# DESIGN OF A WIRELESS POWER AND DATA LINK FOR A CRANIALLY-IMPLANTED NEURAL PROSTHESIS

by

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## ABSTRACT

When a human body sustains damage to a nerve connecting the brain to a muscle, the muscle can no longer be voluntarily stimulated by the individual's nervous system and is therefore permanently paralyzed. Recent studies have shown that it may be possible to restore the connection between the brain and a paralyzed muscle using sophisticated brain-signal analysis algorithms to control the artificial stimulation of a paralyzed muscle. The system requires direct connection to the brain tissue and therefore a wireless link is desired.

The California Institute of Technology, Jet Propulsion Laboratory (JPL) is currently researching the feasibility of creating a complete system. This thesis, funded by JPL, is targeted at researching the feasibility of creating a wireless brain-computer interface (BCI) link that will transmit power and control data to an implanted prosthetic and transmit sampled data from the prosthetic to an external receiver. There are a number of well documented wireless trans-cranial links; however, none have achieved the 10Mbps data rate necessary for JPL's project.

The link designed for this research uses separate pairs of coils for the data and power links and two separate frequencies to inductively couple signals between the implant and the external system. This thesis documents the procedures and equations necessary for optimizing the wireless links as well as the theory behind the operation. The theory and design procedures are validated through the development of a board-level prototype.

The design of the system targets the future implementation in integrated form with the exception of the coils used for inductively coupling signals for the links. A proposed IC implementation of the circuits is provided.

# **TABLE OF CONTENTS**

L	IST	OF FIG	GURES	iii
L	IST	OF TA	BLES	v
A	CKN	NOWL	EDGEMENTS	vi
1	Int	roducti	on	1
	1.1	Motiva	ation	1
	1.2	Resear	rch Goals	2
	1.3	Thesis	Organization	4
2	Bac	kgrou	nd & Literature Review	5
	2.1	Backg	round	5
	2.2	Literat	ture Review	7
3	The	e Propo	sed System	10
	3.1	System	n Overview	10
	3.2	Physic	al Parameters	11
	3.3	System	n-Level Design	12
4	Abs	sorptio	n of Electromagnetic Radiation in Human Tissues	13
	4.1	Electro	omagnetic Radiation	13
	4.2	Quant	ifying Exposure	14
5	Lin	k Desig	gn	18
		5.0.1	System Frequency Spectrum	18
		5.0.2	Mutual Inductance	19
	5.1	Power	Link	20
		5.1.1	Operating Frequency	20
		5.1.2	Inductive Coupling	21
		5.1.3	Coil Alignment	24
		5.1.4	Effect of Q on Power Reception	25
		5.1.5	Rectifier	26
		5.1.6	Transfer Function of the Link	28
		5.1.7	Power Receiver Coil Parameters	30

		5.1.8	Transmit Power and Absorbed Power Calculations	32
	5.2	Data L	.ink	34
		5.2.1	Wireless Transmission Method	35
		5.2.2	Operating Frequency	35
		5.2.3	Modulation Scheme	36
		5.2.4	Geometry	36
		5.2.5	Transmit and Receive Power	38
		5.2.6	Power-Transmitter Interference	39
		5.2.7	Interaction Between the Implanted Coils	. 40
	5.3	Link D	Design Summary	41
6	The	eory Ve	rification	43
	6.1	Goals		43
	6.2	Prototy	ype Circuit Operation	43
	6.3	Measu	red Results	46
7	Pro	posed l	IC Implementation	48
	7.1	Rectifi	er (RF/DC Converter)	48
	7.2	5.0V F	Regulator	49
	7.3	ASK I	Data Demodulator	50
	7.4	RF Lir	niter and Signal Strength Feedback	51
	7.5	Data N	Aodulator, Up-Converter, and Transmitter	51
8	Cor	nclusion	ns and Future Work	53
	8.1	Conclu	usions	53
	8.2	Future	Work	53
R	EFE	RENC	ES	55
A	PPE	NDIX 1	1: Self-Inductance Calculations	57
APPENDIX 2: Mutual-Inductance Calculations58				58
<b>APPENDIX 3:</b> Tissue Properties <b>60</b>				
APPENDIX 4: Magnetic Flux Density Calculations       61				

# **LIST OF FIGURES**

1.	Block diagram of a neural-recording trans-cranial prosthesis	3
2.	System Diagram	11
3.	Transmitted Power and Control Data Signal	11
4.	Primary tissues that separate the implanted device	
	from the external system	12
5.	Circuit model for a lossy inductor in the presence of tissues	16
6.	Q-factor of a 5-turn coil when placed in air and when placed	
	against a human forehead	17
7.	System Frequency Spectrum	19
8.	Mutual inductance interaction between the four coils	19
9.	(a) Placement and dimensions of the two coils in the power link (b) Cross-	
	sectional view showing the direction of the magnetic flux density vector	21
10.	Cross-section of the coils showing the geometry factors	23
11.	Dependence of the coupling coefficient on the coil separation $d$ in cm	24
12.	Cross-section of the coils showing the direction of the magnetic flux	
	density vector	24
13.	Dependence of the normal component of the magnetic flux density vector	
	on the depth of the receiving coil plane	25
14.	(a) Parallel R-L network, (b) Parallel R-L-C network	25
15.	Circuit model of the power link	26
16.	Simplified power link circuit model	27
17.	The effects of loading on the Q of the receiver network	32
18.	$B_z$ at a distance of 1mm from a 4cm diameter transmitting coil	34
19.	Area exposed to the strong B-field of the transmitting coil	34
20.	Block diagram of the data path	35
21.	Baseband BPSK Spectrum	36
22.	Normal flux density vector in a plane at a distance of 2cm from a	
	radiating 1cm diameter coil	37

23.	Circuit example using only one external coil	39
24.	Data Receiver	40
25.	Transformer Model	41
26.	Schematic of prototype	44
27.	Power transmitting coil and data receiver coil	45
28.	Board-level prototype of the power receiver and the data transmitter	45
29.	Spectrum of the tested system.	47
30.	(a) half-wave rectifier (b) full-wave voltage doubling rectifier	
	(c) full-wave bridge rectifier (d) half-wave voltage-doubling rectifier	49
31.	Coil dimensions for computing the inductance of a coil	57
32.	Geometry factor, F	59
33.	(a) Relative Permittivity and (b) Electrical Conductivity ([17])	60

# **LIST OF TABLES**

1.	ISM Frequency Allocations	20
2.	Measured data from FR4 coils	31
3.	Link parameters used for power calculations	32
4.	Data Link Parameters	38
5.	Prototype Component Values	44
6.	Signal strengths measured on prototype system	46
7.	Tissue density and thickness for a human head	60

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# **Chapter 1**

### Introduction

### 1.1 Motivation

A recent study at Brown University successfully developed a Brain-Computer Interface (BCI) that allowed a rhesus macaque (a species of monkey commonly used in medical research) to control a game-like program on a computer by simply thinking about the movements ([1]). The study was conducted in hopes that this technology can someday allow some paralyzed individuals to once again have mental control of the movement of their paralyzed muscles.

The game played by the monkey involved the movement of a cursor on the screen to a red dot that would appear at random locations. The monkey first played the game using a joystick to move the cursor. Using neural sensors that were directly attached to the brain of the monkey, the neural activity was recorded while the monkey moved its arm. Using this data, a system was developed to interpret the signals in real-time. Once this was achieved, the joystick could then be disconnected and the game was then played only by using the BCI.

The problem with the system is that it requires direct connection to the brain tissue. The presence of wires passing through the cranium of a patient can leave affected areas very susceptible to infection and therefore can only be performed in a sterile clinical environment. In order to break the physical link between the brain matter and the outside world while retaining the ability to directly probe the tissues requires the use of a wireless data link. This type of short-range data link can be implemented using inductive coupling to transmit data bi-directionally between an implanted device and the outside world. Wireless transcutaneous data links using inductive coupling have been successfully created in the past. Some examples of such links will be discussed in Chapter 2.

The use of brain signals for the voluntary control of muscles requires the acquisition of a large amount of data. This project targets the transmission of 10Mbps across the data link. Previously researched transcranial data link solutions have not achieved data rates this high.

Another complication with using an implanted prosthetic device is the method of providing power. As with the data link, the use of wires is undesirable due to health risks. The use of an implanted battery would provide a limited supply of power to the prosthetic at the cost of greatly increasing the size. When the battery expires, it would need to be surgically replaced. It is possible to recharge the battery using an inductive link, but the size and toxicity of the battery is still a concern. Instead of a battery, all of the power can be supplied using an inductive link. This method has been successfully used for many related research projects which will be discussed in Chapter 2.

### 1.2 Research Goals

To understand the goals of this research, one must understand the organization of the components in the full system. The fundamental endpoints of the system are the brain tissue and the computer. The system components between the brain and the computer are illustrated in the block diagram in Figure 1. Starting from the left end of the diagram, the computer sends control data to the signal generator to be transmitted to the prosthetic via

2

the forward link. The output of the signal generator is modulated using the control data from the computer. In this project, the forward link signal will be used both to power the prosthetic as well as to send data. Once the prosthesis is powered and is receiving control data it can begin using the sensors to transduce electrical signals in the brain to signals that can be converted to data using a data converter within the prosthesis. This data is then sent from the prosthesis to the computer via the reverse link. When the data gets from the prosthetic to the computer and has been processed, it can then be used to provide information for many possible applications such as the control of a paralyzed limb, initiation of the injection of a drug, or the control of cursor movements on the screen of a computer.



Figure 1: Block diagram of a neural-recording trans-cranial prosthesis

This research focused on analyzing the factors involved in implementing the two wireless links. The primary goal was to show that it is possible to implement a wireless link between the outside world and the inside of the human head that can transmit 15mW of power in to an implanted device and 10Mbps of data out from that same device. In this research, the obstacles in creating a transcranial power and high data rate link have been thoroughly investigated, a solution that addresses all of the obstacles is proposed, a board-level prototype of the link was constructed that validates the theory, and circuits necessary for the development of an integrated system have been described.

### 1.3 Thesis Organization

This thesis will begin in Chapter 2 with some background information that the reader will need to fully understand the thesis material and the challenges met in this research. The background information will be complemented by a review of previous work related to this project. Chapter 3 will introduce the proposed implementation of the system. This information will give the reader a clear picture of the system-level architecture of the proposed system. Chapter 4 will discuss the effects of non-ionizing radiation on the human body. This information will explain the necessary precautions in assuring that this system will not have harmful health effects. Containing the bulk of the thesis, Chapter 5 will discuss the design of the power and data link. Chapter 6 will then discuss the design and results of the board-level prototype. Chapter 7 will discuss the factors involved in designing a fully-integrated version of the system. Concluding the thesis, Chapter 8 will present the conclusions and the recommended future work.

## **Chapter 2**

# **Background &** Literature Review

This chapter provides a brief overview of some applications for implanted prostheses, research challenges, and a review of related prior work. This chapter will also introduce some of the initially known problems with making this project a success. The background information is complemented by a literature review of some researched solutions and their system requirements.

### 2.1 Background

A neural prosthesis is a device that is used to stimulate neurons or record neural signals. Some examples of stimulating systems include those for curing some types of blindness and deafness or providing motor control via the direct stimulation of neurons. Examples of neural signal recording prostheses are much fewer in number and application. The primary use has been in the research of systems such as the one motivating this research.

The most efficient way to transmit the power and data signals over a small distance is to use inductive coupling. With this type of setup, the link can be modeled as a weak transformer (e.g. a transformer with a low coupling coefficient). Also, this means that to induce an arbitrary voltage amplitude in the receiving coil, the transmitting voltage must be inversely proportional to the coupling coefficient implying that the transmit voltage will be very large. This brings about the concern of the power absorbed in the tissues. The power absorbed in the tissues is a function of both amplitude and frequency. Appendix 3 documents the relative permittivity and the electrical conductivity of the tissues involved for the frequency range of interest. From this data, it can be noted that the absorption will increase with frequency. Another challenge in this design is dealing with the undesired interaction between the coils.

### 2.2 Literature Review

This section discusses other neural prosthesis research projects that use inductive coupling for data and/or power transmission. Several articles have been published (e.g. [2], [3]) with details on the design of inductively coupled coils for implanted devices. They do not provide much useful information for this project because they are typically concerned with links that exhibit high coupling coefficients.

Many different links have been designed for many different applications. However, whether they were designed for implantation in the skull or implantation in another part of the body, many of them still applied concepts that can be applied to this application.

Ghovanloo *et al.* documented in [4] the development of a transcutaneous power and forward data link intended for wirelessly transmitting power and data to an implant that would be responsible for stimulating the brain in the visual cortex area. The implanted device is powered by inductively coupling a 4MHz carrier signal into an implanted coil. The induced signal is rectified and regulated to provide 50mW of DC power via regulated  $\pm$ 5V supplies. The IC was designed in a BiCMOS process and therefore had the freedom to use diode-connected BJTs to create a full-wave bridge rectifier. The system also has a 100kbps forward data link meaning that data is transmitted from the outside of the body to the implant. The data link was implemented using a method of data transmission that is very common in transdermal data/power link applications. The data simply performs a low sensitivity amplitude modulation of the power carrier that can be detected by the implant.

In [5], Bashirullah *et al.* documented the design of a bi-directional link for an implantable retinal prosthetic device. Using an inductive link, the unit transmits a 1MHz power signal that is rectified and regulated to provide 250mW of power via  $\pm$ 7V DC supplies to a prosthetic device implanted in the eye. A reverse data link was implemented to indicate to the amplitude of the received signal to the transmitter. Using this feedback, the transmitter, a variable-power-level class-E amplifier, can then have knowledge of exactly how strong of a signal it needs to transmit. This is done to compensate for instantaneous changes in power level due to coil misalignment and load variations. The data link is implemented using a method known as load modulation. The load modulation method modulates the loading of the receiving coil which can then be detected as a change in impedance by the transmitting coil which is why it is also sometimes called reflected impedance modulation.

In [6], Von Arx and Najafi documented the development of a fully-integrated (including the receiving coil) implantable micro-stimulator capable of providing 20mW to implants at depths up to 3cm. The external transmitter inductively couples a 4MHz RF power carrier into an on-chip coil. The on-chip coil is electroplated after the standard BiCMOS fabrication. The inductor is a 17-turn 2x8.7mm<sup>2</sup> coil resulting in a 2.9uH inductance and a Q of 2.6 at 4MHz. Data is transmitted to the chip at 8.3kb/s by pulse-width encoded amplitude modulation of the power carrier. Full-wave rectification

generates a 4V DC supply. It is unclear whether or not this system exceeds maximum permissible exposure (MPE) limits.

In [7], R. J. Betancourt-Zamora developed the BioLink Implantable Telemetry System (BITS) which is a low-power bi-directional data link for recording bio-signals in lab animals. The system is suitable for short range biosensor and implantable use. This digital RF transmitter operates at 100kbps, using QPSK modulation in the 174-216MHz ISM band. Command and control of the unit is achieved through a 433MHz downlink. The implant uses a battery for power and is therefore only for short-term use.

Yu and Najafi developed a system for the wireless connection to neural recording microprobes. The system uses inductive coupling to transmit power to the implanted device which is implanted only a few millimeters deep. [8] documents the design and testing of the front-end circuitry which included the voltage regulator, clock recovery, ASK demodulator, power-on-reset, and the RF limiter. The link uses a 4MHz carrier for the power and control data. The document does not discuss the data transmitter but does indicate that it would use a separate pair of coils and would be at a different frequency.

In [9], Ghovanloo and Najafi analyzed several different BiCMOS integrated rectifier topologies for use with 4MHz powered biomedical implants. Application of the rectifiers is discussed in the implementation of an integrated +/-5V supply. Methods are discussed for decreasing substrate leakage from the diodes. A new CMOS bridge rectifier is analyzed which eliminates substrate the leakage. [9] also provides some data that indicates that some of the rectifiers including the new CMOS bridge rectifier would function efficiently at the higher frequencies used in this project.

Harb *et al.* in [10] developed an implantable CMOS circuit that rectifies and regulates a 40MHz signal received via an inductive link to 5V DC. The 40MHz signal is also amplitude modulated at 200kHz for the transmission of data to the implanted device. The circuit is fully-integrated with the exception of one of the rectifying diodes. The author documents the fabrication and testing of the rectifier, regulator, and data demodulator in a 1.2um CMOS process.

None of the projects described above would work for the current application since the data rates and the power link frequencies are too low.

## **Chapter 3**

## **The Proposed System**

This chapter provides an overview of the proposed system's architecture and primary component functions. The subsequent chapters will then discuss the theory and design of the link.

#### 3.1 System Overview

The proposed link shown in Figure 2 is implemented using four separate coils; one pair of coils for transmitting power and control data into the head and the other pair for transmitting data out of the head at 10Mbps. To power the prosthesis, L1 radiates a large-amplitude 27MHz signal from outside of the head inducing a voltage in L2. The resonating capacitor in parallel with L2 increases the voltage transfer and will be explained in Chapter 5. The induced voltage is rectified, filtered, and regulated to provide 15mW of 5V DC power to the prosthesis. The regulated supply will also provide power to the other components involved in creating the links such as the regulator, amplitude detector, and the data transmitter. This will further increase the load on the regulated supply. The targeted maximum load is 20mW.

Control data will be transmitted to the prosthetic for functions such as starting and stopping the prosthesis activity. Using a low sensitivity modulation scheme, the control data will modulate the amplitude of the 27MHz carrier signal as illustrated in Figure 3. The amplitude variation of the carrier signal can then be detected by the implanted amplitude detector and converted to useful data. The data that is recorded from the prosthesis will be BPSK (bipolar phase-shiftkeying) modulated and up-converted to 310MHz to be transmitted from L3. The BPSK data signal is then received by L4 and down-converted to a base-band data signal.



Figure 3: Transmitted Power and Control Data Signal

### **3.2 Physical Parameters**

The amount of coil separation is determined by the thickness of the tissues. The thickness of the tissues will vary from person-to-person and placement of the link on the head. The separation could be as much as 2cm ([11]) and therefore 2cm been chosen as the targeted thickness in this research. The implanted coil needs to be a flat spiral coil so

that is can reside harmlessly in the cerebrospinal fluid between the brain tissue and the skull. The outside diameter of the implanted coil has been limited to 1cm to reduce the intrusion on the body.



Figure 4: Primary tissues that separate the implanted device from the external system

### 3.3 System-Level Design

The implanted part of the system will be designed with the intent of future full integration with the exception of the two coils used for receiving power and transmitting data. This research will focus on the development of a board-level prototype of the system to investigate the challenges and feasibility of meeting the requirements of the system. Factors affecting the feasibility of system integration will be considered in the development of the prototype.

## **Chapter 4**

# **Absorption of Electromagnetic Radiation in Human Tissues**

This chapter will discuss electromagnetic radiation, its effects on human tissues, and how to determine the maximum allowable transmit coil voltage/current.

### 4.1 Electromagnetic Radiation

Electromagnetic radiation can be divided into two different types: ionizing and non-ionizing. Ionizing radiation is radiation that has sufficient energy to remove electrons from atoms. Ionizing radiation, depending on the level of exposure can cause serious health problems such as increasing the possibility of cancer or harmful genetic mutations.

Non-ionizing radiation is radiation that does not have enough energy to create ion pairs. Non-ionizing radiation includes the spectrum of ultraviolet, visible light, infrared, microwave, radio-frequency, and extremely low frequency. This project deals with nonionizing radiation in the form of a radio-frequency electromagnetic wave. The main effect of non-ionizing radiation is the heating of tissues due to the absorption of electromagnetic energy. This heating of tissues has been known to have side-effects including the inductions of opacities of the lens of the eye, possible effects on development and male fertility, various physiological and thermoregulatory responses to heat, and decreased ability to perform mental tasks ([12]).

### 4.2 Quantifying Exposure

FCC regulations limit the maximum permissible exposure (MPE) to non-ionizing radiation based on the amount of temperature rise that will occur. The regulation limits the temperature rise to 1°C. This limit being energy/°C is determined by the specific heat of the tissue. Approaching the computation of exposure from a heating standpoint is very difficult since many unpredictable factors are involved (i.e. blood flow). Therefore MPE limits have also been defined in terms of W/kg of absorption.

Poynting's Theorem tells us that when a mass is exposed to an electromagnetic field, energy from the field is either stored in the mass in the form of an electric or magnetic field or dissipated via conduction losses. This is put into equation form as the net power flowing out of a volume v using (1).

$$\int_{v} \nabla \cdot \vec{E} x \vec{H} dv = -\frac{1}{2} \frac{d}{dt} \int_{v} (\varepsilon E^{2} + \mu H^{2}) dv - \int_{v} \sigma E^{2} dv$$
(1)

where *E* is the electric field, *H* is the magnetic field, and *e*,  $\mu$ , and *s* are the electric permittivity, magnetic permeability, and the conductivity of the material occupying the volume respectively. The first integral in (1) represents the energy that is stored in the material via electric fields and magnetic fields. The second integral in (1) represents the energy that is dissipated in the material via conduction losses resulting in heating of the material. Therefore the power dissipated in the material is

$$P = \operatorname{Re}\left\{\vec{E}\mathbf{x}\vec{H}\right\} = \int_{v} \frac{\sigma |E|^{2}}{2} dv \,.$$
<sup>(2)</sup>

When this energy dissipation is applied to the heating of a mass it is referred to as the specific absorption rate (SAR) which is

$$SAR = \frac{\sigma |E|^2}{2 density} = \left[\frac{Watts}{kg}\right]$$
(3)

SAR is the time-rate at which energy is absorbed in a unit mass of tissue expressed in Watts per kilogram. SAR limits are defined differently for whole-body (uniform) and partial-body (non-uniform) exposure and for both uncontrolled and controlled environments. A controlled environment is a location/situation such as the one used in this research where the subject is aware of the EM exposure ([13]). In the past, most analyses and SAR limits were for whole-bode applications. However with the rising concern with wireless telephones and their localized radiation close to the head the limitations were made more detailed/specific and more involved analyses were conducted. According to IEEE standard C95.1, spatial peak SAR in a controlled environment should not exceed 8W/kg as averaged over any 1gram of tissue in the shape of a cube ([13]).

The most widely accepted method for the computation of SAR is to use the Finite-Difference Time-Domain (FDTD) method. The FDTD method is a 1, 2, or 3-dimentional method of electromagnetic simulation using Maxwell's curl equations. It is also important to note that the FDTD method works well for heterogeneous media which would make it ideal for its applications within this project.

The development of a 3-dimensional simulator using the FDTD method is too involved for the scope of this project and pre-written software is not currently owned by the Department of Electrical and Computer Engineering at Kansas State University. Instead, this project uses a measurement-based method for making absorption approximations. The method works as follows: First, a coil was made and its impedance measured in open air giving (4).

$$Z_{\rm air} = R_{\rm air} + jX_{\rm s} \tag{4}$$

From the open-air impedance measurement the loss resistance in air is known and is denoted as  $R_{air}$  in (4) and Figure 5. This loss resistance is due to the DC resistance and the proximity effect in the coil conductors. The power dissipated in the inductor can then be calculated using  $R_{air}$ . When the coil is placed near a lossy medium, more power is absorbed in the medium which is interpreted in the circuit model as an increase in the loss resistance as denoted by  $R_{tissue}$  in Figure 5.



Figure 5: Circuit model for a lossy inductor in the presence of tissues

The increase in loss resistance causes a decrease in the Q of the coil where

$$Q = \frac{X_s}{R_s} \tag{5}$$

and

$$R_s = R_{air} + R_{tissue} \,. \tag{6}$$

To ease the calculation of power absorption using a known voltage on the coil, the impedance was converted to its parallel equivalent form of  $R_p$  and  $X_p$ .

$$R_p = \left(1 + Q^2\right) R_s \tag{7}$$

$$X_p = \frac{1+Q^2}{Q^2} X_s \approx X_s \tag{8}$$

Knowledge of  $R_p$  allows the calculation of the power dissipated in the coil for a given voltage amplitude on the coil.

$$P_{air,tissue} = \frac{1}{2} \frac{V^2}{R_p} \tag{9}$$

Therefore, the power dissipated in the coil was calculated while in open-air and while the inductor was placed against human tissue. The Q measurements for both cases are plotted in Figure 6. Figure 6 shows that the Q decreases as frequency increases. The decrease in Q indicates that the absorption in the tissues increases with frequency. The calculation of the power absorbed in both cases, in air and against tissues, provides an approximation of the total absorption in the tissues using (10).

$$P_{absorbed} = P_{tissue} - P_{air} \tag{10}$$

This absorption data will be used for both the calculation of total energy absorption and for the decision of appropriate frequencies for the signals.



igure 6: Q-factor of a 5-turn coil when placed in air an when placed against a human forehead

## **Chapter 5**

## Link Design

The data and power link cannot be discussed completely separately as there is a great deal of interaction between the two. The interactions exist primarily in the overlapping frequency spectrums and the inductive coupling.

#### 5.0.1 System Frequency Spectrum

The frequency spectrums of the transmitted power and data signals have a great influence on the choice of operating frequencies. Ideally, the frequency spectrum of the power transmit signal is a single impulse. The control data will spread this impulse to give a widened bandwidth. Assuming that the data rate of the control data is low, the widening of the impulse will be small. Unfortunately, in practice, the power signal will have many harmonics spreading over much higher frequencies.

The data signal transmitted from the implant has a wide bandwidth but is very weak and needs to be positioned at an appropriate frequency relative to the power signal such that it does not overlap one of the power signal harmonics. The data signal bandwidth is dependent on the type of modulation that is chosen, but as a maximum, if BPSK is used for a 10Mbps signal, the resulting null-null bandwidth is 20MHz. As seen in Figure 7, the data spectrum needs to fit between the power signal harmonics. This gives a theoretical minimum of 20MHz for the frequency of the power signal. The data link can therefore be centered at:

$$f_{data} = n \frac{27}{2} \text{ MHz} = 13.5 n \text{ MHz} \qquad n = 1,3,5,\dots$$
 (11)



Figure 7: System Frequency Spectrum

#### 5.0.2 Mutual Inductance



Figure 8: Mutual inductance interaction between the four coils

As mentioned in Chapter 3, the two links will be implemented using four separate coils. Therefore, in addition to the intended coupling of the power signal into the power receiver  $(k_{1,2})$  and the data into the data receiver  $(k_{3,4})$ , there are also undesired interactions between the coils. For example, coupling  $k_{1,4}$  of the power transmitted signal from L1 into the data receiver coil L4 will likely harm the input of the LNA used to detect the data signal. Also, the coupling  $k_{2,3}$  between the data transmitter coil and the power receiver coil causes a shift in the resonating frequency of the data transmitter. These matters will be discussed in the following sections detailing the design of the individual links.

### 5.1 Power Link

The geometry and operating frequency are the two primary attributes that define the power link. In this section, the link will be designed around these two attributes to maximize the voltage transfer and minimize the power absorption in the tissues.

#### **5.1.1 Operating Frequency**

The operating frequency is limited by three factors: its placement in the RF spectrum with respect to other signals, the bandwidth of the data signal, and the Q of the resonant circuit on the receiver side of the power link. The placement of the operating frequency with respect to other signals is a concern because the power link transmitter will need to create a very strong B-field to couple a usable signal into the implanted coil and therefore the operating frequency should be placed in the RF spectrum where it will not interfere with other systems. A good place is in one of the ISM (Industrial, Science, and Medical) bands shown in Table 1.

Table 1: ISM Freq	uency Allocations
6.78 MHz	$\pm 15.0 \text{ kHz}$
13.56 MHz	$\pm 17.0 \text{ kHz}$
27.12 MHz	$\pm 163.0 \text{ kHz}$
40.68 MHz	$\pm 20.0 \text{ kHz}$
915 MHz	±13.0 MHz
2450 MHz	$\pm 50.0 \text{ MHz}$
5800 MHz	$\pm 75.0 \text{ MHz}$
24.125 GHz	±125.0 MHz
61.25 GHz	±250.0 MHz
122.5 GHz	$\pm 500.0 \text{ MHz}$
245 GHz	$\pm 1.0 \text{ GHz}$

As discussed in Chapter 4, the operating frequency cannot be too high or the absorption will be too high, increasing the potential for health risks. As described in the introduction to this chapter, the operating frequency is limited on the low side by the bandwidth of the data signal with a theoretical minimum of 20MHz. Another limit on the

operating frequency is the Q of the inductors. Figure 6 shows that inductor Q decreases rapidly as the frequency decreases below the frequency where the maximum Q is observed. Considering all of the limiting factors, the 27.12MHz ISM band is a good choice. Above this frequency, the absorption becomes too large and below this frequency the Q of the inductor decreases rapidly and the harmonics of the power signal encroach on the data signal bandwidth.

#### 5.1.2 Inductive Coupling

The transmission of a signal over short distances is most efficiently performed using inductive coupling. It is also important to note that regular antennas at the chosen operating frequency would be far too large for implantation which. Therefore, the link will use two flat spiral inductors to create an inductively coupled link. As seen in Figure 9, the external primary coil will generate magnetic flux field that will penetrate the area of the secondary coil.



(b) Cross-sectional view showing the direction of the magnetic flux density vector

The geometry of the link is described by three dimensions: the radii of the two coils and the spacing between the coils. The maximum diameter of the implanted coil has been limited to 1cm as stated in the project proposal and the separation distance d is determined by the thickness of the tissues. With two of the three attributes pre-defined, the radius of the transmitting coil is the only remaining dimension to be determined.

The radius of the transmitting coil is chosen to maximize the coupling coefficient, k, between the coils. The coupling coefficient represents the fraction of flux, f, created in the primary coil (denoted by f<sub>1</sub>) that is coupled into the secondary coil (denoted by f<sub>21</sub>). When all of the flux generated by coil 1 is only in coil 1 and vice versa the two coils have no common flux (f<sub>12</sub> and f<sub>21</sub> = 0) and k is therefore zero. The coupling coefficient is 1 when all of the flux created by the primary coil is in the secondary coil (f<sub>11</sub> and f<sub>22</sub> = 0).

The coupling coefficient is only dependent on the geometry of the link and therefore independent of the number of turns in either of the coils. This can be seen by reducing an equation relating the inductances to the coupling coefficient.

$$k = \frac{M}{\sqrt{L_1 L_2}} = \frac{M_o n_1 n_2}{\sqrt{n_1^2 L_{1o} n_2^2 L_{2o}}} = \frac{M_o}{\sqrt{L_{1o} L_{2o}}}$$
(12)

where *M* is mutual inductance between the coils,  $L_1$  and  $L_2$  are the inductances of the coils,  $n_1$  and  $n_2$  are the number of turns in  $L_1$  and  $L_2$ ,  $M_o$  is the single-turn mutual inductance, and  $L_o$  is the single-turn inductance. This relationship ignores the fact that for small dimensions, the geometry of the link is changed slightly by changing the number of turns.

From (12) it is obvious that to maximize the coupling coefficient, the mutual inductance needs to be maximized. The mutual inductance of two concentric circles can be calculated using (13)

$$M = F \sqrt{r_1 r_2} \tag{13}$$

where  $r_1$  and  $r_2$  are the radii of the coils in cm, and *F* is a factor dependent on a/A (see Figure 10) that can be found using information provided in Appendix 2.



Figure 10: Cross-section of the coils showing the geometry factors

A first-order approximation to maximizing M that has been used in many previous implementations is to maximize F. F is maximized when a/A is minimized. Therefore,

$$\frac{d(a/A)}{d(r_1)} = 0 \Longrightarrow r_1 = \sqrt{2r_2^2 + 2d^2}$$
(14)

Therefore, when  $r_2=0.5$ cm and d=2cm,  $r_1=2$ cm. A more involved computation that includes the  $\sqrt{r_1r_2}$  term in (13) shows that the maximum mutual inductance occurs when the radius of the transmitting coil is 2.9cm. Figure 11 illustrates how the coupling coefficient changes with coil separation for both the 2cm and 2.9cm transmit coils. Figure 11 also shows that as expected, the link would achieve much larger coupling coefficients for smaller separation distances with the 2cm coil but the 2.9cm coil exhibits a much smaller sensitivity to vertical misalignment. If the radius is increased further, the sensitivity would be further improved but at the expense of efficiency since the coupling coefficient would decrease.



Figure 11: Dependence of the coupling coefficient on the coil separation d in cm

#### 5.1.3 Coil Alignment

This discussion can be aided by an analysis of the magnetic flux density in the effective area of the receiving coil as it gives insight to the horizontal misalignment characteristics. Figure 12 shows the ideal characteristics of the B-field created by the transmitting coil and its orientation with the receiving coil. The desired part of the B-field is the sum of the vectors that are normal to the plane within the enclosed area of the receiving coil.



Figure 12: Cross-section of the coils showing the direction of the magnetic flux density vector

Using equations given in Appendix 4 for the magnetic flux density, Figure 13 is given which illustrates that in the 1cm to 2cm range of coil separation, both the 2cm and 2.9cm coils provide a reasonably uniform  $B_z$ -field in the implanted coil. The 2.9cm coil however provides a lower sensitivity to horizontal misalignment. The design of this

system will assume that the location of the implanted device is well known and the external coil can be accurately positioned over the implanted coil. This implies the horizontal misalignment is not a large concern.



Figure 13: Dependence of the normal component of the magnetic flux density vector on the depth of the receiving coil plane and the radius of the transmitting coil (a) 1cm coil separation (b) 2cm coil separation

For the targeted maximum coil separation of 2cm, the B-field is stronger for the 2.9cm. However, if the separation is any smaller than 2cm, the 2cm coil quickly becomes the better choice. Therefore, the 2cm coil has been chosen as the optimal coil.

#### 5.1.4 Effect of Q on Power Reception

The Q of the resonating receiving network (L1 and parallel C in Figure 14) can help the link by amplifying the received voltage. Consider a parallel R-L network where R is the parallel loss resistance of the inductor. If a voltage is induced in the inductor via flux from another inductor, then the power induced in R is  $P=V_{ind}^2/2R$ . Consider a parallel R-L-C network where R is again the loss resistance of the inductor.



25

If the voltage in the inductor is induced from another coil at the resonant frequency of the L and C, the current through R will be Q times smaller than the current through L and C since the impedance of R is Q times larger than that of L and C, by the definition of Q. If the same flux and same inductor used in both cases, the power induced in the inductor should be the same for both cases. This requires that the voltage for the resonant case,  $V_{ind}$ , be Q times that of the voltage for the non-resonant case.

The Q can only be increased to a point. As we will see later, the inductor Q will have a maximum value of about 30 and the equivalent resistance of all of the circuitry in parallel with the inductor will decrease this value.

#### 5.1.5 Rectifier

The schematic in Figure 15 shows the basic elements involved in the power link, including the power transmitter (V1), the two inductively coupled coils (L1 and L2), the resonating capacitor (C1), the rectifier and filter (D1 and C2), and the DC load resistance ( $R_{Load}$ ).



A half-wave rectifier has been used in this analysis but other types of rectifiers (i.e. full-wave and voltage doubling) use the same analysis. If the rectifier is assumed to be ideal in the sense that there is no voltage drop across D1 and C2 is large enough to completely filter the AC component of the received signal, then the rectifier and load can be replaced by an equivalent AC load (RMS power loading),  $R_{ac}$  using (14) as seen in Figure 16.

$$R_{ac} = \frac{R_{Load}}{2} \tag{15}$$

One could consider greatly over-sizing the filtering capacitor C2 to reduce the sensitivity to intermittent power link failures. However, the capacitor size must allow for the control data to be detected for which the data rate has not been specified. (16) can be used to determine the value of C2 based on the desired rise and fall time for the control data signal.

$$C2 = I_{DC} \left(\frac{dv(t)}{dt}\right)^{-1}$$
(16)

In (16),  $I_{DC}$  is the DC load current and  $\frac{dv(t)}{dt}$  is the fastest desired change in voltage with

respect to time.

In Figure 16, the equivalent load resistance has been combined with the loss resistances of L2 giving  $R_p//R_{ac}$ .



Figure 16: Simplified circuit model of the power link

Using the circuit in Figure 16, the loaded Q of the R-L-C network on the receiving side  $(Q_{\rm rec})$  can be determined using (17)

$$Q_{rec} = \frac{\omega L2}{R_p //R_{ac}}$$
(17)

where ? is the operating frequency of the link.

#### 5.1.6 Transfer Function of the Link

In this section, a transfer function is derived to determine the required voltage on the transmitter coil. This will be used to estimate the power absorbed in the tissues.

Using Faraday's law, the voltage induced in the receiving coil due to the flux, F, from the transmitting coil is

$$v(t) = -Q_{rec} \frac{d\Phi(t)}{dt}.$$
(18)

The flux is the integral of the normal component of the flux density,  $B_z$  over the area of the 1cm coil (*A*). Assuming that  $B_z$  is constant over *A* we can write

$$\Phi(t) = B_z(t)A \tag{19}$$

As seen in Figure 13, this is an appropriate assumption for this application.

The flux density is the product of the magnetic permeability  $\mu$  and the magnetic field intensity *H*. It has been well documented that the relative permeability  $\mu_r$  of tissue is unity ([11]).

$$\Phi(t) = \mu_o H(t) A \tag{20}$$

The computation of off-axis values of H is very difficult, however since the we know that the normal component of the field is relatively constant in the effective area of the receiving coil, the on-axis value can be used. From [14], this value is

$$H(t) = I(t) \frac{r^2}{2(r^2 + h^2)^{3/2}}$$
(21)

where I(t) is the time varying current in the transmitting coil, r is the radius of the transmitting coil, and h is the axial spacing between the coils.

The other term in (20) for computing the flux is the area of the coil. The effective area of a multi-turn spiral coil can be approximated using the average radius of the coil

and multiplying by the number of turns or it can be more accurately computed using integration.

$$A_{eff} = \int_{0}^{2\pi N} \int_{0}^{r(\theta)} r dr d\theta$$
(22)

where, for a coil with a trace width W and trace spacing S

$$r(\theta) = r_i + \frac{(W+S)\theta}{2\pi}$$
(23)

and  $r_i$  is the starting radius of the innermost turn of the coil given by

$$r_i = 0.5cm - \frac{W}{2} - (W + S)N$$
(24)

Using (19)-(21), the flux density penetrating the effective area of the receiving coil can be written in terms of the current in the transmitting coils and the geometry factors of the link.

$$B_z(t) = \mu_o G(I_m \cos(\omega t))$$
<sup>(25)</sup>

where G is a factor

$$G = \frac{r^2}{2(r^2 + h^2)^{3/2}}$$
(26)

which is strictly a function of the geometry of the coils and their positions with respect to each other.

Combining (18), (20), and (25) the received voltage is

$$\left|v_{received}\right| = Q_{rec} \left|\frac{d\Phi}{dt}\right| \tag{27}$$

$$\left|v_{received}\right| = Q_{rec} \left|\frac{d}{dt} \left[A_{eff} \mu_o G I_m \cos(\omega t)\right]\right|$$
(28)

$$\left|v_{received}\right| = Q_{rec} \left|-A_{eff} \mu_o GI_m \omega \sin(\omega t)\right|$$
<sup>(29)</sup>

$$\left| v_{received} \right| = Q_{rec} A_{eff} \mu_o G I_m \omega_o \tag{30}$$

(30) is therefore the function relating the peak received (induced) voltage to the current in the transmitter coil. To convert this to a voltage transfer function, we need to relate the current in the transmitting inductor to a voltage. Assuming the transmitting coil has very high Q and the mutual inductance is negligible,

$$\left|v_{transmit}(t)\right| = L_{primary} \left|\frac{dI(t)}{dt}\right|$$
(31)

$$\left. \frac{dI(t)}{dt} \right| = I_m \omega_o \tag{32}$$

$$\frac{\left|v_{received}\left(t\right)\right|}{\left|v_{transmit}\left(t\right)\right|} = \frac{Q_{rec}A_{eff}\mu_{o}GI_{m}\omega_{o}}{L_{primary}I_{m}\omega_{o}} = \frac{Q_{rec}A_{eff}\mu_{o}G}{L_{primary}}$$
(33)

It was determined in section 5.1.2 that for a 2cm coil spacing and a 1cm diameter receiving coil, the optimal value for G is obtained with a 2cm diameter transmitting coil. Therefore, to completely maximize the voltage transfer function,  $Q_{\text{rec}}$  and  $A_{\text{eff}}$  must be maximized. However,  $Q_{\text{rec}}$  and  $A_{\text{eff}}$  are dependent variables which will be described and determined in the next section.

#### 5.1.7 Power Receiver Coil Parameters

As stated in the previous section,  $Q_{\rm rec}$  and  $A_{\rm eff}$  are dependent variables. This means that the voltage transfer across the link cannot be maximized by independently maximizing  $A_{\rm eff}$  and  $Q_{\rm rec}$ . Therefore, the product of  $Q_{\rm rec}$  and  $A_{\rm eff}$  is the quantity that must be maximized.

Table 2 provides measured data for a collection of 1cm (outside diameter) printed coils using 1oz copper (1.38mil thick) on FR4. FR4 is a material commonly used for the fabrication of printed circuit boards. While FR4 and copper are not biocompatible

materials, they are an inexpensive and easy material to use for validating the equations derived in the previous sections. Table 2 shows that with a 5mA load from a 5V regulated supply, the 10mil coil with 3 turns exhibits the best  $A_{eff}Q_{rec}$  product. The methods applied here for choosing an optimal copper coil on FR4 can also be used for other materials.

Table 2: Measured data from FR4 coils (Frequency = 27MHz, Load Current = 5mA)

Turns	L (nH)	Q <sub>coil</sub>	Q <sub>rec</sub>	$A_{eff} (mm^2)$	A <sub>eff</sub> Q <sub>rec</sub>
2	70	31	17	121	2054
3	122	35	13	161	2163
4	174	37	11	192	2069
5	248	38	8.3	213	1777
6	258	37	8	227	1825
7	282	35	7.5	234	1734

Turns	L (nH)	Q <sub>coil</sub>	Q <sub>rec</sub>	$A_{eff} (mm^2)$	$A_{eff}Q_{rec}$
2	64	33	18	104	1910
3	104	35	15	131	1932
4	138	38	13	146	1863
5	158	36	11	152	1739

Trace Width (W) = Trace Space (S) = 15 mil

Figure 17 illustrates the effects of the load current on  $Q_{\text{rec}}A_{\text{eff}}$  versus the number of turns in the receiving coil. This figure indicates that at low loading (e.g. 1mA) there is no benefit to increasing the number of turns above 4. At a normal loading value of 3mA,  $Q_{\text{rec}}A_{\text{eff}}$  is harmed by increasing the number of turns above 4 and is not any better than 3. One should also note that while 3 or 4 turns might maximize the received voltage, using 1 or 2 turns results in the lowest sensitivity to loading.



Figure 17: The effects of loading on the Q of the receiver network

#### 5.1.8 Transmit Power and Absorbed Power Calculations

This section will determine the required transmitter signal power and the amount of power absorbed in the tissues using the equations derived in the previous sections and the coil chosen to be optimal. If the load current is chosen to be at least 3mA (15mW) and not much more (assume 5mA), the 3-turn receiving coil is the optimal coil. Using the 3-turn (10mil) coil as the optimal coil, the parameters in Table 3 can be used for the power calculations.

Effective Area	$161 \times 10^{-6} \text{m}^2$
Number of turns in secondary coil	3
Primary Inductance	122nH
Q <sub>coil</sub>	35
Load Current	5mA
R <sub>ac</sub>	500?
Q <sub>rec</sub>	13

Table 3: Link parameters used for power calculations

Using (25),

$$G = \frac{(0.02)^2}{2((0.02)^2 + (0.02)^2)^{3/2}} = 8.84$$

From this information, the required  $I_m$  can be determined using (30).

$$I_m = \frac{5}{(13)(161 \times 10^{-6} m^2)(4\pi \times 10^{-7})(8.84)(2\pi)(29 \times 10^{6})} = 1.18$$

Therefore, using (31) and (32) the peak voltage required on the transmit coil is

$$|v_{primary}(t)| = L_{primary}I_m\omega_o = (120\text{nH})(1.18)(2\pi)(29\text{x}10^6) = 25.8\text{V}$$

The power dissipation in the tissues can now be found using (10). Q measurements similar to those taken for Figure 6 were taken using a single-turn 4cm coil with a 15 mil trace width. This is the power transmitter coil that was used for the prototype. The inductance of the coil is 130nH; the Q of the inductor in air is 45; and the Q of the inductor when pressed against a human head is 40. The corresponding  $R_p$  values are 1.066k? and 948? respectively. Using the  $25.8 V_{\text{peak}}$  on the transmitting coil calculated above, the resulting power dissipated in the coil in air is  $P_{air} = 312$  mW and in the presence of human tissue is  $P_{\text{tissue}} = 351 \text{mW}$ . The difference in power is 39mW which is the amount of power absorbed by the tissue as approximated using the methods described in section 4.2. The absorption limit is specified as 8mW/gram as averaged over any 1gram of tissue in the shape of a cube. The average density for the tissues involved is about 1g/cm<sup>3</sup>. The next step is to determine the amount volume that is exposed to the transmitted field. While theoretically every point in space is exposed to the field, most of the energy is concentrated in a small area near the coil as seen in Figure 18 which shows the magnitude of  $B_z$  in the plane close to the coil. As previously stated, the absorption is to be computed over any 1 gram of tissue in the shape of a cube. Using the average density, 1 gram is 1cm<sup>3</sup> which means that the exposed number of grams can be approximated by determining the number of exposed cm<sup>2</sup> on the surface of the skin. From Figure 18 it can be concluded that the area on the surface of the skin enclosed by a

circle of radius 2.5cm is exposed to most of the B-field. Figure 19 shows the location of the referenced area with respect to the transmitting coil.



Figure 19: Area exposed to the strong B-field of the transmitting coil

Therefore, the approximate exposed area is 19cm<sup>2</sup>. This implies that approximately 2mW is being dissipated in each gram of tissue in the exposed area.

### 5.2 Data Link

The data link is responsible for transmitting the digitized data that is collected from the neural sensors to a receiver on the outside of the tissues. The targeted data rate for the link is 10Mbps which is the rate assumed to be necessary to provide enough data to make the system usable for muscle stimulation applications as discussed in Chapter 1.



Figure 20: Block diagram of the data path

#### 5.2.1 Wireless Transmission Method

In previous designs of transdermal wireless data links, the data link implements what is commonly called reflected impedance modulation or load impedance modulation. This method of data transmission uses the data signal to modulate the Q of the power receiving resonant circuit. This modulation can be detected through the power transmitting coil as a modulated load impedance. Therefore, when using reflected impedance modulation the same pair of coils are conveniently used for the data and power links. To implement this type of data link, the Q of the power receiving resonant circuit must be low enough to support the bandwidth of the data signal. However, as discussed in section 5.1.2, the power receiving resonant circuit must have a high Q. The frequency of the link can not be increased to suit the data signal bandwidth since the absorption is too high for the power link to function.

#### **5.2.2 Operating Frequency**

Using the conditions stated in the introduction to this chapter, the data signal is optimally centered at 310MHz. In the band between 300MHz and 322Mz, the FCC has allocated this band to Government Exclusive, Fixed, Mobile Satellite, and Mobile. These signals should not cause enough interference to cause concern in this system.

#### 5.2.3 Modulation Scheme

The simplest modulation scheme to implement is BPSK. However, it also occupies the widest bandwidth. A larger PSK constellation could be used to decrease the bandwidth without drastically harming the bit-error-rate, but increasing the order will increase the complexity and the power consumption of the transmitter. This will be discussed in more detail in section 7.5 when addressing the factors involved in the design of the integrated version of the implant.



#### 5.2.4 Geometry

As stated in the design of the power link, the most efficient form of short-range transmission is inductive coupling. Therefore, the data link will use a geometry similar to that of the power link with planar circular loops. An implanted transmitter coil with a diameter of approximately 1cm will transmit the data signal to an external coil with a diameter that will be determined in this section.

The transmitter power is limited by the DC supply voltage and the voltage limitations of the transistors used in the chosen IC process. Given these voltage limitations of the system, the most efficient transmit coil is a 1-tun coil with maximum allowed area since it maximizes the amount of flux per volt. Ideally this single-turn loop would be used as an inductive load to a driving PA to get the greatest bandwidth out of the antenna. However, the data transmitter must use a resonant circuit so that the power signal does not couple in to its output. Care must also be taken so that the bandwidth of the L-C tank is large enough to support the necessary bandwidth.

For the data-receiving coil, a large capture area is desired to capture a large amount of flux. A large number of turns is also desired to multiply the flux/voltage conversion. However, satisfying both conditions would give a low self-resonant frequency (SRF). Figure 22 shows the pattern of the normal flux density vector at a distance of 2cm from a transmitting 1cm diameter coil. As seen in the figure, maximum flux can be captured using a coil with a radius of 1cm. Absorption of energy in the tissues may cause a distortion of the field. Therefore this is not completely valid, but it is a good starting point for determining the radius of the receiving coil.

The next parameter to be determined is the number of turns for the data receiving coil. Using 15mil printed coils with a radius of 1cm, 1 and 2 turn coils are the only options with acceptable SRFs.



2cm from a radiating 1cm diameter coil

#### 5.2.5 Transmit and Receive Power

The required transmit power for the data link can be determined by analyzing the link budget. Analyzing the link budget allows the transmit power to be computed based on the noise specifications of the link and the receiver. Therefore, the required transmit power is

$$P_{t} = Noise \ power + SNR + NF + PL$$

$$P_{t} = 10\log(kT) + 10\log(BW) + SNR + NF + PL$$
(34)

where k is Boltzmann's constant, T is temperature in Kelvin, BW is the transmission bandwidth, SNR is the required signal-to-noise ratio, NF is the noise figure, and PL is the path loss. The targeted SNR for this link is 60 dB (10 bits resolution). An appropriate assumed noise figure for the receiver is 8 dB. Thus, the required power becomes:

$$P_t = -174 \text{dBm} + 70 \text{dB} + 60 \text{dB} + 8 \text{dB} + PL$$

$$P_t = -36dBm + PL$$

Effective Area	$314 \times 10^{-6} \text{m}^2$
Number of turns in secondary coil	1
Primary Inductance	23nH
Q <sub>rec</sub>	5
G	1.43

Table 4: Data link parameters

Using (24) and the parameters in Table 4, the path loss is  $12dB^*$  giving a required transmit power of -24dBm (4µW).

<sup>&</sup>lt;sup>\*</sup> This assumes no excess loss from tissue absorption, which has been validated by experiments discussed in section 6.3

#### 5.2.6 Power-Transmitter Interference

As mentioned in Chapter 3, one of the challenges in the link design is the undesired interaction between the four coils involved in the link. This section will describe the how the power link causes interference in the data link and how it can be reduced.

It is feasible to implement the two links with a single pair of coils. For example, Figure 20 provides a schematic with the necessary filtering to implement the two external coils (power transmitter and data receiver) with a single coil. In the circuit in Figure 23, the power transmitter works by radiating a signal from L5 which is driven by the 27MHz source through L1 and C1 which resonate at 310MHz. The radiated signal is induced in L3. L2 and C2 resonate at 27MHz to block the power signal from getting into the data receiver. The data link works by radiating a signal from L4 that can be received by L5. L5 resonates with C3 at 310MHz to enhance the reception. L1 and C1 resonate at 310MHz to increase the impedance looking into the output of the PA. This is done to improve the Q of the data receiving network.



Figure 23: Circuit example using only one external coil

While the circuit in Figure 23 will theoretically work, there is not a significant benefit to combining the external coils, and the isolation between the links can be

improved by using separate coils. Even if the coils are separated, there is still a need for filtering to block the power signal from driving the LNA into compression. Figure 24 shows the schematic used for the receiver prototype that will be discussed in more detail in the next chapter. In Figure 24 as in Figure 23, C1 resonates with L2 at 27MHz to block the power signal from reaching the LNA. L3 and C3 provide an impedance transformation to increase the Q of the receiver to amplify the received voltage by increasing the resistance seen looking toward the LNA. In the case of the prototype design, a spectrum analyzer is used as the receiver which has a 50? input which must be increased to obtain a reasonable Q.



A circuit identical to the one in Figure 23 could be used for the implanted coils. However, the filtering would need to be performed with on-chip passive methods that would too greatly degrade the efficiency of the system. The interaction between these two coils is still important to consider and is discussed in the next section.

#### 5.2.7 Interaction Between the Implanted Coils

The interaction between the power receiving coil and the data transmitting coil is very large and cannot be ignored. The interaction is both in the capacitive coupling and the coupling coefficient. The capacitive coupling can be reduced by reducing the trace overlap in the coils. This will be discussed further in the next chapter. The coupling coefficient problem is not easily avoided. However, the transformer created by the two coils can be easily simulated using two-port S-parameter data from a network analyzer. This method provides very reliable results which can be used for the design of the resonant circuits on the input of the power receiver and the output of the data transmitter. When using Kapton as a substrate, prior work has exhibited coupling coefficients of about 0.5. If the coupling coefficient is well known, a simple circuit analysis method can be used with the model in Figure 25.



Figure 25: Transformer Model

### 5.2 Link Design Summary

Both of the links will transmit over a 2cm separation of tissue. When using a 1cm diameter implanted (2cm deep) power receiver, the optimal transmitter coil is a single-turn coil with a radius of 2.0cm. The use of a single turn maximizes the flux/volt ratio. When using coils printed on FR4, the optimal number of 10mil turns for the power receiver is three. The data link also optimally transmits from a 1cm diameter single turn coil to maximize the flux/volt ratio. When using coils printed on FR4, the optimal number of 15mil turns for the data receiver is one. This was determined based on the SRF of the coils.

Some of these results would be different when fabricated with methods other than printing coils on FR4. For example, if the power receiver coils are fabricated using wire or coils printed on Kapton, the Q-Area relationships are substantially different resulting in different optimal solutions. However, the design procedures presented previously in this chapter are all the same.

## **Chapter 6**

## **Theory Verification**

As stated in the objective, the feasibility of this link can be verified via the development of a board-level prototype. This chapter will explain the goals of the testing, the operation of the circuit, and then provide measurement results.

### 6.1 Goals

To validate the theory presented in the previous chapters, the following must be tested with the prototype:

- An RF signal can be received by a 1cm coil displaced from a transmitting coil by 2cm of tissue that can be rectified and regulated to provide 5V@3mA.
- A second coil sharing the same axis and spaced very near to the power receiving coil can properly transmit the data signal
- The data signal can be received by a coil that is placed within the power transmitting coil.
- Tissue absorption is not great enough to block the transmitted signals

### 6.2 Prototype Circuit Operation

A schematic of the prototype is shown in Figure 26. The operation of the circuit is as follows: an AC voltage is induced in L1 by the power transmitter. The induced voltage is then rectified and filtered by D1 and C2 and regulated by U1.



Figure 26: Schematic of prototype

C1	270pF
C2	1µF
C3	220nF
C4	2.2µF
C5	1nF
C6	12pF
L1	122nH

Table 5: Prototype Component Values

L2	27nH	
L3	23nH	
D1	HBAT-540	
R1	50k?	
R2	1k?	
R3	1k?	

U1 is a low-dropout, efficient 5V regulator that provides the voltage for powering other circuits in the system. As stated previously, the goal was to supply 15mW of power. In the prototype circuit, the 5V supply is used to power a VCO. The VCO is used to transmit the 310MHz signal for testing the functionality of the data link. U2 transmits a 310MHz signal by driving L3. The actual achievable data rate is not tested because sending a constant wave to test the attenuation against the theoretical value and having knowledge of the bandwidth capabilities of the transmitting coil is enough to establish validity.



Figure 27: Power transmitter coil and data receiver coil



Figure 28: Board-level prototype of the power receiver and the data transmitter

Figure 27 is a photograph of the 4cm power transmitter coil and the 2cm data receiver coil prototypes. Figure 28 shows a photograph of the implanted part of the prototype. In the figure labeled "top", the data transmitting coil can be seen and in the figure labeled "bottom" the power receiving coil can be seen. It is important to note that while the coils lie on top of each other, the turns do not overlap. This is done to reduce the coupling capacitance between the coils which reduces the interaction between the coils. For this prototype fabricated on 60mil FR4, it is not a major concern, but it is a design factor that would certainly need to be considered when using a thin film substrate such as Kapton.

### 6.3 Measured Results

To test the prototype, a high-voltage RF transmitter was used to provide the power transmit signal and a spectrum analyzer was used to detect the transmitted data signal. The board made for the external system seen in Figure 27 was placed 2cm away from the prototype of the implanted part shown in Figure 28. Measurements were then taken to determine the required transmit power for the power signal. This value and the theoretical value are compared in Table 6. The signal power transmitted from the data transmitter was then measured as well as the received power. These values and the theoretical values are provided in the table below.

Table 6: Signal strengths measured on prototype system

	Measured	Calculated
Power Link		
Power transmitter voltage:	30V	26V
Power transmitter current:	1.26A	
On-axis magnetic flux dens	$14 \times 10^{-6} \text{Wb/m}^2$	
in the power receiver coil		
<u>Data Link</u>		
Transmitter voltage:	15mV	20mV
Receiver voltage:	3.2mV	5mV
Attenuation:	-13.5dB	-12dB

The amplitude of the data transmitter signal cannot be controlled making it difficult to analyze the SNR of the data link. However, the 15mV measured on the data transmitter coil is coincidentally close to the desired 20mV amplitude. This means that the SNR can be approximately validated. More importantly, the path loss can be validated since it is independent of the signal amplitudes.

The effect of the presence of tissue between the coils was then tested. This was done by placing a hand between the coils. As expected, the power signal was not attenuated. Attenuation in the data link was expected but was not observed. Figure 29 shows the measured spectrum of the received data signal. The SNR observed in Figure 29 is much lower than the desired 60dB SNR. This is because the 50dB noise figure of the spectrum analyzer used to produce this figure is much higher than the assumed 8dB noise figure for the link. Therefore, using a 10MHz bandwidth, the actual noise floor is approximately 40dB lower than what is observed in Figure 29. This results in a SNR of approximately 56dB.

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## **Chapter 7**

## **Proposed IC Implementation**

This chapter will describe some of the necessary attributes of a fully integrated version of the implanted part of the link. The IC design must be very robust since if the chip is damaged or fails, a skull must be opened to replace it.

The integrated version will include the following sections: Rectifier, Regulator, RF Limiter, Power Level Detector, ASK Demodulator, BPSK Modulator, 310MHz Up-Converter, and a 310MHz Power Amplifier. As mentioned in Chapter 3, the coils used for receiving the power signal and transmitting the data signal will not be put on the IC due to their size and Q constraints.

### 7.1 Rectifier (RF/DC Converter)

Many innovative integrated rectifier solutions have been explored. However, the research shows that they would not function efficiently at 30MHz as is needed for this implementation [9]. [9] also indicates that simpler rectifiers such as a single diode half-wave rectifier or a two-diode voltage-doubling rectifier will most likely have the best high frequency performance.



Figure 30: (a) half-wave rectifier (b) full-wave voltage doubling rectifier (c) full-wave bridge rectifier (d) half-wave voltage-doubling rectifier

All of the rectifiers in Figure 30 are drawn so that the p-doped half of the diodes are connected to the lowest potential in the circuit. This makes it possible to use the substrate as the p-doped half of the diode (if a p-substrate process is used).

The best solution is the full-wave voltage-doubling rectifier. This rectifier obviously has the benefit of doubling which decreases the requirements on the power transmitter. The use of a full-wave bridge rectifier would obviously decrease the peak power (and peak current through the diodes) and decreases the total area needed for the filtering capacitor. However, with four diodes instead of one or two, the efficiency is degraded due to power loss in the diodes and leakage to the substrate. [9] provides an analysis of current capacity and substrate leakage current vs. forward bias voltage for diode-connected BJTs. These are parameters that are needed in determining how efficiently the rectifier will operate.

### 7.2 5.0V Regulator

The regulator will regulate the rectified and filtered received signal to 5.0V DC. It needs to provide 15mW of power to the prosthetic as well as provide power to the data transmitter.

The input of the regulator may be exposed to large voltage amplitudes and therefore must be designed to withstand voltages much higher than the regulated output. The input voltage will be limited using the RF limiter that will be discussed in section 7.4. However, depending on the quality of the RF limiter, the regulator might still be exposed to short periods of higher voltages.

For efficiency purposes, the regulator should be able to operate with input voltages very close to that of the output. This decreases the necessary input voltage and decreases the voltage drop across the pass transistor in the regulator, effectively decreasing the power loss in the regulator.

The regulator may needs to have a high power supply rejection ratio PSRR. This is necessary to reduce the amount of filtering needed in the rectifier. As mentioned in Chapter 5, a lack of filtering is also necessary to allow the ASK data signal through the rectifier. The large PSRR spec will allow large AC signals to be present on the input of the regulator without degrading quality of the regulated output.

Another common component of a regulator is a bandgap reference. A bandgap reference is not necessary in this design since the circuit is in a very temperature-stable environment.

Some regulator circuits for implantable devices can be found in [8] and [15].

50

### 7.3 ASK Data Demodulator

The ASK demodulator will be used to detect the control data that modulates the amplitude of the 27MHz power signal. It is dangerous to put the amplitude detector on the front side of the rectifier because it would then need to be able to withstand strong negative voltages. An ASK demodulator circuit for implantable devices is presented in [8].

### 7.4 RF Limiter and Signal Strength Feedback

The amplitude of the received signal will be greater than 5.0V so that a regulator can be used to provide a stable 5.0V supply. To protect the circuitry on the chip (primarily in the regulator), an RF limiter should be implemented. This will basically use a power level detector and pull current from the supply reducing the Q of the circuit and therefore reducing the voltage on the coil. A standard loading limiter is presented in [8]. This however will cause power to be wasted and dissipated in the form of heat and therefore can only be done for a limited period of time. To avoid this problem, information must be fed back to the transmitter to tell it to turn down the amplitude. The power level detector and feedback also act to protect from too little power.

### 7.5 Data Modulator, Up-Converter, and Transmitter

The implanted part of the data link includes the modulator, up-converter, and transmitter. When considering power consumption, circuit complexity, and chip real estate BPSK is the optimal modulation method. The BPSK modulation can be easily performed on the base-band signal before up-conversion.

The up-conversion can be performed using a standard analog mixer circuit. The complication in the up-conversion is in the generation of a local-oscillator signal. In an implanted design, there is no room for a crystal resonator for generating a stable reference signal. However, this device is in a very temperature-stable environment. This implies that once the reference frequency is tuned to the correct frequency, the frequency should remain stable. This type of tuning can be performed via digital-control using the control data from the external system.

The transmitter needs to transmit a -21dBm signal from the data transmitter coil. This power level is very small when compared to the total power consumption of the implanted device and therefore the efficiency is not of paramount importance. Thus a simple linear amplifier will suffice.

## **Chapter 8**

## **Conclusions & Future Work**

### 8.1 Conclusions

The theory and results documented in this thesis have shown that it is feasible to create a wireless 15mW power link and 10Mbps data link across a 2cm thickness of human tissue using implanted coils no larger than 1cm diameter. Design methods have been documented in this thesis that can be applied to several different methods of fabrication (e.g. printed coils, wire coils).

### 8.2 Future Work

The primary focus of future work should be to design the integrated system described in Chapter 7. This thesis has shown that the link will work. A logical next step in the testing of the theory would be to design and fabricate an integrated version to test the integrated rectifier and data link.

Another area of research that needs more attention is testing the coils in the environment where they are intended to be used (i.e. sandwiched between real tissues). A related topic is the concern of biocompatibility. While Kapton is a biocompatible material, the copper that is printed on the Kapton is not. A more suitable metal for implantation may be Platinum ([16]). Coils have been implanted before in human subjects. Therefore, future investigation on this topic should include the investigation of commercially available products and possibly design around this coil. In addition,

methods of bonding the coil to the IC and packaging of the entire implant should be investigated.

Finally, the original 15mW power specification needs to be validated through an IC-level circuit design with the prosthesis. Decreasing this power to 5mW would be helpful to reduce the localized heating of the tissues from the implant's power dissipation.

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## **APPENDIX 1**

### **Self-Inductance Calculations**

Equation (35) is intended for "thick coils: coils approximately pancake form" ([17]). It has been verified that this equation works well for the type of coils used in this project.

$$L = \frac{N^2 a_{cm} FP}{1000}$$
(35)  
$$P = 4\pi \left[ 0.5 \left\{ 1 + \frac{1}{6} (x)^2 \right\} \ln \frac{8}{x^2} - 0.84834 + 0.2041x^2 \right]$$
$$x = \frac{c}{2a}$$

Figure 31: Coil dimensions for computing the inductance of a coil

F is a parameter dependent on b/c and c/2a found in [17]

## **APPENDIX 2**

### **Mutual-Inductance Calculations**

(36) from [18] is the fundamental integral for computing the mutual inductance between two conductors.

$$M = \frac{\mu_o}{4\pi} \oint \oint \frac{dl_1 \cdot dl_2}{r_{12}}$$
(36)

When applying (36) to two filament loops of radius A and a and separation distance h, the equation reduces to (37) ([19]).

$$M_{12} = \mu_o \sqrt{Aa} \frac{2}{k} \left[ \left( 1 - \frac{k^2}{2} \right) K(k) - E(k) \right]$$
(37)  
$$k = \sqrt{\frac{4Aa}{(A+a)^2 + h^2}}$$

K(k) = elliptic integral of the first kind

E(k) = elliptic integral of the second kind

(37) can also be applied using (38) which uses data provided in Figure 32 from [17].

$$M = F\sqrt{Aa} \text{ uH}$$
(37)

*F* is a parameter dependent on r2/r1 which can be found in table form in [20]. The numbers in the table are closely approximated by the following equations:

$$F = -0.012\ln(r2/r1) - 0.006 \quad [0.1 \le r2/r1 \le 0.3]$$
(38a)

$$F = 0.0288x^{4} - 0.0939x^{3} + 0.1266x^{2} - 0.0875x + 0.026 \quad [0.3 \le r2/r1 \le 0.998]$$
(38b)



(39)



# **APPENDIX 3**

### **Tissue Properties**

	Density	Average Thickness	
	Kg/m <sup>3</sup>	mm	
Skin	1120	1	
Fat	940	0.7-1.5	
Bone (Cranium)	1610	2-4	
Dura (Grey matter)	1039	0.5	
CSF	1007	2	
Brain (Grey)	1039		
Density [21], Thickness [22]		•	

Table 7: Tissue density and thickness for a human head



Figure 33: (a) Relative Permittivity and (b) Electrical Conductivity ([23])

## **APPENDIX 4**

### **Magnetic Flux Density Calculations**

[24] provides (38) which is used to determine the magnitude and direction of the magnetic flux density vector due to a current I flowing in a filament loop.

$$\vec{B}_{\phi} = 0$$
(40)  
$$B_{\rho} = \frac{\mu NI}{2\pi} \frac{z}{\rho \sqrt{(r+\rho)^{2} + z^{2}}} \left( -K(k) + \frac{r^{2} + \rho^{2} + z^{2}}{(r-\rho)^{2} + z^{2}} E(k) \right)$$
$$B_{z} = \frac{\mu NI}{2\pi} \frac{1}{\sqrt{(r+\rho)^{2} + z^{2}}} \left( K(k) + \frac{r^{2} - \rho^{2} - z^{2}}{(r-\rho)^{2} + z^{2}} E(k) \right)$$
$$k = \sqrt{\frac{4r\rho}{(r+\rho)^{2} + z^{2}}}$$

K(k) = elliptic integral of the first kind E(k) = elliptic integral of the second kind r = radius of the coil ? = ?-dimension (perpendicular distance from the z-axis) z = z-dimension (coil axis)