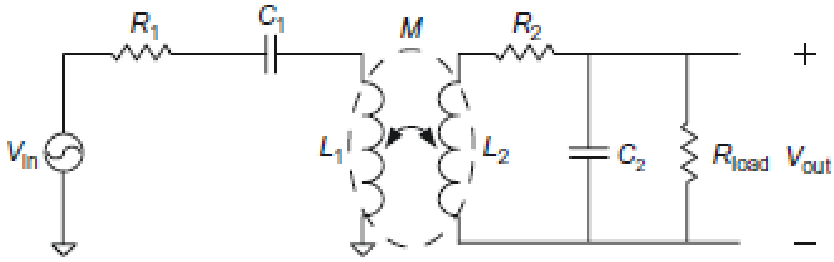


Figure 3.20: Inductively coupled RF power transfer architecture

Where M is the mutual inductance between L_1 and L_2 and ω is the operating frequency.

3.8 Signal-to-Noise Ratio(SNR)

SNR is a parameter which indicates the performance of the communication system and it is defined as:

$$SNR = P_{Signal}/P_{Noise} \tag{3.16}$$

Where P_{Signal} is the signal power and P_{Noise} is the noise power. SNR is used in evaluating the reliability of the link between Tx and Rx [23]. SNR decreases as the noise level increases in the channel which result in a poor performance for signal strength. As can be seen from equation**??**, there are two methods to improve SNR, which is 1).Increase the signal power or 2). Decrease the noise level. SNR in dB can be calculated by equation 3.17:

$$SNR(dB) = 10log_{10}(P_{signal}/P_{noise}) \tag{3.17}$$

3.9 Path Loss

In a communication system that contains one transmitter and one receiver, there will be a loss in the strength of an EMF as the signal is transferring from Tx to Rx due to many reasons such as reflection, diffraction, free-space loss and coupling loss [23]. This phenomenon is called path loss. Pass loss is one of the signal propagation characteristics that is essential in designing telecommunication system. The formula for calculating path loss in dB is given by 3.18:

$$L = 10nlog_{10}(d) + C \tag{3.18}$$

where

- L = Path loss in dB

- d = Distance between Tx and Rx
- n = Path loss exponent
- C = Constant that accoounts for system losses

3.9.1 Free Space Model

In free space model, it assumes that a signal is transmitting between Tx and Rx in free space without considering any reflection and diffraction caused by obstacles. The signal power decreases as the distance between Tx and Rx increases for any wireless communication system. This is true even if there is no obstacles or any attenuation. The equation 3.19 proves the point that signal disperses with a larger distance. Figure 3.21 is a simple model for free space path loss.

$$P_L = \frac{P_t}{P_r} = (\frac{4\pi d}{\lambda})^2 = (\frac{4\pi df}{c})^2 \tag{3.19}$$

where

- λ = Carrier frequency
- f = Carrier frequency
- d = Distance from the Tx in meters
- C = Speed of light (2.99792458× 10^8 $m/s \approx 3\times\ 10^8$)

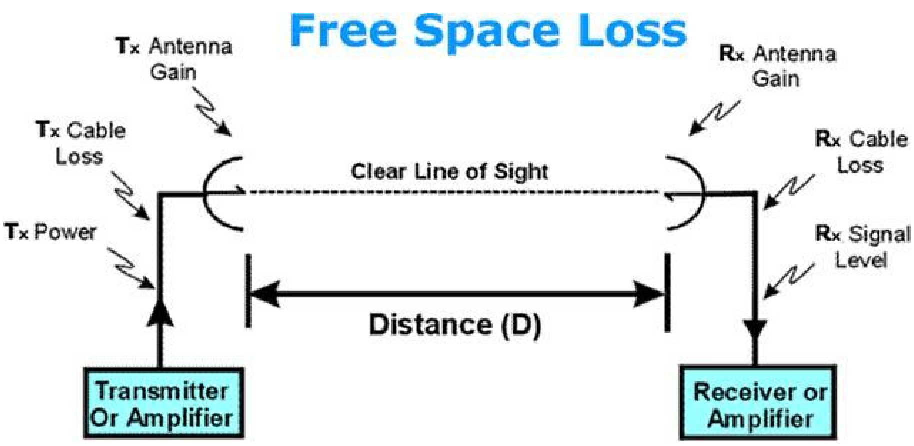


Figure 3.21: Path loss in free space [45]

3.10 Shannon Capacity

In a given wireless communication system, Shannon Capacity indicates the maximum error-free rate at which data can be transmitted through a channel. It is the theoretical upper limit of the communication system which is not achievable [23]. Shannon capacity is a parameter that can be used to evaluate the performance of the wireless communication system. The system will have better performance as its close to Shannon upper bound. Figure 3.22 is a Shannon capacity in AWGN model. Shannon Capacity is given based on

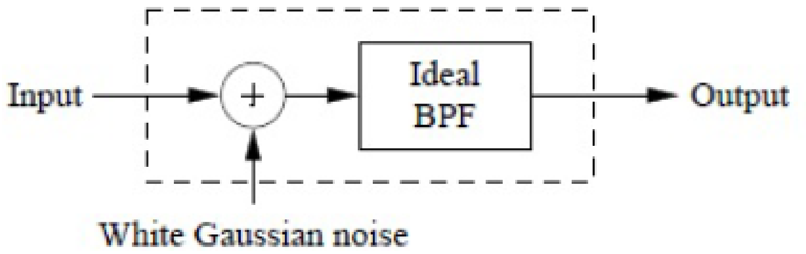


Figure 3.22: Shannon-Hartely theorem [2]

the Shannon-Hartley theorem:

$$C = B * log_2(1 + SNR) \tag{3.20}$$

Where

- C = Capacity(bits/s)
- B = Bandwidth(Hz)
- SNR = Signal-to-Noise ratio

3.11 Modulation Techniques

Modulation is an essential process for data transmission no matter the types of telemetry link. Data needs to be modulated onto a carrier for wireless transmission. Hence, some of the conventional modulation technologies will be introduced here including amplitude modulation (AM), frequency modulation (FM), quadrature phase-shift keying (QPSK) and quadrature amplitude modulation (QAM). Figure 3.23 shows some typical digital to analog modulations.

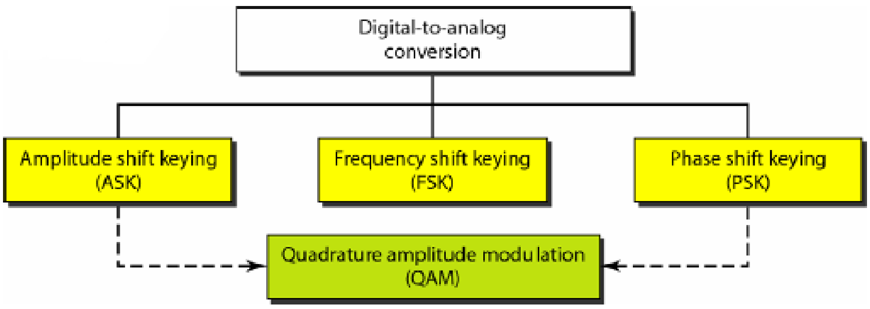


Figure 3.23: Types of Digital to Analog Modulation [23]

The details of each modulations and the mathematical formulations will be given as part of the design process.

Chapter 4

System Modelling and Simulation Results

4.1 Wireless charging system

In this chapter, the wireless charging system will be design and the trade-offs will be mathematically formulated. Then, the test benches will be designed in AWR along with its simulations. A wireless power transmission system should have three parts, namely wireless power transmission system, telemetry system and sensor system. In this thesis, wireless power transmission system and telemetry system will be the focus. System level design for this wireless charging system will be given. Since the objective of this project is to define and quantify the tradeoffs between power requirement, data rate, frequency and lifetime, therefore, we will need to look at different functionalities(e.g.,wireless transmission system, telemetry system and sensor system). This thesis will only focus on the system level design of wireless transmission and telemetry system.

4.2 Design for power requirements

To start the design, we need to understand the power requirement for some of the common biomedical implants. Table 4.1 summarizes several power requirements for IMDs.

IMD	**Power Requirement**	**Lifetime**	**Energy Source**
Pacemaker	<100 μW	10 years	Primary battery
Hearing Aid	100 to 2000 μW	1 week	Rechargeable battery
Cochlear Implant	20 to 100 mW	N/A	Inductive power
Retinal implant	40 to 250 mW	N/A	Inductive power
Neural recorder/stimulator	1 to 100 mW	N/A	Rechargeable battery
Artificial heart	10 to 100 W	N/A	inductive power

Table 4.1: Power Requirement and Energy Source for IMDs

4.3 Wireless link design

The background knowledge and comparison between multiple wireless power transmission technologies have been introduced in the literature review. Inductive coupling seems to be the most suitable choice for implantable biomedical devices. This has been proved in a variety of applications where the inductive coupling is employed. Therefore, this section will provide the system level design for a wireless link which will be implemented for power and data transmission. Figure 4.1 is the proposed design for an inductive power link. In the coupling part, two coils are tuned to the same frequency to enhance the efficiency. From the block diagram, we can see that the system consists of three main parts which are TX, inductive link and RX. At the power transmitter side, a power amplifier is used to drive L1 at the carrier frequency fc. Then the signal is induced onto L2 through the electromagnetic flux coupling and generates AC voltage across the resonance circuit. Finally, this signal is passing through an AC-DC converter to provide constant DC voltage for the rest of the receiver [41].

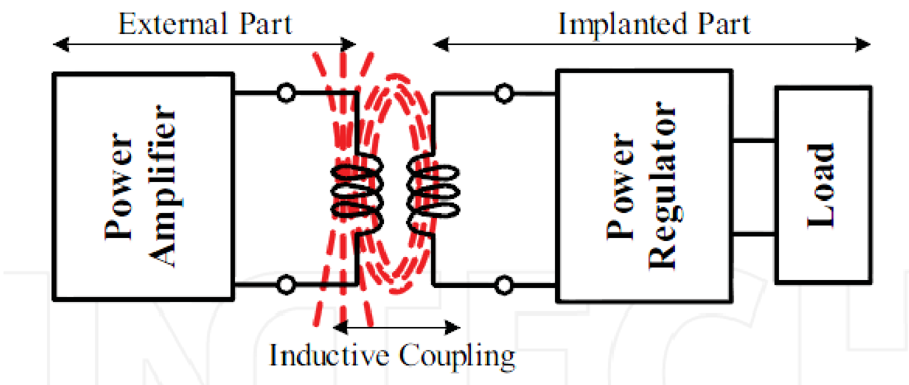


Figure 4.1: Inductive power link

The left-hand side of the coil (primary coil) is driven by a class-E amplifier to convert DC voltage to the magnetic field. The reason why the class-E amplifier is chosen is that class-E amplifier has high efficiency compare to other types of amplifiers which are approximately one hundred percent in theory [1]. The coil coupling coefficient can be mathematical formulated as:

$$K = \frac{M}{\sqrt{L_1 L_2}} \tag{4.1}$$

In order to calculated the power efficiency of the wireless power link, a simplified circuit is designed.

As shown in the circuit 4.2, two capacitors are inserted to produce resonance to the link.

$$\omega_0 = \frac{1}{\sqrt{L_1 C_1}} = \frac{1}{\sqrt{L_2 C_2}} \tag{4.2}$$

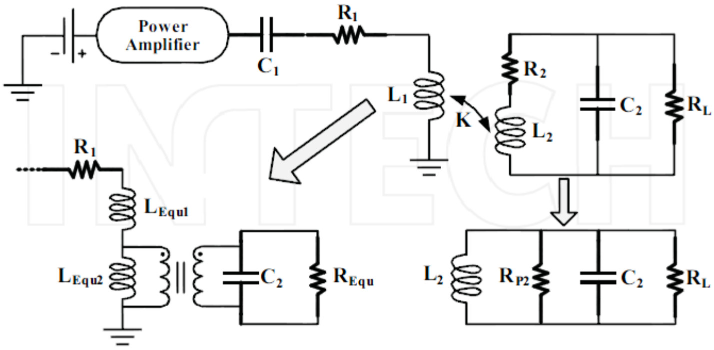


Figure 4.2: Simplified equivalent circuit

According to the equivalent circuit, the efficiency for secondary coil can be calculated by:

$$\mu_2 = \frac{R_{P_2}}{R_{P_2} + R_L} \tag{4.3}$$

In order to calculate the efficiency for primary coil, we will use the ideal transformer model in figure 4.2.

$$\mu_1 = \frac{K^2 Q_1 Q_2}{1 + K^2 Q_1 Q_2 + \frac{R_{P_2}}{R_L}} \tag{4.4}$$

Where Q_1 and Q_2 are the quality factors of two sides of coil resectively. Hence the total power efficiency can be calculated as:

$$\mu = \mu_1 \mu_2 = \frac{K^2 Q_1 Q_2}{1 + K^2 Q_1 Q_2 + \frac{Q_2}{Q_L}} * \frac{1}{1 + \frac{Q_L}{Q_2}} \tag{4.5}$$

From this equation 4.5, we can see that the total power efficiency of the inductive wireless link is determined by both K and Q. Hence, the design of primary coil and secondary coil is essential to improve the total efficiency. However, coils are weakly coupled, and usually have a coupling coefficient of less than 0.4 for implantable biomedical devices as reviewed. A relationship can be discovered from equation 4.3 and 4.4, primary coil efficiency decreases as Q_L decreases. In the contrast, the secondary coil efficiency increases when Q_L decreases. Therefore, there must be a value for Q_L that maximise the total efficiency of the link.

$$Q_L, optimum = \frac{Q_2}{\sqrt{1 + K^2 Q_1 Q_2}} \tag{4.6}$$

As a result, the maximum achieveable link efficiency can be calculated by:

$$\mu_{Max} = \frac{K^2 Q_1 Q_2}{(1 + \sqrt{1 + K^2 Q_1 Q_2})^2} \tag{4.7}$$

Compare to another near field wireless power transfer technologies; inductive coupling has the best performance. Additionally, power is transfer via magnetic field in inductively coupled coils which makes it safe for the human tissue because human body only absorbs electric current generated from an electric field.

4.4 Total Power Transfer Efficiency of the System

As introduced above, the proposed design has several components including a rectifier, amplifier and regulator. The total power efficiency is a combination of each efficiency. Total power transfer efficiency can be calculated by:

$$\eta_{Total} = \eta_S * \eta_1 * \eta_T * \eta_2 * \eta_L \tag{4.8}$$

Where η_S, η_1, η_T, η_2, η_L are the efficiencies of the power amplifier, primary LC circuit, transcutaneous powering through the tissue, secondary LC circuit and AC-DC converter and the regulator. Where $\eta_1 {*} \eta_T {*} \eta_2$ is the inductive link efficiency η_{Link} [18]. As can be seen from equation 4.8, the total efficiency decreases as we add more units to the system. This is because there are power losses at each stage which result in a decrease in total efficiency. Therefore, improving power efficiency for each stage is a key to enhancing the power receiving efficiency of the system. When designing an inductively coupled system, higher power transfer efficiency is essential. This can be achieved by increasing the efficiencies for AC-DC converter and regulator. This thesis is not focused on the circuit design for each stage. Therefore, the details design for the amplifier, rectifier and regulator will not be addressed here. A system level design will be given in the next section.

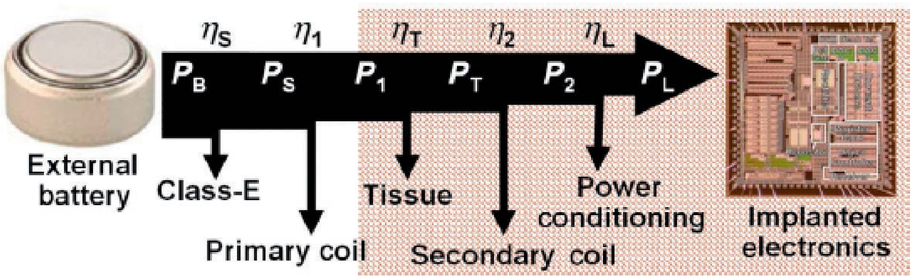


Figure 4.3: inductive power transmission flow from TX to the IMD

4.5 System Level Design for Different Modulations

Figure 4.4 is a block diagram of a wireless communication system. The signal is transmitting through a channel with the presence of noise. The signal might be attenuated as passing through the channel. The signal will be mixed with channel noise at the receiver. As can be seen from the figure, S_0 and N_0 are the signal power and noise power at the receiver output. This output SNR can be used to evaluate the performance of the communication system or the quality of the received signal. From equation 3.16, the value for SNR can be increased by increasing the transmitted power. However, it is often hard to increase the transmitted power in practice due to certain limitations such as channel capacity, cost and interference. Therefore, finding the optimal value for SNR is a figure of merit in wireless communication system design. In this section, we will look at SNR for two proposed systems with different modulations.

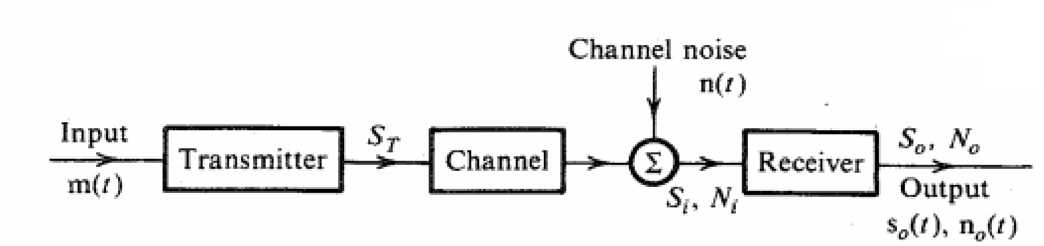


Figure 4.4: Wireless communication system model

4.5.1 QPSK

QPSK is digital modulation which uses 90-degree phase shift. With QPSK, the carrier undergoes 4 changes in phase, each representing 2 bits. QPSK allows high data transmission compare to PSK in the same bandwidth.

$$s(t) = \begin{cases} A\cos(2\pi f_c t), & \text{binary } 00 \\ A\cos(2\pi f_c t + \frac{\pi}{2}), & \text{binary } 01 \\ A\cos(2\pi f_c t + \pi), & \text{binary } 10 \\ A\cos(2\pi f_c t + \frac{3\pi}{2}), & \text{binary } 11 \end{cases}$$

Figure 4.5: QPSK

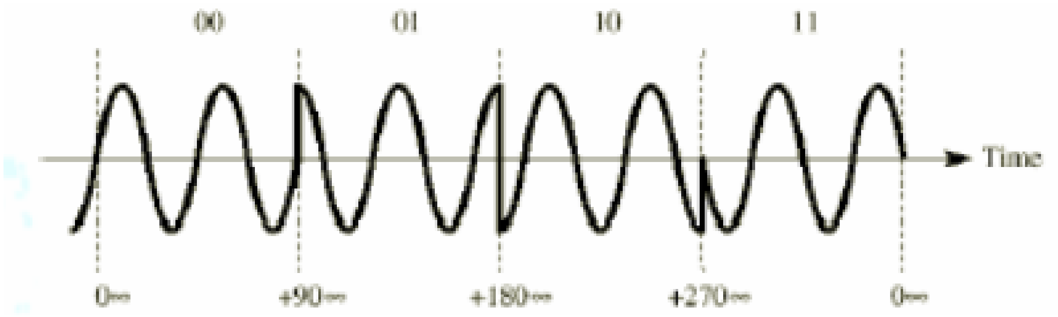


Figure 4.6: QPSK Modulated Signal

Additive White Gaussian Noise Channel

For QPSK, the channel can be modeled as:

$$y = ax + n \tag{4.9}$$

Where

- y = Received signal at the input
- x = Modulated signal
- n = Additive White Gaussian Noise
- a = Scaling factor = 1

For Additive White Gaussian Noise, the noise variance is defined as:

$$\sigma^2 = \frac{N_0}{2} \tag{4.10}$$

The symbol energy is calculated as:

$$E_s = R_m R_c E_b \tag{4.11}$$

Where

- Es = Symbol energy per modulated bit
- Rm = $log_2(M)$, M equals 4 for QPSK
- Rc = 1
- Eb = Energy per information bit

Then, Eb/No is given as:

$$\frac{E_b}{N_0} = \frac{E_s}{R_m R_c N_o} = \frac{1}{2 R_m R_c \sigma^2} \tag{4.12}$$

4.5.2 Analytical Solutions

We will look at the proposed model between the frequency range of 0 to 10 GHz, which is where the data/voice transmission usually be done. For wireless powering of an implantable biomedical device, large charging distance is not essential. Therefore, the distance between Tx and Rx can be assumed as 1 meter. The distance between Tx and Rx is dependent on path loss as well. Hence, we assume it to be constant during the operation. The transmitter gain and receiver gain is equal to 1 for the purpose of this project. The free-space path loss is calculated as:

$$FEPL = 20log_{10}(d) + 20log_{10}(f) + 20log_{10}(\frac{4\pi}{c}) - G_t - G_r \tag{4.13}$$

Substitute all the values into equation 4.13,

FSPL(dB)	**Frequency(GHz)**	**Distance(meter)**	**Gt(dB)**	**Gr(dB)**
24.42	0.5	1	1	1
30.44	1	1	1	1
33.96	1.5	1	1	1
36.46	2	1	1	1
38.46	3	1	1	1
42.48	4	1	1	1
44.42	5	1	1	1
46	6	1	1	1
47.34	7	1	1	1
48.5	8	1	1	1
49.52	9	1	1	1
50.44	10	1	1	1

Table 4.2: Path loss vs frequency

Theoretically, path loss vs frequency can be plotted as:

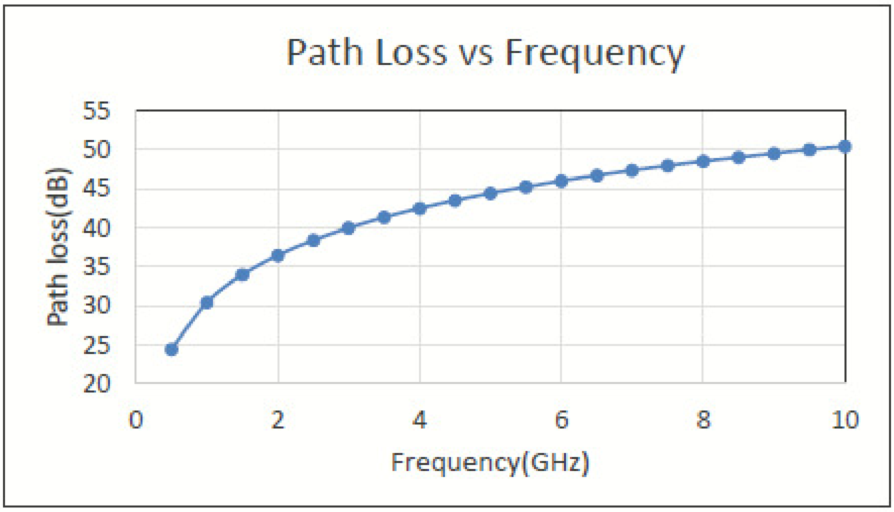


Figure 4.7: Path loss vs frequency(0.5GHz to 10GHz)

The similar plot can be done to show the relationship between path loss and distance if simulating at the same frequency. According to the equation 4.13, the distance between Tx and Rx is proportional to path loss. Path loss increases quickly as distance increases. Therefore, the transmitted signal gets attenuated by the channel loss. Hence, the poor quality of received signal at the receiver output is expected. The BER can be calculated and plotted as a function of SNR according to Shannon capacity theorem and SNR for QPSK. BER is given as

$$BER = \frac{1}{2} erfc(\sqrt{\frac{E_b}{N_o}}) \tag{4.14}$$

BER plot is often used in evaluating the performance of a digital communication system while BER vs SNR plot is used in describing wireless communication.

4.5.3 AWR Test Bench Setup

Figure 4.8 is a system level design of a transceiver with QPSK modulatiuon technique. The simulation was done in 10 GHz with the presence of additive white gaussian noise between transmitter and receiver. The path loss in the AWGN channel has been modified as a tunable variable. The changes in power spectrum can be obtained by varing the path loss in the AWGN channel.

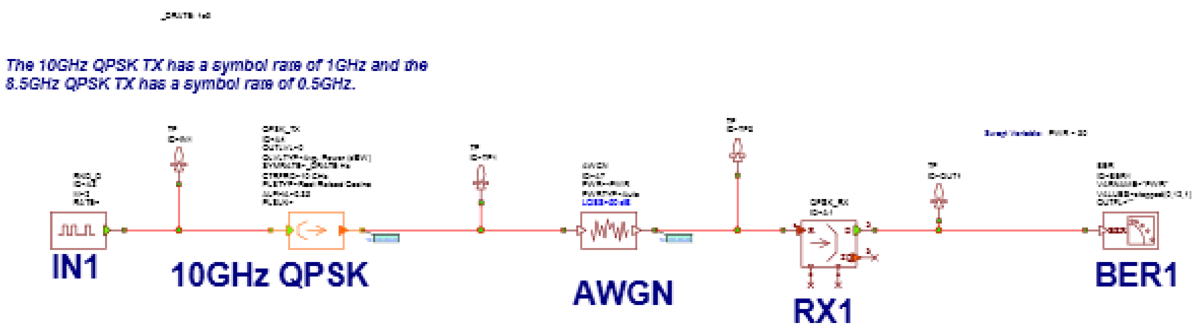


Figure 4.8: AWR Test Bench with QPSK Modulation

4.5.4 Simulations Results

Figure 4.9 is the QPSK power spectrum at 0 dB path losses.

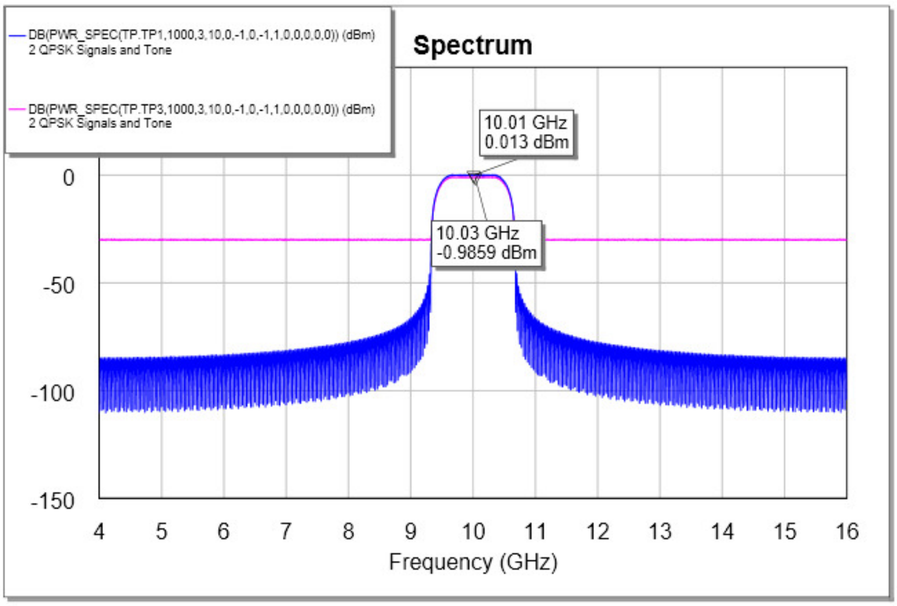


Figure 4.9: QPSK Power Spectrum at 0 dB Losses

Figure 4.10 is the QPSK power spectrum at 20 dB path losses.

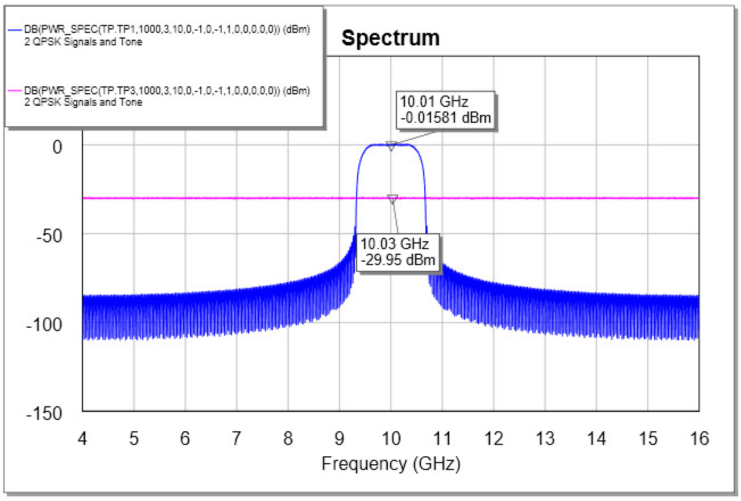


Figure 4.10: QPSK Power Spectrum at 20 dB Losses

Figure 4.11 is BER vs SNR plot for QPSK at frequency of 10GHz.

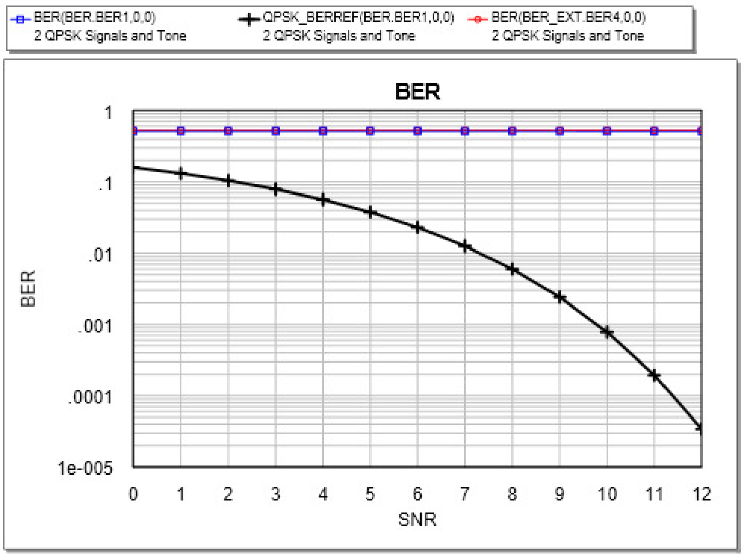


Figure 4.11: BER vs SNR for QPSK

4.5.5 Discussion on the Proposed QPSK Model

Based on the data generated by AWR simulation of QPSK model, several observations can be obtained. As we can see from figure 4.9, the transmitted signal is approximately equal to the received signal at the receiver output when there are no losses in the AWGN channel. This means that the transmitted signal gets attenuated or even distorted as the signal passing through channels. The signal strength or quality falls off with the increasing in path loss. Pass loss is proportional to distance and frequency. From figure **??**, two signals are getting mismatched as path loss increases which indicate a signal distortion at the output. Therefore, the frequency and wireless link design are essential for designing a wireless power/data transmission system. In wireless communication, BER indicates the quality of the link. The acceptable BER value of a data link is 10^{-9}. For a coherent QPSK system, the value for SNR should be greater than 12dB for a BER of better than 10^{-3} [10]. As shown in Figure 4.11, BER decreases as SNR increases. This means that there is less error in the signal when SNR is low. More error is expected to occur as SNR reduces. Consider figure 4.11, BER is approximate 10^{-4} at 12dB SNR. Therefore, the proposed design with QPSK is acceptable referring to the requirements. Normally, the data rate, noise level and path loss are relatively low at the low-frequency range. In contrast, higher frequencies give high data rate, hence the more in noise and path loss. In wireless communications, the BER can be improved by adding more bandwidth. More bandwidth can increase the BER and can allow complex modulations. The tradeoffs can be defined as more complex modulations can be used to enhance the system data rate at the same frequency if we increase the value of SNR. To increase the value of SNR, the output power has to be increased. However, the increase in output power has huge effects on the system power consumption. Therefore, several parameters need to be considered when designing a wireless power transmission system. Hence, the design of frequency, data rate and power consumption in a wireless powering system is an appropriate figure of merit.

4.6 QAM

QAM can be used for both digital and analog modulations. In QAM, two signals are sent simultaneously on the same frequency. The output contains both amplitude and phase variations as two carriers are shifted by 90 degrees and modulated [42]. Figure 4.12 is a representation of QAM.

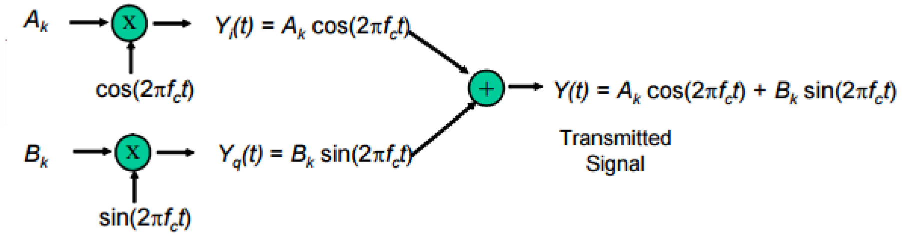


Figure 4.12: Mathmatical Representation of QAM [46]

QAM is also a combination of ASK and PSK modulations.

$$Y(t) = A_k \cos(2\pi f_c t) + B_k \sin(2\pi f_c t) = \left(A_k^2 + B_k^2\right)^{\frac{1}{2}} \cos(2\pi f_c t + \tan^{-1}\frac{B_k}{A_k})$$

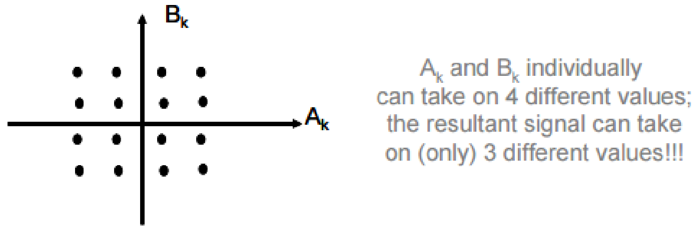


Figure 4.13: A combination of ASK and PSK [46]

4.6.1 BER Calculations in AWGN channel

In a general M-QAM model where M is $M = 2^b$, the symbol error rate can be defined as:

$$P_M QAM(SNR) = 4\frac{\sqrt{M}-1}{\sqrt{M}} Q(\sqrt{\frac{3}{M-1}\frac{kE_b}{N_0}}) - 4(\frac{\sqrt{M}-1}{\sqrt{M}})^2 Q^2(\sqrt{\frac{3}{M-1}\frac{kE_b}{N_0}}) \quad (4.15)$$

Where Q is the defined as:

$$Q(x) = \frac{1}{\sqrt{2\pi}} \int_x^{\infty} exp(-\frac{x^2}{2}) dx \quad (4.16)$$

4.6.2 Path Loss in the Channel

In this section, the path loss will be looked with the presence of AWGN. The specified frequency for the designed model is at 900MHz, 2.4 GHz and 5.8GHz band which is the industrial, scientific and medical (ISM) radio bands. The free space path loss can be calculated by:

$$FEPL(dB) = 10log_{10}(\frac{4\pi df}{c})^2 \tag{4.17}$$

Table **??** indicates the free space path loss with respect to different frequencies.

	Free Space Path Loss(dB)		
Distance(Meter)	**900MHz**	**2.4GHz**	**5.8GHz**
0.1	9.525	18.04	25.71
0.2	15.55	24.06	31.73
0.3	19.07	27.59	35.25
0.4	21.57	30.09	37.75
0.5	23.50	32.02	39.69
0.6	25.09	33.61	41.27
0.7	26.43	34.95	42.61
0.8	27.59	36.11	43.77
0.9	28.61	37.13	44.79
1	29.52	38.04	45.71

Table 4.3: FSPL

4.6.3 SNR

SNR is the most important aspect to evaluate the level of reliability for the received signal at the receiver output. It is given in terms of BER:

$$SNR = ReceivedPower - ChannelNoise \tag{4.18}$$

Figure 4.14 shows the minimum SNR required for different modulations.

Modulation & Encoding Scheme	Data Rate (Mbps)	SNR (dB)
BPSK 1/2	6	8
BPSK 3/4	9	9
QPSK 1/2	12	11
QPSK 3/4	18	13
16-QAM 1/2	24	16
16-QAM 3/4	36	20
64-QAM 2/3	48	24
64-QAM 3/4	54	25

Figure 4.14: Data Rate vs Minimum SNR [46]

4.6.4 System Level Design in AWR for QAM

Figure 4.15 is a system level design of a transceiver with QAM modulation technique. The simulation was done in the industrial, scientific and medical (ISM) radio bands 900MHz, 2.4GHz and 5.8GHz respectively with the presence of additive white gaussian noise between transmitter and receiver. The path loss in the AWGN channel has been modified as the tuneable variable. The changes in power spectrum can be obtained by varying the path loss in the AWGN channel.

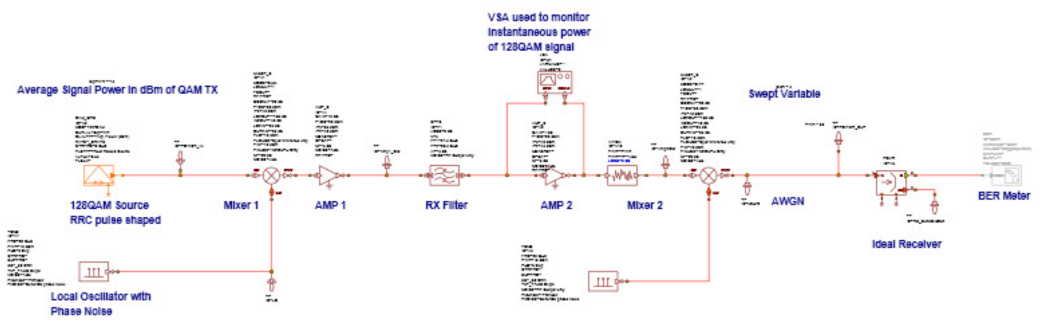


Figure 4.15: Test bench for QAM

4.6.5 Simulation Results

Figure 4.16 is the QAM power spectrum at 0 dB path losses.

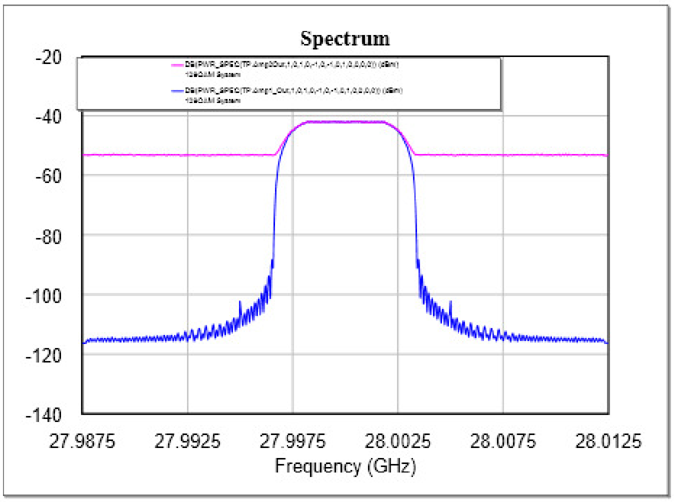


Figure 4.16: QAM Power Spectrum at 0 dB Losses

Figure 4.17 is the QAM power spectrum at 20 dB path losses.

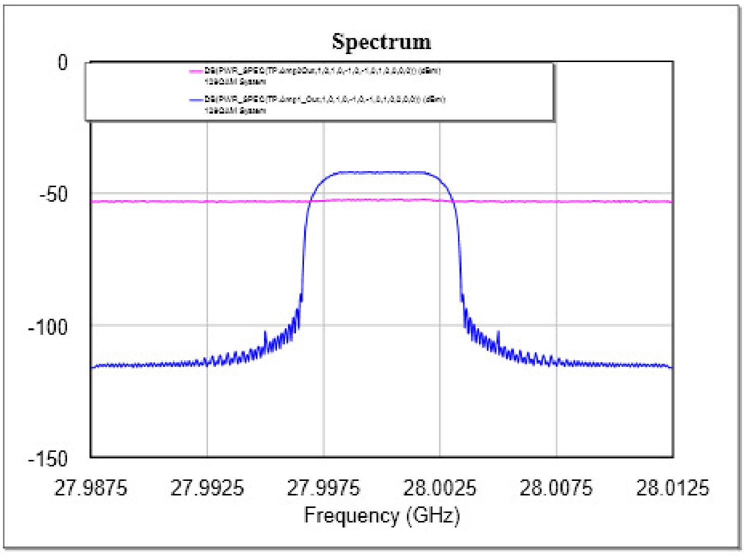


Figure 4.17: QAM Power Spectrum at 20 dB Losses

Figure 4.18 is BER vs SNR plot for QAM.

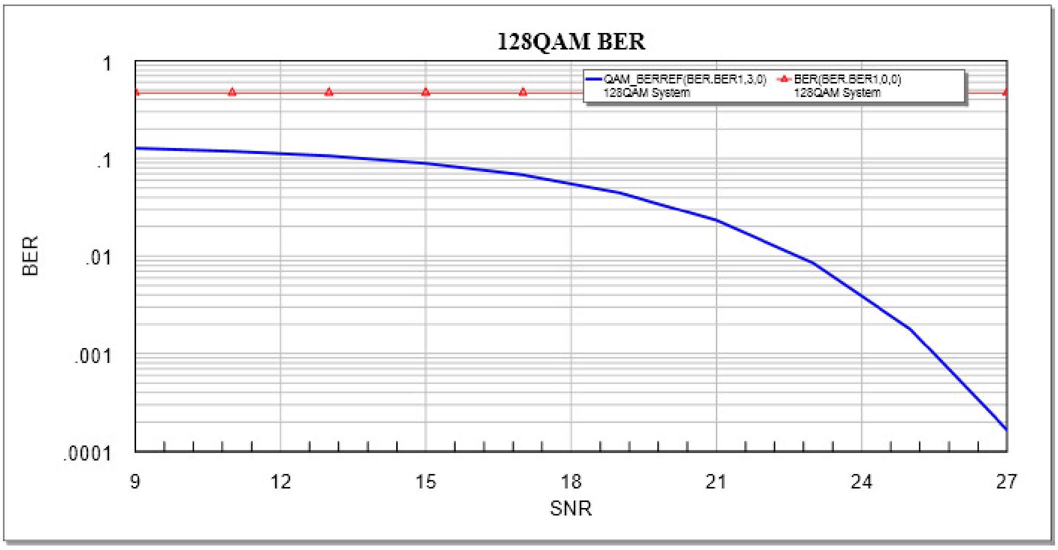


Figure 4.18: BER vs SNR for QAM

4.6.6 Dicussion

The necessity of these simulation results is to show the relationship between BER, SNR, channel capacity, data rate, frequency and distance. Those parameters are essential in designing a coherent wireless power/data transmission system for implantable biomedical devices. The desired BER and SNR graphs are plotted in AWR for simulation in Additive White Gaussian Noise channel. From the simulation results, a comparison can be provided between QPSK and QAM. QAM is less affected by noise compare to QPSK. QAM can provide more data in a bandwidth than QPSK. However, this requires more power. Therefore, QAM is efficient in bandwidth but not in power consumption. Table 4.4 shows the comparison between two modulations.

Modulation	C/N Ratio(dB)	SNR(dB)
QPSK	13.6	10.6
QAM	28.6	19.8

Table 4.4: Performance Comparsion between QPSK and QAM

Therefore, the tradeoffs between power consumption, BER and bandwidth can be summarized as:

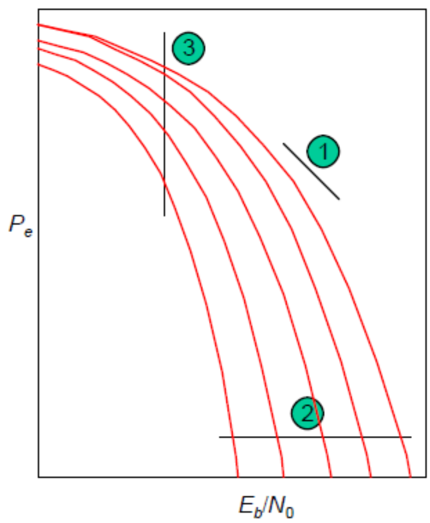


Figure 4.19: Graphical Representation of the Tradeoffs

- (1). Trade BER for Power when data rate is fixed.
- (2). Trade data rate performance for power when BER is fixed.
- (3). Trade BER performance for data rate when power is fixed.

Chapter 5

Future Work

The thesis discussed researches and activities performed for this project. However, due to the time constraints and the lack of experience, some of the aspects were not deeply studied. Further research on the topic is expected in higher degree studies. The circuit design and implementation of a wirelEss power transfer system for an implantable biomedical device may be chosen for future projects since it is beyond the scope of this project. The design for rectifier, amplifier and regulator in the wireless powering system needs to be investigated in the future. The development for wireless power transfer technology has been enormous. Therefore, the demand for providing robust, reliable and efficient power supply to the IMDs is expected to be huge in the following years. There are some directions that can be investigated in the future.

5.1 Small Size Receiver

The size of the receiver inside the human body is always the limitation. In the future, the ultra small size receiver is required for micro implants. Both sizes of the transmitter and receiver are going to be minimised. Therefore, more circuit implementation technologies and packaging methods are required in the future.

5.2 Feedback Control

In this project, a telemetry system is introduced in the review. The telemetry system provides a feedback loop between the external unit and internal unit. Data can be transmitted through the telemetry system. Some of the implants require constant data transmission and fast data rate for monitoring purpose. For example, a feedback loop can be used to send blood pressure, glycemic index and blood oxygen content to the external transmitter to trigger external devices to deliver actions. This could be an interesting field to look at in the future.

Chapter 6

Conclusions

6.1 Conclusions

This document has reported the research, formation and solution of the problem which is to design a wireless power transfer system for a biomedical implant. The primary objective is to define and quantify the tradeoffs between power, frequency and data rate in a wireless power transfer system. This thesis has looked at different wireless charging technologies including indictive coupling and capacitive coupling. A comparison is given to select the better charging tehcnologies. Two system level designs of the system were provided along with simulations. Different modulation technologies are also studied in the project. Based on the theoretical and experimental data obtained during the project, the tradeoffs are studied at the end of Chapter 4. The project can be summarised as:

- Wireless charging technologies are studied. Inductive coupling is chosen as the solution to provide wireless power transfer. Additionally, Inductive coupling works better in near field transfers.
- Wireless charging standards and regulations are stated in the thesis. The design of wireless power/data transmission is strictly under the standards. Safety must be concered as part of the design procedure.
- Two system models with different modulations are designed based on the analytical solutions. The QPSK modulations is compared with QAM modulation. More complex modulation methods allow high data rate. The tradeoffs between parameters are an appropriate figure of merit.
- The AWR simulation results were discussed based on the theory. Plots are used to demonstrate the findings and outcomes.
- Finally, the last section discussed some of the directions for future research.

Chapter 7

Abbreviations

RFID	Radio frequency identification devices
WPT	Wireless power transmission
ISM	Industrial,Scientific and Medical
AC	Alternating current
DC	Direct current
IMDs	Implantable biomedical devices
EMP	Electromagnetic powering
RF	Radio Frequency
TX	Transmitter
RX	Receiver
FCC	Federal Communications Commission
ASK	Amplitude-shift-keying
PSK	Phase-shift-keying
FSK	Frequency-shift-keying
QPSK	Quadrature Phase Shift Keying
QAM	Quadrature amplitude modulation
SNR	signal-to-noise ratio
BER	bit error rate
AWGN	Additive White Gaussian Noise

Appendix A

Project plan and deliverables

A.1 Overview

This table addressed some of the milestones and deliverables of the project.

A.2 Project Plan

Task	Start	End	Days	% Done
Literature Revirew	1/03/2016	4/04/2016	34	100
Project specification and plan	5/03/2016	15/03/2016	10	100
Understanding the scope of the project	1/03/2016	4/04/2016	34	100
System level design	25/03/2016	1/05/2016	35	100
Analytical solutions	25/03/2016	10/05/2016	44	100
Simulations through AWR	10/04/2016	15/05/2016	33	100
Comparison between the results	15/05/2016	17/05/2016	2	100
Thesis report draft	18/05/2016	25/05/2016	8	100
Thesis report final	25/05/2016	5/06/2016	11	100

Table A.1: Planned start day and end day for each of the activities

Appendix B

Curren Medical Telemetry Bands

Standard	Frequency	Data Rate	Range
Inductive Coupling Devices	< 1 MHz	1-30 kbps	<1m
Wireless Medical Telemetry System	608-614 MHz 1395-1400 MHz 1427-1429.5 MHz	>250 kbps	30-60m
Medical Device Radiocommunication Service	401-406 MHz	250 kbps	2-10m
802.11a Wi-Fi	5 GHz	54 Mbps	120m
802.11b Wi-Fi	2.4 GHz	11 Mbps	140m
802.11g Wi-Fi	2.4GHz	54Mbps	140m
802.11n Wi-Fi	2.4/5GHz	248 Mbps	250m
802.15.1 Bluetooth Class I	2.4 GHz	3 Mbps	100m
802.15.1 Bluetooth Class II	2.4 GHz	3 Mbps	10m
802.15.4 (Zigbee)	868, 915 MHz, 2.4 GHz	40 kbps 250 kbps	75m
World Interoperability for Microwave Access (WiMAX)	2.5 GHz	70 Mbps (fixed) 40 Mbps (mobile)	Several km

Figure B.1: Current Medical Telemetry Bands [29]

Appendix C

Radio Wave Spectrum

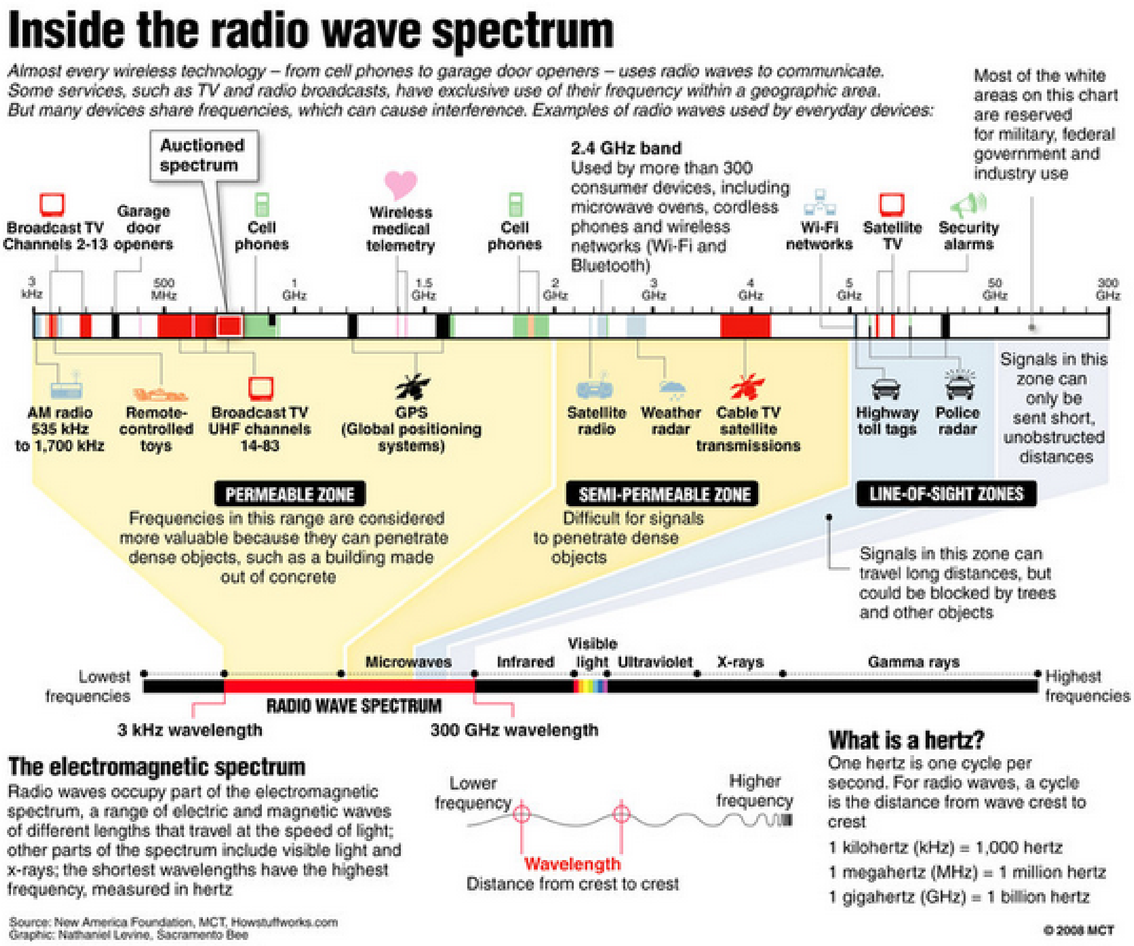


Figure C.1: Radio Wave Spectrum [3]

Appendix D

Frequency Range and Standards for Short Distance Wireless Communications

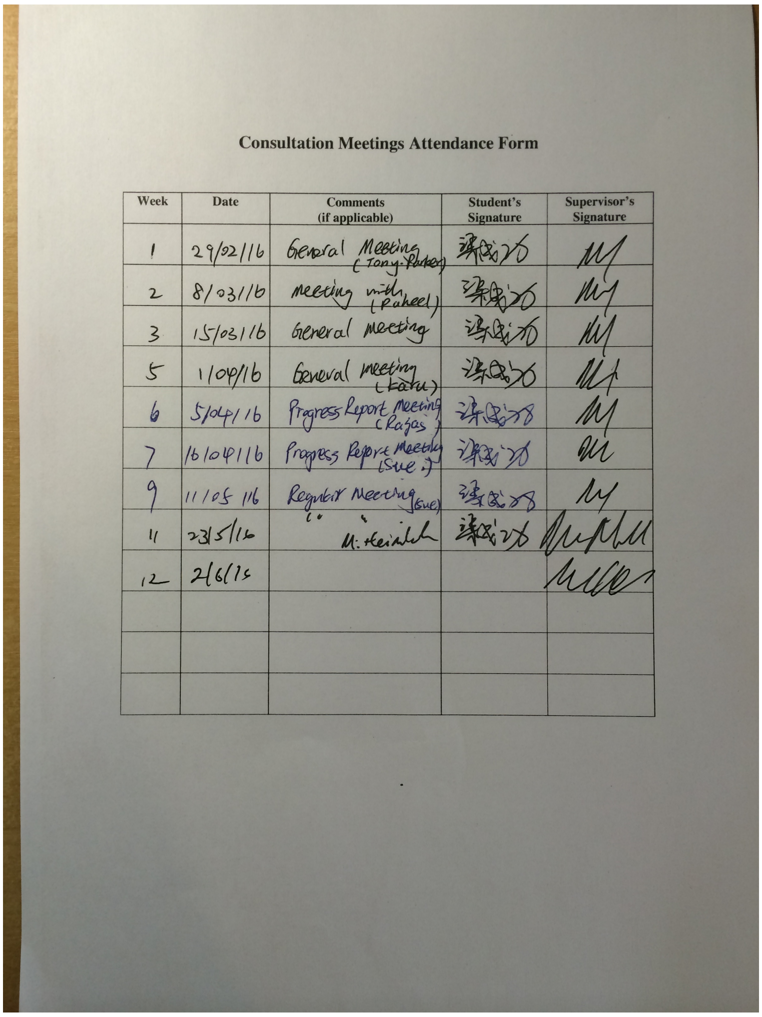
TABLE 6: MOST POPULAR SHORT-RANGE WIRELESS INTERFACES

Interface	Standard	Frequency	Speed (max)	Modulation	Range (max)	Application
802.15.4	IEEE	868, 902-928 MHz, 2.4-2.835 GHz	20, 40, 250 kb/s	DSSS with BPSK, or O-QPSK	10-100 m	Industrial, consumer, IoT, utility
Bluetooth	Bluetooth SIG	2.4-2.4835 GHz	1 Mb/s, 2.1 Mb/s, and 3 Mb/s	FHSS with GFSK, π/4-DQPSK and 8DPSK	1-100 m	Speakers, headsets, , medical, fitness, smart-phones and watches
Digital Enhanced Cordless Telecommunications	ETSI	1880-1930 MHz	Up to 2 Mb/s	GFSK, π/2 -DBPSK, π/4-DQPSK, π/8-D8PSK	200 m	Cordless phones, home automation
EnOcean	ISO/IEC	315, 868, 902-928 MHz	125 kb/s	ASK	30 m	Building or home auto-mation, industrial
ISA100-11a	ISA, IEC, IEEE, WCI/ASCI	2.4-2.4835 GHz	250 kb/s	DSSS with O-QPSK	10-100 m	Process automation, industrial, IoT
Near-field communications	ISO/IEC, ECMA, GSMA	13.56 MHz	106 - 424 kb/s	ASK	< 20 cm	Payments, access, pairing
Ultra Wideband (UWB)	IEEE, WiMedia Alliance	3.1-10.6 GHz	480 Mb/s, 1.3 Gb/s	OFDM, BPSK, pulse	< 10 m	Video, docking, military
Wi-Fi (802.11)	IEEE	2.4 - 2.4835 GHz, 5.725-5.875 GHz, 60 GHz	11 Mb/s to 7 Gb/s	DSSS, mostly OFDM	100 m	LAN, Internet access, IoT, industrial
WirelessHART	HART Comm. Foundation, IEEE	2.4-2.4835 GHz	250 kb/s	DSSS with O-QPSK	10-100 m	Process and building automation, sensor nets
ZigBee	IEEE, ZigBee Alliance	868, 902-928 MHz, 2.4-2.4835 GHz	20, 40, 250 kb/s	DSSS with O-QPSK	10-100 m	Industrial, home automation, IoT
Z-Wave	Z-Wave Alliance	908.42 MHz	9.6 and 40 kb/s	GFSK	30 m	Home automation, IoT

Figure D.1: Some of the popular short distance wireless interfaces [12]

Appendix E

Attendance Form



Consultation Meetings Attendance Form

Week	Date	Comments (if applicable)	Student's Signature	Supervisor's Signature
1	29/02/16	General Meeting (Tony Parker)	[illegible]	[illegible]
2	8/03/16	meeting with (Raheel)	[illegible]	[illegible]
3	15/03/16	General meeting	[illegible]	[illegible]
5	1/04/16	General meeting (Karu)	[illegible]	[illegible]
6	5/04/16	Progress Report meeting (Rajas)	[illegible]	[illegible]
7	16/04/16	Progress Report Meeting (Sue)	[illegible]	[illegible]
9	11/05/16	Regular Meeting (Sue)	[illegible]	[illegible]
11	23/5/16	" " M. Heinrich	[illegible]	[illegible]
12	2/6/16			[illegible]

Figure E.1: Attendance Form for Weekly Meetings

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Thesis Preparation Assignment: Wireless powering of RFID devices

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***Abstract*— This document provides the preliminary research prior to the final year thesis project. The purpose of this document is to carry out the background research for Wireless powering of RFID devices. RFID stands for Radio-Frequency Identification. RFID are quite popular for a variety of applications. This report will focus on RFID devices in the use of biomedical implant. This project will develop a wireless charging system for a biomedical implant that is compatible with the human body for wireless sensor networks and will build a breadboard prototype of such a system.**

I. INTRODUCTION

RFID(Radio-Frequency Identification) device provides a unique identifier, which can be scanned to retrieve the indentifying information, for an object. It serves the same purpose as a bar code. However, unlike bar code, RFID device does not need to be positioned precisely relative to the scanner. This means that RFID device is not needed to be scanned like bar code or swiped through a special reader like credit cards. RFID devices are widely used in many applications nowadays. One of the most interesting field is biomedical use. This project will focus on RFID devices in the use of biomedical implant. Electronics devices can make remarkable things in human body. Examples like Microchip implant which is an RFID transponder implanted in the body of a human being that can be used to retrieve personal identification, medical history, medications, allergies and contact information from external database. However, these devices need to be made as small as possible in order to easily implanted into human body. In biomedical devices, battery is the largest part of implants. Wireless powering technology can be used to make the battery smaller or to expand its life time. This project aims to investigate the wireless powering of RFID devices through research and experiment in the lab. A prototype of the system needs to be made at the end of the project.

II. PROJECT SPECIFICATION

The title of the project is Wireless Powering of RFID devices. The project is carried out by the Electronic Engineering Department. In the course of this thesis project I will be supervised by Professor Michael Heimlich from Macquarie University. All reporting, consultation and communication regarding the project will be carried out with Professor Michael Heimlich. The supervisors will be responsible for creating required milestones and deliverables for successful completion of the project. Additionally, weekly meeting need to be conducted with Professor Michael Heimlich to keep in track of the completion of tasks.

According to the project description, the project will be based on both research and lab work. A great part of the project time will be spent on researching wireless powering technologies and how to make it compatible within a human body. A reasonable time will be spent on understanding biomedical implants of RFID devices with a greater emphasis on the size, power and compatibility. Analyses of transformer links for coupling power and information through skin is also needed to be made. Most importantly, a prototype of wireless charging system for a biomedical implant that is compatible with the human body for wireless sensor networks will be built on a breadboard.

The thesis project is a 12 credit points unit and it will be completed within a whole semester. Some of key milestones of the project are listed below:

- Research of Wireless technologies that can be used with human body.
- Literatures review on an RFID-Based Closed-Loop Wireless Power Transmission System, Inductive Wireless Power Transfer for RFID and Electromagnetic waves as a potential wireless power source for medical devices(far-field electromagnetic waves and near-field waves).
- Study results from existing research paper
- Study what other solutions for powering a RFID medical implant.
- Design Wireless charging system for biomedical implants.
- Building prototype of the system
- Carry out testing and simulations.
- Conduct the final report using obtained research and results.

The student has to produce periodic presentation to the supervisors regarding the improvements of the thesis project. Midterms report followed by a final term report along with a full presentation about the thesis work will be used as a marking criterion.

III. BACKGROUND LITERATURE REVIEW

A. Wireless powering transmission:

Wireless powering is the transmission of electrical power from a power source to a consuming device without using wires. Wireless power techniques can be divided into two categories, non-radiative(near-field) and radiative(far-field)[1]. Near-field techniques than can be further categorized into four types, Inductive coupling, Resonant inductive coupling, Capacitive coupling and Magnetodynamic coupling.

B. Far-field waves

Far-field waves can travel over long distance. But when they encounter biological tissue, they either reflect off the body harmlessly or get absorbed by the skin as heat. Therefore, far-field waves have been ignored as a potential wireless power source for medical devices.

C. Near-field wireless power technique

In near-field wireless power techniques, power is transferred over shot distances by magnetic fields using inductive coupling between coils of wire or in a few devices by electric fields using capacitive coupling between electrodes. This technique is used widely in many applications such as RFID tags, smartcards and chargers for implantable medical devices.

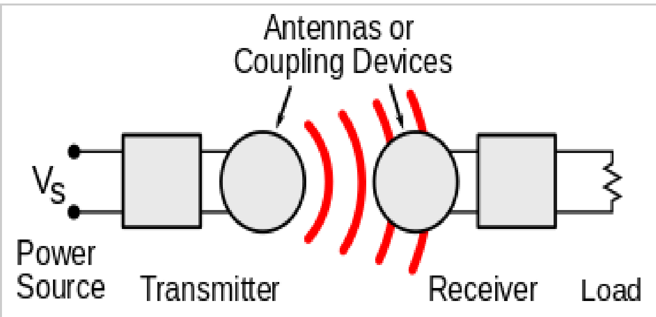

Figure 1: Block diagram of a wireless power system[2]

Technology	Range	Directivity	Frequency	Penetrability	Uses
Inductive coupling	Short	Low	Hz-MHz	Strong	Electric tooth brush
Resonant inductive coupling	Mid	Low	MHz-GHz	Medium	Biomedical implants, RFID
Capacitive coupling	Short	Low	KHz-MHz	Strong	Charging portable devices

Table 1: Wireless power technologies[3]

Near-field has low resonate frequency and short transfer distance and it has higher power transfer efficiency over the far-field. Using longer wavelength, the near-field transfer is easier to generate diffraction when the electromagnetic wave encounters human body. Therefore, we will focus on the near-field transfer technology in this project.

Biomedical applications often require directivity and transfer range. However, the penetrability and power efficiency are extremely important. As a consequence, the inductive coupling and the capacitive coupling seem to be more suitable for biomedical applications because of high efficiency and strong penetrability. Moreover, the inductive coupling delivers energy using alternating magnetic field which causes much less adverse effects on human body compare to the electric field. Therefore, inductive coupling technology suits best for biomedical applications. Passive radio-frequency identification tags and certain implantable devices such as cochlear implants do not include batteries due to cost, size, weight, safety, and lifetime limitations.

D. Inductive Coupling

In inductive coupling, power is transferred between coils of wire by a magnetic field. The transmitter and receiver coils together form a transformer. An alternating current through the transmitter coil creates an oscillating magnetic field by Ampere's law. The magnetic field passes through the receiving coil, where it induces an alternating voltage by Faraday's law of induction(i.e. a basic law of electromagnetism predicting how a magnetic field will interact with an electric circuit to produce an electromagnetic induction), which creates an AC current in the receiver. The induced alternating current may either drive the load directly, or be rectified to DC current by a rectifier in the receiver.

In inductive coupling, the coupling coefficient k between transmitter and receiver coils is the main factor in determining the amount of power that can be delivered to the implantable microelectronic devices. Any changes in k can drastically change the received power. The distance variations between the coils as a result of their relative movements are the main causes of variation in k. Changes in the received power can cause large voltage variations across the receiver coil. Load changes as a result of stimulation, for example, can also cause variation in the receiver coil voltage[4]. such variations are highly undesired in implanted medical devices. Because too little power can cause a malfunction, and extra power can increase heat dissipation within the implant and damage the surrounding tissue.

Possible solution for stabilizing the implantable microelectronic devices received power: Change the transmitted power in a closed loop in a way that the received power stays slightly above the minimum level that keeps the implantable microelectronic devices operational[5].

A recent study developed by Assistant Professor Ada Poon from Stanford University shows that there is a new way of transfer power deep inside the body wirelessly by using electromagnetic waves. The research is carried out on the fact that waves travel differently when they come into contact with different materials such as air, water or biological tissue. By considering this fact, she designed a power source that generated a special type of near-field wave. When this special wave moved from air to skin, it changed its characteristics in a way that enabled it to propagate. It is been called "mid-field wave" by Professor Ada Poon. In her experiment, Poon used her mid-field transfer system to send power directly to tiny medical implants[6]. It is possible to build tiny batteries into microimplants, and then recharge these batteries wirelessly using the mid-field system.

The power transferred to the coil is given by:

$$P_{SC} = \int d^3r \, \mathbf{M}_C(t) \cdot \frac{d\mathbf{B}_S(t)}{dt},$$

Bs is the magnetic field generated by the source and Mc the induced magnetization due to current in the coil.

IV. PROJECT PLAN

While allocating a thesis project to a student, the department assumed that the student is equipped with skills and knowledge gained during the course of last 3 years in engineering degree. Some of the generic skills would include measurement and recording experimental data, writing report in international standard with correct format and referencing, presenting and demonstrating the experimental results.

Projects are allocated to the students on the basis of their corresponding engineering degree in my case its electronics engineering.

The entire thesis project revolves around the concept of research, design and implementation and testing. The thesis project will be conducted in S1 2016. The thesis will be start off with the basic literature review about the important concept of wireless powering of a RFID device especially in biomedical implants. As the project involves lab work, it is also necessary to have a firm idea about the existing wireless powering system for biomedical implants system and how it can be modified or extended to design better systems. A project plan Gantt chart will be provided in this document. At the start of the thesis, project specification and the plan will be agreed between student and supervisor. Any changes regarding the project specification will be noted on the revised plan. After the scope of the project is verified, a student will explore what other people are proposing as potential solution to overcome the problem. The research and initial developed designs will be reflected on the report. The next stage of the project is to go on to further developing the wireless charging system and simulation will be carried out using simulation software such as AWR in the lab. Then a prototype of the proposed wireless charging system needs to be made on breadboard. The project will come to an end with the submission of final report that will contain all the results obtained during the course of this project. This will be followed by presentation where the student will be required to summarize the project to an audience of academic. At the end of the presentation student will required to demonstrate what they did during their thesis to the audience present to give the audience a better understanding of the corresponding project. The project life cycles along with important deadlines are included in the Gantt chart below.

V. CONCLUSIONS

This document has briefly outlined some of the initial research based on my final year thesis. A project plan is provided to complete the thesis as part of my engineering degree. The report gives the reader about the specifications, requirements and the scope of the project. Changes regarding the specifications and scope of the project will be introduced in results of the further meetings with Professor Michael Heimlich. Completing the background literature review is a good starting point for the project and further research will results in a successful project and report.

WBS	Task	Start	End	Cal. Days	% Done	Work Days
1	[Task Category]					
1.1	Basic literature review	Wed 3/23/16	Mon 4/11/16	20	100%	14
1.2	Project specification and plan	Wed 3/23/16	Fri 3/25/16	3	50%	3
1.3	Understanding the scope of the project	Thu 4/14/16	Sat 4/16/16	3	75%	2
1.4	Design	Sat 4/18/15	Sun 5/24/15	37	50%	25
1.5	Improvement	Mon 5/25/15	Fri 5/29/15	5	50%	5
1.6	Testing and simulation	Mon 6/01/15	Wed 6/03/15	3	50%	3
1.7	Thesis report draft	Thu 6/04/15	Sat 6/13/15	10	50%	7
1.8	Thesis report final	Sun 6/14/15	Sat 6/20/15	7	50%	5

Figure 2: Project timeline

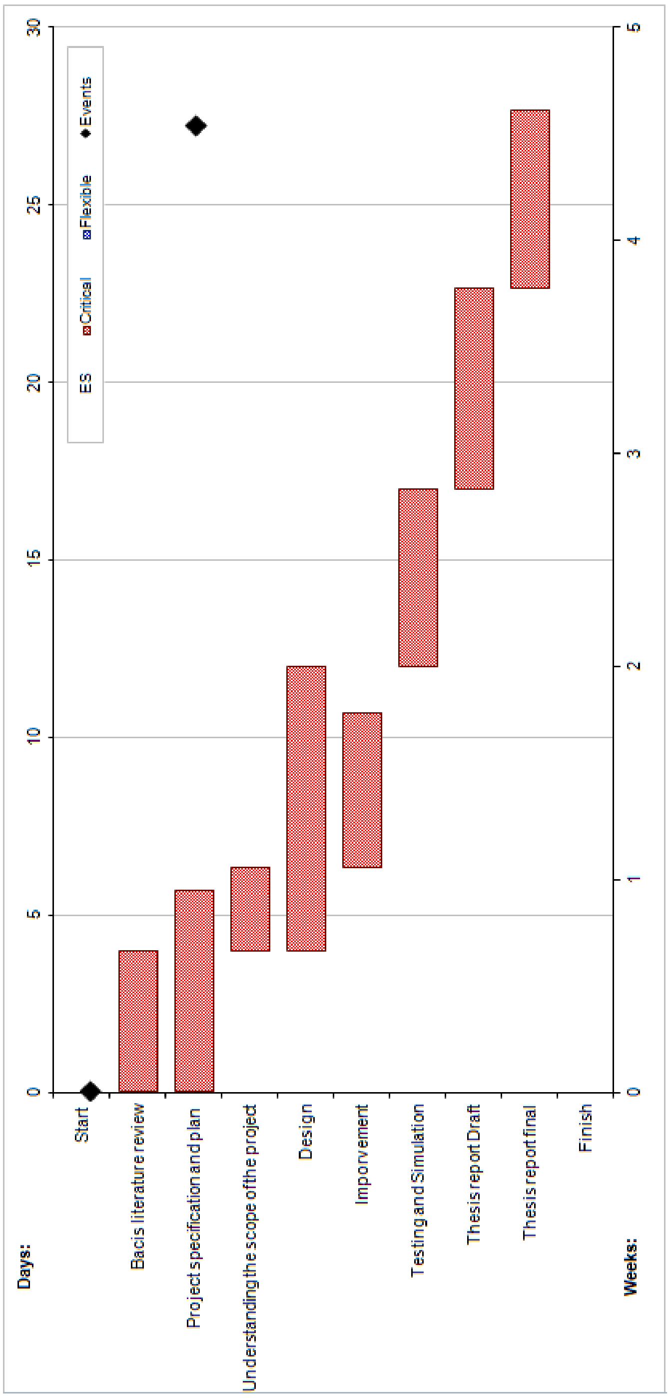

Figure 3: Gantt chart of project plans

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