Implanted Wireless Sensing for Monitoring Joint Replacements

By

Stephen Peter Mehaffey

A thesis submitted to Macquarie University

for the degree of Master of Research

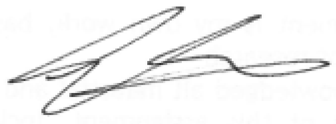
Department of Engineering

June 2016

MACQUARIE
University
SYDNEY · AUSTRALIA

Typeset in $\LaTeX\,2_\varepsilon$.

Except where acknowledged in the customary manner, the material presented in this thesis is, to the best of my knowledge, original and has not been submitted in whole or part for a degree in any university.



Stephen Peter Mehaffey

Acknowledgements

First and foremost, I would like to acknowledge my research supervisor, Professor Michael Heimlich. Without his assistance and guidance, this project would have never been accomplished.

I would also like to show gratitude to my family for supporting me throughout this year, as this was invaluable to my success.

Abstract

Post-operative failures of joint replacement can only be detected by MRI after 50 microns of displaced positioning. Surgeons need to be alerted of failure earlier, while this displacement is as small as 10 microns.

This document details the specifications of an implanted wireless sensor system, which utilises cylindrical wireless, externally powered sensors implanted within orthopaedic knee replacements, to monitor health complications over time. The wireless sensor uses an electromagnetic wave to measure the distance from the sensor to the bottom plate of the knee replacement. This distance is monitored regularly via an external reader device which reports the data and any complications to the clinician. This system allows wireless, non-invasive monitoring of the health of the implant, to eliminate the need for invasive procedures such as surgery. This initial implementation is designed specifically for knee replacements, but can be applied in the future for other joint replacements.

This document will support the development of a prototype for the implanted wireless sensor system, along with further testing and final implementation.

Contents

Acknowledgements **v**

Abstract **vii**

Contents **ix**

List of Figures **xiii**

List of Tables **xv**

1 Introduction **1**

1.1 Project Overview 1

1.2 Report Structure 2

2 Background and Related Work **3**

2.1 Total Knee Arthroplasty 4

2.1.1 The Knee 4

2.1.2 Reasons for Knee Replacement 4

2.1.3 How Knee Replacement Works 5

2.1.4 Implant Complications 6

2.2 Implanted RFID Communication 7

2.3 Radio Frequency Identification 9

2.3.1 RFID Power Sources 10

2.3.2 RFID Operating Frequencies 11

2.4 RFID Sensors 13

2.5 Near Field Communication 14
2.5.1 Inductive Coupling 14
2.6 Far-Field RF 16

3 Proposed System **17**
3.1 Overview 17
3.1.1 Implanted Sensors 18
3.2 System States 19
3.2.1 Calibration State 19
3.2.2 Charging State 19
3.2.3 Measurement State 19
3.3 Functionality 20
3.3.1 Overview 20
3.3.2 Measurement 21
3.4 System Structure 23
3.4.1 Geometric Parameters 24
3.4.2 Link Performance versus the User Build 25
3.4.3 Reader-Tag Alignment 29
3.4.4 Maximum Achievable Read Distance 30
3.5 Operating Frequency 30
3.5.1 Overview 30
3.5.2 Medical Implant Communication Service (MICS) 31
3.5.3 RF Characteristics of the Human Body 32
3.5.4 Carrier Frequencies 33
3.5.5 Sensing Frequency 34
3.6 Power 34
3.6.1 Powering Method 35
3.6.2 Electromagnetic Exposure 35
3.6.3 Analysis 36

4 Future Work **37**

5 Conclusion **39**

A Dielectric Properties of Body Tissues in the Frequency Range 10 Hz - 100 GHz 41

List of Abbreviations 45

References 47

List of Figures

2.1 Normal knee anatomy [1] 5

2.2 Anatomy of knee after total knee arthroplasty (patella not shown) [1] 6

2.3 Overview of an RFID system [2] 9

2.4 RFID capsule tag used to microchip pets [3] 10

2.5 The three classes of RFID tag power configurations [2] 11

2.6 NFC inductive coupling between coils [4] 15

3.1 Implanted wireless sensor system overview [5] 18

3.2 Implanted wireless sensor system states 20

3.3 Displacement measurement using calculated phase difference (assuming there is an issue present) 22

3.4 Functionality of the circulator in the implanted displacement sensor 23

3.5 Geometric parameters of the through-the-body radio frequency link between implanted sensor tag and external reader [6] 24

3.6 Simulated transducer gain for implants in knee-like cylindrical phantoms corresponding to three cases of normal, muscular, and obese builds. Black markers indicate the position of the tag. GT is transducer gain. [6] 26

3.7 Conductivity versus frequency of muscle and fat tissue over a frequency range of 10Hz to 100GHz 27

3.8 Relative permittivity versus frequency of muscle and fat tissue over a frequency range of 10Hz to 100GHz 28

3.9 Measured transducer gain around the phantom with respect to angular alignment. The dark area represents the blind angular region where the reader cannot detect the tag [6]. 29

3.10 Calculated transmission loss through tissue over frequency range of 50 - 700MHz [7]. 33

List of Tables

2.1 Common RFID operating frequencies [2, 8–10] 12

A.1 Dielectric Properties of Muscle Tissue in the Frequency Range 10 Hz - 100 GHz [11] 42

A.2 Dielectric Properties of Fat Tissue in the Frequency Range 10 Hz - 100 GHz [11] 43

1 Introduction

1.1 Project Overview

If a person's knee is severely damaged by arthritis or injury, this can create complications and difficulties when performing even simple day-to-day activities. If non-surgical treatments do not prove effective, total knee arthroplasty (knee replacement surgery) may be considered as a solution [1]. In this procedure, an orthopaedic surgeon removes damaged cartilage and bone from the joint, and positions new metal and plastic implants to restore the alignment and function of the knee [1]. While this process is becoming more effective over time, implant surfaces still wear down over time and the components may loosen [1].

Currently, post-operative failures of joint replacement can only be detected using MRI after 50 microns of displaced positioning. Surgeons need to be alerted of failure earlier, while this displacement is still much smaller, as being able to identify complications early and apply proper treatment could eliminate the need for a second orthopaedic surgery to fix the problem.

The aim of this document is to outline the design specification for a system which can detect knee replacement prosthesis displacements as small as 10 microns. From early analysis, 10 microns is a reasonable target for accuracy, given the technologies to be implemented, and the relative advantage over the accuracy of existing systems, at 50 microns. The proposed system is the implanted wireless sensor system, which utilises cylindrical wireless, externally powered sensors implanted within orthopaedic knee replacements, to monitor health complications over time. The wireless sensor uses an electromagnetic wave to measure the distance from the sensor to the bottom plate of the knee replacement. This distance is monitored regularly via an external reader device which reports the data and any complications to the clinician. This system allows wireless, non-invasive monitoring of the health of the implant, to eliminate the need for invasive procedures such as surgery. This document will support the development of a prototype for the implanted wireless sensor system, along with further testing and final implementation.

1.2 Report Structure

The remainder of this document details the design, system specification and research behind the implanted wireless sensor system. Chapter 2 discusses some background research and existing work related to this project, which was used as guidance and a basis for the design and implementation. Chapter 3 gives an overview of the system specification for the implanted wireless sensor system, including its functionality, structure, and operating parameters. It also discusses some of the research and design decisions that were made, based on the different options that were available for each function. Finally, Chapter 4 provides an overview of some of the possible future work that could be implemented for the implanted wireless sensor system, and Chapter 5 provides a summary of the project and its successes.

2

Background and Related Work

This chapter covers some background research and previous work related to this project, which was used as the basis for my research and implementation. Section 2.1 introduces knee replacement surgery and the issues this project aims to address. Section 2.2 then analyses past experiments into implanted RFID communication within the body, and the associated successes and challenges. Section 2.3 discusses radio frequency identification (RFID) in general, before delving into more detail regarding the power configurations and operating frequencies implemented by RFID technology. Section 2.4 outlines the use of RFID tags as sensors, as this is an important aspect of this sensor system. Finally, Section 2.5 outlines near field communication (NFC), a wireless communication technology linked and related to RFID, and Section 2.6 covers Far-Field RF, another possible radio frequency communication method.

I have selected the literature discussed within for my background research, as it covers a broad selection of material related to this project. Some older, well-known texts have been used as the basis of my research for a historical background and as an overview, while more recent and focused literature has been used to cover specific topics and provide detailed information.

2.1 Total Knee Arthroplasty

If a person's knee is severely damaged by arthritis or injury, this can create complications and difficulties when performing even simple day-to-day activities. If non-surgical treatments do not prove effective, total knee arthroplasty (knee replacement surgery) may be considered as a solution [1]. In this procedure, an orthopaedic surgeon removes damaged cartilage and bone from the joint, and positions new metal and plastic implants to restore the alignment and function of the knee [1].

2.1.1 The Knee

The human knee functions as a type of "biological transmission" whose purpose is to accept and transfer a range of mechanical loads between the femur, patella, tibia, and fibula without causing structural or metabolic damage. Damaged or arthritic knees are like living transmissions with worn bearings that have limited capacity to safely accept and transmit forces [12]. One method of representing the functional capacity of the knee is the "envelope of function", a load and frequency distribution that describes the range of loads the joint can sustain while still maintaining homeostasis of all tissues [12].

The fundamental principle of any orthopaedic treatment is to restore and maintain normal musculoskeletal function. Therefore, the purpose of joint replacement surgery is to maximize the envelope of function for a given joint as effectively as possible [12].

2.1.2 Reasons for Knee Replacement

There are several reasons why knee replacement surgery may be recommended, but the most common reasons for undergoing this procedure are if the patient experiences moderate or severe knee pain which limits everyday activities, chronic inflammation or swelling that does not improve, or knee deformity, which have failed to improve substantially through treatments such as medications, injections, or physical therapy [1].

The most common cause of chronic knee pain and disability is arthritis [13]. Most knee pain is caused by three of the possible types of arthritis:

- Osteoarthritis / Arthrosis – caused by the softening and wearing away of the protective knee cartilage over time, resulting in the bones rubbing against each other and causing pain and stiffness

- Rheumatoid Arthritis – a disease in which the synovial membrane surrounding the joint becomes inflamed and thickened

- Post-traumatic arthritis – follows from serious knee injury, where fractures or tearing of ligaments damages the articular cartilage over time, causing pain and limiting knee function [1]

2.1.3 How Knee Replacement Works

Figure 2.1 provides an overview of the anatomy of a normal knee.

There are four basic steps to a knee replacement procedure:

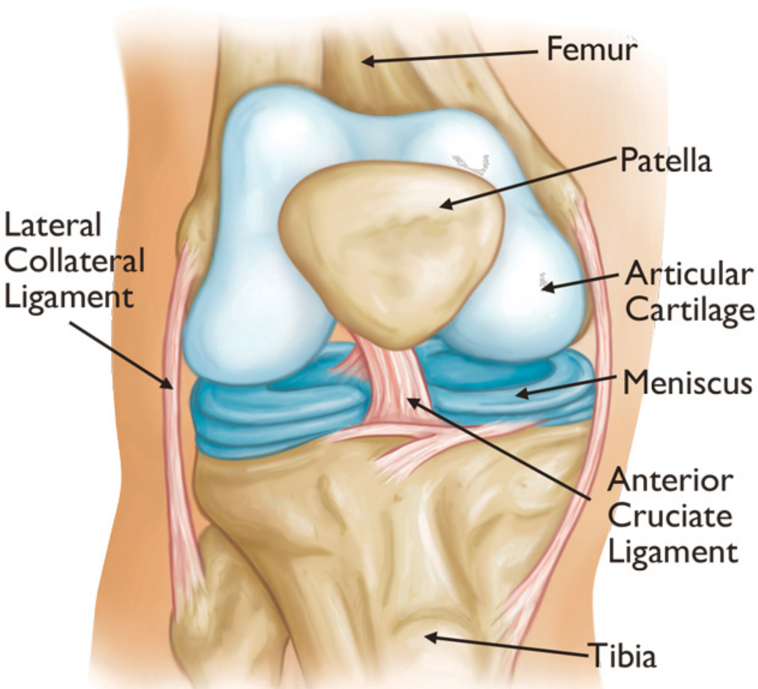


Figure 2.1: Normal knee anatomy [1]

1. Prepare the bone – The damaged cartilage surrounding the ends of the femur and tibia are removed along with a small amount of underlying bone. The distal femur and proximal tibia are cut perpendicular to the mechanical axis of the bones [13].

2. Position the metal implants – The removed cartilage and bone is replaced with metal components to recreate the joint surfaces. These parts may be cemented or fitted into the bone [1, 13].

3. Resurface the patella – If damaged, the under-surface of the kneecap can be cut and resurfaced with a polyethylene button [1, 13].

4. Insert a spacer – A plastic spacer is inserted between the metal components to create a smooth gliding surface [1].

The anatomy of the replaced joint is outlined in Figure 2.2.

The indicators that joint replacement surgery has been successful are the absence of pain

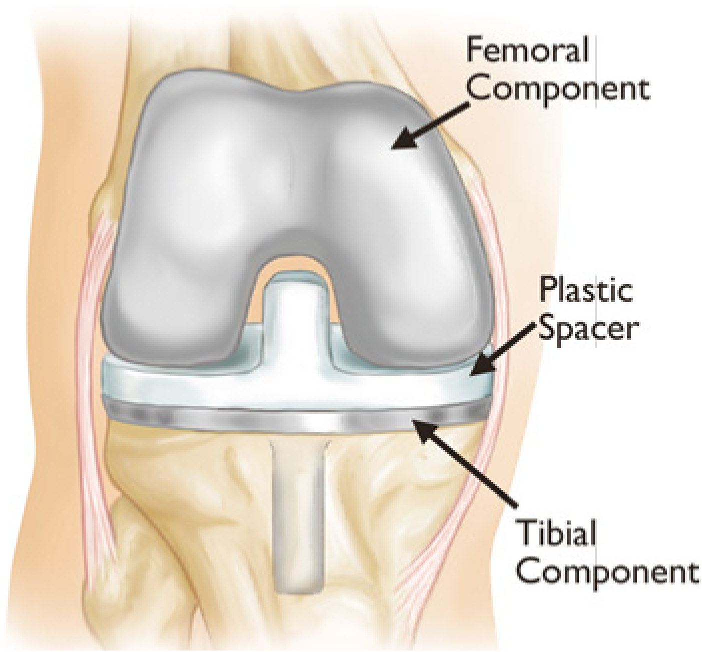


Figure 2.2: Anatomy of knee after total knee arthroplasty (patella not shown) [1]

and swelling, and a large range of motion and muscle control [12].

2.1.4 Implant Complications

Significant advances have occurred in the design of artificial knees, as well as the surgical techniques used to implement them. The type and quality of the metals, polyethylene, and ceramics used in the prosthesis manufacturing process are constantly improving, leading to improved longevity of the joint [13]. However while this process is becoming more effective over time, implant surfaces may still wear down over time and the components may loosen

[1].

Implant wear is the primary mechanical factor limiting the long-term outcome of total knee replacement. The day-to-day motion of the knee is a critical factor influencing wear at the joint [14, 15]. The secondary motions of the knee such as rolling and sliding of the joint can also have a substantial influence on wear, and contribute to fatigue of the polyethylene components [12, 15]. Over time, this motion of the joint can lead to long-term failure modes such as wear and loosening [12].

Currently, post-operative failures of joint replacement can only be detected using MRI after 50 microns of displaced positioning. This system is designed to detect displacements as small as 10 microns.

2.2 Implanted RFID Communication

Various experiments have been carried out, testing the feasibility of implanting tags within the human body, and communicating wirelessly using RFID to obtain information gathered and stored within the body. The key idea is to add communication and sensing capabilities to implanted biomedical devices in order to collect data about the health state of the device. Orthopaedic prostheses, in particular, would greatly benefit from these capabilities because of the many possible pathological complications to be monitored along years, such as tissue regrowth, infections and displacement [6, 16]. They also offer an effective surface to host electronics for sensors and communication. Due to their low profile and size, RFID tags are suitable for integration inside polymeric/metallic parts of orthopaedic prostheses. Being able to identify complications early and apply proper treatment will eliminate the need for a second orthopaedic surgery to fix the problem, and so it is believed that RFID tags with sensor applications will have a huge market value [16].

When creating an implanted wireless radio system, the critical issues to focus on in the design are the feasibility of a reasonable link range [17], low power consumption [18], biocompatibility [19] and miniaturization [20]. The final selection of an appropriate tag structure, power source, carrier frequency and communication bandwidth are dependent on many properties of the human body. In the case of an implanted RFID tag within an orthopaedic implant, some important factors to be considered include the surrounding materials that make up the implant, and how deep within the body the tag would be implanted.

In [6], Lodato, Lopresto, Pinto and Marrocco reported on the proposition of using RFID to develop implanted radio-sensors to be integrated into orthopaedic prostheses. They investigated the feasibility of wireless links for UHF RFID (860-960 MHz) tags implanted into human limbs, which are interrogated by a non-contacting reader's antenna, with the purpose to label and, in the near future, to collect data about the health status of an implanted orthopaedic prosthesis [6]. Performance tests were carried out using electromagnetic simulations over an anthropomorphic phantom as well as experimentation with a real RFID communication link involving a simplified in vitro setup, using minced meat and a bovine bone segment to reproduce a human limb [6]. The required power level, maximum achievable read distance, and link sensitivity to reader-tag alignment were analysed to determine feasibility.

The results of these experiments suggest that using current RFID technology, and the specific tag (loop antenna) and reader antenna (SPIFA) considered, a stable communication link with tags implanted inside limbs may be already feasible up to 10-35 cm from the body, whilst successfully complying with health constraints regarding electromagnetic exposure [6]. Another important result from this experiment is that the addition of an aluminium coating to the bone surrounding the tag, simulating a metallic prosthesis, had a negligible effect on signal power, proving it is still possible to establish a stable RFID link in close proximity to metallic surfaces [6].

In [16], Liu, Stachel, Sejdic, Mickle and Berger aimed to solve the issue of implanted RFID tag communication interfering with pacemakers and other sensitive medical equipment, by communicating with an implanted tag via transcutaneous near field communication (TNFC) based on capacitive coupling between the reader and the tag. A maximum reading range of 4.1 cm was achieved through pork skin using 30 dBm power from the reader. This proposed UHF RFID system is a feasible solution to provide high efficiency for transcutaneous operation while can eliminating any issues of interference with other medical or RFID devices [16]. The main issue with this proposed system is that it requires its reader and tag to be in direct contact with the tissue to work properly and respond to each other.

Experiments into implanted wireless sensor networks, such as those carried out in [6] and [16], are what we intend to use as a foundation for the design and production of this system. We will use the information and results from studies such as these to assist with design decisions when developing a specification for our system.

2.3 Radio Frequency Identification

Radio frequency identification (RFID) involves small transponders (tags) attached to physical objects, which are wirelessly interrogated by RFID transceivers (readers), to retrieve identifying information [8]. Typically, RFID systems have been used as automatic identification systems, for applications such as electronic toll collection, proximity cards for access control, and contactless payment systems [8, 9]. While initially developed in the 1970s, recent advancements in chip technology have allowed the expansion and use of RFID technology in countless new applications in many industries.
RFID systems involve three core components:

- Tags (transponders), which carry data
- Readers (transceivers), which read and write tag data
- Host systems / Databases, to provide information or operational function [9]

Figure 2.3 outlines a typical RFID system.

RFID tags are typically composed of an antenna and an integrated circuit (IC). The

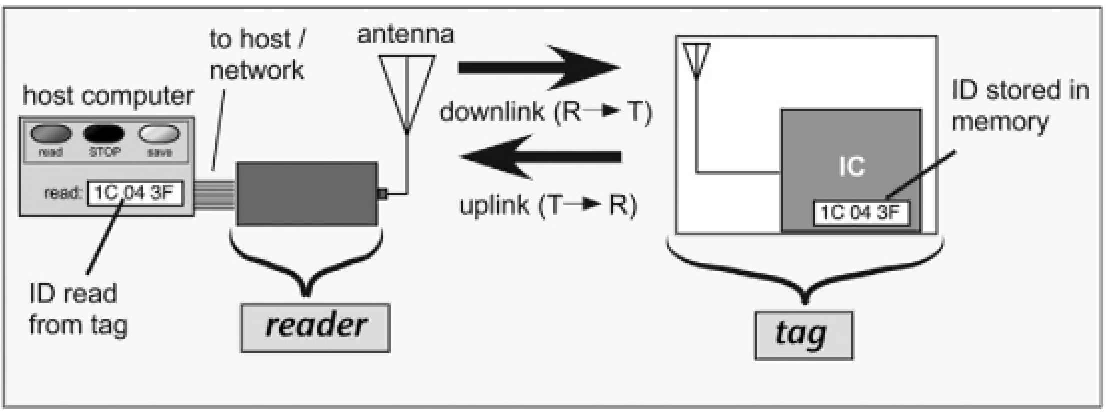


Figure 2.3: Overview of an RFID system [2]

antenna is responsible for receiving and transmitting signals, while the IC is responsible for computation, storage and collecting power from an incident reader signal [8]. Once manufactured, the antenna and IC are commonly packaged into a glass capsule or foil inlay, to be incorporated into the final product [8, 21]. Figure 2.4 shows an example of a capsule design RFID tag, used to microchip pets.

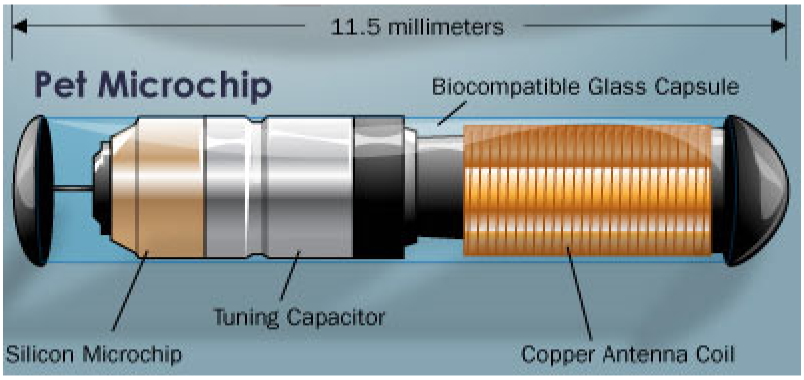


Figure 2.4: RFID capsule tag used to microchip pets [3]

RFID readers communicate with tags through a radio frequency (RF) channel to obtain identifying information. The reader transmits radio signals at a pre-set frequency and interval to interrogate the tag. Any tag within range receives this signal via its antenna, and responds with a signal containing its identification and other information [21].
Readers come in many forms and may offer a wide range of functionality. They may also be integrated into hand-held mobile devices [8].

2.3.1 RFID Power Sources

RFID tags can obtain their power in many different ways. The method they use for this will determine the tag's potential read range, lifetime, cost, and possible functionalities, along with the physical design of the tag [8].
There are three main classes of tag power sources; active, semi-passive, and passive [9]. These classes are summarised in Figure 2.5.
Active tags are equipped with an on-board power source such as a battery, and may initiate communication with a reader or periodically transmit signals. Having their own power source means these tags typically have a much longer operating range than passive tags [8, 21]. However an in-built battery results in a bulkier tag structure, and a requirement for recharging or replacing the battery, which are inconvenient for implanted devices, where miniaturisation and a long lifetime are paramount.
By contrast, a semi-passive (or semi-active) tag is one which has an internal battery, but is unable to initiate communications with a reader. This ensures that semi-passive tags are only active when queried by a reader [2, 8]. Semi-passive tags have the advantages of a long read distance [2] and the ability to remain idle most of the time to save power, but are more

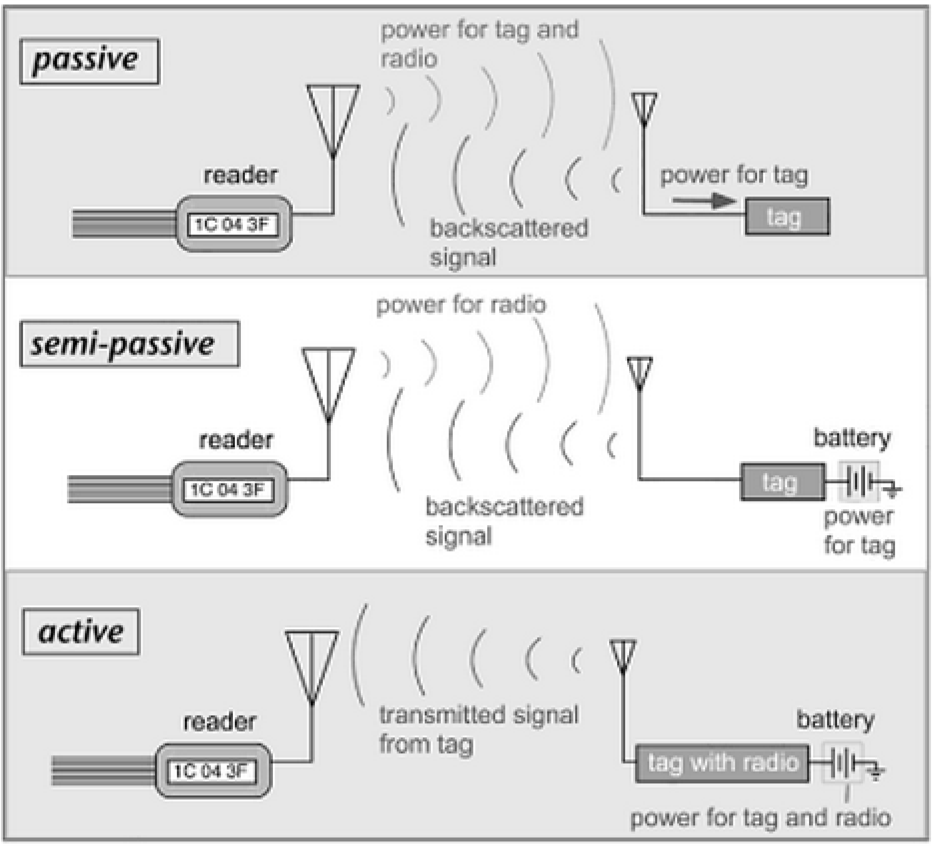


Figure 2.5: The three classes of RFID tag power configurations [2]

expensive to produce [8].

Passive tags have no on-board power source, and rely solely on a reader to operate. They obtain energy by harvesting it from an incoming RF communication signal from a reader [8]. At lower frequencies, this energy is typically harvested inductively, while at higher frequencies it is harvested through capacitance. The read range of an inductive tag is roughly comparable to the size of the reader antenna, and is dependent on the relative orientation of the tag and the reader [2]. While they have the shortest read range, passive tags are the cheapest to manufacture and the easiest to integrate into products, and are therefore the most commonly used RFID tags [8]. Due to their simplicity, passive tags may also rely on a reader to perform computation as well.

2.3.2 RFID Operating Frequencies

RFID systems can operate at a variety of different radio frequencies. These are classified into ranges, each with their own operating range and power requirements. Some ranges are

Frequency Range	Frequencies	Data Rate	Passive Read Distance
Low Frequency (LF)	120-140 KHz	Low	10-30cm
High Frequency (HF)	13.56 MHz	Low to Moderate	10cm-1m
Ultra-High Frequency (UHF)	860-930 MHz	Moderate	3m
Microwave	2.45 and 5.8 GHz	High	3-5m
Ultra-Wide Band (UWB)	3.1-10.6 GHz	High	10m

Table 2.1: Common RFID operating frequencies [2, 8–10]

also subject to regulations and restrictions that limit the applications that they can be used for [2, 8]. These available frequency ranges are outlined in Table 2.1.

The selected operating frequency determines which physical materials will effectively propagate the radio frequency signals. It can also have an effect on the physical dimensions of an RFID tag, as different sized and shaped antennae will operate at different frequencies [8]. Selection of a carrier frequency for an implanted RFID tag within an orthopaedic implant is dependent on many properties of the human body and the implant, such as the surrounding materials that make up the implant, and how deep within the body the tag would be implanted. The human body is not an ideal medium for radio frequency wave transmission. It consists of materials of different dielectric constants, thickness, and impedance. Therefore depending on the radio frequency, body tissue can lead to high losses caused by power absorption, central frequency shift, and radiation pattern destruction, varying with frequency and tissue characteristics [22].

Low frequency (LF) RFID tags are normally passively powered through induction, and therefore have short read ranges. They are able to operate in close proximity to metals and liquids, but have a low data rate compared to other operating frequencies [8].

High frequency (HF) RFID tags offer a higher data rate than LF tags, but are still limited to a short read range. They also do not perform as well as LF tags in proximity to metals or liquids [8]. The HF frequency range lies on a heavily regulated part of the radio spectrum, in a worldwide industrial, scientific and medical (ISM) radio band. Broadcasts must operate within a narrow frequency band, which presents an issue due to interference with other signals [8]. This has been considered as a possible issue for medical applications, due to concerns regarding interference with sensitive medical devices or other wireless implants [16, 23]. The availability of high power at short range means HF tags can support large amounts of memory, allowing users to record and retrieve substantial amounts of information from a tag [2].

Ultra high frequency (UHF) RFID tags offer a longer read range than lower frequency tags. When considering implanted RFIDs, tags in the UHF band may offer some attractive advantages over HF tags because of the higher data-rate and longer activation distances [6]. As attenuation through tissue increases with frequency [24, 25], UHF systems experience higher power losses than HF and LF systems. However, UHF RFID systems have the advantages of less signal degradation due to metallic implants than HF and LF systems, and the tag can be made smaller in size because of the much shorter wavelength [16]. Furthermore, UHF systems follow standards with more industry support for medical devices such as MRI compatible RFID integrated circuits [16].

Microwave RFID tags are typically semi-passive or active, and offer higher read ranges than UHF tags. However, microwave tags consume more energy than comparable UHF tags, and generally cost more to produce [8].

Ultra-wideband (UWB) RFID communication uses low-power signals on a very broad range of frequencies, rather than sending a strong signal on a particular frequency [8]. UWB has a long read range, relatively low power consumption, and is compatible with metals and liquids. Since the signal on any particular frequency is very weak, UWB also does not interfere with sensitive equipment [8]. Compared to other RFID frequency bands, UWB technology is highly attractive for implantable devices due to its ability to transmit less energy per Hz within a wide spectrum, thereby moderating radiative tissue absorption [25].

2.4 RFID Sensors

One novel idea when designing an RFID system to monitor the health state of an orthopaedic implant is the possibility of passive sensing, using the RFID tag itself as a sensor. By processing the response signal of an implanted tag, RFID technology has been recently recognized as potentially capable to produce additional information about the hosting object or nearby environment, such as its physical state and time-evolution, without any specific embedded sensor or local power supply [26, 27].

In [26], this idea is examined with the purpose of using implanted RFID tags to sense changes in some human physiological and pathological process, by investigating local variations of effective permittivity inside the body. This theory relies on the relationship between the tag's input impedance and its surrounding environment. Using periodic data acquisition, changes in the geometrical or chemical features of tissues could be detected, thus providing an indirect way to monitor the health state of the orthopaedic implant and possible complications [26]. The big challenge here is controlling the sensitivity of the sensor tag with regard to biological processes, to correlate with the process under observation. This is further complicated by the high permittivity of human tissue. Early experimentation suggests this method of passive tag sensing is feasible, by observing the small-scale effects on the electromagnetic response of the tag-sensor produced by biological processes [26].

2.5 Near Field Communication

Near field communication (NFC) is a technology for contactless short-range communication between NFC enabled devices [28]. Based on RFID, NFC uses inductive coupling between the NFC initiator and target to enable communications, making it possible to support passive operation, where the target tag does not have its own power supply [4, 28]. NFC's specifications are outlined in ISO/IEC 14443 and IOS/IEC 18000-3 [4, 29]. ISO/IEC 18000-3 is a standard for NFC devices communicating wirelessly at 13.56MHz, the same frequency as HF RFID readers and tags [4, 30]. NFC technology uses the same working standards and is compatible with RFID infrastructure [28].

2.5.1 Inductive Coupling

Passive operation in NFC is accomplished by using inductive coupling to transfer power from the NFC initiator to the tag over the air [4].

When current moves in a wire, a magnetic field is created around the wire, with strength proportional to the amount of current. When this wire is bent into a coil, the magnetic field grows stronger and increases in proportion to the number of coil turns [4]. If a second coil is placed in the magnetic field, a current is induced in it [31].

During passive NFC operation, the initiator (which can be an NFC capable phone) creates magnetic fields, which reach the passive tag and induce a current in the tag's coil, which is then rectified by a diode, and powers the circuit. The target tag then transfers data back to the initiator using the magnetic field channel [28, 32, 33]. To modulate data from the tag to the reader, the tag circuitry changes the load on its coil (load modulation) and this is detected by the reader as a result of the mutual coupling [33].

Figure 2.6 represents the inductive coupling between two coils. The magnetic field is created by a Read/Write antenna, and powers the Data Carrier antenna.

The current induced in the passive NFC tag is proportional to the mutual inductance between

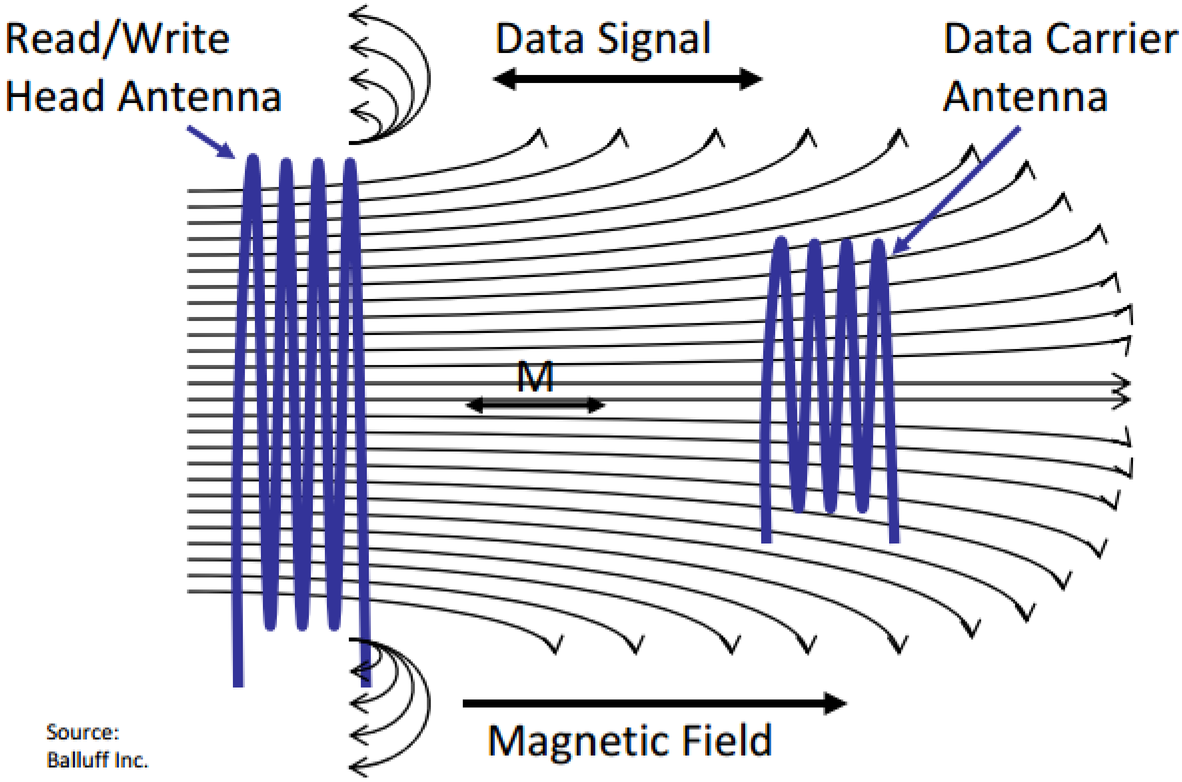


Figure 2.6: NFC inductive coupling between coils [4]

the coils and the strength of the magnetic field [33–35]. Based on the mutual inductance of the coils, it is possible to calculate the voltage at the passive tag. This information is useful for determining the maximum read distance between the tag and reader [35].

Inductive coupling is a near field effect, so for the NFC tag to work, the distance between the coils must be less than the wavelength of the magnetic field divided by two times pi [33, 36]. Many smartphones are embedded with NFC technology, with this number increasing in recent years [32]. Smartphones are able to transfer power to passive NFC tags, with the amount of power transferred depending on the distance and angle of alignment between the NFC phone and tag, and the environmental conditions in which the NFC is operating [4]. One issue, highlighted in a study by Vena and Roux, is that as RFID technology advances and tags are manufactured smaller, smaller antennas pose challenges related to maintaining acceptable reading range and interoperability [36]. This leads to a requirement for more efficient antenna design.

2.6 Far-Field RF

Inductive coupling, which is used for wireless communication in technologies such as NFC, is effective at powering and communicating with low-power passive tags, however it is very sensitive to alignment of tags and readers, as well as depth [33, 36, 37]. As outlined earlier in [16], a UHF RFID system using NFC was only able to achieve a maximum read range of 4.1cm, when the reader was in direct contact with the tissue. For this reason, inductive coupling is not a good candidate for powering or communicating with multiple devices implanted in the body. The development of far-field power and communication is essential to the development of implantable RFID sensor networks [37].

3

Proposed System

This chapter details the specifications and the decision making behind the proposed design for the implanted wireless sensor system. Section 3.1 gives a brief overview of the system and the sensors used. Section 3.2 outlines the different system states involved in the use of this system, and how each performs. Section 3.3 details the actual functionality of the system, and how the displacement measurements are made. Section 3.4 discusses the physical structure of the system, and how physical parameters such as angles and distances affect operation. Section 3.5 discusses the factors considered when selecting the different operating frequencies for each function of the sensor system. Finally, Section 3.6 discusses the power transfer method implemented by this system, how this method was chosen, and the results of its use.

3.1 Overview

As outlined in Chapter 2, there are many possible options for frequency, power transmission, and circuit design when producing an implanted wireless sensor network. The selection of techniques is dependent on each particular case and the environment that the system will be operating within.

The main focus of this system specification is to utilise existing literature and basic radio frequency theory to determine how little energy the sensor circuitry can run on, and what is the best way to deliver this energy to and from the wireless sensor. This is expressed in Figure 3.1.

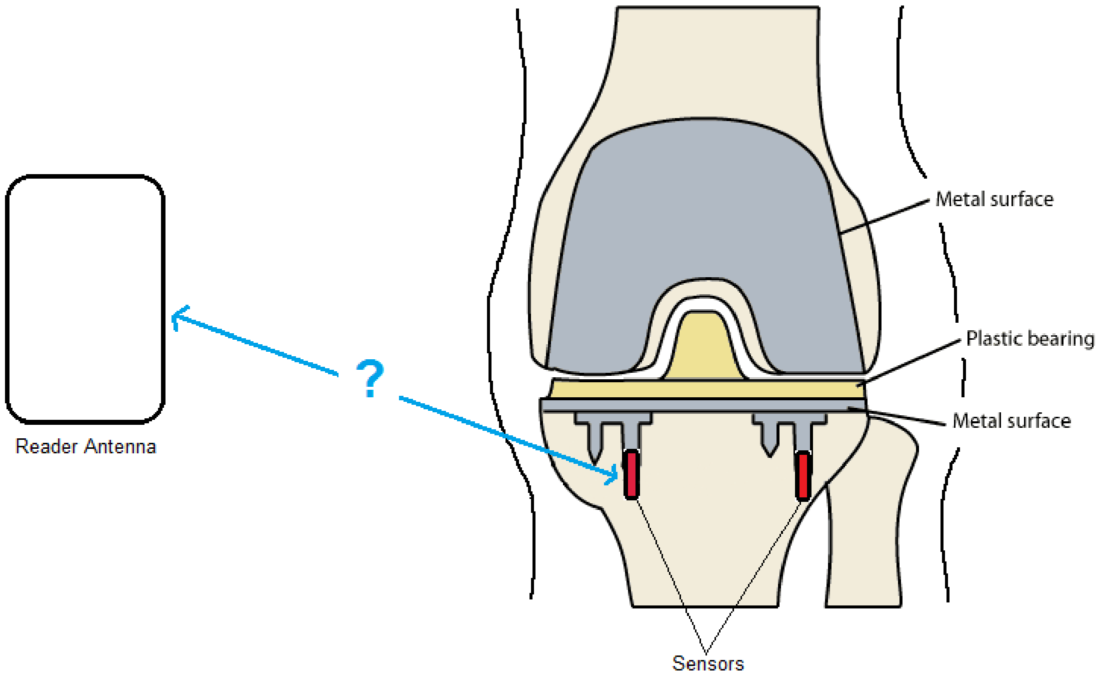


Figure 3.1: Implanted wireless sensor system overview [5]

3.1.1 Implanted Sensors

The sensors to be used in the wireless implanted sensor system are integrated circuits encased within a 3mm by 10mm cylindrical capsule. One sensor is inserted into each of the three guide holes used to screw the knee prosthesis to the existing bone. As discussed in Section 2.3, the sensors are passive circuits with no local power supply, so they are powered wirelessly by the external reader device. Each sensor measures the distance between itself and the metal plate of the knee prosthesis, to detect displacement. Together, the three sensors define a plane, so the angle of displacement of the metal plate can also be calculated using further computation. The functionality of the sensors to detect prosthesis displacement is discussed in more detail in Section 3.3.

3.2 System States

When in operation, the wireless sensor system cycles through multiple system states which each serve a different purpose for functionality. The three states that are involved are the calibration state, the charging state, and the measurement state.

3.2.1 Calibration State

- Occurs once, after tags are implanted into the knee.

- Involves calibrating each of the sensor tags so that they are all equally sensitive to movements in the knee. This is important for calculating displacements as small as 10 microns. Calibration for each sensor will be saved to the reader for future measurements.

3.2.2 Charging State

- Occurs each time measurement is required.

- Involves wirelessly transferring power from the reader to the implanted tags to activate them, before measurement.

3.2.3 Measurement State

- Occurs each time measurement is required.

- Involves transmitting signals from the sensor tags to the reader device to provide information regarding the knee implant.

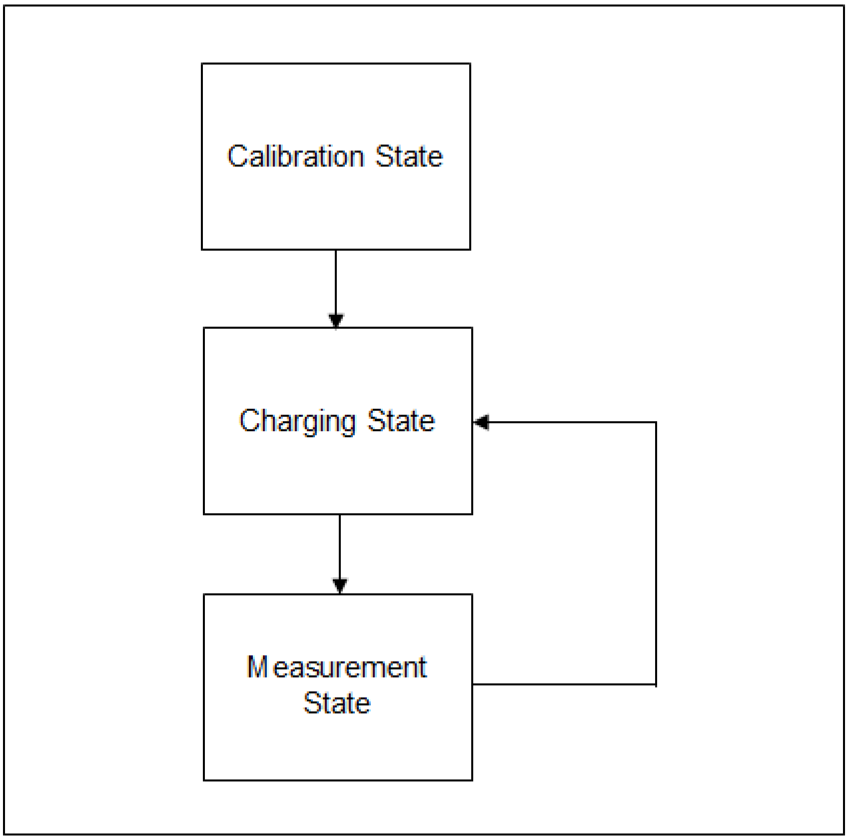


Figure 3.2: Implanted wireless sensor system states

3.3 Functionality

3.3.1 Overview

The wireless implanted sensor system uses an electromagnetic wave to measure the distance from the sensor to the bottom plate of the knee replacement. This distance is monitored regularly via an external reader device which reports the data and any complications to the clinician. The sensors used are displacement sensors, and they are designed to make distance calculations. There are two possible methods to detect whether there are any issues with the joint replacement:

- Measuring the distance between the sensor and the metal plate once, statically, on numerous occasions over a period of time. This involves calculating any relative change in distance between current and previous measurements.
- Measuring the distance between the sensor and the metal plate continuously over a

short interval of time, while moving the knee joint. This tests whether there is any relative movement between the knee and the prosthesis while using the knee, indicating a loosening of the plate.

Because we are testing for such small displacements (approximately 10 microns), we are forced to proceed with the second method of issue detection. If we were to use the first option, small changes in the sensor's environment such as water content changes in the knee would produce slightly different responses, so measuring relative displacement over multiple days or weeks would be impossible. This means the system must function by detecting relative movement between the knee joint and the metal prosthesis plate during the measurement.

3.3.2 Measurement

The actual measurement of distance from the implanted sensor to the prosthesis metal plate is calculated using signal phase difference. This involves the sensor using a built-in antenna to broadcast a directional radio wave at a very specific wavelength towards the metal plate. The signal will reflect off of the plate and return to the sensor antenna for computation.
During initial calibration of the sensors (after joint replacement), we will calculate the expected phase of the signal once it is reflected and returns to the sensor chip.
During measurement, the patient will swing their knee back and forwards while the sensor is constantly transmitting the signal and receiving the reflected signal. If there is any relative movement between the sensor and the metal plate when the knee is moved, the distance that the signal travels will change, and this will result in a reflected signal with a different phase to that calculated during the initial calibration. This is summarised in Figure 3.3. This is easier to calculate if the wavelength used for transmission is similar to the expected distance between sensor and metal plate. For example, if the distance is normally half a wavelength, any relative movement will result in a reflected signal with a different phase to the original transmitted wave.

A circulator will be used to separate the original transmitted signal and the reflected signal for comparison. The input to port 1 of the circulator is our original input signal. This signal outputs to port 2, which is reflected off the metal plate and returns back into port 2. This reflected signal outputs to port 3. So now port 1 is the original input, and port 3 outputs the returned signal. This is summarised in Figure 3.4. If there is relative movement between the

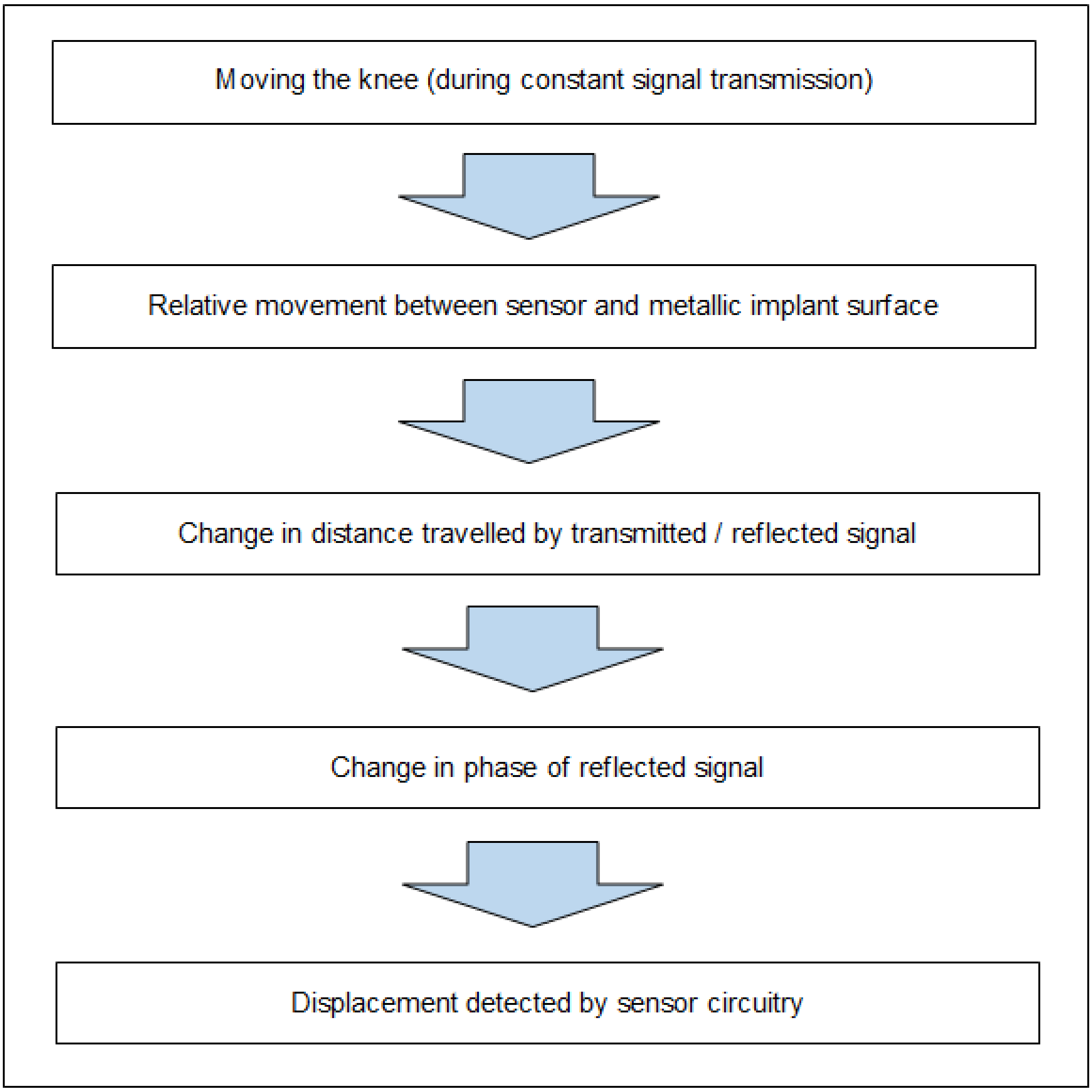


Figure 3.3: Displacement measurement using calculated phase difference (assuming there is an issue present)

sensor and the plate, this returned signal will have a different phase to the original input. The original transmitted signal and reflected signal are both fed into an IQ modulator, which produces a DC value as its output. If there is no relative movement, then transmitted signal Tx should be equal to reflected signal Rx, and the output DC should be zero. However, if there is a phase difference between these signals, then there should be a resultant non-zero DC value. For example, if the selected transmission signal has a wavelength of 1mm, then a 10 micron displacement during measurement would result in a phase difference of 0.01 wavelengths in the reflected signal, which equals a 3.6 degree phase difference.

The complications with this measurement method come from the resolution of the components used, and how small of a displacement can be detected by the IQ modulator. The resultant phase difference in the reflected signal from a 10 micron movement depends on the selected frequency for transmission by the sensor. Therefore the selected transmission frequency and required component resolution are very closely linked. This is discussed more in Section 3.5.

Displacement measurements can be taken either daily or weekly. The basis of this system

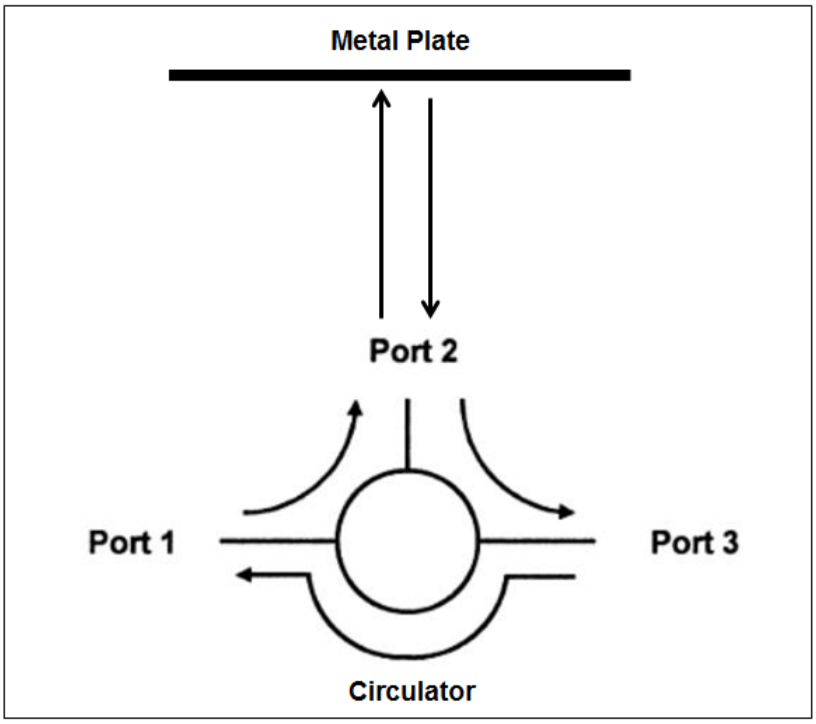


Figure 3.4: Functionality of the circulator in the implanted displacement sensor

design is to detect knee replacement complications very early on, before they begin to cause problems such as joint movement restriction. Also, each measurement is discrete and does not depend on the history of previous measurements or the interval between each. Therefore, measurement frequency is not a vital factor for the operation of this system.

3.4 System Structure

As outlined earlier, the implanted wireless sensor network is composed of two main components; an implanted radio frequency displacement sensor, and an external reader antenna.

These two components communicate wirelessly, with the effectiveness of communication largely reliant on the physical parameters of the system and the surrounding environment of the knee. The physical arrangement of the sensor and reader, as well as the build of the patient, both largely affect system efficiency, and factors such as transmitted power and therefore maximum achievable read distance.

3.4.1 Geometric Parameters

With reference to Figure 3.5, the features of this radio frequency channel involving implanted sensors depends on the particular configuration, which is a function of the body region where the sensor is implanted, the distance between the reader's antenna and the body surface (d), and the depth of the implant from the body surface (h).

Experimental results have shown significant differences in the reliable available power levels

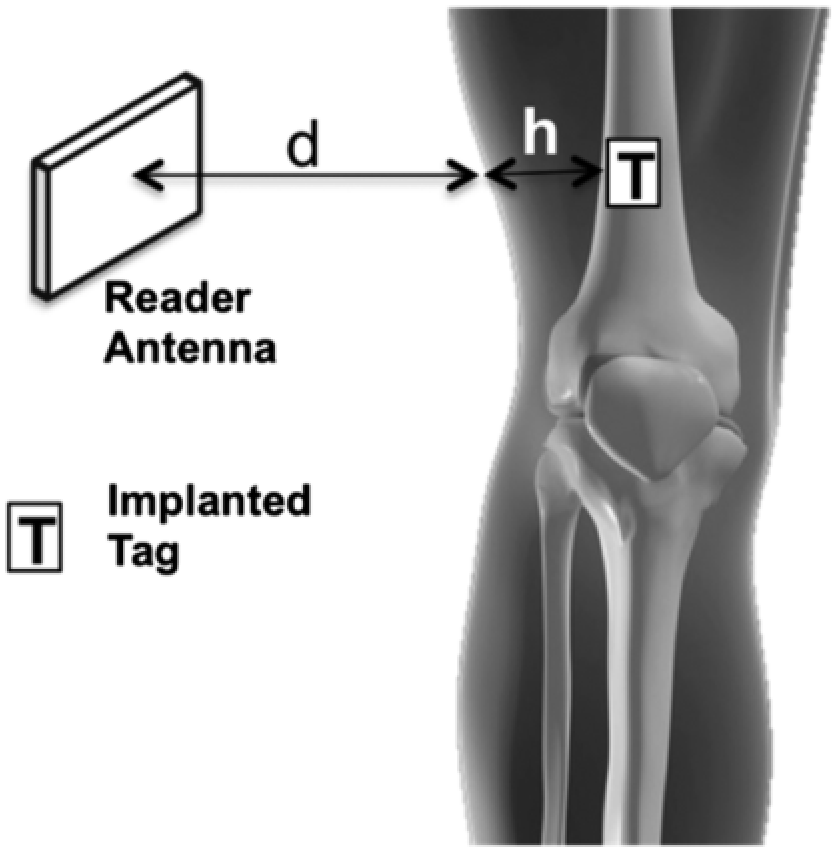


Figure 3.5: Geometric parameters of the through-the-body radio frequency link between implanted sensor tag and external reader [6]

when implants are located at differing tissue depths. For example, in [6], identical sensor arrangements implanted at depths of 70 and 27 millimetres in the knee and elbow respectively, resulted in a difference of 8dB. This corresponds to the tissue layers and overall power loss of the body tissues between sensor tag and skin.

As the implanted sensor system is initially intended for use in knee replacements only, we can simplify this function for consideration within this paper. The body region considered will only be the knee joint, and an average depth of h = 70mm for the considered implant will be used, as obtained from [6]. This means the only largely variable factors are the biological make-up of the particular knee joint, and the placement of the external reader (distance and angle).

3.4.2 Link Performance versus the User Build

Another factor to consider is the effect of user build (percentage of muscular and fat mass) on link performance. This variation can be due to normal physiological build as well as pathological conditions such as obesity. In [6], computations were performed using a simplified cylinder model of the knee (with implanted tag), using different thicknesses of fat and muscle layers. The corresponding electromagnetic parameters used in these experiments are derived from [38].

The results concluded that in the three cases investigated (normal, muscular and obese build), and at a constant distance d = 15cm from the skin, the difference in transducer gain within the implanted tag between the worst case (muscular/obese) and the normal build is about 1 dB, due to the different absolute depth of implant and to the different percentage of surrounding muscle/fat [6]. These results are summarised in Figure 3.6.

Overall, the absolute depth of the implant due to a muscular or obese build had a much greater effect on transducer gain than the percentage of muscle or fat present in the limb. Another interesting result is that at lower frequencies, the high fat percentage in the obese case resulted in a larger gain than the muscular case, while at higher frequencies, the muscular build performed better. This is due to the differing electromagnetic properties of different human tissue types, and how they vary with frequency. Appendix A contains the results of experiments into the calculation of dielectric properties of body tissues for the design of implanted electrical systems. The conductivity and relative permittivity of both muscle and

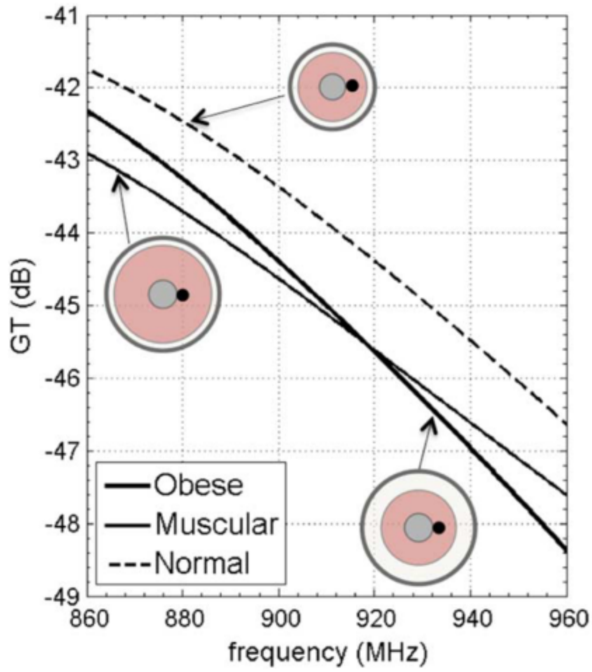


Figure 3.6: Simulated transducer gain for implants in knee-like cylindrical phantoms corresponding to three cases of normal, muscular, and obese builds. Black markers indicate the position of the tag. GT is transducer gain. [6]

fat tissues are calculated over a frequency range of 10Hz to 100GHz. The results contained within this appendix are summarised in Figures 3.7 and 3.8.

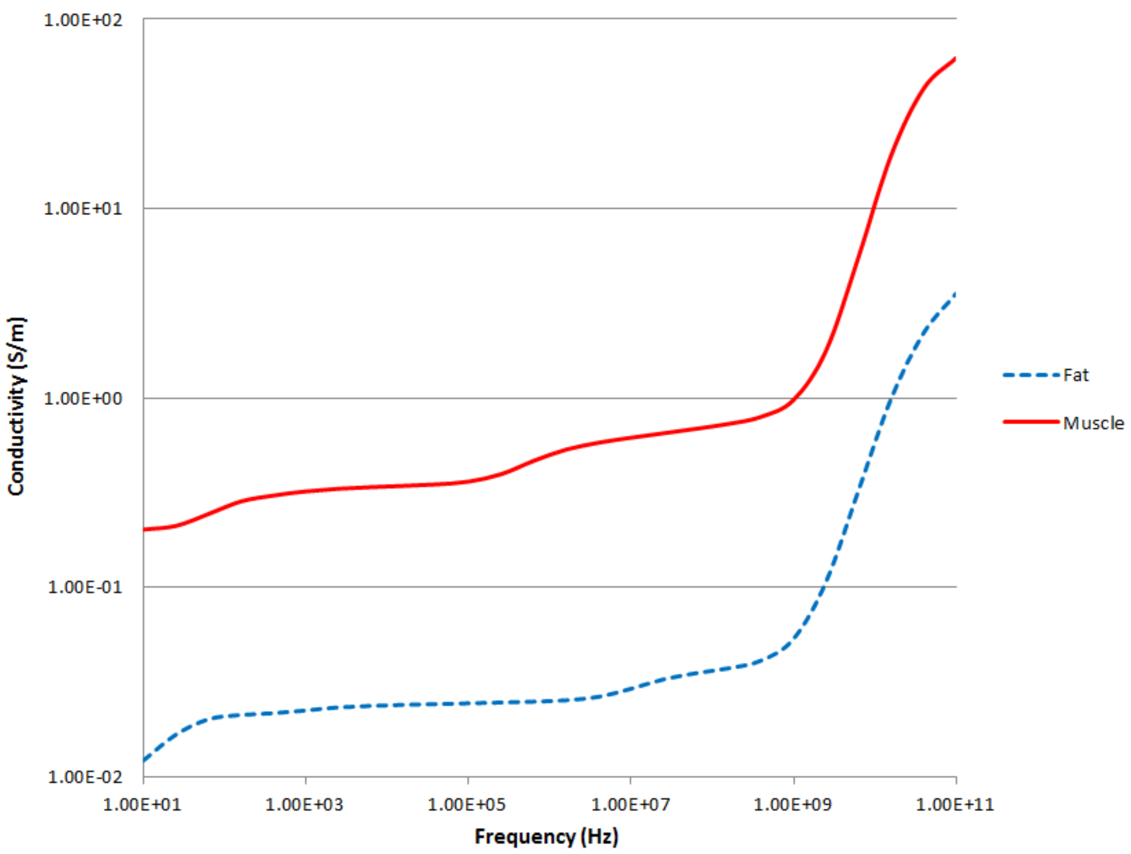


Figure 3.7: Conductivity versus frequency of muscle and fat tissue over a frequency range of 10Hz to 100GHz

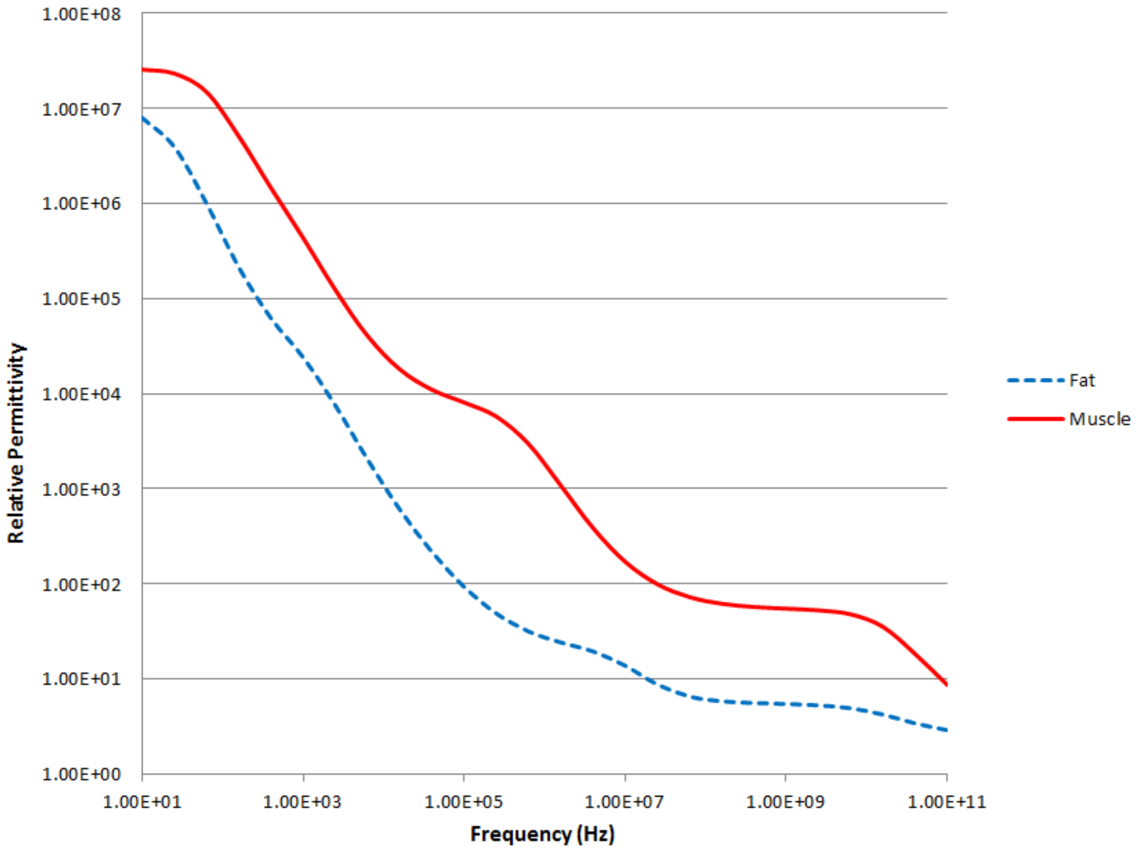


Figure 3.8: Relative permittivity versus frequency of muscle and fat tissue over a frequency range of 10Hz to 100GHz

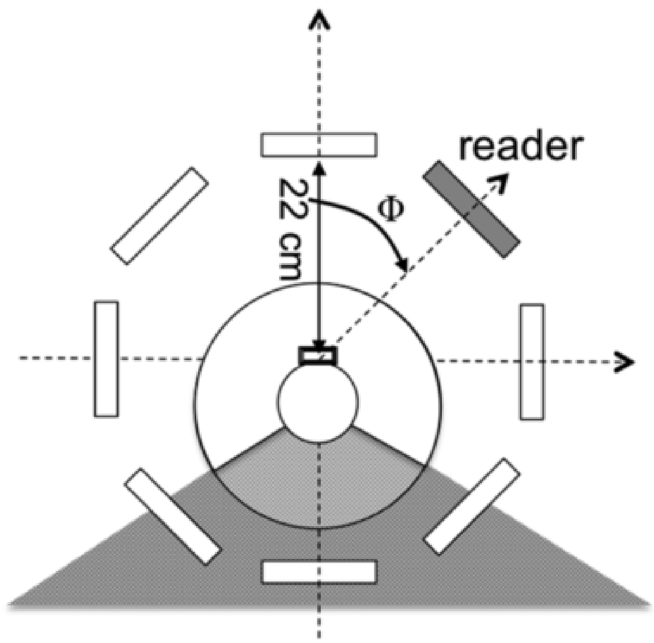
As can be seen from Appendix A and Figures 3.7 and 3.8, while fat tissue has an overall lower conductivity value than muscle for all frequency values, it also has a lower relative permittivity across this frequency range. However, muscle tissue's conductivity increases much more rapidly as frequency increases, resulting in the better performance in a muscular build when communicating at higher frequencies.

3.4.3 Reader-Tag Alignment

Along with geometric parameters such as implant depth, and user build, a third factor which affects the power transfer between the reader and tag is the angular alignment between the reader and tag.

In [6], the effect of reader-tag alignment on transducer gain is tested using a meat-bone phantom with bone diameter equal to 4cm, implant depth of h = 40mm, and reader at a fixed distance of 22cm from the tag. The reader was rotated at this fixed distance around the phantom, and the transducer gain was monitored over 360 degrees. The results of this experiment (displayed in Figure 3.9) found that angular offset had only a small effect on transducer gain, except when the reader is placed in an opposite angular position with respect to the implant [6]. Due to the thicker layers of bone and tissue to transmit through, larger power attenuation contributed significant losses, and the tag was unable to activate at these angles.

These findings have recommended that the exact alignment between the implanted tag and the reader device is not expected to be a significant issue, provided that the reader-tag misalignment does not exceed ±130 degrees [6].



$\Phi(°)$	G_T(dB)	P_{to} (dBm)
0	-44.5	27
45	-43.5	26
90	-44.5	26
135	-46.5	29
180	≤-47.5	≥30
225	≤-47.5	≥30
270	-44.5	26
315	-44.5	26

Figure 3.9: Measured transducer gain around the phantom with respect to angular alignment. The dark area represents the blind angular region where the reader cannot detect the tag [6].

3.4.4 Maximum Achievable Read Distance

When transmitted from one antenna to another over free space, it is quite easy to estimate the power received from one antenna, when transmitted from another antenna. This is estimated using the Friis Transmission Equation (see Equation 3.1).

$$\frac{P_r}{P_t} = G_t G_r \left(\frac{\lambda}{4\pi * R}\right)^2 \tag{3.1}$$

In this equation, the ratio of power available at the receiving antenna, P_r, to output power of the transmitting antenna, P_t, can be found using the respective antenna gains, G_t and G_r, the wavelength of transmission, λ, and R, the distance between the antennas [39]. Unfortunately, this model is not as accurate at predicting transmitted power once the space between the antennas is obstructed, especially in this case, where the receiver antenna is implanted within the human body.

In a similar wirelessly powered system in [6], a meat-bone phantom with bone diameter equal to 4cm and an implanted depth of h = 40mm, the reader antenna was gradually moved away from the model, and the attenuation with distance variance was monitored. The resulting gain-distance dependence exhibited a 0.3 dB/cm attenuation with a 1/d profile, as found by linear regression of measurement data. The achieved maximum read distance of the tag was d = 35cm, after which the implanted circuit stopped responding [6]. Measured and simulated data from this experiment was compared with the values predicted by the Friis Transmission Equation. As predicted, the formula clearly overestimated the gain, especially when the reader is placed in the close proximity of the body (more than 5 dB in excess for $d \leq 15$cm) [6].

3.5 Operating Frequency

3.5.1 Overview

The frequency to be used for powering and communicating with implanted RF devices is one of the most important considerations when designing a system. At low frequencies, electromagnetic energy has a significant penetration depth, and the body can therefore be used to support communications channels. For example, at 10MHz, the penetration depth is about 200mm for muscle and over 1 metre for fat. However at 2.45GHz, the depths are

25mm and 120mm respectively [40].

It is for this reason that many existing implantable radio frequency devices currently operate in low-frequency ranges, such as the widely accepted 13.56 MHz industrial, scientific, and medical (ISM) band. Unfortunately, using frequency bands like this has the disadvantage of requiring relatively large implanted antennas, and it also imposes difficulties in designing efficient high data rate transceivers [41].

Recent work considering the effects of tissue on miniature implants has demonstrated that millimetre-sized antennas in biological tissue achieve optimal power transfer efficiency in the sub-gigahertz to low-gigahertz range [42], depending on the dimension of the external reader antenna, which we have relative freedom in designing. Using this higher frequency range simplifies the design of the data link, as well as desensitizing the link gain to antenna orientation [41].

In the design of the implanted wireless sensor system, there are three separate radio communication functions, which each require selection of an appropriate and effective operating frequency. There are two separate radio waves which travel between the implanted sensor tag and the external reader, which function to power the passive sensor, and to query it for results of the measurements for computation, and the wave which will be transmitted by the sensor towards the metal plate, used for the measurements.

3.5.2 Medical Implant Communication Service (MICS)

The Medical Implant Communication Service (MICS) is a specification outlining the use of the frequency band from 402-405MHz for communication with medical implants [22]. This frequency range was selected by the Federal Communications Commission (FCC) as the most appropriate range for use with medical implants for a number of reasons:

- The 402-405 MHz frequencies have propagation characteristics conducive to the transmission of radio signals within the human body.
- Equipment designed to operate in the 402-405 MHz band satisfies the requirements of the MICS with respect to size, power, antenna performance and receiver design.
- The use of the 402-405 MHz band is compatible with international frequency allocations.

- The use of the 402-405 MHz frequency band for the MICS does not pose a significant risk of interference to other radio operations in that band. [43]

The maximum power level for operating within the MICS is limited to 25μW of effective radiated power (ERP), to reduce the risk of interfering with other devices operating in the same band [22]. The maximum used bandwidth at any one time is 300 kHz [40].
The main advantage of using the MICS is the possibility of communication range up to a couple of meters, compared to previously used inductive technologies discussed in Section 2.5.1, which required the external transceiver to touch the skin of the patient.

3.5.3 RF Characteristics of the Human Body

The main tissues that need to be considered in the selection and design of an implanted communication system are muscle and fat. The electrical properties of these tissues depend strongly on the water content of the particular tissue [7]. As can be seen in Appendix A, for both muscle and fat, with higher frequencies, average conductivity rises slowly and permittivity gradually declines. The differences in these values for muscle and fat are due to the high water content in muscle, and low water content in fat [7].
While the electrical properties of tissues vary over a large range of frequencies, there are some frequencies which can be immediately ruled out for applications such as implanted communications. For example, frequencies above 2.4GHz result in massive absorption of power by the water in human tissues, which in turn heats up the tissue above safe levels. For this reason, these frequencies are not applicable. Furthermore, as can be seen by the gradient of permittivity and conductivity of human tissues in Figures 3.7 and 3.8, operation above approximately even 1GHz is not reasonable, as conductivity sharply increases. From the dielectric properties of human tissues, it can be concluded that a suitable carrier frequency for communication with an implanted antenna would be between 50 MHz and 1 GHz, corresponding to free-air wavelength from 6m - 30cm [7].

3.5.4 Carrier Frequencies

The wireless sensor is powered and queried by the external reader device on potentially separate frequencies to optimize both. In [7], an investigation of transmission losses is carried out through 10cm of tissue, over a frequency range of 50MHz to 700MHz. As can be seen in Figure 3.10, losses increase almost monotonically with frequency increase from about 43dB at 50MHz to 63dB at 700MHz. Two minor resonances are observed that correspond fairly to the size of the tissue cube and the permittivity [7]. The results of this experiment show that transmission is possible within the whole frequency range. For this reason, usage of the 402 - 405MHz MICS band is the best options for communication between the external reader and implanted sensor, due to the advantages of this frequency range outlined above. While a higher frequency could be used to transmit more information or power the implanted circuit easier, it would require more power, and would result in more power absorption by water in the tissues. Therefore, it is more advantageous to transmit using the 402 - 405MHz MICS band, and to focus the sensor circuit design on operation using the minimum required power possible.

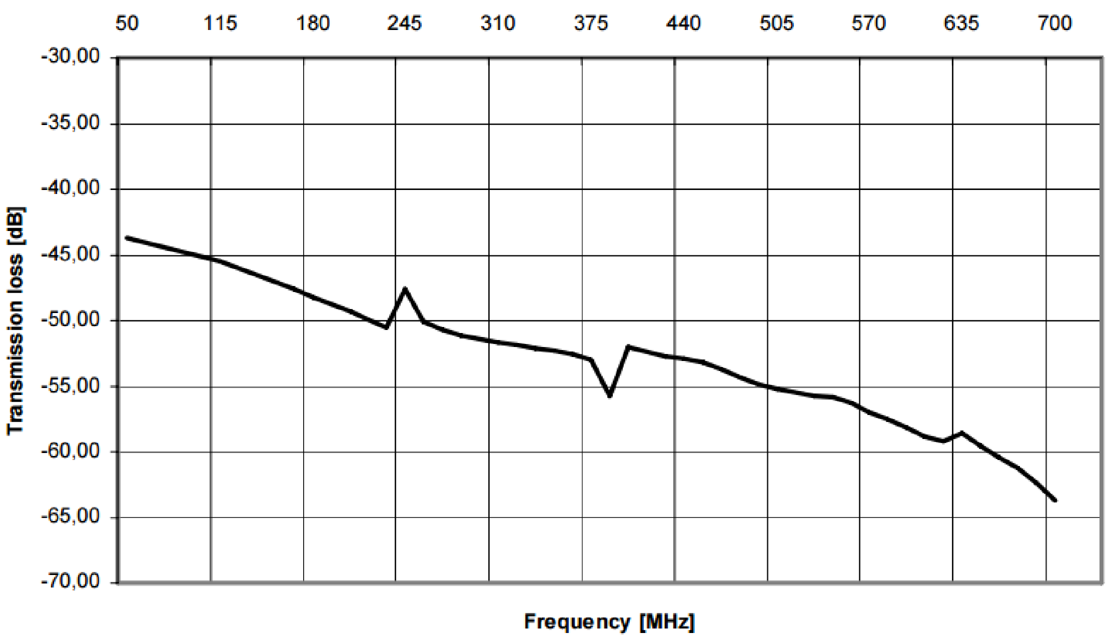


Figure 3.10: Calculated transmission loss through tissue over frequency range of 50 - 700MHz [7].

3.5.5 Sensing Frequency

For the sensing wave, the important factors to be considered when selecting an operating frequency are the design of the implanted antenna, which will be required to fit within a 3mm by 10mm cylindrical capsule, the resolution of the computation components used, and how small of a displacement can be detected by the IQ modulator, and losses due to power absorption by water in the tissues.
In terms of appropriate antenna design, the available sensing frequency range is approximately 1 - 15GHz. At the lower frequencies within this range, sourcing or making an appropriate integrated circulator with the required accuracy may be impossible, as most circulators only have a resolution of approximately 50dB on average.
Once transmitted by the sensor antenna, the sensing wave travels 1.5cm from the sensor to the metal plate, and 1.5cm back. This causes losses in the tissue. As frequency increases, the phase difference for a 10 micron movement (and therefore the minimum resolution we require from our components) increases, but losses are also increased dramatically. This rules out the highest frequencies within our workable range, as losses due to absorption will become too high, and small displacement measurements will not be possible.
Therefore, the window of possibility for an appropriate sensing frequency is approximately 4 - 9GHz. Below this, the resultant phase difference from a 10 micron movement is too small to measure, and above this, losses by absorption are too high. Within this range, the selection of a sensing frequency depends on the available resolution of the selected / designed circulator and IQ modulator. Based on the smallest phase difference that can be detected, this should correspond to a 10 micron displacement. This way, the lowest possible frequency is being utilised, resulting in minimum losses.

3.6 Power

Section

When designing a passive implanted system to be powered wirelessly, there are two major factors that will limit the success of the power transfer method. These are the received open voltage at the implanted antenna, and the amount of power that can be safely delivered to the implanted circuit while still complying with the SAR (specific absorption rate) regulation, which determines the safety of wirelessly transferring power through the human body [41].

To successfully activate the implanted device, the received open voltage at the tag antenna must be above a certain value. While complying with the IEEE safety regulations regarding the maximum allowable SAR, the received open voltage is likely to be the factor determining how deep the implant can be located within the body [41]. Successfully achieving a high enough activation voltage while still satisfying safety regulations relies of the selection of an appropriate power transfer method.

3.6.1 Powering Method

As discussed in Chapter 2, the two main methods for wirelessly powering a passive circuit from an external reader are inductive coupling and far-field RF. Inductive coupling has the advantage that the sensor could be powered via an NFC-enabled smartphone, which would eliminate the need to design an external reader. Also, inductive coupling allows for a single connection to both power the sensor and read data from it, via load modulation. The disadvantage of inductive coupling is that it is a near field effect, so for an implanted circuit to work, the distance between the reader and sensor must be less than the wavelength of the magnetic field divided by two times pi [33, 36]. This severe limit on the maximum achievable read distance is the main reason why inductive coupling is not appropriate for this system. Inductive coupling is also very sensitive to the alignment of the reader and implanted tag [33, 36]. Far-field RF is necessary to wirelessly power an implanted sensor network within knee replacements.

3.6.2 Electromagnetic Exposure

While it is technically possible to operate and implanted sensor system and power it wirelessly, the social acceptance of such devices cannot avoid the concerns about the compliance of radio emission with safety issues for an antenna placed within the human body [6]. Specific absorption rate (SAR) is a measure of the rate at which the human body absorbs energy, when exposed to a radio frequency electromagnetic field [41]. For a system such as this, the parameter that must be analysed is the localized SAR averaged over 10 g of tissue. SAR is required to be smaller than 4W/kg for the exposure of limbs and less than 2W/kg for head and trunks averaged over 10 g of tissues and time-averaged over 6minutes of exposure [6]. For an implanted system to be approved for human use, it must comply with these regulations [41].

The challenge with any implanted passive system is wirelessly transferring enough energy through the body to activate the implanted circuit, while still complying with IEEE safety regulations regarding the maximum allowable SAR.

3.6.3 Analysis

In similar implanted systems in [6] and [37], received available power at the implanted antenna has been easily sufficient to power the circuitry, with SAR regulations easily satisfied. In [6], an average of 1W was available at the reader antenna for the worst case scenario, with 15cm between external reader and implanted circuit. In [37], up to 100μW of power was available for on chip circuit operation with losses. Only 35μW after losses was required for signal conditioning, spike detection, analog to digital conversion, and control and communication circuits for a viable circuit. In both these cases, far-field RF within the same frequency ranges that this system considers, successfully powered an implanted circuit, while complying with maximum allowable SAR regulations.

4

Future Work

This report is a system specification, which serves as the basis for the design and implementation of the implanted wireless sensor system. Using this report and continued research into existing literature and general RF theory, the first aim for the future is to complete the circuit design for the implanted sensors and external reader devices, and to use this to develop a prototype for the implanted wireless sensor system. Before physical construction of a prototype, system simulations can be used to verify any calculations and support decisions.

Once a circuit prototype has been designed, it will be important to vigorously test this using meat-bone analogues, as actual human testing will not initially be possible. This stage is important to verify that the predicted results for transmitted power and displacement measurements can be achieved. The important results to test for here are the resolution of the displacement measurements by the sensor (must be at least 10 microns), the power required by the passive sensors for on-chip circuit operation, and the power received by the implanted circuit once transmitted by the external reader, after all losses. If the received power is not equal or higher than the required circuit activation power, design changes will be necessary.

It is also important at this stage to test the final SAR results, to check that they comply with the necessary safety regulations.
Once the prototype is successful and the results are up to standard, an industry partner will most likely be approached to help develop the final commercial product. The system's final commercial use in medicine will then just depend on safety approval by governing bodies.

5

Conclusion

This document outlines the design specification of the implanted wireless sensor system, which can detect knee replacement prosthesis displacements as small as 10 microns, using implanted wireless, externally powered sensors to monitor health complications over time.

The wireless sensors use electromagnetic waves to measure the distance between implanted sensors and the metal plate of the knee replacement. This distance is monitored regularly via an external reader device which reports the data and any complications to the clinician. This system allows wireless, non-invasive monitoring of the health of the implant, to eliminate the need for invasive procedures such as surgery.

As outlined in Chapter 2, there were many possible options for frequency, power transmission, and circuit design when designing an implanted wireless sensor network. Extensive research into existing literature and basic radio frequency theory was required for me to make calculated decisions about wireless power transfer, circuit design, and system structure. My processes for conducting this research and designing the system based on my findings are summarised in Chapters 2 and 3.

To support my design decisions outlined within this report, I looked to existing literature outlining similar systems, and analysed their success with regard to this system. Systems such as those detailed in [6] and [37] support the specifications selected for this system, as they have reported similar success in terms of implant depth and alignment, operating frequency, received power and achievable read distances. The details of this analysis is contained within Chapter 3.

This report is a system specification, which serves as the basis for the design and implementation of the implanted wireless sensor system. Using this report and continued research into existing literature and general RF theory, the aim for the future is to complete the circuit design for the sensors and external reader, and to use this to develop a prototype, and further on, a final implementation. Once developed, the goal is for this system to provide early detection of joint replacement failure, so that proper treatment can be applied, eliminating the need for a second orthopaedic surgery to fix the problem. The development of a final commercial version of the implanted wireless sensor system, to be implemented within the healthcare system, is the end goal of this project.

A

Dielectric Properties of Body Tissues in the Frequency Range 10 Hz - 100 GHz

This Appendix contains the results of experiments into the calculation of dielectric properties of body tissues for the design of implanted electrical systems. These values are obtained from [11]. Table A.1 contains the conductivity and relative permittivity of muscle tissue over a frequency range of 10Hz to 100GHz. Table A.2 contains the conductivity and relative permittivity of fat tissue over a frequency range of 10Hz to 100GHz.

Frequency (Hz)	**Conductivity (S/m)**	**Relative Permittivity**
10	0.20197	2.57e+07
25.119	0.21122	2.3337e+07
63.096	0.24424	1.4944e+07
158.49	0.28446	5.1102e+06
398.11	0.30557	1.4618e+06
1000	0.32115	434930
2511.9	0.33162	123590
6309.6	0.33813	40496
15849	0.34348	18055
39811	0.34973	11009
100000	0.36185	8089.2
251190	0.39591	5749.1
630960	0.46549	2962.4
1.5849e+06	0.53413	1082.7
3.9811e+06	0.58063	386.74
1e+07	0.61683	170.73
2.5119e+07	0.65126	99.06
6.3096e+07	0.68761	72.482
1.5849e+08	0.73007	61.765
3.9811e+08	0.79582	57.144
1e+09	0.97819	54.811
2.5119e+09	1.7812	52.654
6.3096e+09	5.5818	47.801
1.5849e+10	19.114	35.36
3.9811e+10	43.034	18.313
1e+11	62.499	8.6307

Table A.1: Dielectric Properties of Muscle Tissue in the Frequency Range 10 Hz - 100 GHz [11]

Frequency (Hz)	Conductivity (S/m)	Relative Permittivity
10	0.012207	7.9735e+06
25.119	0.016711	3.9402e+06
63.096	0.02012	1.0017e+06
158.49	0.021175	214090
398.11	0.021658	62295
1000	0.022404	24104
2511.9	0.023218	7529.4
6309.6	0.02368	2028.4
15849	0.023959	605.43
39811	0.024191	216.23
100000	0.024414	92.885
251190	0.024644	48.701
630960	0.024906	31.565
1.5849e+06	0.025325	24.204
3.9811e+06	0.026418	19.454
1e+07	0.029152	13.767
2.5119e+07	0.032668	8.7363
6.3096e+07	0.035242	6.5246
1.5849e+08	0.037407	5.822
3.9811e+08	0.041088	5.5806
1e+09	0.053502	5.447
2.5119e+09	0.10725	5.2737
6.3096e+09	0.32677	4.9087
1.5849e+10	0.99337	4.2154
3.9811e+10	2.2044	3.4086
1e+11	3.5624	2.8891

Table A.2: Dielectric Properties of Fat Tissue in the Frequency Range 10 Hz - 100 GHz [11]

List of Abbreviations

The following list is neither exhaustive nor exclusive, but may be helpful.

dBm .... decibel-milliwatts

ERP .... Effective radiated power

GHz .... Gigahertz

HF.......... High frequency

IC .... Integrated circuit

IEEE........ Institute of electrical and electronics engineers

ISM .... Industrial, scientific and medical

KHz .... Kilohertz

LF.......... Low Frequency

MHz........ Megahertz

MRI .... Magnetic resonance imaging

NFC .... Near field communication

RF.......... Radio frequency

RFID .... Radio frequency identification

SAR .... Specific absorption rate

SPIFA . . . . . . Stacked planar inverted-F antenna

UHF . . . . Ultra high frequency

UWB . . . . Ultra wide band

References

[1] American Academy of Orthopaedic Surgeons. *Total knee replacement* (1995). URL http://orthoinfo.aaos.org/topic.cfm?topic=a00389.

[2] D. M. Dobkin. *The RF in RFID: UHF RFID in Practice* (Newnes, 2012).

[3] J. Mcgrath. *How pet microchipping works* (2008). URL http://science.howstuffworks.com/innovation/everyday-innovations/pet-microchip1.htm.

[4] M. Mareli, S. Rimer, B. Paul, K. Ouahada, and A. Pitsillides. *Experimental evaluation of nfc reliability between an rfid tag and a smartphone*. In *AFRICON, 2013*, pp. 1–5 (IEEE, 2013).

[5] P. Tran. *Total knee replacement* (2014). URL http://www.phongtran.com.au/patientinfo/knee/knee-replacement/index.html.

[6] R. Lodato, V. Lopresto, R. Pinto, and G. Marrocco. *Numerical and experimental characterization of through-the-body uhf-rfid links for passive tags implanted into human limbs*. Antennas and Propagation, IEEE Transactions on **62**(10), 5298 (2014). URL http://ieeexplore.ieee.org/ielx7/8/6915774/06872801.pdf?tp=&arnumber=6872801&isnumber=6915774.

[7] D. Werber, A. Schwentner, and E. Biebl. *Investigation of rf transmission properties of human tissues*. Advances in radio science **4**(13), 357 (2006).

[8] S. A. Weis. *Rfid (radio frequency identification): Principles and applications*. MIT CSAIL **2**, 3 (2007).

[9] C. M. Roberts. *Radio frequency identification (rfid)*. Computers & Security **25**(1), 18 (2006). URL http://ac.els-cdn.com/S016740480500204X/1-s2.0-S016740480500204X-main.pdf?_tid=e9da9956-b726-11e4-aec8-00000aacb35d&acdnat=1424234029_a502ada65b7ba0a2d0648076be13887a.

[10] D. Sen, P. Sen, and A. M. Das. *RFID for energy & utility industries* (Pennwell Books, 2009).

[11] D. Andreuccetti, R. Fossi, and C. Petrucci. *An internet resource for the calculation of the dielectric properties of body tissues in the frequency range 10 hz - 100 ghz* (1997). URL http://niremf.ifac.cnr.it/tissprop/.

[12] J. Bellemans, M. D. Ries, and J. M. Victor. *Total knee arthroplasty* (Springer, 2005).

[13] S. H. Palmer and M. Cross. *Total knee arthroplasty*. Available on http://emedicine.medscape. com (ultimo accesso: 15 maggio, 2009) (2009).

[14] D. L. Bartel, V. Bicknell, and T. Wright. *The effect of conformity, thickness, and material on stresses in ultra-high molecular weight components for total joint replacement.* J Bone Joint Surg Am **68**(7), 1041 (1986).

[15] M. D. Ries, A. Salehi, K. Widding, and G. Hunter. *Polyethylene wear performance of oxidized zirconium and cobalt-chromium knee components under abrasive conditions*. J Bone Joint Surg Am **84**(suppl 2), S129 (2002).

[16] L. Xiaoyu, J. R. Stachel, E. Stachel, M. H. Mickle, and J. L. Berger. *The uhf gen 2 rfid system for transcutaenous operation for orthopedic implants*. In *Instrumentation and Measurement Technology Conference (I2MTC), 2013 IEEE International*, pp. 1620–1623 (2013). URL http://ieeexplore.ieee.org/ielx7/6548505/6555364/06555688.pdf?tp=&arnumber=6555688&isnumber=6555364.

[17] A. Sani, M. Rajab, R. Foster, and Y. Hao. *Antennas and propagation of implanted rfids for pervasive healthcare applications*. Proceedings of the IEEE **98**(9), 1648 (2010).

[18] A. Alomainy and Y. Hao. *Modeling and characterization of biotelemetric radio channel from ingested implants considering organ contents*. Antennas and Propagation, IEEE Transactions on **57**(4), 999 (2009).

[19] F. Merli, B. Fuchs, J. R. Mosig, and A. K. Skrivervik. *The effect of insulating layers on the performance of implanted antennas*. Antennas and Propagation, IEEE Transactions on **59**(1), 21 (2011).

[20] W. Xia, K. Saito, M. Takahashi, and K. Ito. *Performances of an implanted cavity slot antenna embedded in the human arm*. Antennas and Propagation, IEEE Transactions on **57**(4), 894 (2009).

[21] B. Glover and H. Bhatt. *RFID essentials* (" O'Reilly Media, Inc.", 2006).

[22] K. Y. Yazdandoost and R. Kohno. *Wireless communications for body implanted medical device*. In *Microwave Conference, 2007. APMC 2007. Asia-Pacific*, pp. 1–4 (2007). URL http://ieeexplore.ieee.org/ielx5/4542564/4554523/04554534.pdf?tp=&arnumber=4554534&isnumber=4554523.

[23] A. Ogirala, J. R. Stachel, P. J. Hawrylak, D. Rong, R. K. Yalamanchili, M. A. Rothfuss, X. Liu, S. Saba, and M. H. Mickle. *Impact of iso 18000 series rf signals on crmds: a unified approach*. International Journal of Modelling and Simulation **31**(3), 250 (2011).

[24] A. Iji, X. Zhu, and M. Heimlich. *High gain/power quotient variable-gain wide-band low-noise amplifier for capsule endoscopy application*. Microwave and Optical Technology Letters **54**(11), 2563 (2012). Cited By :1 Export Date: 19 January 2015, URL http://www.scopus.com/inward/record.url?eid=2-s2.0-84865469100&partnerID=40&md5=14791a826f67ad0220eaeb0661eeebfe.

[25] A. Iji, X. Zhu, and M. Heimlich. *A 3-5 ghz lna in 0.25Îijm soi cmos process for implantable wbans*. In *Midwest Symposium on Circuits and Systems*, pp. 766–769 (2012). Export Date: 19 January 2015, URL http://www.scopus.com/inward/record.url?eid=2-s2.0-84867319727&partnerID=40&md5=bb5eba7cc8cdbcfcb51cb97311f7637a.

[26] C. Occhiuzzi, G. Contri, and G. Marrocco. *Design of implanted rfid tags for passive sensing of human body: The stentag*. Antennas and Propagation, IEEE Transactions on **60**(7), 3146 (2012). URL http://ieeexplore.ieee.org/ielx5/8/6230681/06196180.pdf?tp=&arnumber=6196180&isnumber=6230681.

[27] G. Marrocco, L. Mattioni, and C. Calabrese. *Multiport sensor rfids for wireless passive*

sensing of objects: Basic theory and early results. Antennas and Propagation, IEEE Transactions on **56**(8), 2691 (2008).

[28] A. Paus. *Near field communication in cell phones*. Seminararbeit Ruhr-Universitat Bochum (2007).

[29] J. Thrasher. *Rfid vs. nfc: What's the difference?* (2013). URL http://blog.atlasrfidstore.com/rfid-vs-nfc.

[30] NearFieldCommunication.org. *Near field communication technology standards* (2015). URL http://www.nearfieldcommunication.org/technology.html.

[31] C. Reinhold, P. Scholz, W. John, and U. Hilleringmann. *Efficient antenna design of inductive coupled rfid-systems with high power demand*. Journal of communications **2**(6), 14 (2007).

[32] H. Al-Offishat and M. Rababah. *Near field communication*. Int. J. Comput. Sci. Netw. Secur.(IJCSNS) **12**(2), 93 (2012).

[33] B. Bilginer and P.-L. Ljunggren. *Near field communication* (2011).

[34] P. Csurgai and M. Kuczmann. *The mutual inductance effective permeability and its application*. Acta Technica Jaurinensis **5**(1), pp. 67 (2012).

[35] R. Volpato, F. Ramos, P. Crepaldi, M. Santana, and T. C. Pimenta. *Evaluation of coupling factor in rf inductively coupled systems*. World Academy of Science, Engineering and Technology **1**(64), 1066 (2012).

[36] A. Vena and P. Roux. *Near field coupling with small rfid objects*. Session 3AP p. 153 (2009).

[37] C. Hutchens, R. L. Rennaker, S. Venkataraman, R. Ahmed, R. Liao, and T. Ibrahim. *Implantable radio frequency identification sensors: wireless power and communication*. In *Engineering in Medicine and Biology Society, EMBC, 2011 Annual International Conference of the IEEE*, pp. 2886–2892 (IEEE, 2011).

[38] C. Gabriel. *Compilation of the dielectric properties of body tissues at rf and microwave frequencies.* Tech. rep., DTIC Document (1996).

[39] J. A. Shaw. *Radiometry and the friis transmission equation*. American Journal of Physics **81**(1), 33 (2013).

[40] P. S. Hall and Y. Hao. *Antennas and propagation for body centric communications*. In *Antennas and Propagation, 2006. EuCAP 2006. First European Conference on*, pp. 1–7 (IEEE, 2006).

[41] A. Yakovlev, S. Kim, and A. Poon. *Implantable biomedical devices: Wireless powering and communication*. Communications Magazine, IEEE **50**(4), 152 (2012).

[42] A. S. Poon, S. O'Driscoll, and T. H. Meng. *Optimal frequency for wireless power transmission into dispersive tissue*. Antennas and Propagation, IEEE Transactions on **58**(5), 1739 (2010).

[43] Computer Support Group Inc. *Medical implant communications service frequency table* (1973). URL http://www.csgnetwork.com/micsfreqtable.html.