RF Inductive Link Data and Power Communication for Medical Implants

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Declaration

I certify that this research work titled *"RF Inductive Link Data and Power Communication for Medical Implants"* is my own work. The work has not been presented elsewhere for assessment. The material that has been used from other sources it has been properly acknowledged / referred.

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Abstract

To improve the quality of life of a patient, implantable stimulator systems such as cardiac pacemakers, defibrillation systems, implants based on orthopedic instruments, cochlear implants, implants used for stimulating deeper brain and spinal cord regions are widely used by clinicians in medical practices. There are two major challenges in the development of these implantable microelectronic devices i.e. size and consumption of power because the procedure of implantation to the patient should be minimally invasive. The optimal choice for these implants is not the utilization of batteries due to limited battery time, large size, and leakage current that can damage the body tissues. The long-term implantation of these devices in the body needs an external transcutaneous wireless connection that charges the internal battery to be utilized as a power supply.

This thesis presents an inductive link power and data communication for medical implantable devices. The initial and final designs of the project are presented and efficiently implemented on a PCB sheet. The power communication is tested in different media such as air, polyurethane (skin) and wood (bone). The results show that power is efficiently transmitted from the primary to secondary side without any power loss. Similarly, data communication is also tested in these three-media using UART. The accuracy is 100 percent on 100 bytes' transmission but on transmitting 1000 bytes 99.9 % results are achieved. Additionally, a comparison of the data transmission in these media is also presented.

Key Words: Medical implants, PCB sheet, Air, Polyurethane, Wood, UART.

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CHAPTER 1: INTRODUCTION

1. Overview

To improve the quality of life of a patient, implantable stimulator systems such as cardiac pacemakers, defibrillation systems, implants based on orthopedic instruments, cochlear implants, implants used for stimulating deeper brain and spinal cord regions are widely used by clinicians in medical practices [1]. A broader research is in progress for developing new instruments of several application areas such as retinal implants, devices for recording internal body activities, and controllers for neuromuscular stimulations [2]. There are two major challenges in the development of these implantable microelectronic devices i.e. size and consumption of power because the procedure of implantation to the patient should be minimally invasive. The optimal choice for these implants is not the utilization of batteries due to limited battery time, large size, and leakage current that can damage the body tissues. The long-term implantation of these devices in the body needs an external transcutaneous wireless connection that charges the internal battery to be utilized as a power supply [3]. Most of the circuits for biomedical implants require wireless connection for their operation. An inductive coupling is used for powering many of these devices within the range of centimeters on the surface of skin. A few implants specifically interfaced with Central Nervous System (CNS), such as retinal and cochlear implants, require huge amount simultaneous data interface with larger neuronal activity through many channels [4]. Thus, a receiver circuit with higher data rate is required for establishing an effective wireless connection between the external units and the implantable device [5].

However, there are two ways of making communication with implantable devices i.e. wired connection and wireless link. There are several disadvantages of a wire connection i.e. they can be broken, infected or noise can be introduced by them in the recordings by movements and artifacts or by effects of antenna. It has also been reported that stimulation of deep brain regions, cardiac pacemakers, and defibrillation systems face many complications [6]. Therefore, to avoid such complications, wireless communication techniques have been introduced in many medical implants [7]. Although, significant amount power is required for wireless communication and suffered from weak data transmission across the biological tissues. A larger antenna is usually required for wireless communication that limits how smaller the implant might be and prevents embedding in heart, spinal cord and brain deprived of producing appropriate damage in the tissues.

Additional approaches of communication wirelessly are examined for communicating with implantable devices such as optics [8] and ultrasonic [9]. But, the transmission efficiency of these methods is very low in the body and their miniaturization would be difficult.

1.1. Problem Statement

Implantable medical devices need bidirectional data transmission and their processor is powered wirelessly. There are several processors that are not used in medical implants. For example, ASIC is specifically used in integrated circuits (IC) and designed for specific applications. Additionally, ASIC is expensive and not readily available in the market. Similarly, commercially available microcontrollers are not used in medical implantable devices due to high power consumption, unwanted general purpose inputs/outputs (GPIOs), inappropriate serial peripheral interface bus (SPI) and $\mathrm{I}^2\mathrm{C}$ protocol for communication. Additionally, the patient's safety and comfort should be kept in mind while implanting a medical device in the body.

1.2. Significance of Research

The significance of data and power transfer is found in studying the active implantable devices, data link where required, and low power constraints applied on these devices.

1.3. Aims and Objectives

- To design, fabricate and test a multifunctional device on different medical implants;
- Power transmission through inductive link run by microcontroller in passive side.
- Data transmission through inductive link in simplex (sending data to passive one), half duplex (data sending and receiving only one at a time), and full duplex (simultaneous sending and receiving of data) manners.

1.4. Thesis Outline

Chapter 2 gives the literature review of the communication in medical implants, design consideration medical implants transceivers, existing wireless medical implantable devices, regulating authorities, standards, and increased miniaturization of medical implants.

Chapter 3 covers the entire methodology starting from the initial design, materials utilized in final design, improvements made in initial design, and other important things required for implementing the project.

Chapter 4 discusses the results and analysis of the wireless power and data transfer in different media such as air, polyurethane (skin) and wood (bone).

Chapter 5 provides conclusion of thesis and future recommendations for getting more better results of project.

CHAPTER 2: LITERATURE REVIEW

2.1. History of Communication in Medical Implants

The RF communication in medical implants uses lower frequency near the magnetic coupling of field. It is possible to couple the signal in the region of its operation i.e. 125 kHz across a 1mm titanium housing used in many implants. It should be noted that the skin depth in titanium is 900 µm at the operating frequency of 125 kHz. The power might also be transferred in systems having shorter requirements range such as cochlear implants. There is no encapsulation of such systems in metal and the frequency range of communication might be at higher frequencies of many M Hz. A close coupling is needed between the implantable device and programmer in near filed systems.

During the mid of 1990s, the largest medical implantable devices manufacture known as Medtronic petitioned the United States federal communication commission (FFC) for communication in medical implants. A communication spectrum was dedicated for medical implantable devices [10]. In 1998, ITU-R Recommendation SA1346 recommended medical implant communication service (MICS) band of frequency range between 402 to 405 MHz. Also, during 1999, a band of similar standards was developed in Europe. The allocated band supported the utilization of relatively wireless links with longer range and greater speed [11].

In March 2009, the development of finalized MICS band was announced by FCC after establishing the service of MedRadio under the 95^{th} part of the rules of commission. It was basically followed by the approval of European regulatory authorities during 2007 [12]. The currently available services of MICS core band at frequency range between 402 to 405 MHz were incorporated by this new service including the addition of 2 MHz in the adjacent wing band of spectrum at 401 to 402 MHz and 405 to 406 MHz. The MICS core band can be utilized for completely implantable devices whereas the MICS wind band are utilized for both the body worn devices and implants. The channels of MICS band contain a maximum emission bandwidth of 300 kHz whereas the channel frequency of wind band is about 100 kHz. Both the devices should be prescribed to the patient by an authorized medical doctor [13].

Several limitations of dated inductive frameworks were overcome by the band of MedRadio and facilitated the next generation medical implants with improved patient's care delivery. It is particularly significant as raising wellbeing costs drive appropriation of locally situated monitoring of patients. Figure 1 shows the health benefits of this novel technology.

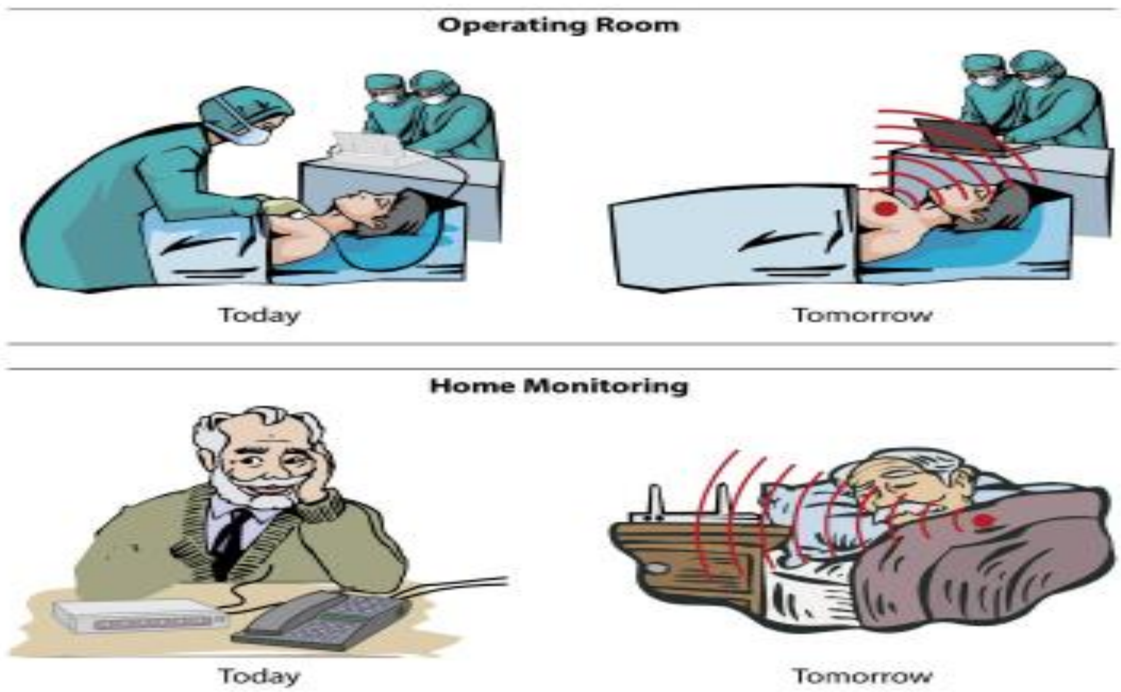


Figure 1. MedRadio band technology benefits in Patient's healthcare [13]

Because of characteristics of signal propagation in the patient's body, compatible with incumbent users, and its availability across the globe, the band between 401 to 406 MHz is well suitable for this service. It should be noted that the greater attenuation of body suffers high frequencies that is up to some extent compensated by the improved gain of antenna. Additionally, some other systems use the industrial scientific, and medical bands (ISM) and are not suitable for medical data communication.

2.2. Design Considerations of Transceiver Design for Medical Implants

Following are the basic requirements that challenge the design of transceiver of a medical implantable device:

- There is requirement of lower power during communication at the frequency of 400 MHz. The power of implant battery is reduced and has a higher battery impedance. Because of this factor the peak currents are limited usually drained from the battery. The current is kept less than 6 mA for most of the implants during the process of communication.

- There is requirement of lower power during sleep and periodically sniffing or observing for a wakeup signal.
- The important factors are the utilizing of minimum external components and smaller physical size. A pacemaker's radiofrequency (RF) module should a size up to approximately $3\times5\times10$ mm^3 to fit in usual cans of pacemaker. Moreover, components of implant grade are very expansive and the cost might be reduced by sophisticated components integration. This components integration also increases the whole reliability of the system.
- There is demand of reasonable data rates. In pacemaker applications, the current demand of data rate is greater than 20 kbps for future projection of the device.
- The system should have higher reliability of both the system and data.
- There is need of selectivity and interference rejection from European TETRA radio waves.
- Usually, the MICS band of greater than 2 m range is considered better on the shorter inductive link. There is requirement of better sensitivity over longer ranges because smaller antennas and losses in the body have influence on link budget and allowable range. The overall losses including antenna, impedance matching, fading and losses in the body are much higher up to 40 to 45 dB.

There are two types of medical devices i.e. first, those devices having a non-rechargeable battery located inside the body such as pacemakers, and second, those devices having inductive coupling of power such as cochlear implant. The previous intensely duty cycle the operation of frameworks to ration control. The transceiver is off often, accordingly the off-state current and the current required to occasionally search for a conveying gadget must be to a great degree low between 1 to 2 µA. In both cases, the power should be less than 6 mA for transmission process is likewise needed.

Minimally, the whole consumption of power is defined by Joules/Bits, and recommended that transceivers in implants must utilize the greatest expected data rate, which satisfies the sensitivity requirements of application receiver. The lower data rate based system should utilize data buffering and work at the greatest possible data rate and exploit the duty cycle of the states of power for reducing the consumption of average current and timing window for interference. It should be noted that the only possible condition is when there is availability of sensible link budget conditions. The maximum utilizable data rate is usually limited by poor antennas and body losses.

2.3. Examples of Existing Wireless Medical Implants

This section gives an overview of the commercially available medical implants that meet the regulatory and standards requirements of the system. From literature, most of the systems are observed to be incomplete in their design and are not meeting the medical standards due to lack of major components utilized in the design.

2.3.1. MICS band Transceiver (ZL70103)

The ZL70103 is a RF transceiver that consumes very low power and a RF link with a very high data rate is supported for communication applications in medical implants. The unique design of chip permits health of a patient and performance data of device to be transmitted in a quick manner having smaller effect to the useful battery life of the implant. This chip uses the MICS band spectrum of 402 to 405 MHz and is designed for both the medical implants and base stations.

The flexibility of chip is very high and can support many options of wakeup with greater power consumption. By utilizing the ISM band wakeup option of receiver operated at 2.45 GHz, an ultra-lower power consuming operation is achieved. There is also a high-level media access controller (MAC) integrated in the system that encodes and decodes the RF messages, forward error correction (FER) and cyclic redundancy check error (CRC) for getting a link with greater reliability. The application can have an easy access through a standard serial peripheral interface (SPI) [14]. The general block diagram of ZL70103 wireless transceiver is shown in Figure 2.

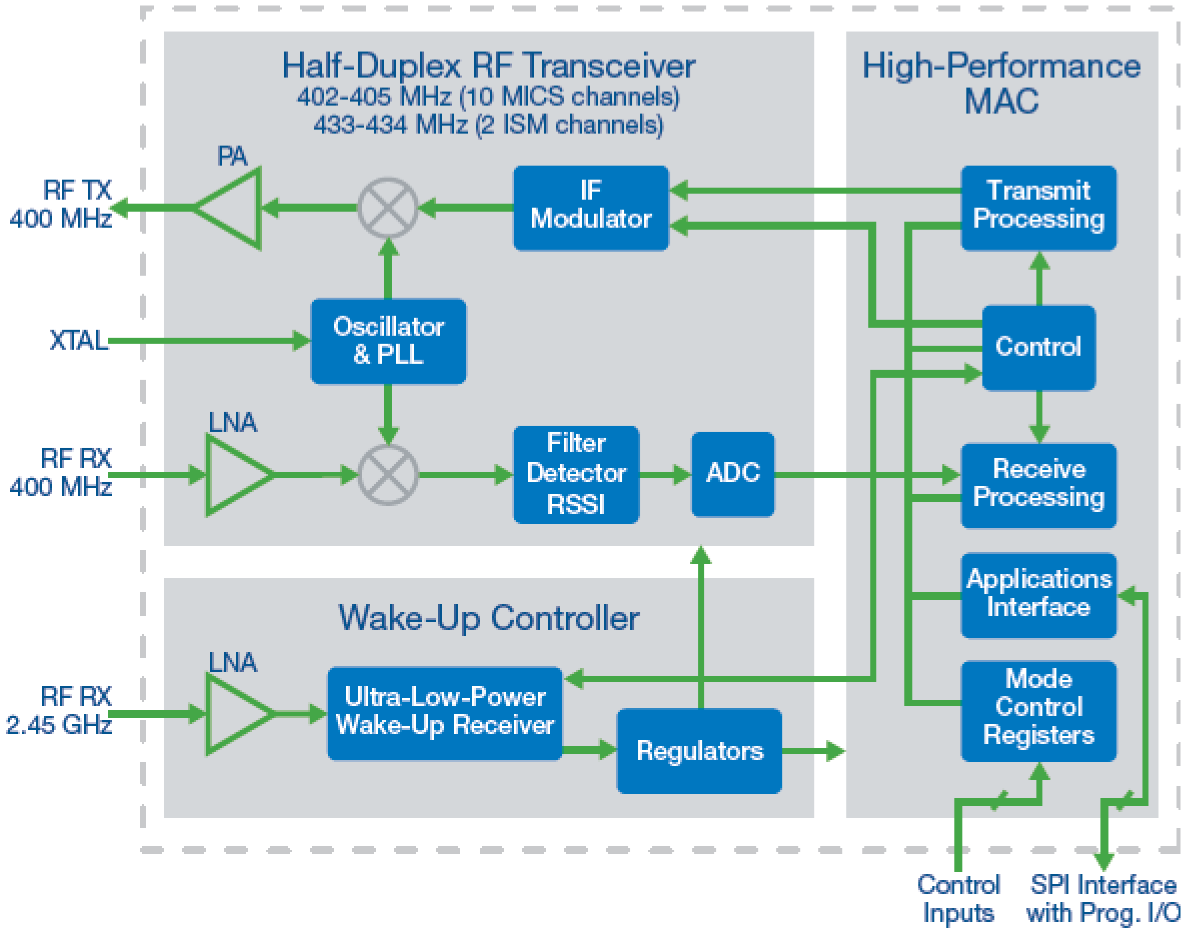


Figure 2. Block Diagram of ZL70103 Wireless Transceiver [14]

As mentioned before that the ZL70103 works in the MICS band spectrum of 402 to 405 MHz for enabling the wireless links to achieve communication between the control instrument and the medical implant. The device previous versions were using the inductive links for communication with the implantable systems. The data rates of such systems were very low and their range of operations were also limited. Additionally, these systems were very complex, difficult to use, and needed the accurate positioning of patient with the inductive wand over the implant.

In contrast, the higher data rate and long range operation of ZL70103 permit the manufacturers of medical devices for enabling novel value-added facilities to improve the care of patient. Through higher data rate support, events of patient can be stored in the memory of device and uploaded in a quick manner to the base station for analyzing it in a short duration of time. In sterile surgical situations, the longer operating range permits the base station to be kept outside this environment.

This usually simplifies the use of device in remote monitoring applications. As outlined, an ultra-low-power RF handset in a pacemaker can remotely transmit persistent occasion and gadget execution information to a base station in the home. Information is then sent via phone or web to a doctor's office. If an issue is identified, the patient goes to the clinic where the fast, two-way RF connection is utilized to screen and change gadget execution under a specialist's supervision.

There are two new data rates offered by ZL70103 i.e. 18.18 and 40 k-bits/ sec to be used in wide range of applications along with data rates of 200 and 400 k-bits/sec utilized in the previous version of device ZL70103 chip. This provides an increased link budgets for cases with greater difficulties such as deep implantable devices or long-range applications.

In medical implantable devices, the critical parameter that measures the performance of a device is its batter life. The ZL70103 handset consolidates a 2.45 GHz wake-up receiver that enables the chip to work with a normal current of 290 nA while sniffing once per second. Correspondence amongst embedded and base station handsets is then started utilizing a uniquely coded wake-up flag from the 2.45 GHz base transmitter. Elective wake-up systems utilizing 400 MHz or direct wake-up by the embedded medicinal gadget are additionally upheld.

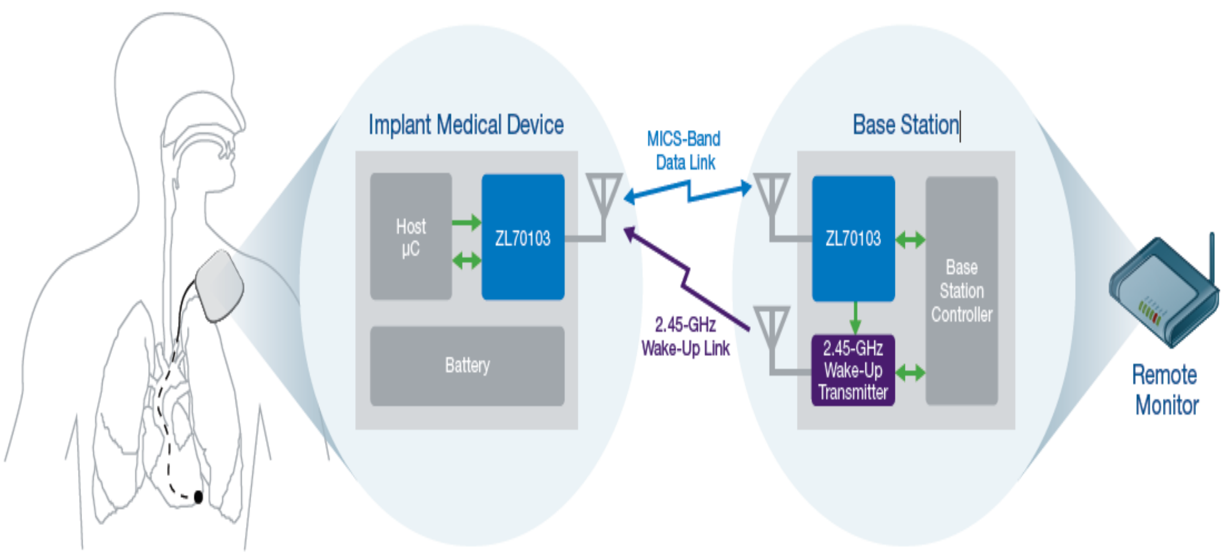


Figure 3. Uses of ZL70103 Wireless Transceiver [14]

2.3.2. Imaging Pill Camera Transceiver

An imaging pill camera transceiver is a low power device that can transmit data at a rate of 2.7 Mbps and 5.2 mW power consumption. The frequency band spectrum for operation of this device is 400 to 440 MHz. The transmission ability of this device is minimally 2 video-images/ second for about 8 hours due to transverse manner of it in the gastrointestinal tract (GI). This device has a simple architecture mainly composed of a voltage controlled oscillator (VCO) with a small coiled antenna. Obviously, the application of this device uses very small components and the constraints applied on its size are very tight. An ultra-lower power camera pill transceiver is shown in Figure 4.

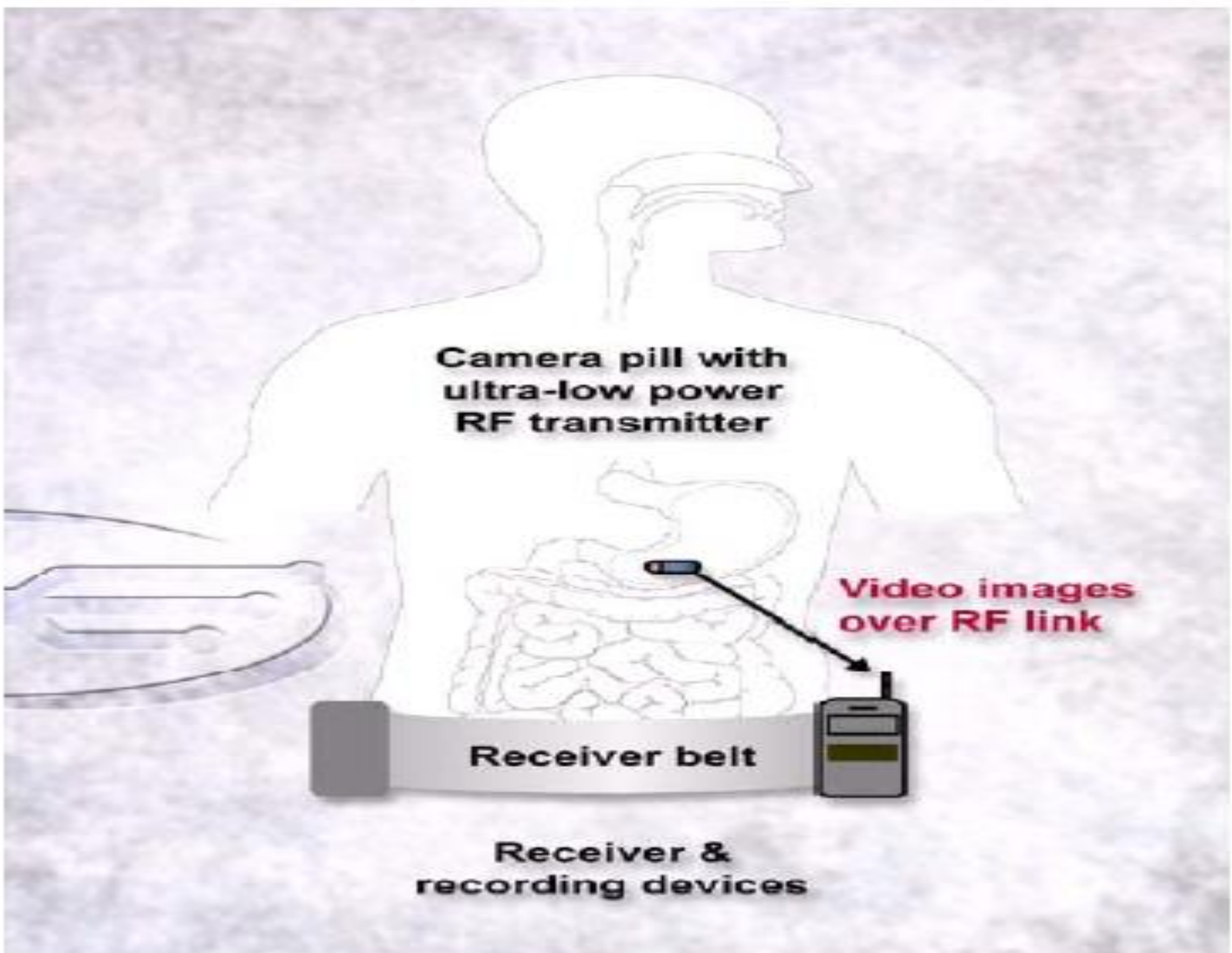


Figure 4. An Ultra-Low Power Camera Pill Transceiver Utilized in the Diagnosis of Gastrointestinal Tract [13]

2.4. Regulations for Wireless Medical Implants

There are various developments to be occurred in the United States regulatory bodies that have effect on medical devices. In 2009, MedRadio service was approved by FCC [12]. Additionally, there are two active proceedings of notices proposed rulemaking (NPRM). First, NPRM 09-20 presented the feasibility comments that allowed a spectrum of 24 MHz in 413 to 457 MHz band to be utilized as a secondary option under the MedRadio service in 95th part of the commission rules. The FCC was petitioned by the Alfred Mann Foundation for utilization of this spectrum in wideband smaller power mesh of medical devices. The uses of this application included electrical stimulation in artificially neural system for sensory restoration, movement, and rehabilitation of paralyzed organs and limbs. Additionally, a frequency of 5 MHz bandwidth emission would be need for the reliable working of these medical micropower networks [15]. Second, NPRM 09-59 commented on the spectrum allocation and established technical rules and services for operating medical body area network (MBAN) frameworks utilizing body sensor instruments [16]. Similarly, during the beginning of 2011, the Aerospace & Flight Test Radio Coordinating Council (AFTRCC), Philips Healthcare, and GE Healthcare showed a proposal to the FCC with an aim of resolving sharing of thorny spectrum issues and permit MBANs instrument to work on secondary basis in 2360 to 2390 megahertz and 2390 to2400 MHz bands. It was the best decision for communication in externally used medical devices [17].

In Europe, the Task group 30 of European Telecommunications Standards Institute (ETSI) is developed for making the standards for medical devices. This group currently works on the development of a draft standard (EN301-559) to operate power efficient medical implantable device communication within the range of frequencies between 2483.5 to 2500 MHz having a power of about 10 mW, 9 minimum channels and a 1.833 MHz bandwidth of maximum emission. The major issue in this band selection is its comparison with the MedRadio Band of frequency range 401 to 406 MHz cannot be considered as global allocation of band. However, in Europe, it will offer a spectrum with greater usability for medical devices and additional areas that select to obey the standards of ETSI. MedRadio band protection from expected interfering sources needs regulatory vigilance of continuous nature. For instance, communications of power line might have interference of harmonics and need significant guidelines for regulation to make sure the operation reliability both in home and at clinic. Additionally, the most necessary thing is that devices made by MedRadio should be expelled from the process of consideration because of globally harmonized short range devices (SRD) and there is great difficulty in maintaining a relative free surrounding from interference under the SRD category [13].

2.5. Standards for Wireless Medical Implants

The Task group 6 (BAN) is assigned under the section of IEEE 802.15 to develop a standard of communication with power efficient and workable device optimization either in or outside the human body for serving a huge amount of applications such as clinical, personal entertainment or consumer electronics. In 2007, the foundation of IEEE 802.15 TG6 was made and worked as TG6 in Taipei during the month of January in 2008. The process of transmission and reception must be supported a compliant device in on the shown below bands of frequency; 402 to 405 MHz, 420 to 450 MHz, 863 to 870 MHz, 902 to 928 MHz, 950 to 956 MHz, 2360 to 2400.5 MHz and 2400 to 2483.5 MHz [18].

There is great success obtained by the 802.15.6 standard for medical devices used externally such as blood oximeter device particularly in the environment of hospitals. The target of this standard is especially for the data rates, service quality and networking required by such applications and interoperability benefits along with cost reduction will provide facility to the standard adoption. The deployment of IEEE 802.15.6 standard is an open question for medical implantable devices. Many obstacles are faced by this standard introduction in medical implantable devices that are not manifesting as stronger candidates for external use. The major objective of medical standards development is to facilitate interoperability, many production companies and cost reduction of the systems. While reducing the cost of systems are very important in medical implantable devices and customized needs still continuously driving the market. Recently, manufacturers setup networks of communication and systems of remote monitoring that are not specifically designed for achieving interoperability. The available devices and networks utilize modified formats of modulation and protocols and the compatibility is backward with such an infrastructure and the currently available devices are seldom significant [18].

Lastly, the competition among manufacturers of device results into the emerging needs. Whereas. IEEE 802.15.6 standard provides the potential for significant performance in few applications the drive towards additional improvement in performance, such as miniaturization creates difficulty in the application of standard. Although, there is much feasibility in the evolution of standard to cope with demands of the market. Eventually, market will decide the overall success of the standard and the medical implant devices.

2.6. Increased Miniaturization in Wireless Medical Implants

The currently available RF transceiver technology is permitting the remote monitoring development and communication of longer ranges for existing medical implantable devices. A new variety of medical devices could be facilitated by the future technology that are smaller in size with cost-effectiveness and least invasiveness. Figure 5 shows an example of such medical devices [19]. Medtronic presented the miniature concept of a pacemaker that would eradicate the utilization of leads that usually result in complications in their failure. Further innovations are still required by the system particularly in power supplies that is major need in all efforts towards miniaturizing devices.

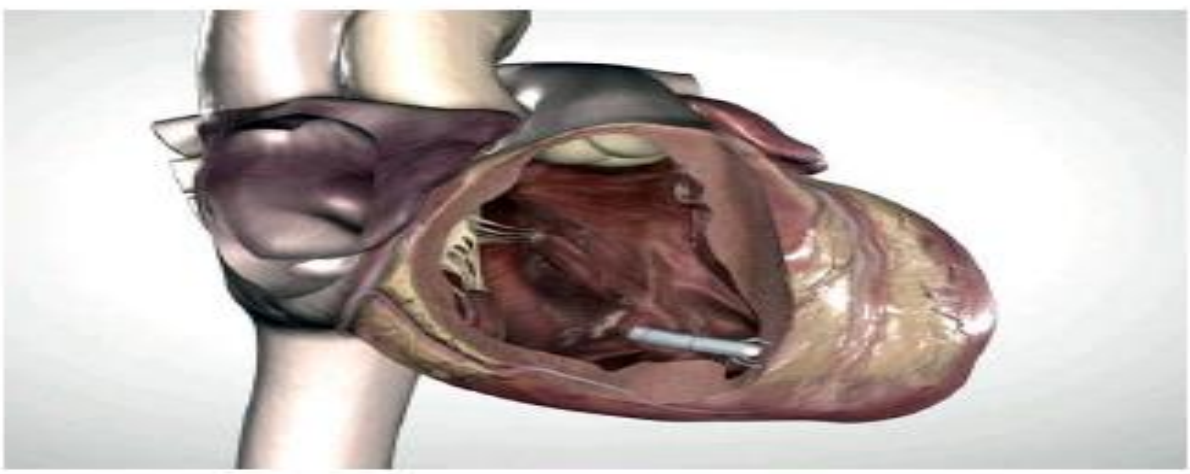


Figure 5. Example of Miniatured Pacemaker Concept Proposed by Medtronic [19]

The miniaturization of devices enables completely a novel application. For example, a cubic millimeter size device was presented to record the pressure of interocular area in the cornea of eye. A frequency shift keying (FSK) scheme of transmission was utilized with a capability of transmitting 2 tones over a larger bandwidth of 570 to 690 MHz. The transmitter contained LC tanks connected in series that enabled the larger separation of frequency than a single tank transmitter thereby requirements of phase noise relaxation and area reduction comparable with two isolated LC tanks. A thin film lithium battery of capacity 1 μAh was utilized by the system from

Cymbet. Its life is about 28 days without having energy loss. The lifetime of the system is extended by using power from cells and battery recharging [20].

Innovation in techniques of circuits and architecture can be utilized for power reduction and thus the needs of battery size. For example, a new transmitter based on MICS band with energy efficiency of 450pJ/bit and power consumption of less than 100 uW. Because it was energy efficient system, the range of the device might be sacrificed because of lower output power. Thus, the miniaturization challenge is to retain useful range i.e. distance between transmitter and implantable device provided limited abilities of power and body attenuation effects.

Miniaturizing devices will require additionally progresses in low power plan methods, framework advancement, packaging, new battery as well as power storing approaches and watchful determination of the correspondence groups to use. A multilayered all-encompassing methodology is expected to push the limits of scaling down. Nonetheless, if these difficulties are met the open doors for propelling human services later seem exceptionally encouraging.

This thesis proposes an RF inductive link for both power and data transfer. The design of the circuit, coils, and other requirements are discussed in detail. The implementation of both the primary and secondary coils and circuits are implemented and discussed. The testing of circuits for different mediums such air, polyurethane (skin) and wood (bone) are done and the results are compared. Additionally, hardware results are also presented and future recommendations are given for improving the design of the entire system.

CHAPTER 3: DESIGN & METHODOLOGY

3.1. Materials

3.1.1. DC Power Supply

The Block diagram of a general DC power supply is given in Figure 6.

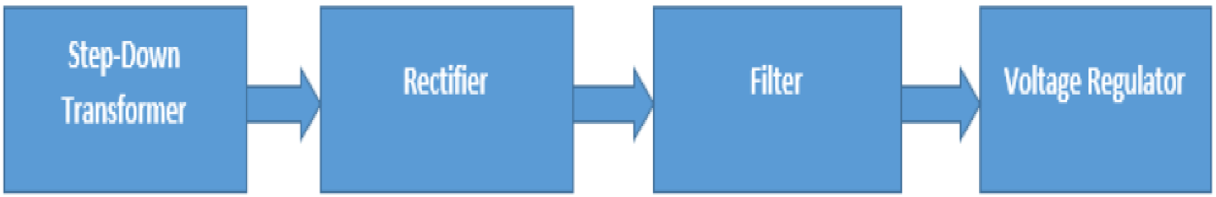


Figure 6. Block Diagram of a DC Power Supply Design

The main purpose of a DC power supply is to provide power to the subparts of the system. The main aspects of DC power supply design are as under:

- **Step Down Transformer:** It is usually required to stepdown the 220 V alternating current (AC) into the desired levels of AC i.e. 12 or 24 V.
- **Rectification:** This step usually done to convert the stepdown AC voltage into DC voltage by either half-wave or full-wave rectification process. In half-wave rectification, only positive half cycles of the AC voltage are appeared whereas both positive and negative half cycles appear in full-wave rectification.
- **Filtering:** After rectification, the pulsating DC is obtained that contains ripples in the output. These ripples are removed by polarized capacitors to give smooth DC voltage.
- **Voltage Regulation:** Finally, the levels of DC voltage are maintained by a voltage regulated that are necessary to be kept fixed. Because, the variable DC can affect the overall performance of the system. It is usually achieved by a Zener diode or some other voltage regulator circuits.

However, in this project 9V DC batteries are used to power-up the subsections of the circuit.

3.1.2. Bipolar Junction Transistor

A bipolar junction transistor (BJT) is a current controlled amplifier that contains three terminals i.e. emitter, base and collector. The emitter is heavily doped; base is lightly doped and collector is moderately doped. A BJT can be in two forms i.e. NPN and PNP depending on the direction of emitter currents in both the configuration. The PNP and NPN BJT circuits are shown in Figure 7.

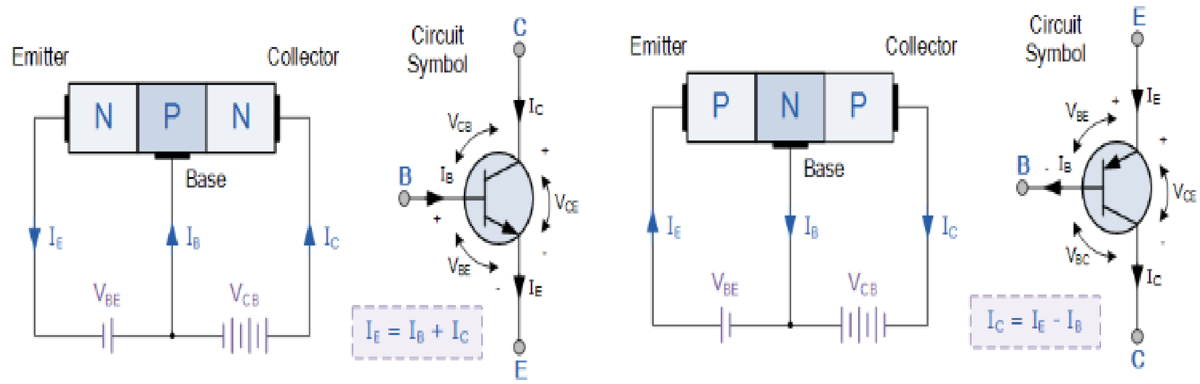


Figure 7. BJT in NPN and PNP Configurations [21]

There are various purposes of using a BJT in a circuit. Some of the uses of BJT are discussed below:

- **Switch:** To use BJT as a switch it is usually operated in the cutoff or saturation region. In cutoff region, it behaves like an open switch whereas in saturation region it works as a closed switch. The BJT as a switch is shown in Figure 8.

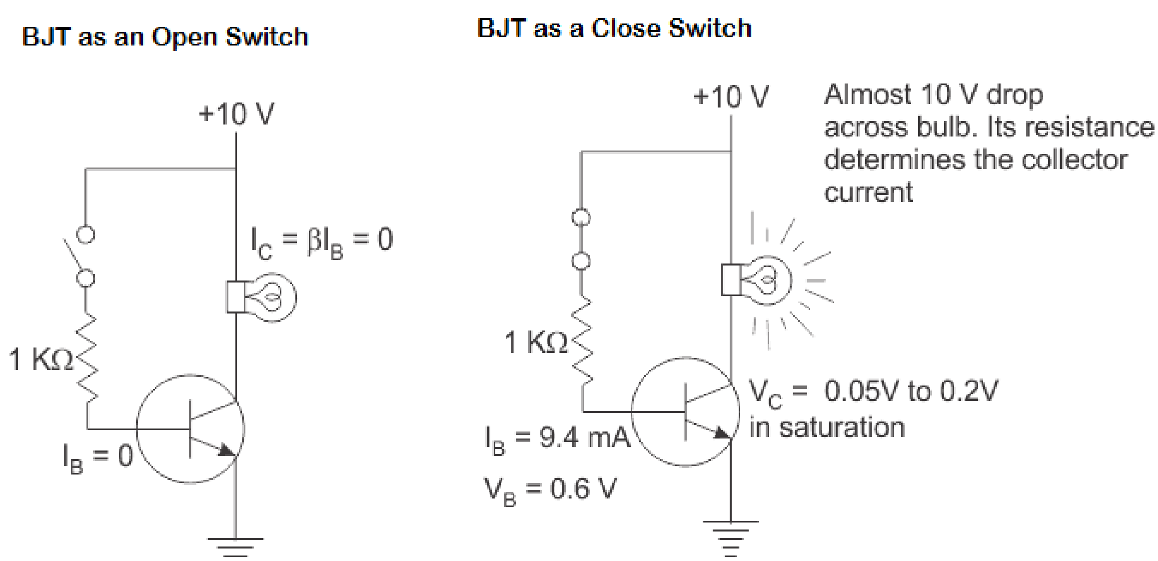


Figure 8. BJT as a Switch [22]

- **Current Follower:** To use as a current follower circuit, BJT is operated in common-base configuration. Because, the current gain $\alpha = \frac{I_C}{I_E} \equiv 1$ in this configuration. A common-base configuration of BJT is shown in Figure 9.

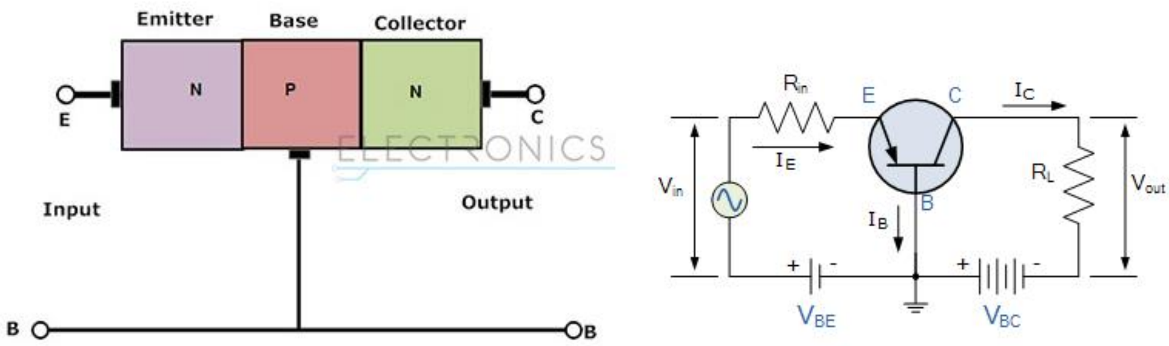


Figure 9. A BJT in Common Base Configuration [23]

- **Amplifier:** The BJT can be used as an amplifier when connected in common-emitter configuration. Because, the gain $\beta = \frac{I_C}{I_B} > 1$. The common-emitter configuration is shown in Figure 10.

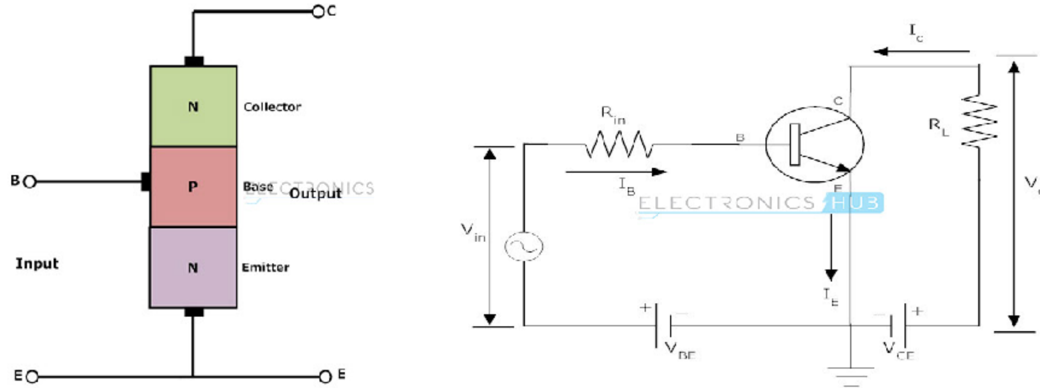


Figure 10. Common Emitter Configuration of a BJT [24]

- **Voltage Follower:** BJT is used as a voltage follower circuit when connected in common-collector mode. The output voltage follows the input voltage and can be used as a buffer. The common-collector configuration of BJT is shown in Figure 11.

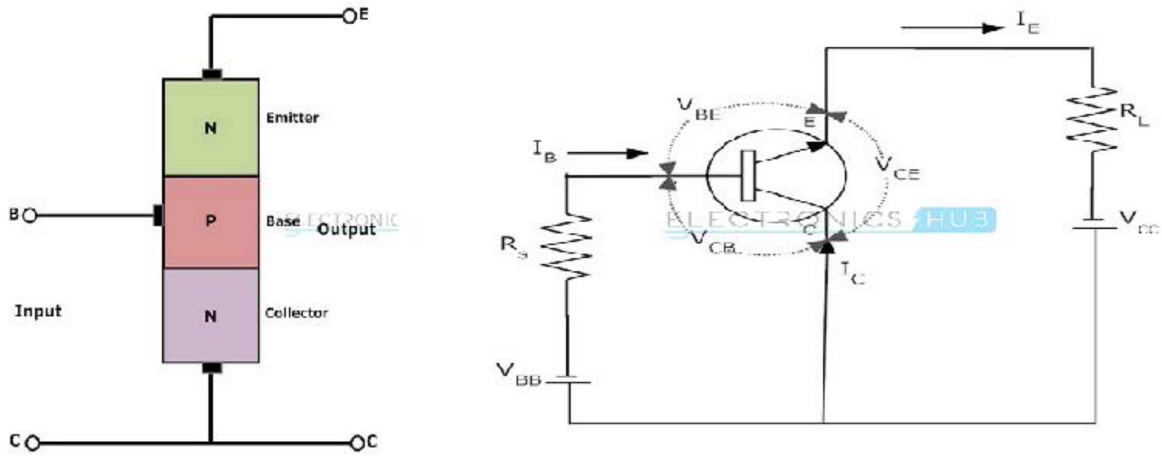


Figure 11. A BJT in Common-Collector Configuration [24]

In the proposed circuit, PN2222 BJTs are used in common-collector configuration.

3.1.3. Coil Inductor

An inductor is also known as a coil, which is a two-terminal passive device that stores electrical energy in a magnetic field after the passage of current through it. The inductor is composed of wire that is wound into a coil [25].

According to Faraday's law of electromagnetic induction, the current changes in an inductor within a time varying magnetic field induces an electromotive force (EMF) that opposes its cause. Therefore, any changes in current across an inductor are being opposed by it. The value of inductor is characterized by its inductance and is measured in Henry (H).

3.1.3.1. Applications of Inductor

There are various applications of inductor and some of them are discussed below:

- **Analog Circuits:** Inductors are used in analog circuits and signal processing applications. The use of larger inductors is in power supplies that is in combination with the filter capacitors to remove the main supply hum or other DC fluctuations, smaller inductors are used to remove the interference of RF signals in the wires. Additionally, inductors are used in switch-mode DC power supplies and store electrical energy.

- **Tuned Circuits:** Inductors are extensively used in tuned circuits such as filters, oscillators, and resonators.
- **Transformer:** When inductors are used in such a way that it is combined with the mutual flux known as mutual induction, then, they make a transformer. The transformer can either be stepdown or step-up depending on the number of coils in primary and secondary windings.

3.1.3.2. Inductor Designing

Inductors designing depends on various factors that is given in the following equation:

$$L = \frac{N^2 UA}{l} \tag{3.1}$$

Where, N shows the number of turns, U shows the core material, A represents the coil area, and length of the coil is represented by *l*.

The effect of these factors on the inductance of a coil is shown in Table 1.

No. of truns		Coil Area		Coil Length	
less inductance	more inductance	less inductance	more inductance	less inductance	more inductance

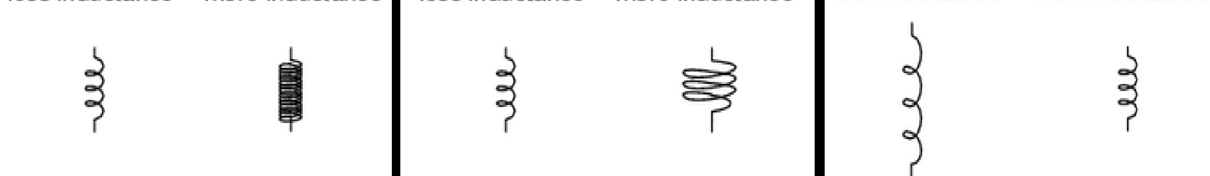


Table 1. Effect of Number of Turns, Coil Area and Coil length on the Inductance of a Coil

Based on the winding of inductor, it is classified in following types.

- **Single-Layer Wound Coil Inductor:** A single-layer wound coil inductor is described by following equation:

$$\boldsymbol{L} = \frac{N^2 r^2}{9r + 10l} \tag{3.2}$$

Where, N is the number of turns, r shows the wire pitch and l shows the length of the coil.

- **Multi-Layer Wound Coil Inductor:** This type of inductor is described by following equation:

$$L = \frac{0.8N^2r^2}{6r + 9l + 10c} \quad (3.3)$$

- **PCB Mounted Square Coil Inductor:** This is the inductor that is proposed in the project and is given by following equation:

$$L = 85\times 10^{-10} \times D \times N^{\frac{5}{3}} \quad (3.4)$$

The inductor is designed in Proteus 8.0 circuit design software and was mounted over a printed circuit board (PCB). It is shown in Figure 12. The details of coils are shown in the following table.

Wire gauge	No. of turns	Coil type	Internal dia	External dia	Ampere
24	10	Center taped	4.5 cm	5cm	20mA
34	50	Single layer	2.5cm	2.5cm	40mA
34	100	Multi-layer	0.8cm	1.5cm	60mA
34	150	Multi-layer	0.8cm	2.1cm	100mA

Table 2. Design of Inductor Coil

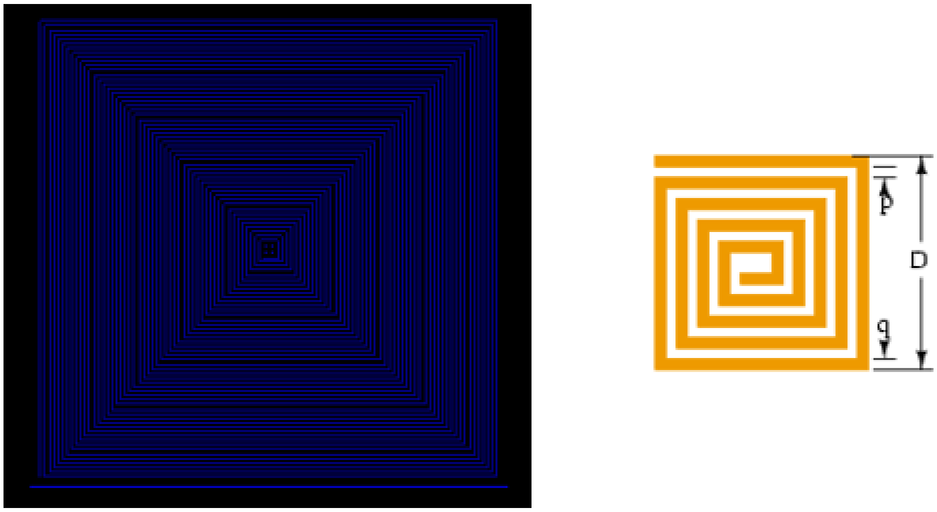


Figure 12. PCB Mounted Square Coil Design in Proteus

3.1.4. Arduino Board Nano

A typical Arduino board Nano is shown in Figure 13.

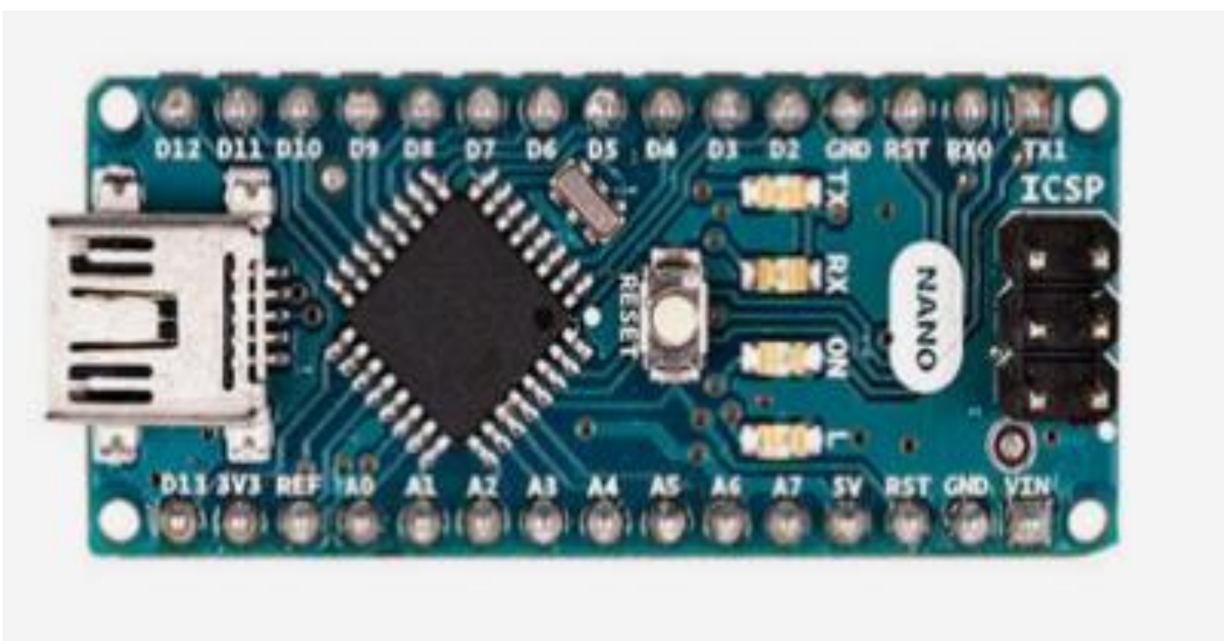


Figure 13. A typical Arduino Board Nano [26]

The specifications of an Arduino Board Nano are given in Table 2.

Microcontroller	ATmega328
Architecture	AVR
Operating Voltage	5 V
Flash Memory	32 KB of which 2 KB used by bootloader
SRAM	2 KB
Clock Speed	16 MHz
Analog I/O Pins	8
EEPROM	1 KB
DC Current per I/O Pins	40 mA (I/O Pins)
Input Voltage	7-12 V
Digital I/O Pins	22
PWM Output	6
Power Consumption	19 mA
PCB Size	18 x 45 mm
Weight	7 g
Product Code	A000005

Table 3. Specifications of an Arduino Board Nano [26]

3.2. Methodology

The methodology of the whole system is divided into two parts i.e. initial design that is implemented at the early stages of the project to check the implementation of idea and final design was developed after amendments in the initial design.

3.2.1. Initial Design

The block diagram of initial design is shown in Figure 14.

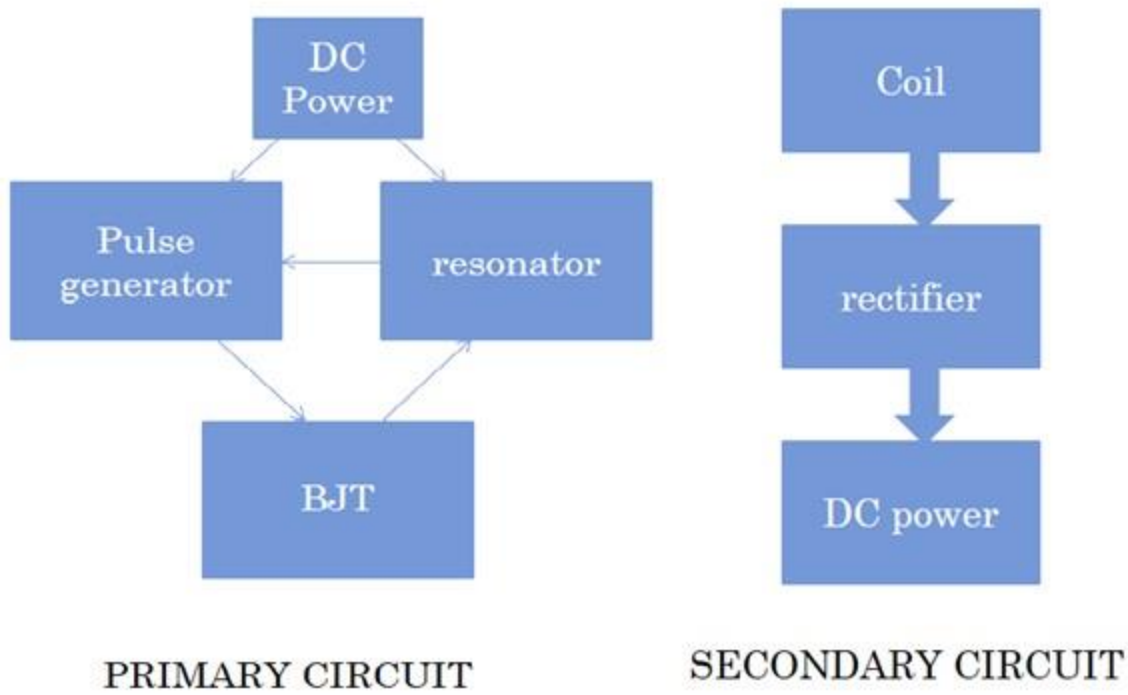


Figure 14. Block Diagram of Initial Design of the system

The initial design was composed of a primary and secondary circuits. A 9V DC power supply was used to power up the primary circuit. A square wave was given from the functional generator at the base of a BJT connected in common-emitter configuration. The BJT is used for switching purpose and the collector of BJT is connected to the primary coil of the circuit. The number of turns in primary coil are N=10, 4.5 cm center tapped coil made up of 24-gauge enameled copper wire. The change in flux in the primary coil induces an EMF in the secondary coil. The secondary coil is also made up of N=10 number of turns, 4.5 cm coil made up of 24-guage enameled copper wire. The voltage in the secondary coil is of alternating nature that is rectified by a half-wave rectifier circuit composed of a diode. This gives a pulsating DC voltage and an LED is on/off by it. The glowing of LED proves the idea that made the foundation of the final design. The initial circuit diagram is shown in Figure 15.

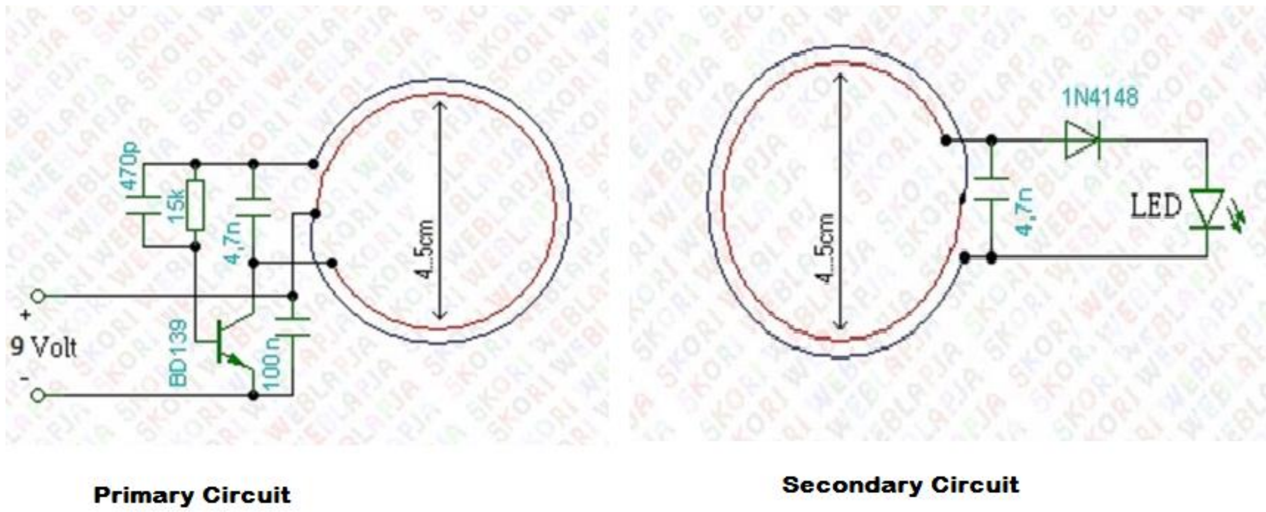
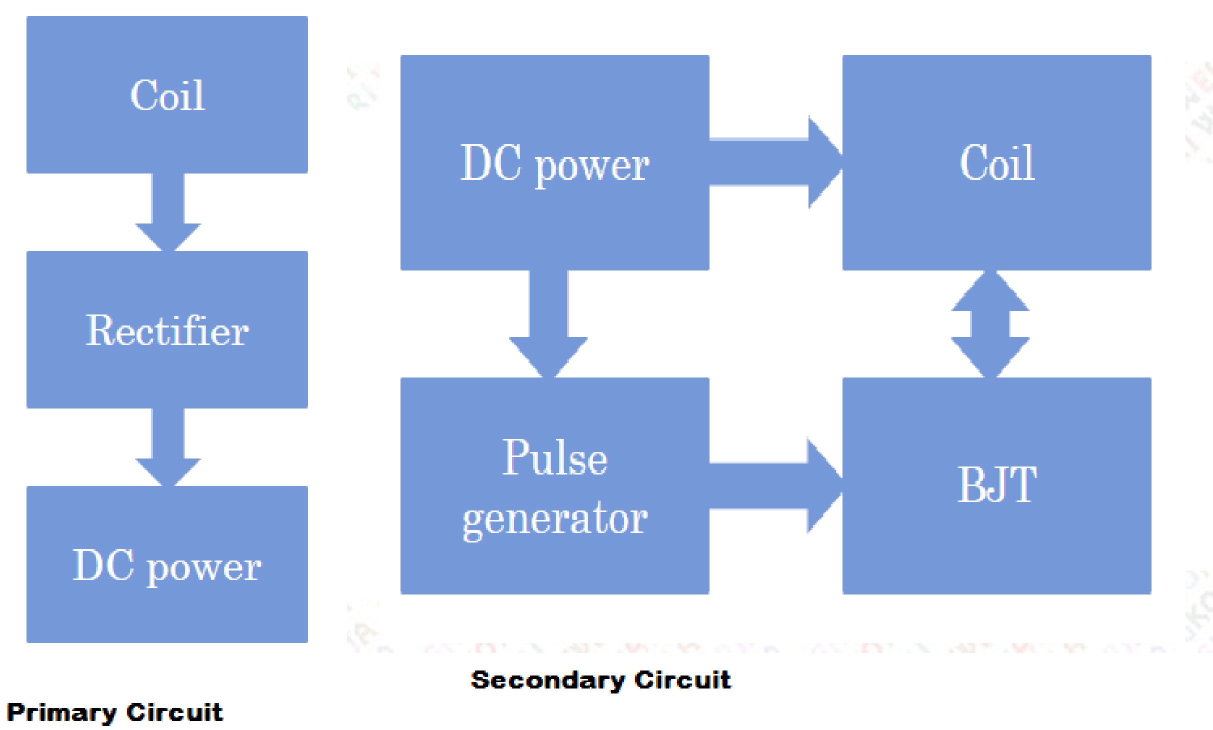


Figure 15. Initial Circuit Diagram of the System

3.2.2. Final Design

3.2.2.1. Modification in Initial Design

After getting a successful initial design of the circuit, the design was modified and final design were constructed and implemented practically. The block diagram of the final design is shown in Figure 16.



After modification in the initial circuit diagram, the new circuit design is shown in Figure 16.

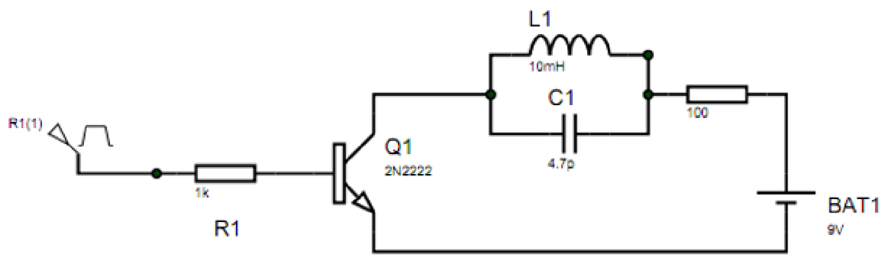


Figure 16. Change in the Initial Circuit Diagram

This design was made due to very lower power transfer in the initial circuit design. The 9V battery is connected to the LC circuit through a 100 Ω resistor. The LC circuit is connected to the collector of a common-emitter BJT and the base of the BJT gets pulses from the functional generator. When there is a high pulse found by the base of BJT, the collector opens and generates a pulsating DC with a high amplitude. The primary side of the circuit connected to DC that power up the circuit components, primary coil containing single layered N=54 turns made up of 54-guaged enameled copper wire. Similarly, the secondary coil also contains the same number of turns and other parameters of that having the primary coil. When the secondary coil is brought closer to the primary circuit, an EMF is induced in it and is rectified by a halfwave rectifier and the LED glows.

3.2.3. Data Transmission through UART

The whole data transmission through UART is given by the following block diagram:

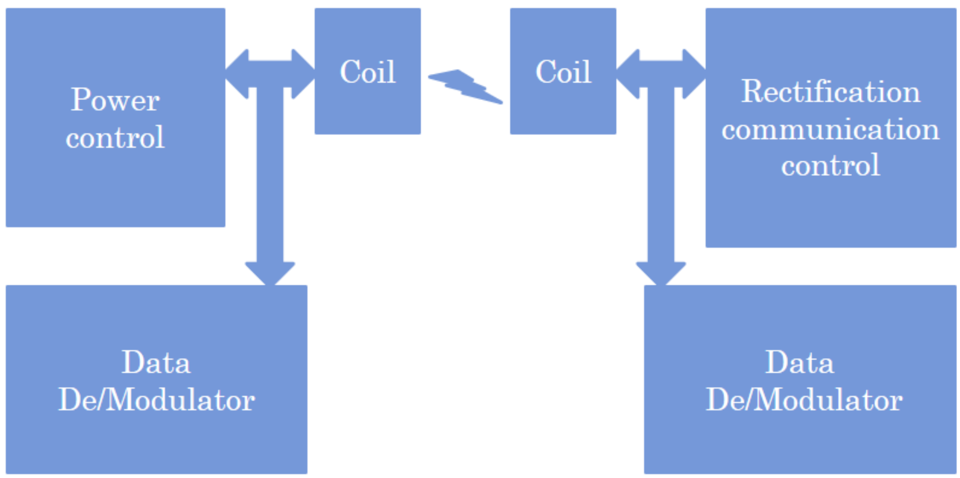


Figure 17. Block Diagram of Data Transmission Through UART

- **Data Transmission from Primary to Secondary:** Similar circuit used for power transmission is modified with modulator circuit is attached between the LC and power controlling circuit. Basically, for modulation BJT is used, connected in common-emitter configuration. The base of the BJT is connected to the TX (serial port) of the Arduino Board Nano that sends data in bytes. When each bit is sent then it shows the voltage is high and emitter drops the transmitting level of voltage. On the other side the (secondary), the voltage is collected and it is step-down, filtered for removal of noise and connected to the analog pin of secondary Arduino board Nano. The secondary Arduino Board Nano receives each bit and converts it through defining the threshold for low and high bits. To convert it into byte, one digital pin is kept high and low as per the sequence received on analog pin. This digital pin to RX of the Arduino is shortened to make it as if it is receiving the serial data in RX.
- **Data Transmission from Secondary to Primary:** For secondary to primary data communication, the BJT is connected in common-emitter configuration, and the base of BJT is connected to the LC to the TX of the secondary Arduino Board Nano. When TX transmit data in bytes the voltage goes high and low as per the bit sequence. The BJT accordingly switches, which results in change in the voltage, that change in the voltages is received at the analog pin of the primary side's Arduino Board Nano, and each bit converted it through defining the thresholding for high and low bits. Then, to convert it into byte, one digital pin is kept high and low as per the sequence received on analog pin. This digital pin is shortened to RX of the Arduino to make it as if it is receiving the serial data in RX.

3.3. Final Circuit Diagram

The final circuit diagram of the proposed system is shown in Figure 18 and Figure 19.

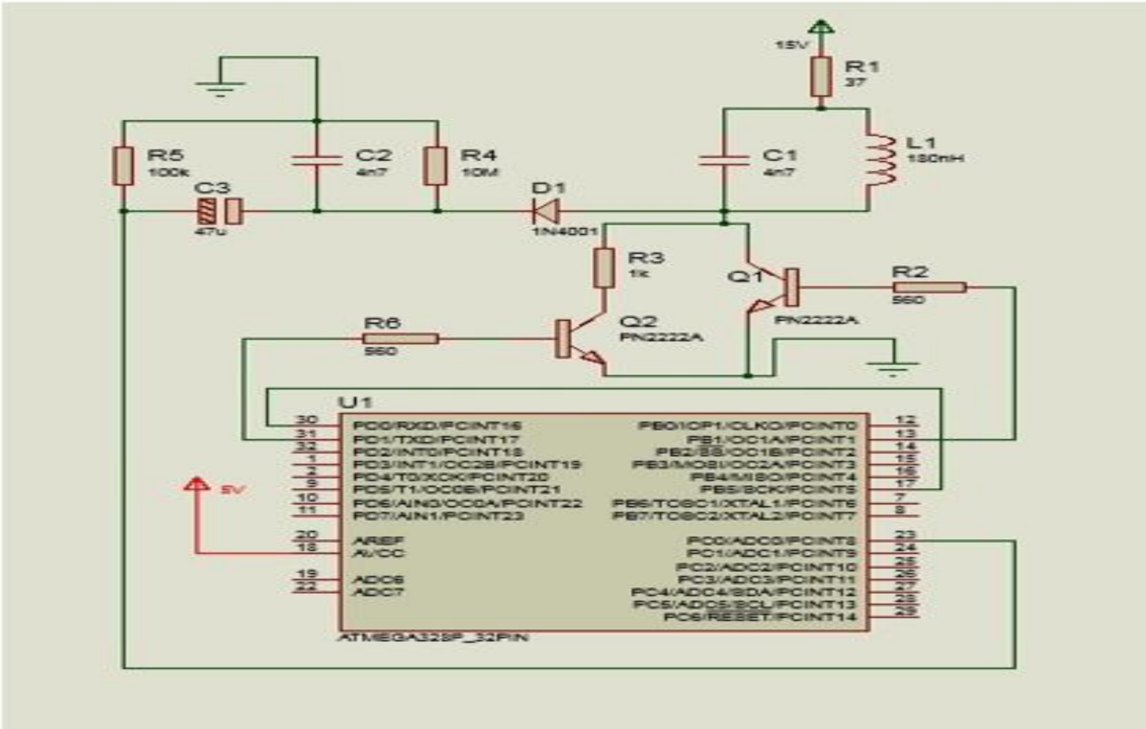


Figure 18. Final Primary Circuit in Proteus

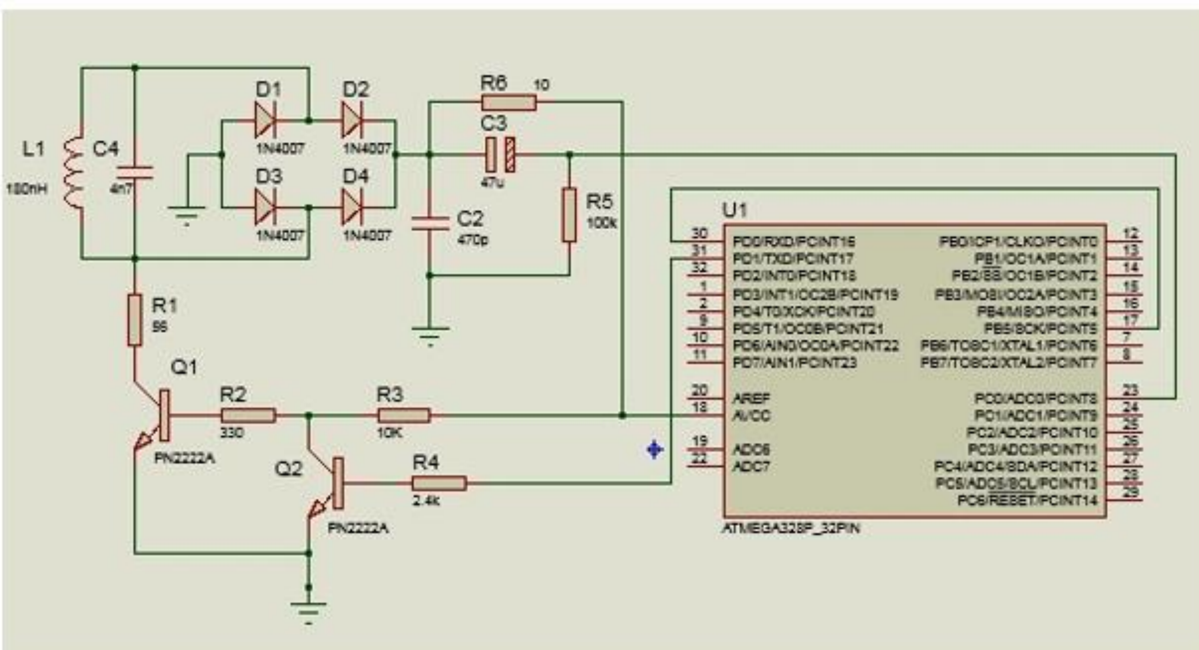


Figure 19. Final Secondary Circuit in Proteus

CHAPTER 4: RESULTS AND ANALYSIS

4. Results

4.1. Power Transmission

4.1.1. Power Transmission in Air

When the medium between the primary and secondary circuit is air, power is efficiently transmitted from primary to secondary circuit. A maximum power of 500 mW is transmitted from primary side and with an increase in 1 mm distance, the power received by secondary circuit was 480 mW. This power was reduced within increase in distance between the primary and circuits. By reaching at 5 mm, the received power was about 380 mW.

The details of the power transmission are shown in Table 3.

Power Transmitted	**Distance between Primary and Secondary Circuits**	**Power Received**
500 mW	1 mm	480 mW
500 mW	2 mm	450 mW
500 mW	3 mm	420 mW
500 mW	4 mm	400 mW
500 mW	5 mm	380 mW

Table 4. Power Transmission in Air

The results were then plotted distance (mm) vs power transmission (mW) and are shown in Figure 20.

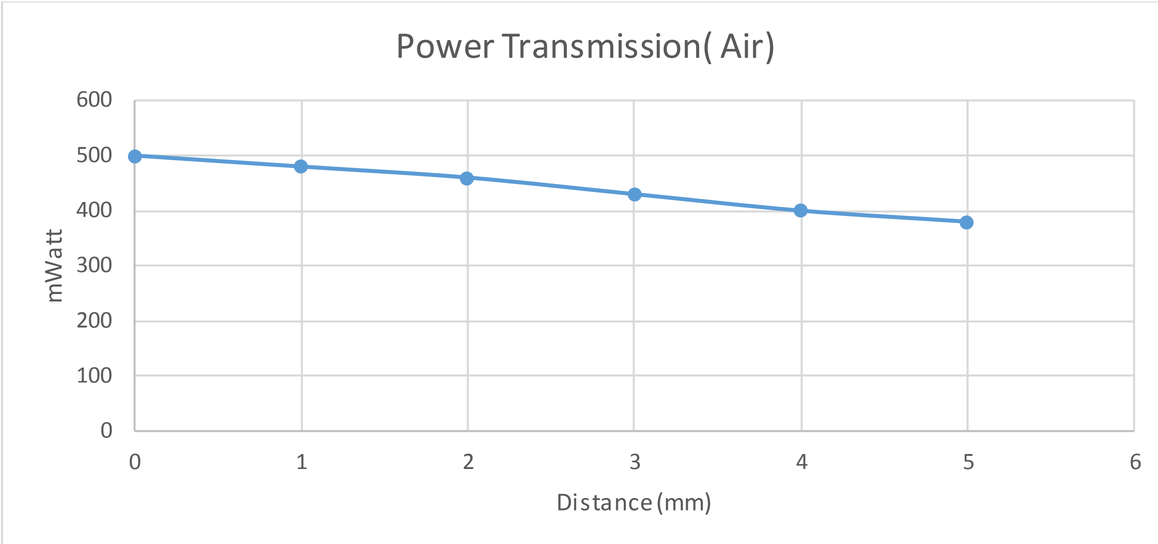


Figure 20. Plot of Power transmission in Air

4.1.2. Power Transmission in Polyurethane

When polyurethane was kept as a medium between the primary and secondary circuit, similar results was obtained as observed in the case of air. Table 4 shows the detail of the transmission in polyurethane.

Power Transmitted	**Distance between Primary and Secondary Circuits**	**Power Received**
500 mW	1 mm	480 mW
500 mW	2 mm	450 mW
500 mW	3 mm	420 mW
500 mW	4 mm	400 mW
500 mW	5 mm	380 mW

Table 5. Power Transmission in Polyurethane

The results were then plotted distance (mm) vs power transmission (mW) and are shown in Figure 21.

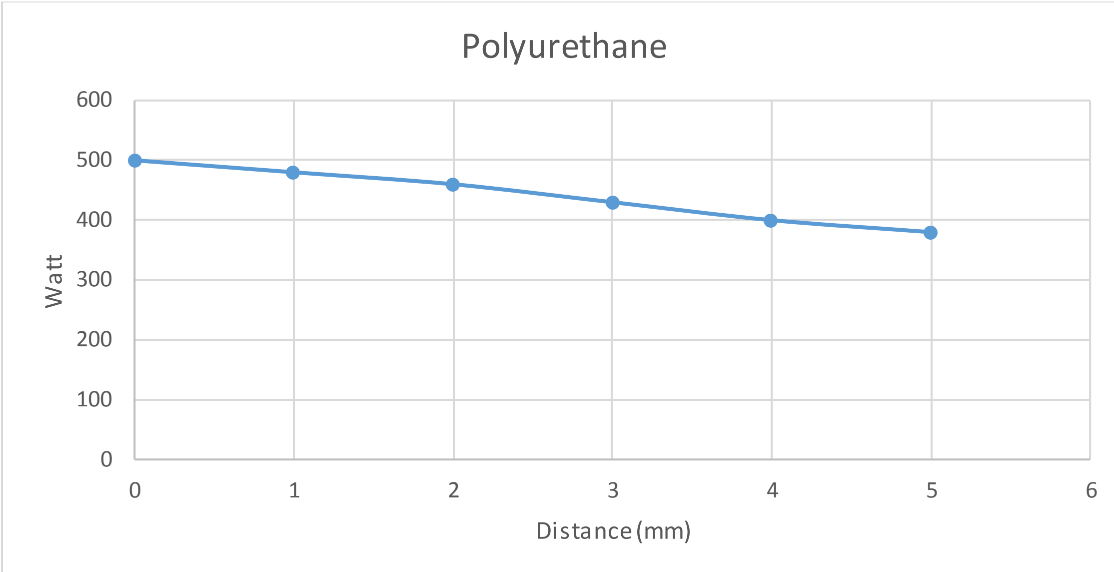


Figure 21.Plot of Power transmission in Polyurethane

4.1.3. Power Transmission in Wood

Similarly, when wood was kept as medium between primary and secondary circuits, there was no effect on the transmission process. Table 5 gives the detail of results.

Power Transmitted	**Distance between Primary and Secondary Circuits**	**Power Received**
500 mW	1 mm	480 mW
500 mW	2 mm	450 mW
500 mW	3 mm	420 mW
500 mW	4 mm	400 mW
500 mW	5 mm	380 mW

Table 6.Power Transmission in Wood

The results were then plotted distance (mm) vs power transmission (mW) and are shown in Figure 22.

Figure 22. Plot of Power Transmission in Wood

4.2. Data Transmission

A single byte (52 in decimal 0r 00110100) was sent from the primary side, and received (52) accurately on the secondary side. Then, the same byte was sent 10 times over the inductive link and 10 bytes was received accurately again. The experiment was repeated for 100 resulting in 100% accuracy then and 1000 bytes sent and 1000 was received but one byte was not received truly so the accuracy was 99.9% which means only one byte was lost i.e.1000 bytes, with a loss of a single data byte during the transmission. Table 6 gives the details of the whole data transmission process.

No. of Bytes Transmitted	No. of Iterations	Accuracy of Receiving
1 byte	10	100 %
10 bytes	100	100 %
100 bytes	100	100 %
1000 bytes	1000	99.9 %

Table 7. Data Transmission Process

The number of bytes were plotted vs accuracy of byte reception on secondary side is given by the following plot.

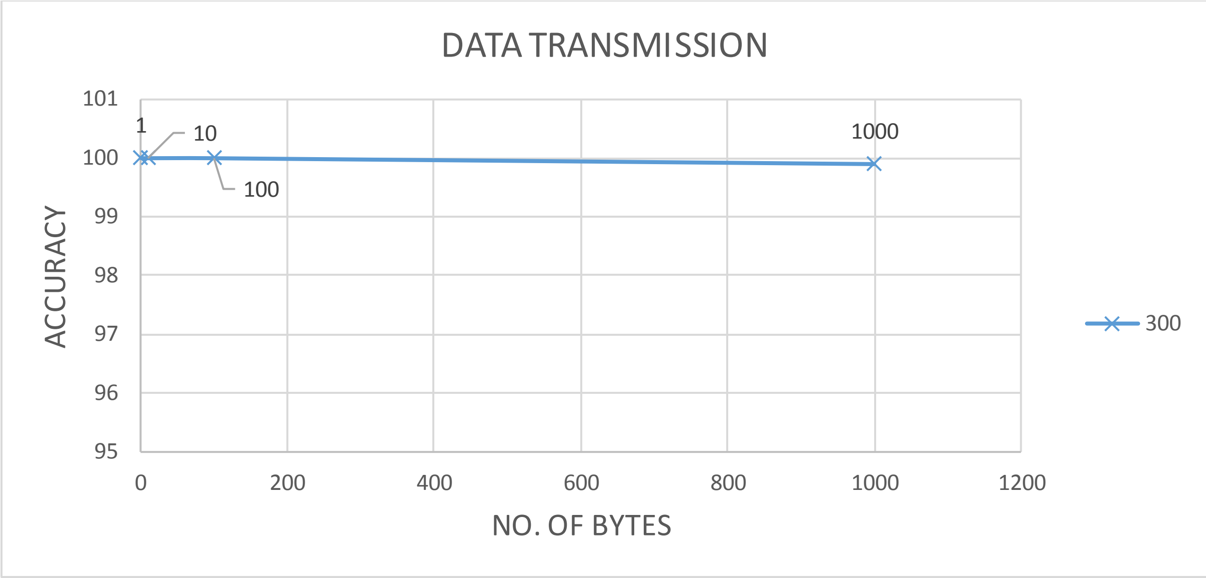


Figure 23. Plot of Number bytes versus Accuracy of Bytes reception on Secondary side

4.2.1. Transmission of Same Bytes

The data was transmitted again on varying baud rates of 300, 1200, 2400 and 4800. The results showed the gradual decrease in the accuracy of data transmission with the increasing baud rate as the bytes were received with 99.9%, 98%, 93.1% and 50% for the baud rates 300, 1200, 2400 and 4800 respectively. The process is shown in Figure 24.

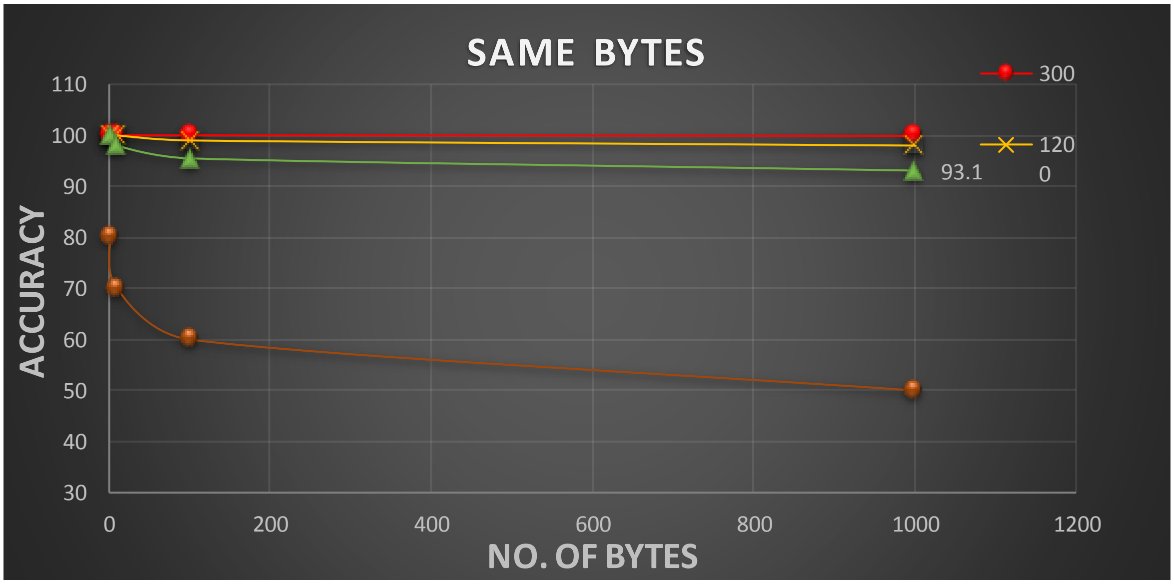


Figure 24. Transmission of Same Bytes

4.2.2. *Transmission of Random Bytes*

The experiment was repeated for random bytes on similar baud rates and the results obtained are given in Figure 25.

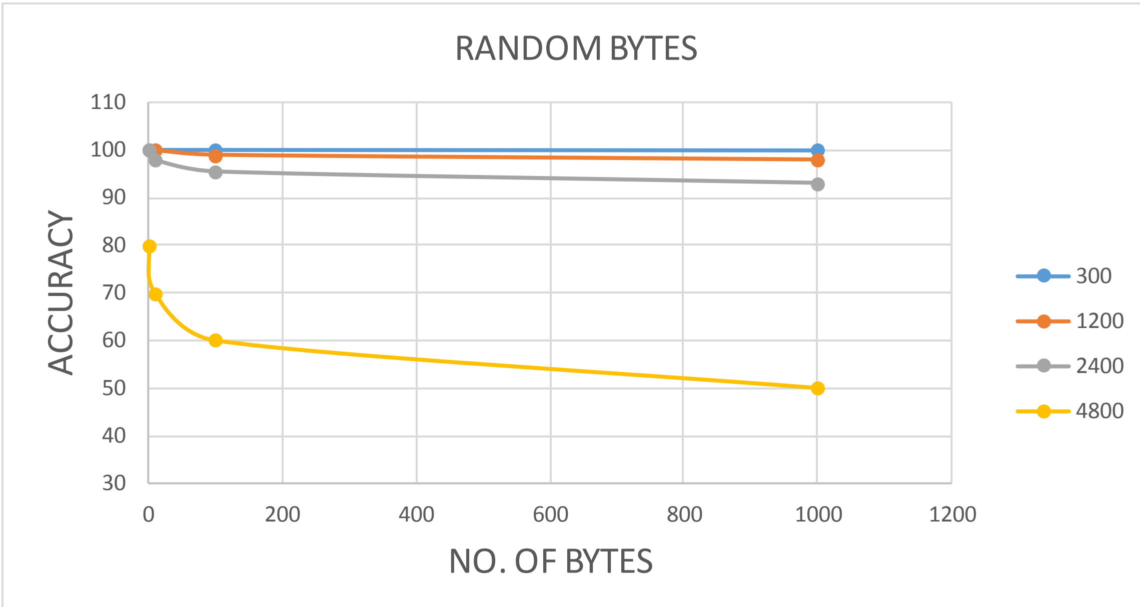


Figure 25. Transmission of Random Bytes

4.2.3. *Data Transmission Over Distance*

The results also showed decrease in accuracy with increase of distance between the two coils. It was also observed that the accuracy varies a little over 2mm distance (100%-98%) and then drastically falls to 42% on 5mm distance. This is shown in Figure 26.

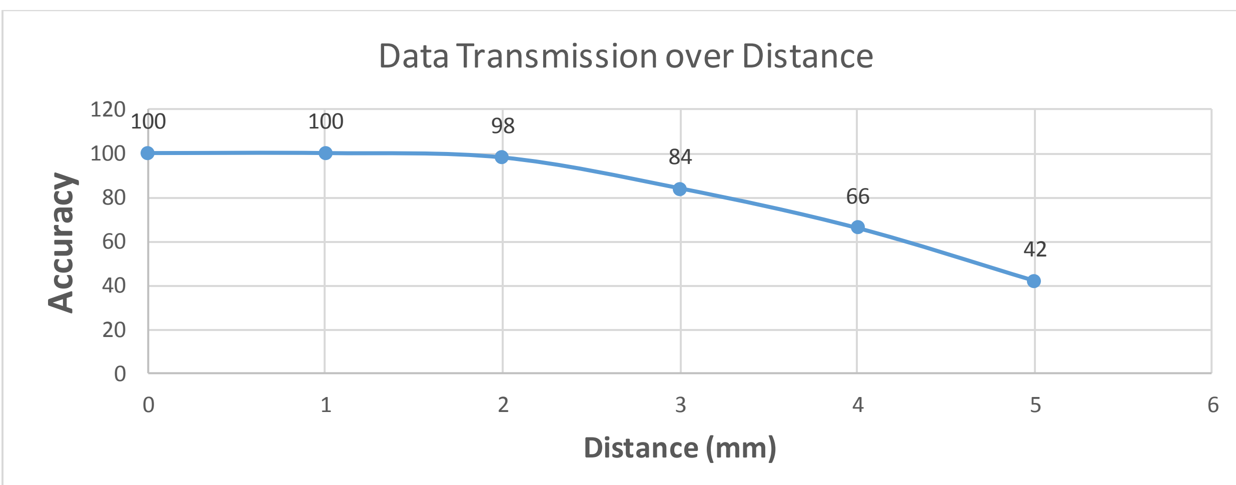


Figure 26. Data Transmission Over Distance

4.2.4. Data Transmission in Air

The experiment was repeated by changing baud rate and results are as below:

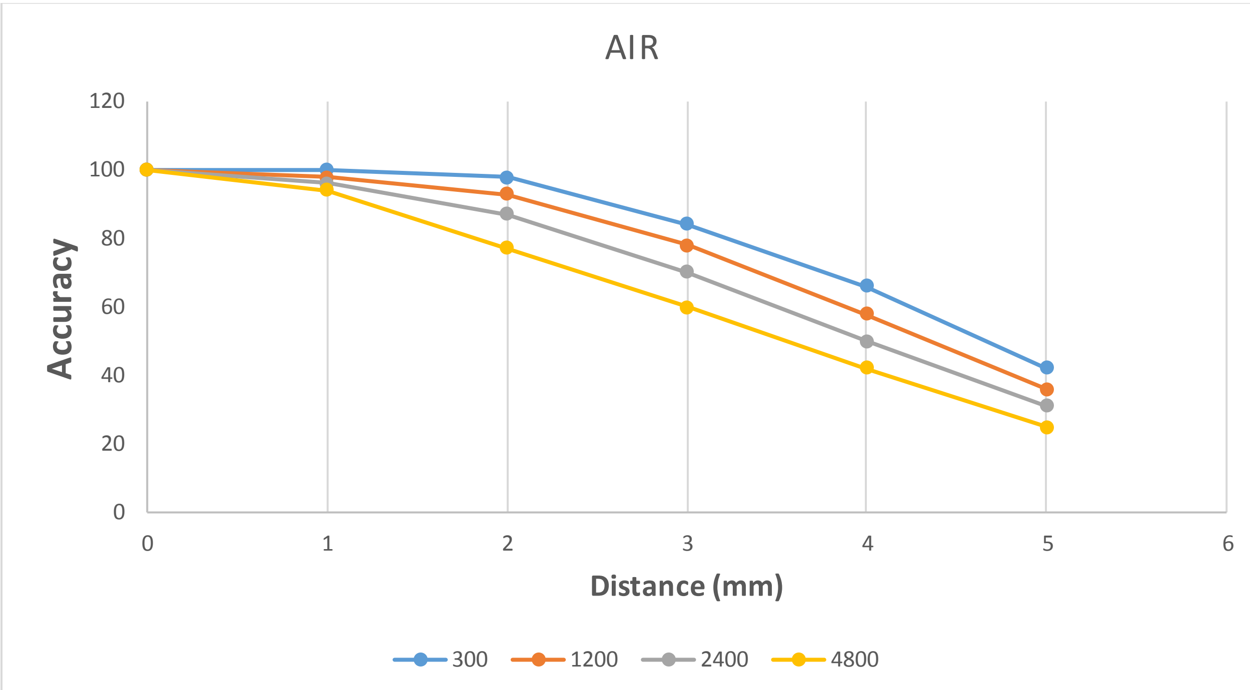


Figure 27. Data Transmission in Air

4.2.5. Data Transmission in Polyurethane

When polyurethane was put in between the coils to check whether it is being effected by other material or not. Polyurethane was used as it is most likely used in experiments for skin properties. The results are shown in Figure 28.

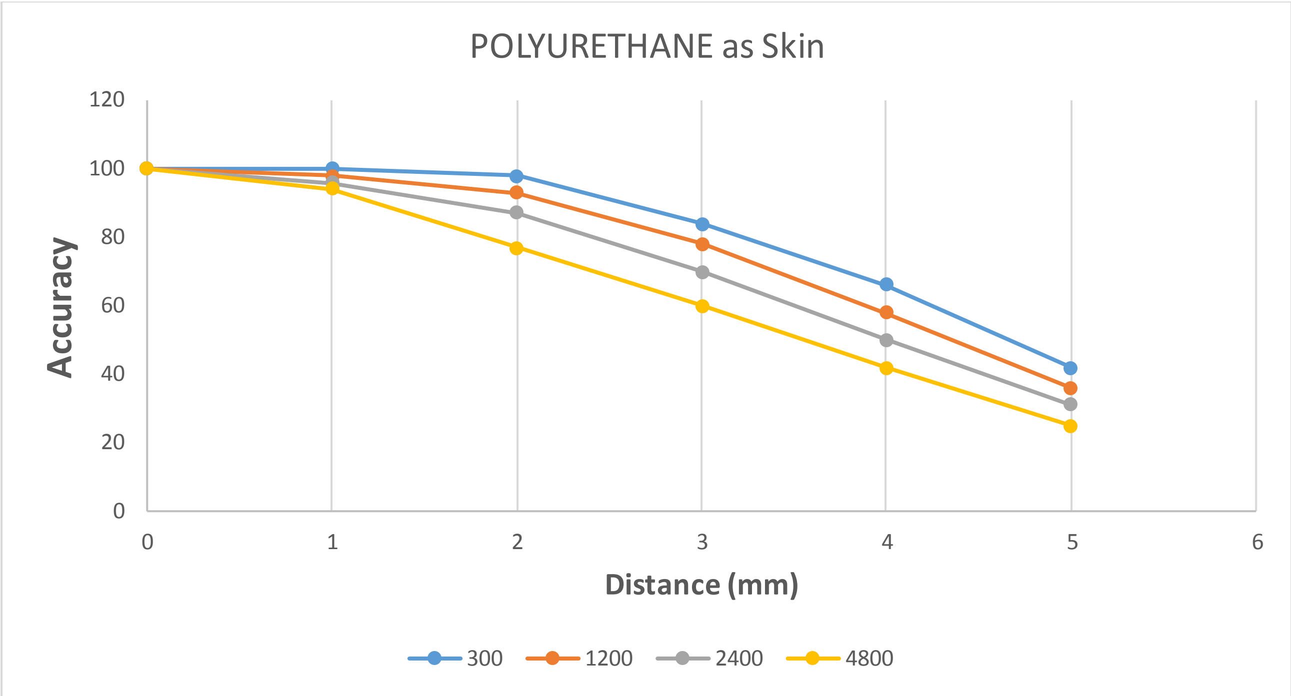


Figure 28. Data Transmission in Polyurethane

4.2.6. Data Transmission in Wood

Wood was put in between coils to check the effect of bone, and the results are as below:

Figure 29. Data Transmission in Wood

4.3. Hardware Results

The project was initially implemented on breadboard and then implemented on a PCB sheet. The PCB layout was designed in Proteus 8.0 software. The circuit implementation, coil and its results on a digital oscilloscope are shown in Figure 30 and Figure 31 and Figure 32.

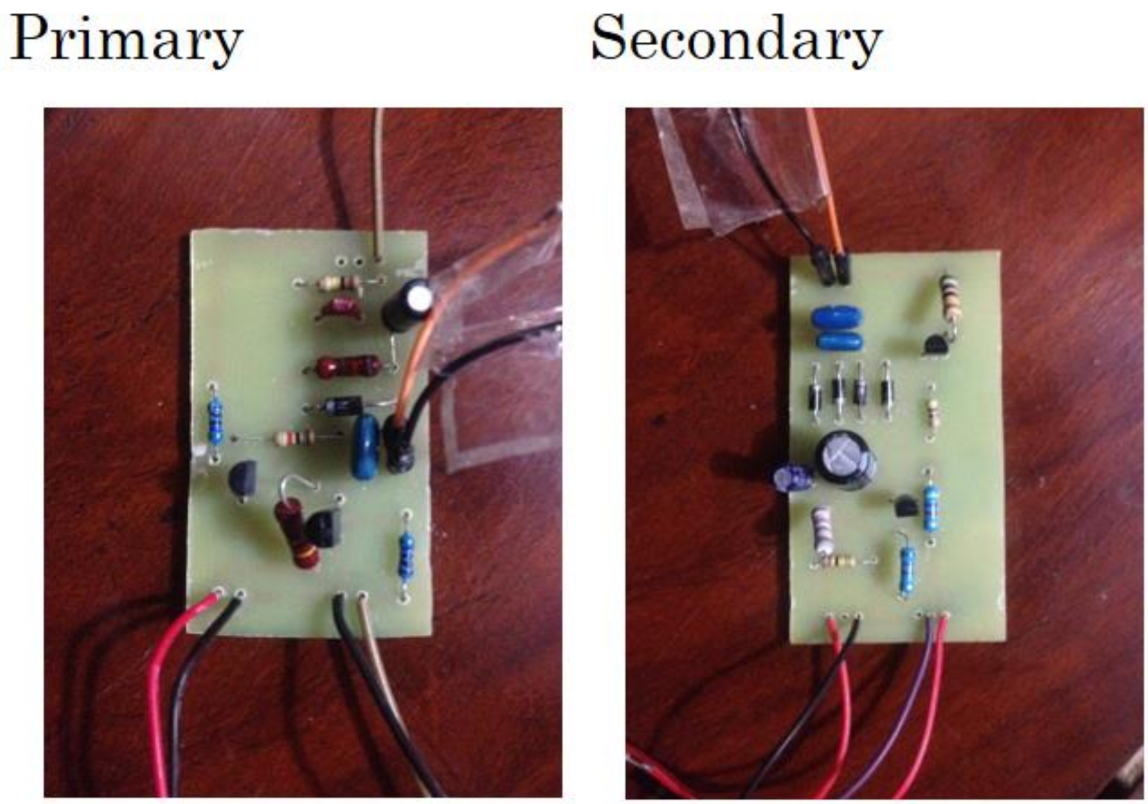
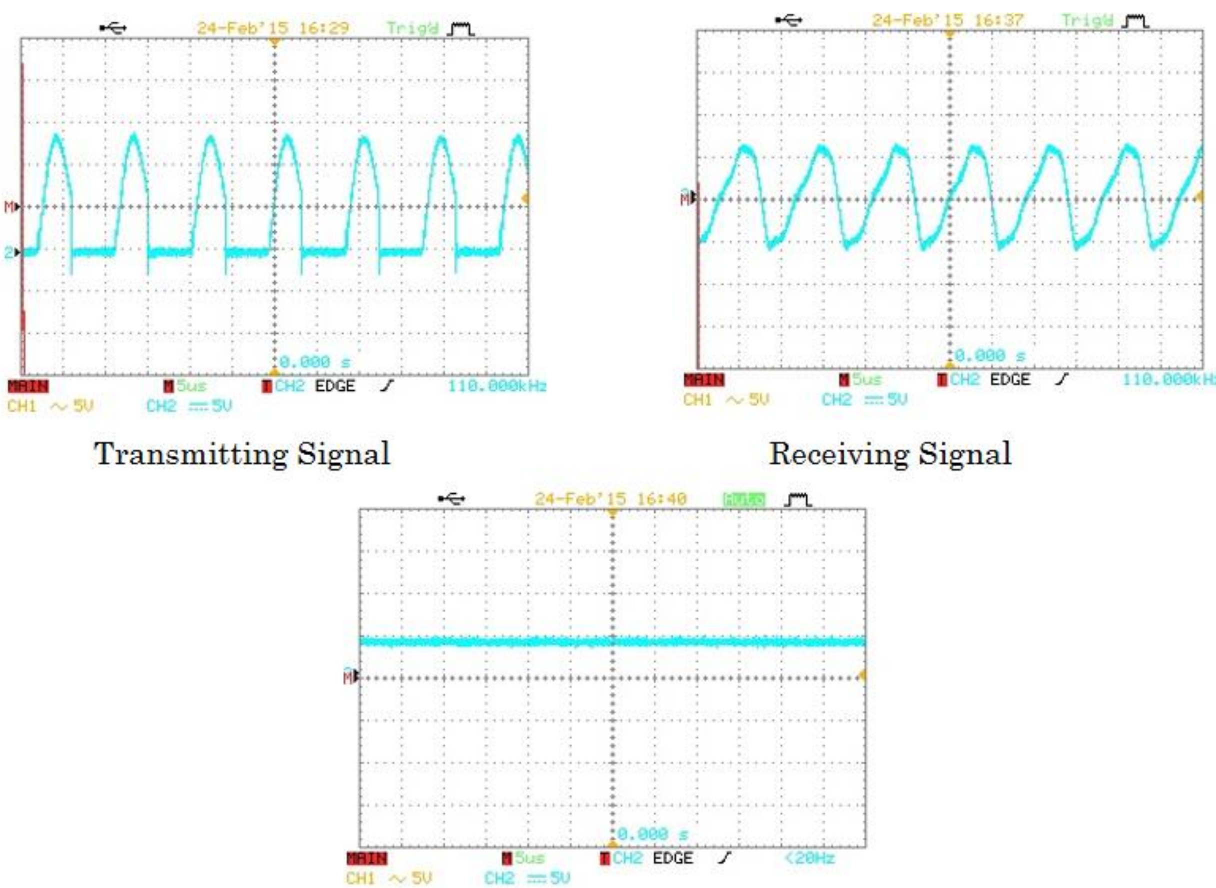


Figure 30. Implementation of Primary and Secondary Circuits on PCB sheet

Primary Coil **Secondary Coil**

Figure 31. Primary and Secondary Coils



Transmitting Signal

Receiving Signal

Rectified Signal

Figure 32. Hardware Results of Power Transmission on Digital Oscilloscope

CHAPTER 5: CONCLUSION AND FUTURE RECOMMENDATIONS

5.1. Conclusion

The project was aimed to develop an RF inductive link for medical implantable devices for both the power and data transmission. As mentioned before in the literature review section that most of the circuits for biomedical implants require wireless connection for their operation. An inductive coupling is used for powering many of these devices within the range of centimeters on the surface of skin.

The project started from an initial design that was powered by a 9V battery and rectangular pulses were provided from a functional generator. The pulses were given at the base of the common-emitter BJT and were rectified by a halfwave rectifier that glows the LED. After, successful power transfer between the primary and secondary circuit, modification was made in the circuit and pulses were provided to the BJT by Arduino board Nano using pulse width modulation. The BJT switches between on and off state and produced alternating signal that was rectified and filtered on the secondary side. The power was efficiently transferred and excellent data communication between the primary and secondary circuit.

To test the power transfer and data communication between the primary and secondary circuits, different mediums were used i.e. air, polyurethane (skin) and wood (bone). It was observed that there is no effect of medium on both the power and data transmission between both the circuits. Without any power loss, the secondary circuit efficiently received the power transferred by the primary circuit. The data communication between the primary and secondary circuits and the effect of different mediums are compared and shown in graphical form in Figure 33.

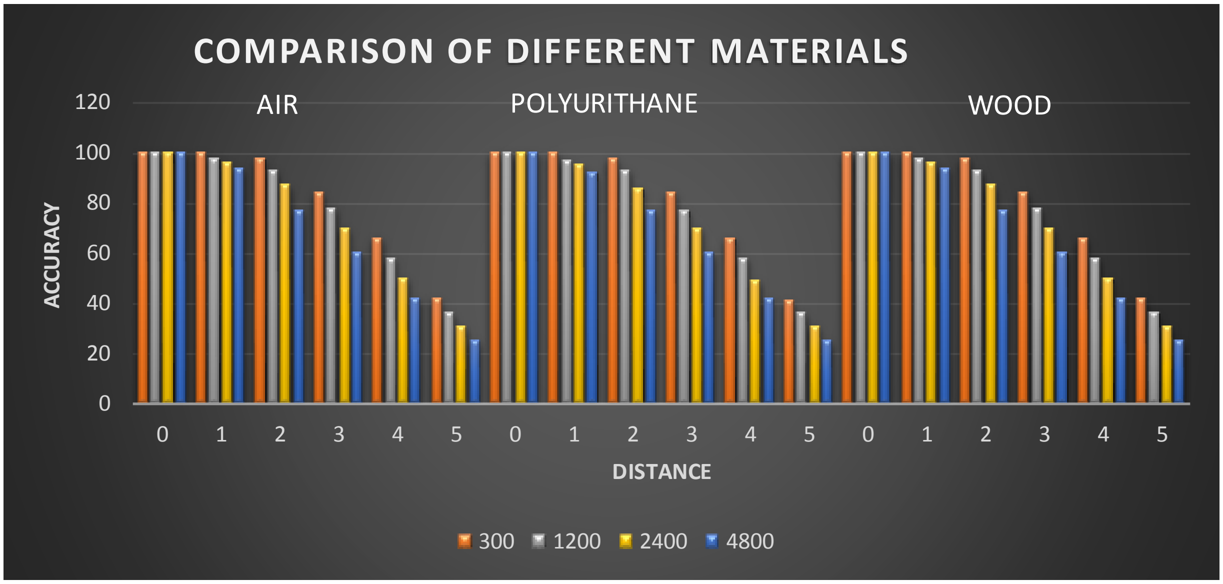


Figure 33. Comparison of Data Communication between Primary in Air, Polyurethane and Wood

5.2. Future Recommendations

Although, the main objectives of the project were achieved but the circuit can be further improved by using the following recommendations:

- The circuit was powered up by a 9V battery that has very limited amount of current and non-rechargeable as well. Thus, rechargeable Lipo batteries can be used that are smaller in size, high current ratings and rechargeable.
- The coils are wound by hand, so to improve them, windings should be made using machine for improving the efficiency of the circuit.
- The number of turns in both the primary and secondary are kept same. If the primary number of turns are increased, then more flux will be generated and will improve the range of power transfer.
- The circuit uses BJT in common-emitter configuration for switching purpose, so, to improve the circuit efficiency, MOSFET can be used.
- The circuit used ordinary components bought from the local market. To make it power efficient, the size of the circuit and utilization of SMD components can be brought into account.
- Further improvements can be made by using efficient coding algorithms and circuit designing techniques.

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