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Title

**Design of a Wireless Power Transfer System for Implantable Bio-
MEMS Microelectronic Devices: Application to the Pacemaker**

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Dedication

First and foremost, I praise Allah Almighty and thank Him for His blessings, guidance, and for granting me the strength to complete this work.

To my beloved parents, a token of my deepest love, affection, and unwavering support since the day I was born. For all the hard work, sacrifices, and dedication they made for my education; they taught me to work hard until the very end to achieve success. Words are simply not enough to thank them, as they fail to express my deep gratitude, love, and utmost respect. I pray to Allah Almighty to grant them long lives, good health, and eternal happiness.

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List of Abbreviations

List of Abbreviations

Abbreviation	Full Meaning
Bio-MEMS	Biomedical -Micro Electro Mechanical Systems
DMI / IMDs	Implantable Medical Devices
WPT	Wireless power transfer
IPT	Inductive power transfer
MRIC	Magnetic Resonant Inductive Coupling
CPT	Capacitive power transfer
MPT	Microwave power transfer
SAR	Specific Absorption Rate
ISM	Industrial, Scientific, and Medical
S-S	Series - Series
P-P	Parallel - Parallel
S-P	Series - Parallel
P-S	Parallel - Series
LT	Transmitter Coil
LR	Receiver Coil
CT	Transmitter Capacitor
CR	Receiver Capacitor
RL	Load Resistance

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Abstract

Abstract

Implantable Medical Devices (IMDs) require a continuous power supply for sustainable operation. However, relying on conventional integrated batteries remains limited, with a lifespan generally not exceeding 10 years, which forces patients to undergo repeated and risky surgical procedures for replacement. To overcome this limitation, Wireless Power Transfer (WPT) via resonant inductive coupling emerges as a highly viable and sustainable solution. This study proposes the design, modeling, and practical development of high-efficiency inductive links tailored for biomedical implants. Operating at the international standard frequency of 13.56 MHz, the proposed system achieves an optimal trade-off between biological safety and transmission efficiency at the required depths.

During the characterization of the electrical links, the work focuses on strict compliance with the miniaturized dimensions of the implant. A rigorous comparative analysis was conducted among three coil geometries: planar, solenoid, and hybrid configurations across various transfer distances. Experimental results demonstrated the superiority of the proposed hybrid planar-solenoid configuration (a 4.5 cm planar transmitter combined with a 1.5 cm miniaturized solenoid receiver) as the optimal solution. This hybrid architecture successfully secured a maximum power delivery of 73.125 mW in free space and maintained an effective power of 23.972 mW at a transmission distance of 1 cm.

Furthermore, biological safety and electromagnetic permeability were successfully validated through experimental tests using human tissue as a direct biological barrier between the coils, ensuring efficient and uninterrupted power flow. The achieved results demonstrate strong alignment with contemporary global research, confirming the scientific and practical feasibility of the proposed hybrid concept for the development of a new generation of safe and sustainable electronic implants.

Keywords: Implantable Medical Devices (IMDs), Wireless Power Transfer (WPT), Resonant Inductive Coupling, Hybrid Configuration, Coil Geometry, Biosafety

Résumé

Les dispositifs médicaux implantables nécessitent une source d'énergie électrique continue pour leur fonctionnement durable. Cependant, la dépendance vis-à-vis des batteries intégrées conventionnelles reste limitée par une durée de vie n'excédant généralement pas 10 ans, ce qui impose aux patients de subir des interventions chirurgicales répétées et risquées pour leur remplacement. Pour surmonter cet obstacle, le transfert d'énergie sans fil (WPT) par couplage inductif résonnant apparaît comme une solution idéale et durable. Cette étude propose la conception, la modélisation et le développement pratique de liaisons inductives à haute efficacité adaptées aux exigences des implants biomédicaux. Opérant à la fréquence standard internationale de 13,56 MHz, le système proposé réalise un compromis optimal entre la sécurité biologique et l'efficacité de transmission aux distances requises.

Lors du calcul des caractéristiques électriques des liaisons, le travail s'est concentré sur le respect strict des dimensions miniaturisées de l'implant. Une analyse comparative rigoureuse a été menée entre trois configurations géométriques de bobines : planaire, solénoïde et hybride, à travers différentes distances de transfert. Les résultats expérimentaux ont démontré la supériorité de la configuration hybride planaire-solénoïde proposée (un émetteur planaire de 4,5 cm combiné à un récepteur solénoïde miniaturisé de 1,5 cm) comme solution optimale. Cette architecture hybride a permis de garantir un transfert de puissance maximal de 73,125 mW en espace libre et de maintenir une puissance efficace de 23,972 mW à une distance de 1 cm.

De plus, la sécurité biologique et la perméabilité électromagnétique ont été validées avec succès par des tests expérimentaux utilisant des tissus humains comme barrière biologique directe entre les bobines, assurant ainsi un flux d'énergie efficace et ininterrompu. Les résultats obtenus montrent une forte convergence avec les recherches internationales contemporaines, confirmant ainsi la faisabilité scientifique et pratique du concept hybride proposé pour le développement d'une nouvelle génération d'implants électroniques sûrs et durables.

Mots-clés : Dispositifs Médicaux Implantables (DMI), Transfert d'Énergie Sans Fil (WPT), Couplage Inductif Résonnant, Configuration Hybride, Géométrie des Bobines, Sécurité Biologique.

ملخص

تتطلب الأجهزة الطبية الإلكترونية القابلة للزرع داخل جسم الإنسان مصدراً موثقاً ومستداماً للطاقة الكهربائية لضمان تشغيلها بصورة مستمرة وفعالة. إلا أن الاعتماد على البطاريات التقليدية المدمجة يظل محدوداً بعمر تشغيلي لا يتجاوز في الغالب عشر سنوات، مما يستلزم إجراء عمليات جراحية متكررة لاستبدالها، وهو ما يزيد من المخاطر الصحية والتكاليف العلاجية على المرضى.

وللتغلب على هذه التحديات، يبرز نظام نقل الطاقة اللاسلكي (WPT) المعتمد على تقنية الاقتران الحثي الرنيني بوصفه حلاً واعداً وفعالاً لتزويد الغرسات الطبية بالطاقة دون الحاجة إلى التدخلات الجراحية المتكررة. وفي هذا السياق، تهدف هذه الدراسة إلى تصميم ونمذجة وتطوير رابط حثي عالي الكفاءة يتوافق مع متطلبات الأجهزة الطبية المزروعة.

ويعمل النظام المقترح عند التردد الصناعي والعلمي والطبي (ISM) البالغ 13.56 ميغاهرتز، لما يوفره من توازن مثالي بين كفاءة نقل الطاقة ومتطلبات السلامة الحيوية، مع ضمان أداء موثوق عند مسافات النقل المناسبة للتطبيقات الطبية الحيوية المزروعة.

ركز هذا العمل على حساب وتحليل الخصائص الكهربائية لروابط نقل الطاقة مع الالتزام الصارم بالقيود الهندسية والأبعاد الميكروية التي تفرضها تطبيقات الغرسات الطبية المزروعة داخل جسم الإنسان. ولتحقيق ذلك، أجريت دراسة تحليلية ومقارنة شاملة لثلاثة تكوينات هندسية مختلفة للوشائع، وهي: الوشائع المسطحة (Planar)، والوشائع اللولبية (Solenoid)، والوشائع الهجينة (Hybrid)، وذلك عند مسافات اقتران متعددة بين المرسل والمستقبل.

وقد أظهرت النتائج تفوق التكوين الهجين المقترح، المكوّن من وشيعة إرسال مسطحة بقطر 4.5 سم ووشيعة استقبال لولبية مصغرة بقطر 1.5 سم، باعتباره الحل الأمثل لتحقيق أداء مرتفع في نقل الطاقة لاسلكياً. حيث تمكن هذا التكوين من توفير قدرة منقولة قصوى بلغت 73.125 ميلي واط (mW) في حالة الاقتران المثالي، مع الحفاظ على قدرة فعالة قدرها 23.972 ميلي واط (mW) عند مسافة نقل تبلغ 1 سم، مما يؤكد ملاءمته لتطبيقات الأجهزة الطبية القابلة للزرع التي تتطلب كفاءة عالية وأبعاداً صغيرة.

علاوةً على ذلك، تم التحقق بنجاح من السلامة البيولوجية والنفاذية الكهرومغناطيسية للنظام من خلال اختباراتٍ تجريبية استُخدمت فيها أطرافٌ بشرية (عائق حيوي مباشر) بين الوشائع، مما أكد استمرارية تدفق الطاقة وكفاءته دون انقطاع. وقد أثبتت النتائج المحققة توافقاً وثيقاً مع الأبحاث العالمية المعاصرة، مؤكدةً في الختام الجدوى العلمية والتطبيقية للمفهوم الهجين المقترح في تطوير جيل جديد من الغرسات الطبية الإلكترونية الآمنة والمستدامة.

الكلمات المفتاحية: الأجهزة الطبية المزروعة (IMDs)، نقل الطاقة لاسلكياً (WPT)، الاقتران الحثي الرنيني، البنية الهجينة، هندسة الملفات الحثية، السلامة الحيوية.

GENERAL

INTRODUCTION

GENERAL INTRODUCTION

The advancements in technology over the past few years have greatly impacted the biomedical field, providing improved health solutions for patients [1, 2]. The progress in the design of implantable medical devices (IMDs) has created a new horizon in the healthcare industry, enhancing the utility of human organs [3]. A variety of these IMDs, including devices like gastric stimulators, biomedical sensors, cochlear implants, pacemakers, and pressure monitoring equipment, are now widely used for the diagnosis, treatment, and monitoring of a patient's health conditions [1]. Typically, batteries with limited lifetimes are considered for these IMDs [3]. However, invasive surgery is required to change the batteries after the end of a lifetime, which is painful, shocking, and life-threatening to the patients [1]. The wireless power transfer (WPT) system provides an ideal solution to this problem by eliminating the need for invasive surgery and also ensures the safety of patients [1, 4, 5]. The development of WPT systems for IMDs has attracted significant attention from the scientific community in the application of biomedical fields [5].

Wireless Power Transfer (WPT) offers an alternative solution by providing an electrical system for biomedical implants [1, 5]. that limits the need for battery replacement. In WPT, an external electrical source outside the human body transmits energy wirelessly to the implanted device

Chapter I provides a general overview and a comprehensive literature review of Wireless Power Transfer (WPT) systems for Implantable Medical Devices (IMDs). It begins by exploring the historical background, architecture, and strict power limitations of medical implants. Then, it establishes a clear classification between far-field radiative and near-field non-radiative transfer techniques (inductive, capacitive, and resonant). Finally, it outlines the critical biological safety standards, tissue characterization, and SAR limits, while justifying the selection of the 13.56 MHz ISM frequency band

Chapter II focuses on the theoretical analysis and mathematical modeling of resonant inductive coupling links. It provides a detailed analytical framework for the equivalent circuit model, calculating key parameters such as self-inductance, mutual inductance, and the coupling coefficient. Furthermore, a comprehensive comparative

study among the four primary compensation topologies is presented to demonstrate the advantages of the Series-Series (S-S) configuration, establishing the governing equations for circuit currents and Power Transfer Efficiency (PTE) at the resonance frequency

Chapter III represents the practical implementation and experimental validation of this research. It begins with a parametric and comparative simulation study of three distinct coil geometries (planar, solenoid, and hybrid) across various transfer distances to identify the optimal configuration. Following this, the chapter presents the experimental setup and laboratory measurements of the proposed prototype operating at 13.56 MHz, confirming the superiority of the innovative hybrid planar-solenoid design. The chapter concludes with the successful validation of the system's biological safety and electromagnetic permeability through a direct human tissue barrier, supported by a comparative analysis with contemporary global literature.

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Chapter I

Wireless Power Transfer for Biomedical Applications

I-1 Introduction

Over the past decade, wireless power transfer (WPT) has played a significant role in a multitude of applications and is widely used in near-field inductive coupling links. Currently, most bio-implanted devices are powered by induction, such as cochlear implants, retinal implants, and implanted microsystems used to stimulate and monitor nerve and muscle activity.

In this chapter, we will present the main architecture of implantable biomedical devices and their principal functions. Then, we will present the different methods of powering electronic medical implants (EMIs) using a wireless power transfer system. Generally, WPT systems can be classified into two methods: electromagnetic induction and electromagnetic radiation. For each method, we will detail the operating principle. We will focus on the study of wireless energy transfer techniques to power these implantable devices.

I-2 Implantable Biomedical Devices

An electronic medical implant is defined as any type of medical device that is implanted inside the body to elicit a physiological response. The first microelectronic medical implant was a pacemaker, introduced in 1970. Many other implants are currently in use, such as neurostimulators, implantable drug infusion pumps, urinary implants, cochlear implants, etc. Medical devices used to treat different diseases have different requirements and specifications; however, there are several characteristics shared by most of them.

Low Power Consumption: Modifying or charging batteries can be uncomfortable, difficult, expensive, and even risky for the patient; therefore, all implantable medical devices need to consume as little power as possible.

- **High Reliability:** A failure of an implantable medical device can lead to discomfort, pain, injury, or even death for the patient.

- **Low Amplitude Signals:** Most physical signals within the human body are on the order of millivolts (mV), which requires careful attention in detection and amplification.

- **Low Frequencies:** The natural frequency range of biological signals varies from a fraction of a hertz to a few kilohertz.

- **Small Size:** Implantable devices must be as small as possible to minimize their invasiveness to the human body.

We distinguish several types of implants devices figure I.3 represents a comprehensive map of modern implantable medical devices. It showcases the versatility of these technologies in treating sensory, cardiovascular, and neurological disorders, emphasizing the essential need for reliable and efficient power sources to sustain their operation

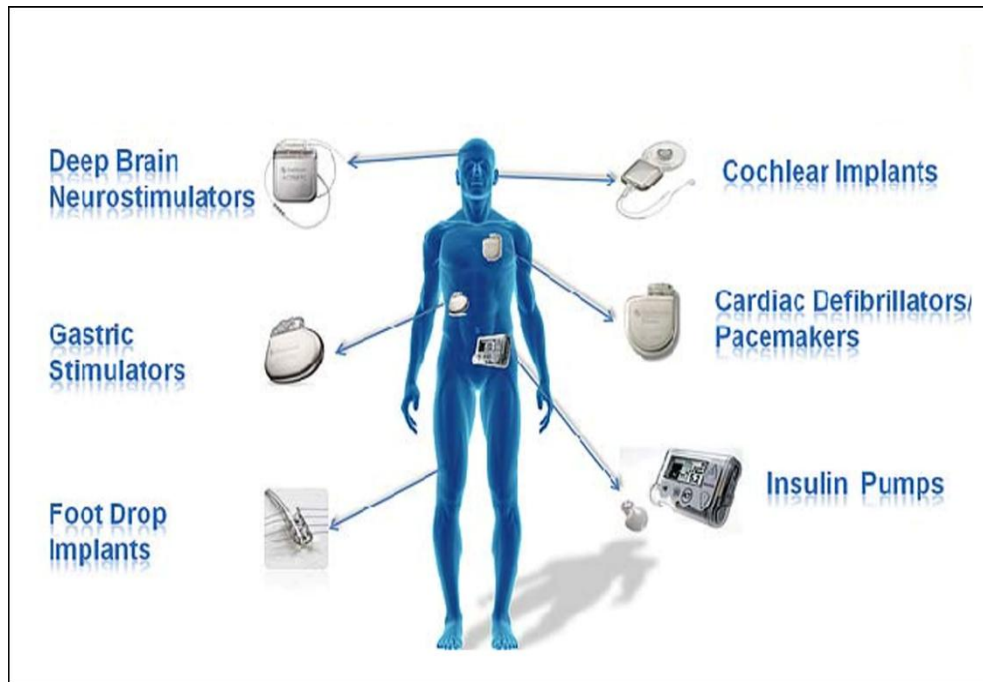


Figure I-1: Representative diagram of various implantable medical and them locations in the human body

I-2-1 Pacemaker Implant

A pacemaker, also called a cardiac pacemaker, is, as its name suggests, a medical device implanted in the human body designed to deliver regular electrical impulses to the myocardium (the heart muscle) and thus stimulate muscle contractions at a regular rhythm. It can deliver an electrical impulse for several years thanks to a battery. This pacemaker, usually implanted under the collarbone, is therefore

A pacemaker consists of two parts: The pulse generator: a small metal case which contains the battery of the pacemaker and the electrical circuit which senses the heart beats and produces the electrical signals.

The leads: thin insulated wires connected with the pulse generator at one end and a chamber of the heart on the other end. The leads are implanted via a vein in the chest below the clavicle and advanced to the heart. The electrodes deliver the electrical signals to the chamber that the pacemaker “paces”, controlling the rate of the heart rhythm. They also “sense” the electrical activity of the heart, sending this information back to the pulse generator

A pacemaker has two different functions. Firstly, it continuously observes the heart rhythm in order to detect abnormal rhythms (sensing). When heartbeats are too slow, the pacemaker stimulates the heart muscle by sending electrical signals which correct the slow heart rhythm (pacing). In other words, the pacemaker “senses” the heart rhythm continuously, but “paces” the heart only when needed. Pacemakers have additional functions in order to optimize

the heart rhythm. One of these functions is the sensor function (so called rate response). A sensor detects physical activity (either via the body motion, the rate or cardiac contractility) and based on that the pacemaker increases the heart rate when needed (e.g., during exercise breathing).

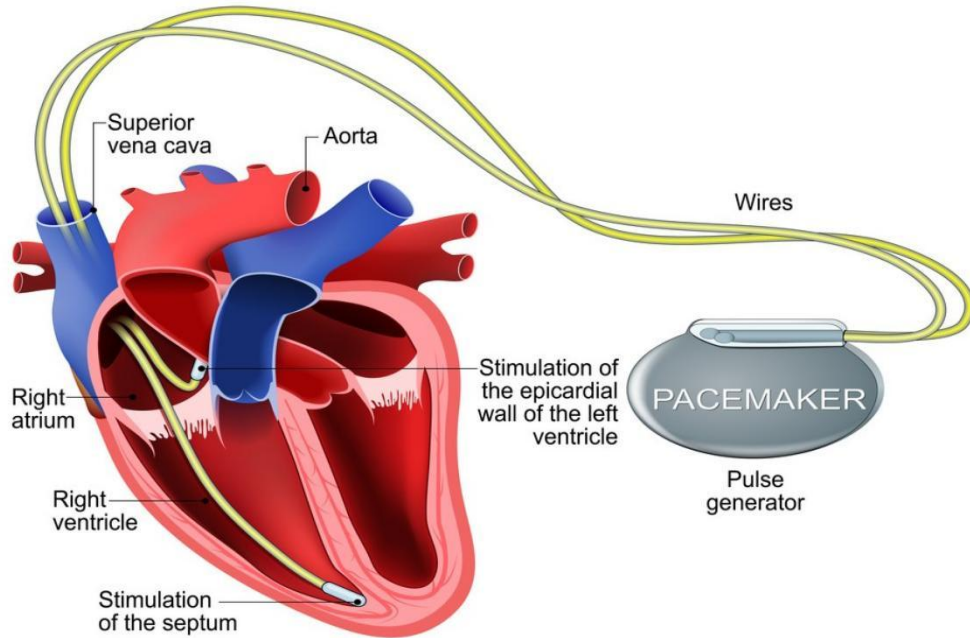


Figure.I-2: Artificial cardiac pacemaker

I-2-2 A Cochlear Implant

A cochlear implant is an electronic implant designed to provide a certain level of hearing for some people with profound or severe deafness and for those suffering from tinnitus. It consists of two distinct parts, one of which cannot function without the other: The external part is removable and worn by the patient behind the ear. It includes the speech processor, which captures sounds through its microphones and analyzes them, and a magnetic antenna that communicates through the skin with the internal part. interne [1].

The internal part is the implant itself, surgically implanted under the skin behind the ear. It receives information from the external part and sends electrical impulses to the auditory nerve via an electrode implanted in the cochlea, the key organ of hearing [1].

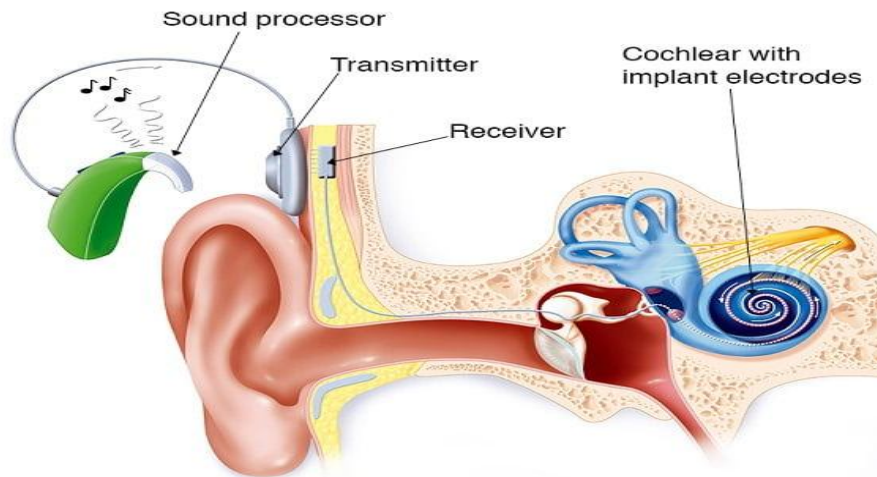


Figure I-3: A cochlear implant

I-2-3 Retinal Prosthesis

Retinal implants are devices used to induce visual function in visually impaired patients by electrically stimulating the remaining intact neurons. A visual prosthesis system comprises two main parts. The first unit is located in the intraocular region and includes an array of microelectrodes and related electronic circuits, while the other part is located outside the eye and includes a processor, a camera, and an external antenna [2], [3]. Images taken from the camera mounted on a lens are processed and transferred into a stimulation pattern that is applied to the electrodes simultaneously. Therefore, the system must be continuously powered [4].

Powering visual prostheses is one of the biggest challenges for high-performance systems. To date, various methods have been used to provide visual prostheses. Because space within the eye is quite limited, using a battery inside the eye is not even an option. Wireless power transfer is an option, and one that many researchers prefer these days. In this near-field power transfer method, inductive coils are used [5], [6]. The simplest coil-link design consists of two coils with a gap between them. The first coil, powered by a source, generates a magnetic field that connects the secondary coil and induces the application of a voltage to the load. Thus, the load is supplied wirelessly via inductive linkage to the coil.

Figure I-6 shows the wireless power transfer system for a retinal prosthesis.

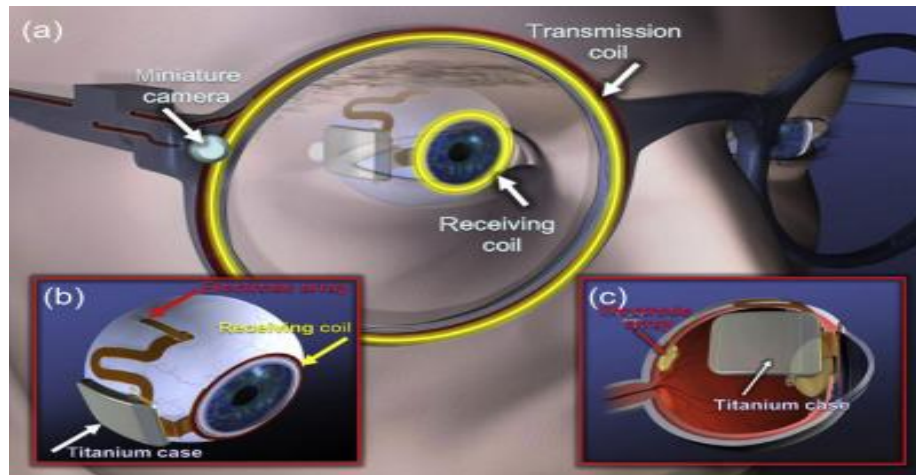


Figure.I-4: Retinal Prosthesis

I-3 Wireless power transfer (WPT)

Wireless power transfer, in its simplest form, is the process of transmitting electrical energy from a power source to an electrical load without the use of conductive wiring. The concept rests on the underlying principles of electromagnetic fields and resonances, where energy is transferred over a distance through various modes inductive coupling, magnetic resonant coupling, and electromagnetic radiation.

The idea behind wireless power transfer (WPT) technology is the elimination of galvanic connections such as wires, cables, traces, etc., to power electrical and/or electronic devices.), as schematically shown in Figure I-4. The coupling path between a transmitter and a receiver is mainly or exclusively in air and therefore can be capacitive inductive, or radiative. Capacitive coupling is based on electric field and is described by quasi-static-electric field equations. Inductive coupling is based on magnetic field and is described by quasi-static magnetic field equations. Radiative coupling is based on electromagnetic field and is described by Maxwell's complete equations. WPT based on the capacitive and inductive couplings is often referred to as near-field WPT, while that based on radiative coupling is referred to as far-field WPT.

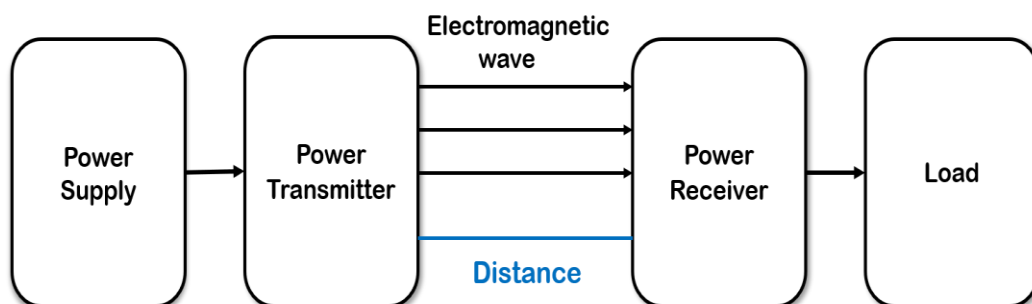


Figure I-5: Block diagram of a WPT system

I-4 Objectives of Wireless Power Transfer

Wireless power transfer technology is used in various fields, for example, biomedical applications (medical implants) or industrial applications (mobile phones), etc. Indeed, it is used for many purposes, such as:

- 1- Removing wires (reducing energy costs) and replacing batteries (protecting the environment from pollution).
- 2- Increasing the reliability of electronic devices by avoiding connector and cable problems.
- 3- Avoiding problems related to electrical devices, such as conductive interconnections.
- 4- Reducing costs by powering multiple devices from a single source.
- 5- In the medical field, it can help avoid surgery and reduce risks to patient health [7].

I-5 Applications of Wireless Power Transfer Systems

A wireless power transfer system offers numerous applications and has become an important part of several fundamental fields. First, in the transportation sector, wireless power transfer for electric vehicles is a promising technology. Second, in implantable medical devices (IMDs), it provides a convenient, reliable, and safe way to provide painless wireless power to patients. Third, in consumer electronics applications, wireless power transfer is used in LED televisions, charging portable devices such as mobile phones, tablets, and other devices. Power transfer technology and systems are used in LED lighting, underwater, detection, military defense systems, space applications, industrial robots, and wireless sensor networks (WSNs).



Figure I-6: Wireless power transmission applications in diverse sectors.

I-6 Classification of WPT Systems

Wireless power transfer can be divided into two main categories: electromagnetic induction and electromagnetic radiation. Electromagnetic induction can be divided into three types: inductive power transfer (IPT) based on magnetic fields [8], capacitive power transfer (CPT) based on electric fields [9], and acoustic power transfer based on acoustic waves [10].

Electromagnetic radiation can be divided into two categories: microwave power transfer (MPT) and laser power transfer. The methods differ in the type of coupling between the transmitting (TX) circuit and the receiving (RX) circuit. For example, inductive link energy transfer uses inductive coupling (mutual inductance between TX and RX), capacitive energy transfer uses capacitive coupling (capacitance between TX and RX), and radiative energy transfer uses

electromagnetic radiation to transmit energy from TX to RX [11], [12]. Most near-field wireless energy transfer methods applied to the biomedical field use operating frequencies between 100 kHz and 50 MHz.

The different methods of the wireless energy transfer system are illustrated in the following Figure I-7:

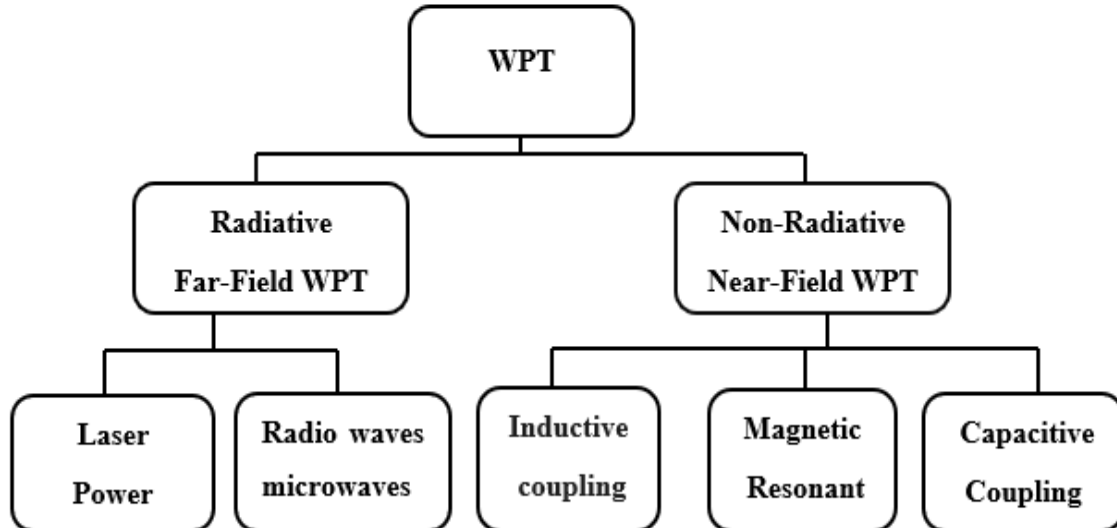


Figure I-7: Different methods of wireless energy transfer [13]

I-7 Near-Field WPT (Non-Radiative)

Near-field Wireless Power Transfer, strictly defined as non-radiative coupling, operates within the reactive near-field region where the transmission distance (d) is much smaller than the wavelength (λ) of the electromagnetic field ($d \ll \lambda/2\pi$) [14].

In this regime, energy is stored in the oscillating field rather than radiating into free space. This category is fundamentally divided into two topologies based on the field component utilized for energy exchange: Inductive Power Transfer (IPT), which relies on magnetic fields (H-field), and Capacitive Power Transfer (CPT), which utilizes electric fields (E-field).

I-7-1 Inductive Coupling Method

This is the fundamental principle of electrical transformers. The primary and secondary coils are electrically isolated; therefore, energy transfer occurs through electromagnetic coupling, typically facilitated by a common permeable magnetic core.

This near-field energy transfer method relies on Ampère's and Faraday's laws. An electric current flow through the transmitter coil, which induces a magnetic field around it. The receiving device contains a receiver coil that, once placed within this magnetic field, converts the field's energy back into an electric current. The efficiency of the energy transmission depends on the coupling factor between the inductors (coils) and their respective quality factors [15][16]

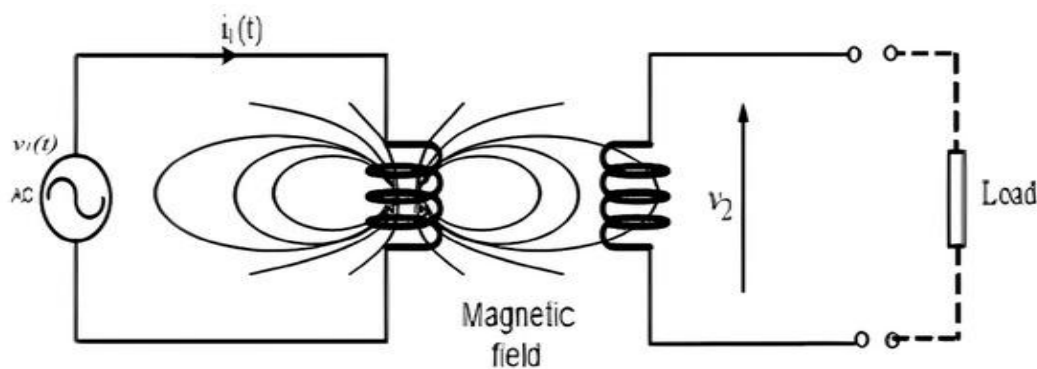


Figure I-8: Schematic diagram of Inductive coupling technology

I-7-2 Magnetic Resonant Coupling Method (MRC)

This method utilizes a coil and a capacitor as a resonator, with electrical energy transmitted via electromagnetic resonance between the transmitter and receiver coils. By matching the resonant frequencies of the two coils with a high-quality factor, energy transmission can be achieved over longer distances, despite weak magnetic coupling. Resonant Wireless Power Transfer (WPT) is considered an improvement over conventional inductive methods regarding distance and efficiency, allowing for electrical energy transmission over several meters. Efficiency is enhanced through resonance, which concentrates the magnetic field toward the receiver by maintaining the same resonant frequency. Furthermore, this

technique offers greater flexibility regarding the positioning of the receiver coil relative to the transmitter coil [17][18].

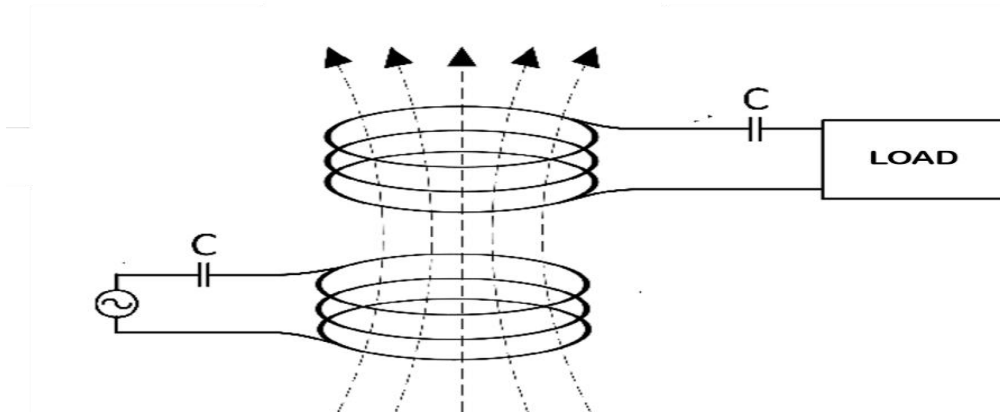


Figure I-9: Schematic diagram of magnetic resonant coupling technology

I-7-3 Capacitive Power Transfer (CPT)

Unlike the previously mentioned near-field Wireless Power Transfer (WPT) technologies, capacitive WPT relies on an electric field. This method utilizes two pairs of metal plates: the first pair is integrated into the power transmitter, while the second is integrated into the receiver. On each side, the two plates are connected across the power source (for the transmitter) and the load (for the receiver) [19- 20]

the plates at each end of the transmitter are parallel to the corresponding plates of the receiver. When the two pairs of plates are sufficiently close, they act as two capacitors, thereby closing the electrical circuit. These capacitors are identified as forward and return capacitors. Under these conditions, capacitive coupling is produced, and an electric field is generated between the plates, inducing an electric current in the power receiver [21]

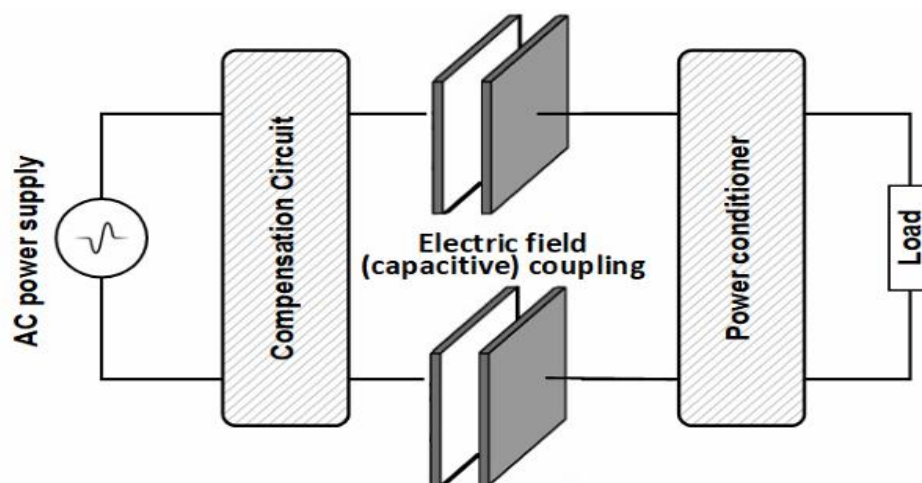


Figure I- 10: The simple structure of a capacitive power transfer system

I-8 Far-Field WPT (Radiative)

Far-field wireless power transfer, scientifically termed radiative power transfer, operates in the Fraunhofer region, where the transmission distance (d) significantly exceeds the wavelength ($d \gg 2D^2/\lambda$), where D is the antenna dimension). Unlike near-field methods that rely on reactive magnetic or electric fields, far-field systems transmit energy via propagating transverse electromagnetic (TEM) waves.

I-8-1 Wireless Power Transfer Laser Beams

Optical WPT systems rely on electromagnetic waves with frequencies in the THz range for power transfer [22-23]. This type of wave requires that the power transmitter and receiver be in line (no obstacles). If this condition is met, the power transfer distance can reach several kilometers. The first experiment with this technology involved moving a mini-rover over a distance of up to 280 m. The laser output power was 5 W, while the power required by the vehicle was 1 W. Advances in laser technology and photovoltaic cells will improve this technology.

A schematic diagram illustrating the operating principle of optical WPT technology is shown in Figure I.10. The laser diode in the transmitter is controlled by the mode of the current in the generator. A beam director is used to adjust the direction of power transfer. In the receiver, a photovoltaic cell converts the received light into energy using the corresponding power converters. The direct current is then used to power a load or a battery.

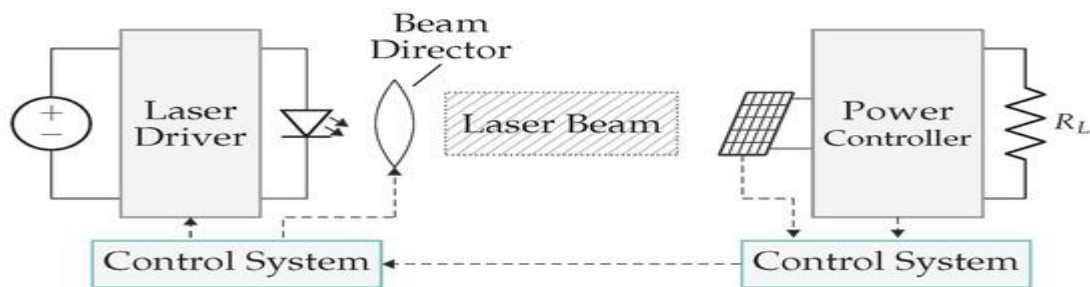


Figure I-11: Schematic Diagram of a Wireless Laser Power Transfer System

I-8-2 Radio waves and microwaves

Radio waves correspond to waves between 1 m and 10^2 m, while microwaves correspond to waves on the order of 10^{-2} m [24]. These waves are electromagnetic waves capable of carrying both energy and information. The source antenna emits an electromagnetic wave that propagates to the target receiving antenna, which opposes the work of the transmitting antenna: it transforms the electromagnetic wave into an electrical signal. The signal is demodulated, either in amplitude or frequency, then amplified and transformed into information by the speakers. These systems allow the transmission of energy over distances much greater

than the diameter of the transmitters [25][26]. Radio frequency or microwave (RFEM) transmission occurs in three stages: the emission of a wave carrying the energy to be transmitted, the propagation of this wave, and the reception and conversion of the wave into usable electrical current. This transmission principle relies on the fact that a wave propagates in a vacuum while conserving its energy, this energy being attenuated depending on the medium through which the wave travels.

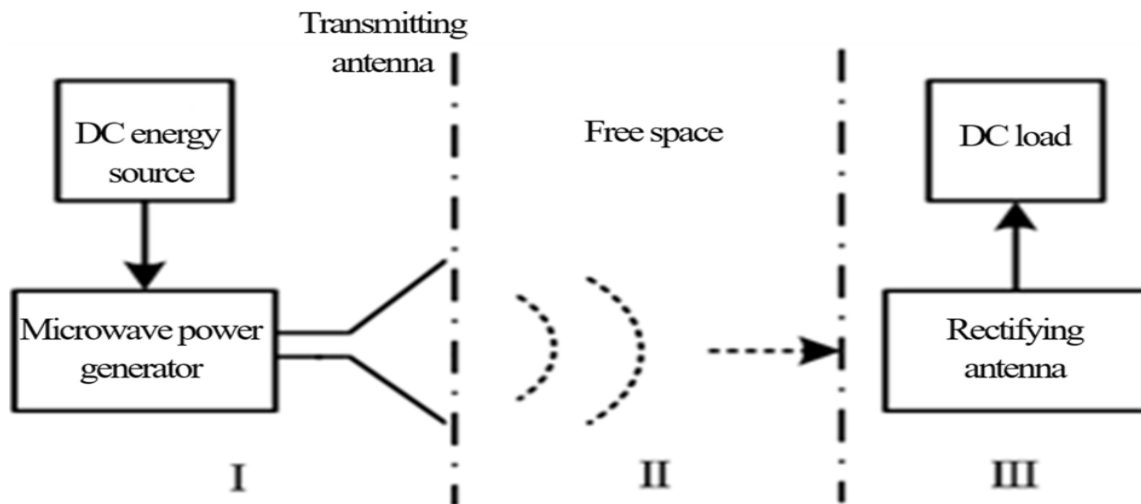


Figure I-12: Schematic diagram of a microwave power transmission system

I-9 Comparison Between Different Wireless Power Transfer Technologies

The characteristics of the five types of transfer technologies are summarized in Table I-1. Depending on the requirements of biomedical applications, directivity and transfer range are of secondary importance. However, penetrability and energy efficiency are extremely critical. Consequently, inductive coupling and capacitive coupling appear to be better suited for biomedical applications due to their superior efficiency and enhanced penetration capabilities

Technology	Distance between Transmitter Receiver	Frequency	Directivity	Efficiency	Penetrability
Inductive coupling	Short	Weak Hz ~ MHz	Weak	High	Forte
Capacitive coupling	Short	Weak Hz ~ MHz	Weak	High	Forte
Microwave	Long	High GHz ~ THz	Forte	Weak	Weak
Laser	Long	High > THz	Forte	Weak	Weak

Table I-1: Comparison between different wireless power transfer technologies [27].

I-8 Conclusion

In this chapter, we began by presenting an overview of implantable devices in the human body and their various functions. After defining the wireless power transfer (WPT) system, explaining its basic principle and objectives, and outlining the different transfer methods, we presented various applications of WPT in implantable biomedical devices. Furthermore, critical considerations regarding human exposure to electromagnetic fields were described. The WPT system stands out as the most efficient and safest technique for the human body's biological environment when charging the batteries of electronic medical implants. The complete elimination of cables and the global deployment of WPT technology across all fields (medical, industrial, etc.) represents a major technological paradigm shift that is currently revolutionizing the scientific world. However, strict compliance with safety limitations regarding human exposure to electromagnetic fields remains imperative to prevent any adverse health effects. In the next chapter, we will study in detail the theory and basic architecture of a resonant inductive link energy transfer system applied to implantable biomedical device

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Chapter II:

Theoretical Analysis of the Resonant Inductive Link (WPT)

II -1 Introduction

Recently, most bio-implantable devices have been powered using implanted batteries, which can lead to infections and health risks for the patient. Because implanted batteries have chemical side effects and a limited lifespan, researchers have developed a safer alternative method for powering implantable devices using inductive links to transfer electrical energy. The magnetic inductive link approach is considered the most promising technique for implantable devices. Its advantages include the continuous availability of high energy levels for an implanted device and the ability to control it with the external device using the magnetic link. Furthermore, the battery's lifespan does not limit the lifespan of the implanted device. In other words, the battery's lifespan and the implanted medical device's lifespan are different. Therefore, it can be used long-term and during the patient's activities. In this chapter, we provided an overview of wireless power transfer via an inductive link. We began by reviewing the different topologies of resonant inductive links and the operating principle of each. Next, we modeled and analyzed a series-parallel (SP) resonant inductive link. Then, we presented the method for calculating wireless power transfer efficiency as a function of various electrical parameters. Finally, we presented the quality factor of the resonant inductive link, including both the power transfer circuit and the implant circuit.

II-2 Operating Principle of an Inductive Link

The behavior of an inductive link comprising a primary circuit (transmitting system) and a secondary circuit (receiving system) can be addressed at two different levels of abstraction: a description of the electromagnetic fields and a description at the system level. These two abstractions are complementary; however, it can be difficult to move from one to the other. The electromagnetic description is based on the geometry and physical composition of the linking elements, but it does not provide direct information on how electrical signals propagate in the energy transfer link. Therefore, a circuit-level description can be used to deduce the characteristics of the link, but the key parameters—that is, the circuit components such as inductances, parasitic elements, and the coupling coefficient—can only be extracted from the description of the Electromagnetic fields. This parameter extraction is not always an easy task due to the complexity of the equations arising from the geometry of the device

. To do this, we will present an analytical method that allows us to calculate these electrical parameters in the case of two coils coupled together, arbitrarily located in space.

II-3 The Architecture of an Inductive Link for Bio-Implanted Devices

In general, the structure of the inductive coupling link for bio-implanted devices consists of two coils: an external coil (primary coil) placed outside the human body and an internal coil (secondary coil) located inside the human body a short distance from the external coil

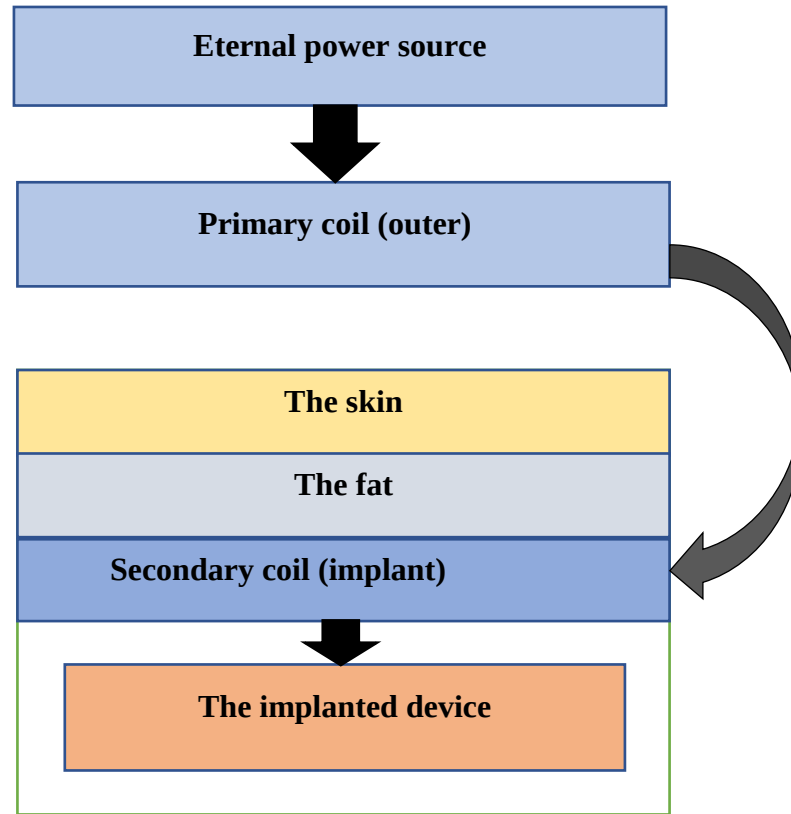


Figure II-1: Diagram of an inductive link of two coupled coils.

Bio-implantable devices optimized for the smallest possible size must be implanted according to the functional depth in human biological tissue. This is generally less than 10 mm deep; for example, microsystems implanted in the human body are placed between 1 and 4 mm deep. Cochlear implants are placed between 3 and 6 mm deep, and for retinal implants, a depth of 5 mm is required [1]. Currently, the study of a wireless power transfer system for bio-implantable devices is focused on the design and optimization of the transmitting and receiving coils (implants) [2].

II-4 Appropriate Frequency Selection for Bio-Implanted Devices

The carrier frequency is important in the design of implantable devices and bio-telemetry systems. Most implantable devices are powered by low frequencies. However, the standard safety level regarding human body exposure to RF electromagnetic fields is between 3 kHz and 30 GHz. Figure II-2 shows the possible frequencies that can be used for wireless

power transfer, called the industrial-scientific and medical (ISM) band. The ISM band is a range of license-free frequencies that are used in scientific and industrial applications without interfering with other communication systems.

Most wireless power transfer (WPT) systems operate at lower frequencies to achieve high efficiency and reduce losses [3].

The industrial, scientific, and medical band deals with frequencies ranging from 125 kHz to 135 kHz, 6.78 MHz, 13.56 MHz, 27.125 MHz, and 40.68 MHz, as well as 433.92 MHz, 869 MHz, and 2.4 GHz in a very high frequency band. A lower frequency is preferred, especially for biomedical implants, because body tissues are lossy media at high frequencies, and WPT systems operating at a lower frequency are not affected by the surrounding environment. The 3 kHz to 30 MHz frequency range is widely used in transcutaneous wireless systems due to its ability to penetrate water and skin over a short distance. Furthermore, the tissue heating effect in this band is negligible or nonexistent. In fact, 13.56 MHz is considered the most suitable frequency for research. Indeed, it provides a compromise between the strength of the permitted magnetic field, the transmission range, and biocompatibility [4].

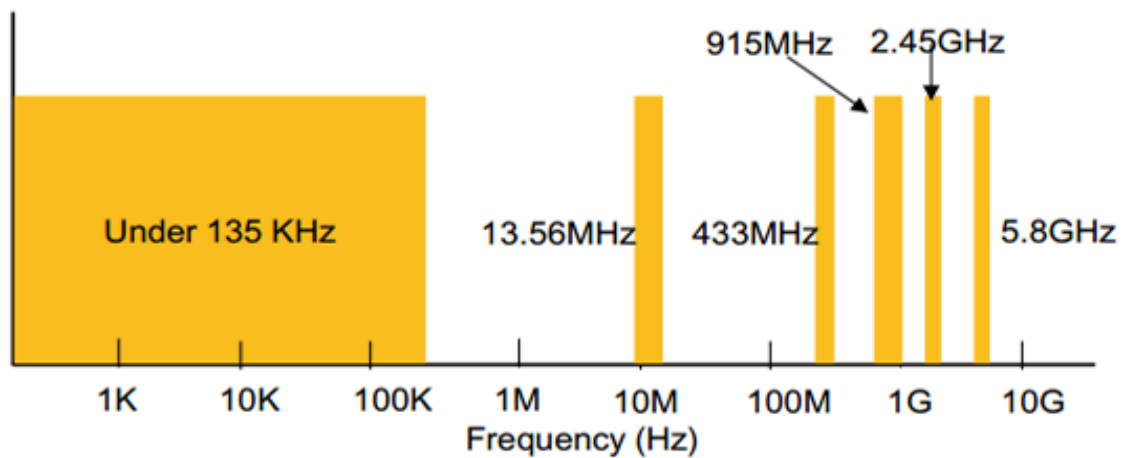


Figure II-2: Distribution of industrial, scientific and medical (ISM) bands [3]

II-5 Magnetic Resonant Inductive Coupling (MRIC)

The Magnetic Resonance method of energy transfer was developed as a solution to the problem of efficiency decline in energy transfer. This occurs when the distance increases between the power transmitter (external coil) and the power receiver (implanted coil). Magnetic resonance is a specific case of inductive coupling, where the energy transfer efficiency between the transmission and reception coils is determined by a value called the coupling coefficient [5].

II-6 Basic topologies of wireless power transfer systems

In order to maximize energy transfer efficiency and reduce losses caused by poor coupling between the transmitter and receiver coils, it is necessary to compensate for the coils' inductance by adding series or parallel capacitors that resonate with them. Fundamentally, adding a series capacitor introduces only negative reactance without modifying the real part of the impedance. On the other hand, adding a parallel capacitor modifies both the real and imaginary parts of the impedance. The configuration of the compensation topology can be selected based on load limitations, source impedances, and the specific application requirements for wireless power devices.

Figure II-3 illustrates the four topologies of a resonant inductive link applied for charging implantable devices. Primary and secondary capacitors (C_T , C_R) are added in series or parallel to resonate with their corresponding inductances (L_T , L_R). The addition of these capacitors improves efficiency over longer distances compared to non-resonant inductive coupling. The four typical topologies for inductive wireless power transfer are: Series-Series (SS), Series-Parallel (SP), Parallel-Parallel (PP), and Parallel-Series (PS).

Several principles guide the selection of the optimal compensation topology, depending on the application's requirements and limitations. In the design methodology, the following rules should be considered:

- For a variable load: A series primary capacitor is added, as C_T in Series-Series (SS) or Series-Parallel (SP) topologies is independent of the load [6].
- For variable magnetic coupling: A series secondary compensation capacitor (C_R) is added, as both C_T and C_R in the Series-Series (SS) topology are independent of the mutual inductance [7].
- For fixed magnetic coupling and load: The most suitable secondary compensation topology is chosen based on the load values and the secondary inductance [8], [9].
- For systems with significant parasitic parallel capacitance: Adding parallel compensation is the best-suited topology, as the parallel resonant capacitor can incorporate the coil's parasitic capacitance [10].
- For systems dominated by coil resistance: Parallel compensation is required when the real part of the input impedance is dominated by the coil's resistance, since adding a parallel capacitor affects the real part of the equivalent impedance [11]

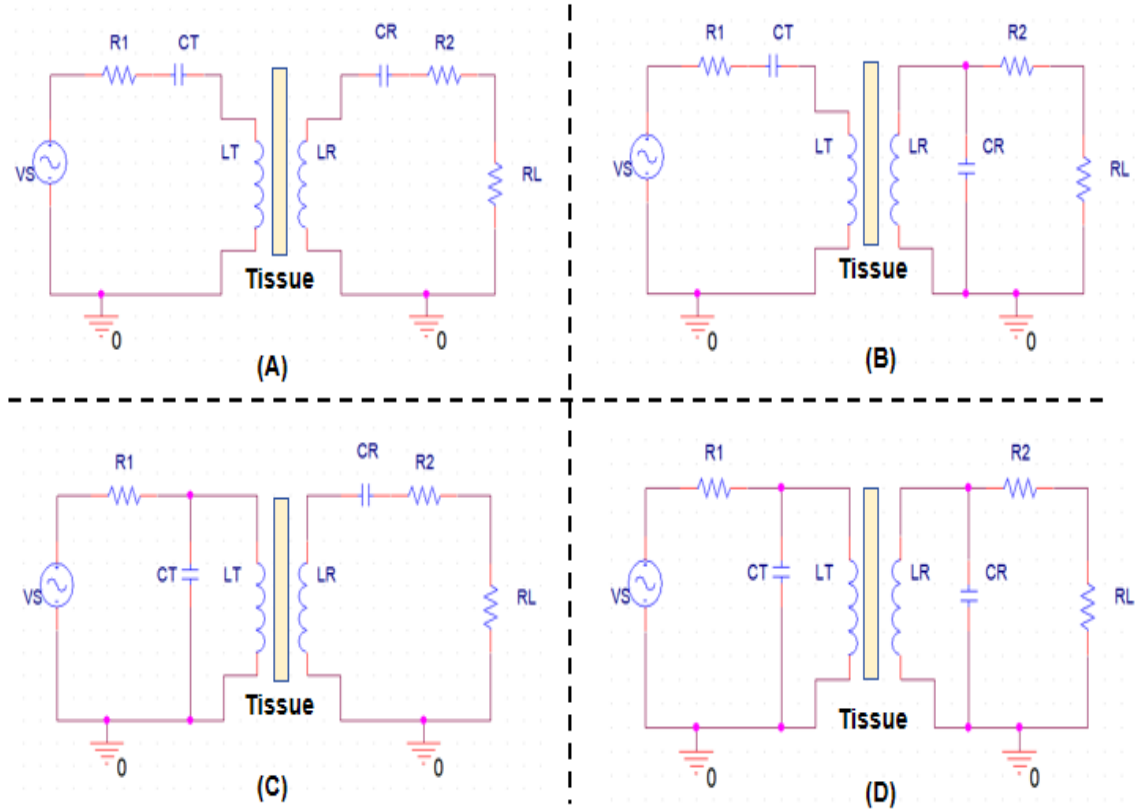


Figure II-3: different topologies of the resonant inductive coupling system:

(A) series - series, (B) series - parallel, (C) parallel - series, (D) parallel - parallel.

II-7 The circuit of a resonant inductive link

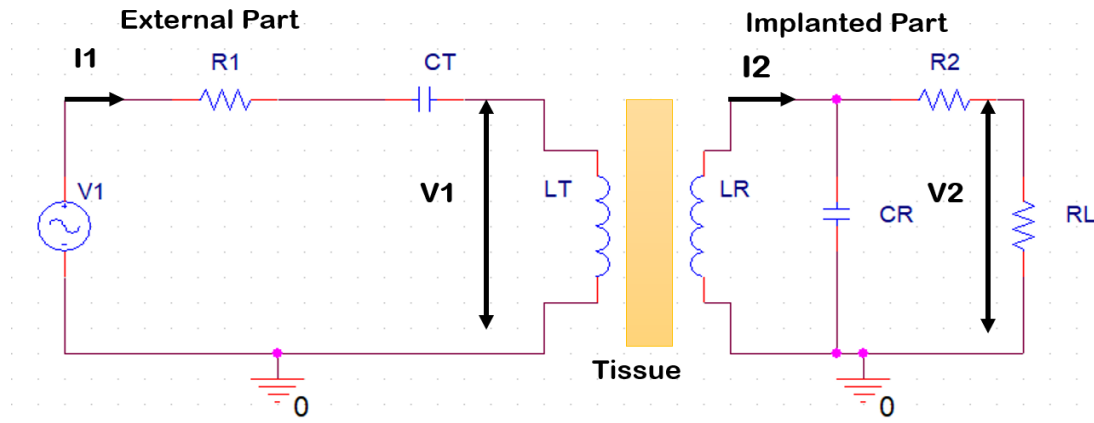
Typically, a Wireless Power Transfer (WPT) system utilizing a resonant inductive link consists of two resonant RLC circuits: the transmitter (primary) circuit, which represents the external unit, and the receiver (secondary) circuit, which is the implanted unit.

The primary circuit is configured in series resonance to provide a low-impedance load, while the secondary circuit is almost always configured in parallel resonance. The operating resonant frequency is set at 13.56 MHz, a frequency chosen within the Industrial, Scientific, and Medical (ISM) band to minimize tissue heating [12].

To achieve optimal energy transfer efficiency, both RLC circuits are tuned to the same resonant frequency, f . The external RLC circuit, located outside the human body, is tuned to series resonance to present a low-impedance load, thereby ensuring maximum power delivery from the source to the implanted circuit. Conversely, the internal RLC circuit is implanted

within the body and tuned to parallel resonance to enhance electrical performance and efficiency across a wide range of load resistances [13], [14].

Figure II-4 illustrates the Series-Parallel (SP) resonant inductive link coupling used to power bio-implanted devices.



Figure

II-4: Resonant inductive link: series-parallel topology

Where L_T and L_R are the self-inductance of the external coil and the implanted coil, respectively. Resistors R_1 and R_2 represent the equivalent series resistance of the primary and secondary coils, respectively. (C_T) and (C_R) are the resonant capacitors of the external and implanted circuits, respectively, and the load resistance R_c represents the battery of the implantable electronic device. The degree of coupling between these two coils can be described in terms of mutual inductance [15]. The mutual inductance between the external and implanted circuits is defined by expression (II-1).

$$M = \frac{\mu_0 \pi (r_{Tx} r_{Rx})^2 N_{Tx} N_{Rx}}{2d^3} \quad \text{II-1}$$

where μ_0 is vacuum permeability; r_{Tx} , r_{Rx} , N_{Tx} , N_{Rx} represent radius and number of turns for the Tx and Rx coils, respectively; and d is the transfer distance between the Tx and Rx coils. In addition, k , L_{Tx} , and L_{Rx} are the resonant network's coupling coefficient and Tx and Rx coil inductance, respectively

The coupling coefficient, denoted as k , is primarily determined by the geometric parameters of the coils, including their shape, dimensions, and relative orientation (distance and angular misalignment). Additionally, the electromagnetic properties of the surrounding medium

significantly influence this value [20]. Physically, k represents the fraction of the magnetic flux generated by the primary (external) coil, L_T , that links with the secondary (implanted) coil, L_R . This coefficient ranges from 0 to 1 and is calculated using expression (II-2) [17], [18].

$$k = \frac{M}{\sqrt{L_T L_R}} \quad \text{II-2}$$

When the coupling coefficient k is equal to 1, the coupling between the coils is maximal, and when the coupling coefficient k is equal to 0, the coupling between the coils is minimal.

The coefficient k is the main factor used to determine the amount of energy that can be delivered to the implanted devices [16], [17].

The external inductance L_T is calculated using expression (II-3)

$$L_T = \frac{1}{C_T \omega_0^2} \quad \text{II-3}$$

Where $\omega_0 = 2\pi f$ is the operating frequency.

The internal inductance L_R is calculated using expression (II-4)

$$L_R = \frac{1}{C_R \omega_0^2} \quad \text{II-4}$$

The series resistance of the implanted secondary coil is expressed by (II-5)

$$R_2 = \frac{R_C}{1 + C_R^2 \omega_0^2 R_C^2} \quad \text{II-5}$$

Figure II-5 represents the equivalent electrical circuit model of resonant inductive coupling (SP), with the primary part (external circuit) placed in air and the secondary part placed in the human body (implanted circuit).

The parasitic resistance R_1 is modeled after the series resistance of one turn of the inductor.

This resistance also includes the reflected resistance in the emitter, which is very small and can be ignored. M is the mutual inductance between the emitter inductance L_T and the receiver inductance L_R .

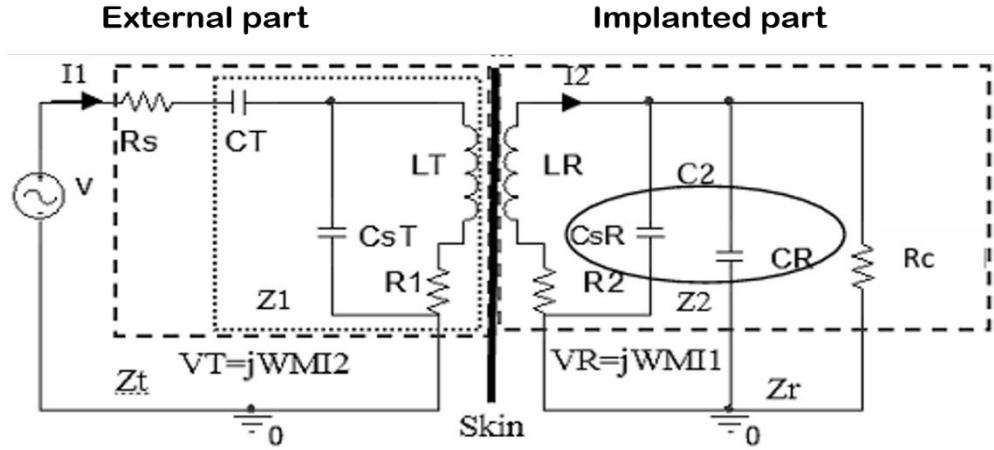


Figure II-5: Simplified model of the resonant inductive coupling link

Each turn is equivalent to an inductance L in series with a resistance R , and both are in parallel with a capacitance C . R_1 and R_2 characterize the parasitic resistances of the inductances LT and LR (loss resistances in the inductors), respectively. C_{ST} and C_{SR} characterize their parasitic capacitances of the primary and secondary coils [23], respectively. Capacitors C_T and C_R are added to the primary and secondary circuits to resonate the coils LT and LR , respectively, and should be chosen to have approximately the same values. R_c represents the load resistance, and a simpler analysis is $C_2 = C_{SR} + C_R$. The energy transfer efficiency is calculated using the conceptual analysis of the reflected impedance of the receiver and transmitter (Z_t and Z_r). In general, the electrical parameters (Z_t and Z_r) of an inductor depend on the frequency expressed by [(II-6) - (II-7)]:

$$Z_T = \frac{V_T}{I_1} = \frac{\omega^2 M^2}{Z_2} \quad (\text{II-6})$$

$$Z_r = \frac{V_R}{I_2} j\omega M \frac{I_1}{I_2} \quad (\text{II-7})$$

Where I_1 and I_2 are respectively the currents in the primary circuit and the implant circuit, and $\omega = 2\pi f$ is the angular frequency measured in radian/s

II-8 Load resistance for an IPT system

In resonant inductive link wireless power transfer systems, there is an optimal load that maximizes energy transfer efficiency for each distance between the transmitter and the

receiver. Furthermore, there is an optimal load that maximizes the energy transferred to the load [10].

The electrical circuit power efficiency (PTE) of the system, which is the ratio between the input power of the external circuit P1 and the power of the embedded circuit P2, is determined as follows

$$PTE = \frac{P_2}{P_1} \quad \text{II-8}$$

The equivalent impedance of the secondary circuit is expressed by expression (II-9) [19]:

$$Z = \frac{M^2 \omega^2 (R_C + \sqrt{R_C^2 + 4\omega^2 L_R^2})}{2\omega^2 L_R^2 + R_2 R_C + R_{LR} \sqrt{R_C^2 + 4\omega^2 L_R^2}} \quad \text{II-9}$$

To ensure that (II-9) is true, the following condition must be met [25]

$$R_C^2 + 4\omega^2 L_R^2 > 0, \omega = 2\pi f \quad \text{and} \quad R_C \geq 4\pi f L_R \quad \text{II-10}$$

According to equation (II-10), we calculate the implemented load resistance R_C . The load R_C satisfies the requirement:

$$R_C = 300 \, \Omega > 170.4 \, \Omega$$

II-8 Quality Factor (Q)

The Quality Factor (Q) of a coil is a critical parameter that significantly influences the energy transfer efficiency in inductive coupling systems. It characterizes the performance of the inductor by representing the ratio between its reactance and the real part of its impedance (parasitic resistance).

In high-frequency applications, such as WPT for implantable devices, the skin effect must be taken into account as it increases the AC resistance of the coil, thereby affecting the Q factor. Physically, a higher Q factor indicates a lower rate of energy loss relative to the energy stored within the coil.

The quality factors for the external (transmitter) and implanted (receiver) coils, denoted as QT and QR respectively, are defined as follows [20]:

$$Q_T = \frac{2\pi f L_T}{R_1} \quad \text{II-11}$$

$$Q_R = \frac{2\pi f L_R}{R_2} \quad \text{II-12}$$

$$\text{With } f = \frac{1}{2\pi(L_t C_T)^{\frac{1}{2}}} = \frac{1}{2\pi(L_R C_R)^{\frac{1}{2}}} \quad \text{II-13}$$

II-9 Properties of Certain Biomedical Devices Using Wireless Power Transfer

Biomedical implants are artificial devices implanted in the human body to replace damaged organs responsible for vital bodily functions. Due to the location of these devices and the risk of damaging surrounding tissues, establishing a wired connection is not suitable for long-term use. Therefore, devices implanted in the body must be charged wirelessly [21]. Implants are used in many different areas of the body for specific applications, such as deep brain stimulation used to treat movement disorders, chronic pain, cardiac pacing (pacemakers), and prostheses such as cochlear implants [23]. Implantable biomedical devices differ in certain parameters, such as the size of the receiving coil, the distance between the transmitting and receiving parts, and the energy required for each implanted device to function. Table II-1 presents the properties of certain parameters of implanted biomedical devices using wireless power transfer as their power source.

Type of Implant	Energy	Distance Between Coils [mm]	Coil Size Implant
Cardio	15 mW– 20 W	10 – 25	400 mm ²
Cochlear	15 mW	10 – 25	100 mm ²
Neural	20 mW	5 – 25	100 mm ²
Retinal	50 m W	10 – 25	64 mm ²

Table II-1: Properties of some biomedical devices using wireless energy transfer via resonant inductive link [22]

II-10 Conclusion

In this chapter, we first presented a wireless power transfer system using a resonant inductive link suitable for bio-implantable devices. Initially, a general overview of the different topologies of a resonant inductive link was presented, along with its characteristics.

Next, we presented a model of the electrical parameters of a series-parallel resonant inductive link. To do this, we developed an analytical model allowing for the accurate evaluation of the electrical parameters. Finally, we propose a simple and accurate method for calculating the energy transfer efficiency. The maximum energy transfer efficiency in the inductive link is obtained for a maximum coupling coefficient and a high-quality factor.

Wireless energy efficiency depends on the resonant frequency (or operating frequency), distance, alignment, and the suitability of the coupling between the transmitter circuit and the receiver circuit (implant circuit).

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Chapter III:

**Electrical Optimization and
Experimental Validation of an Inductive
Link**

III-1 Introduction

Wireless power transfer (WPT) systems based on resonant inductive links have attracted significant interest for powering implantable biomedical devices due to their ability to transfer energy efficiently without physical connections. The performance of such systems mainly depends on several factors, including power transfer efficiency, coupling coefficient, transmission distance, coil geometry, and load conditions. In this chapter, both the electrical optimization and the experimental validation of the proposed wireless power transfer system are presented. First, an electrical analysis is carried out to investigate the influence of key circuit parameters, particularly the coupling coefficient and load resistance, on the power transfer efficiency and output power of a series–parallel resonant inductive link operating at 13.56 MHz.

The electrical behavior of the system is analyzed using PSpice simulations. Subsequently, an experimental study is conducted to validate the theoretical and simulation results through a dedicated testbench. Different coil configurations, including solenoid, planar, and hybrid structures, are evaluated and compared in terms of transmitted power and operating distance.

In addition, the practical feasibility of the proposed system for biomedical applications is assessed through experiments involving biological obstacles, demonstrating its suitability for powering implantable medical devices.

III-2- Electrical Optimization

III-2-1 Analysis of the Influence of Electrical Circuit Parameters on Inductive Link Performance

In this section, we presented an electrical analysis of a series-parallel circuit for a wireless power transfer system using a resonant inductive link to operate an implantable medical device with a resonant frequency of 13.56 MHz. Power transfer efficiency is an important element in the study of this wireless power transfer system [1]. We investigated the effect of the circuit's electrical parameters on the power transfer efficiency between the transmitting circuit (external circuit) and the receiving circuit (implanted circuit), as well as the output energy of the secondary circuit (implanted circuit) [2].

The table III-1 displays the electrical parameters of the circuit meter and the circuit receiver (implant) displayed on this site.

Parameter	Value
L_T	5.1 μH
C_T	27 PF
L_R	1 μH
C_R	138 PF
F	13.56 Mhz

Table III-1: Electrical parameters of the circuit meter and the circuit receiver (implant)

The electrical circuit of the resonant inductive link is shown in Figure III-1. The primary and secondary circuits are tuned to the carrier frequency of 13.56 MHz. Using simulation with the PSpice software, we plotted the energy transfer efficiency and the output energy of the inductive link circuit as a function of frequency for certain values of the coupling coefficient and the load resistance with a source voltage $V_s = 10\text{ V}$.

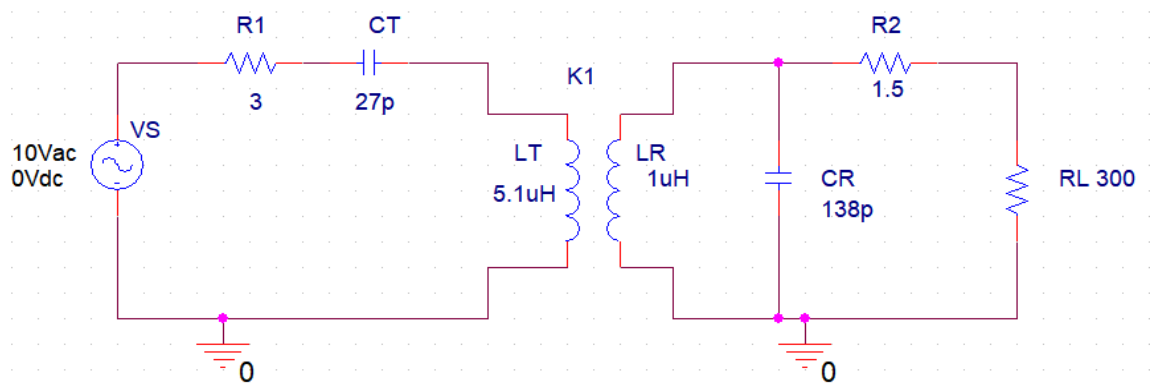


Figure III-1: Resonant inductive link circuit

III-2-2- Effect of the coupling coefficient K on the energy efficiency of the link

The coupling coefficient (K) plays an important role in determining the energy transfer efficiency of a resonant inductive link circuit. Mutual coupling depends on the inter-coil distance.

To determine the effect of the coupling coefficient (K), we simulated the resonant inductive link circuit with a series-parallel topology using the PSpice software. The other circuit

parameters, such as the load resistance R_C , the series resistances of the coils R_1 , R_2 , and the resonant capacitors C_1 and C_2 , are constant. When the coupling coefficient between the coils is variable, the effect on the wireless power transfer characteristics of the system can be explored. We have fixed a value of $300\ \Omega$ for the load resistance R_C , and the electrical circuit under PSpice is constructed as illustrated in Figure

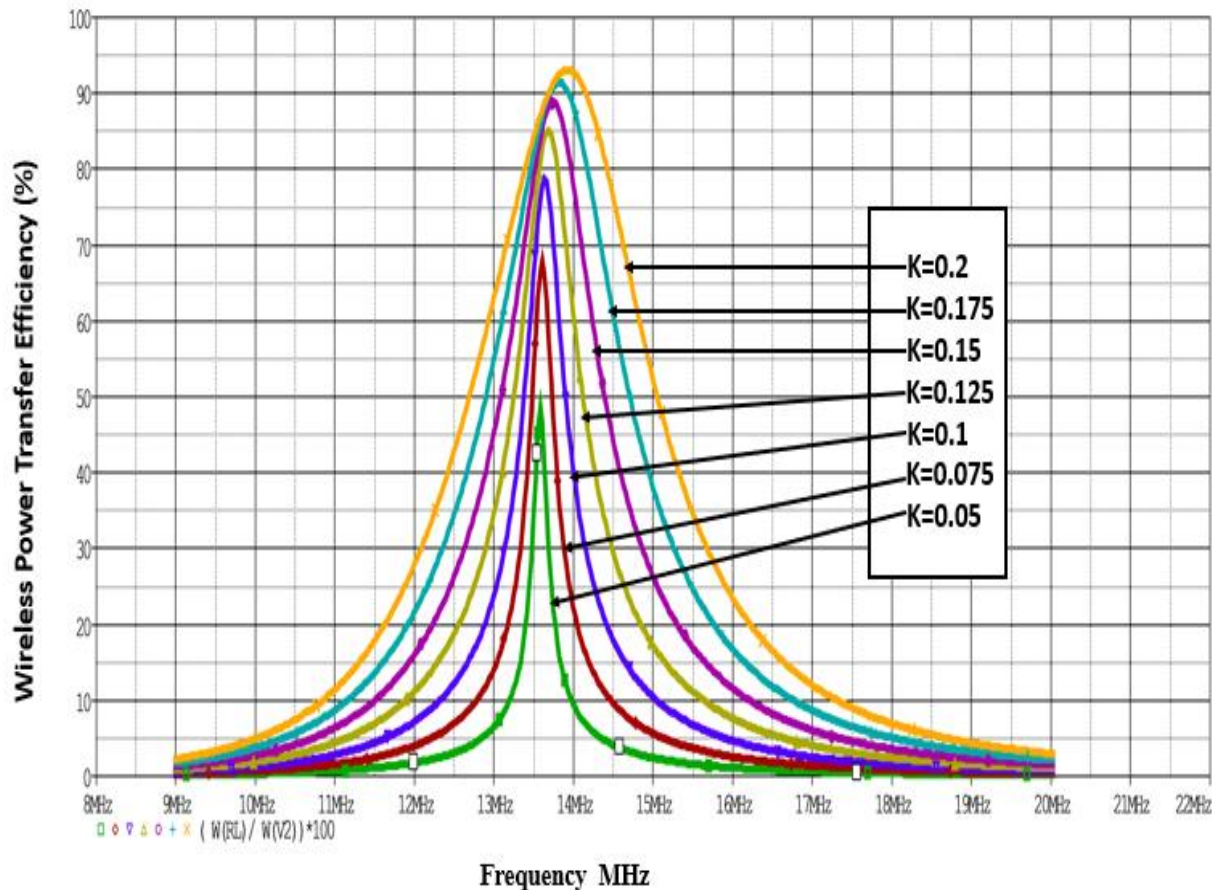


Figure III-2: Energy transfer efficiency as a function of frequency for different values of the coupling coefficient K

Figure III-2: shows the variation of energy transfer efficiency as a function of frequency for different values of the coupling coefficient (K). Note that energy efficiency is always maximum for the same resonant frequency of the circuit, equal to 13.56 MHz

Figure III-2 also shows the effect of varying the coupling coefficient between the transmitting and receiving coils. It is worth noting that the circuit achieves higher energy transfer efficiency for a larger value of the coupling coefficient (K). If the two coils are placed side by side, the input impedance is independent of frequency variations, but the energy transfer efficiency increases considerably. We can conclude that when the mutual distance between the

coupled coils is smaller, the mutual inductance (M) is higher and the coupling coefficient (K) is higher, resulting in higher energy transfer efficiency.

III-2-3- Effect of Load Resistance R_L on Link Energy Efficiency

An increase in the load resistance (R_L) value reduces the impedance of the secondary coil because they are inversely proportional to each other.

This decrease in impedance subsequently increases the energy transfer efficiency. To verify the effect of varying the load resistance on the energy transfer efficiency, we plotted the circuit while keeping all circuit components constant, except for the value of R_L , which varied between $100\ \Omega$ and $500\ \Omega$.

Figure III-3 shows the variation of the energy transfer efficiency as a function of frequency for different load resistance values with a constant coupling coefficient of 0.075

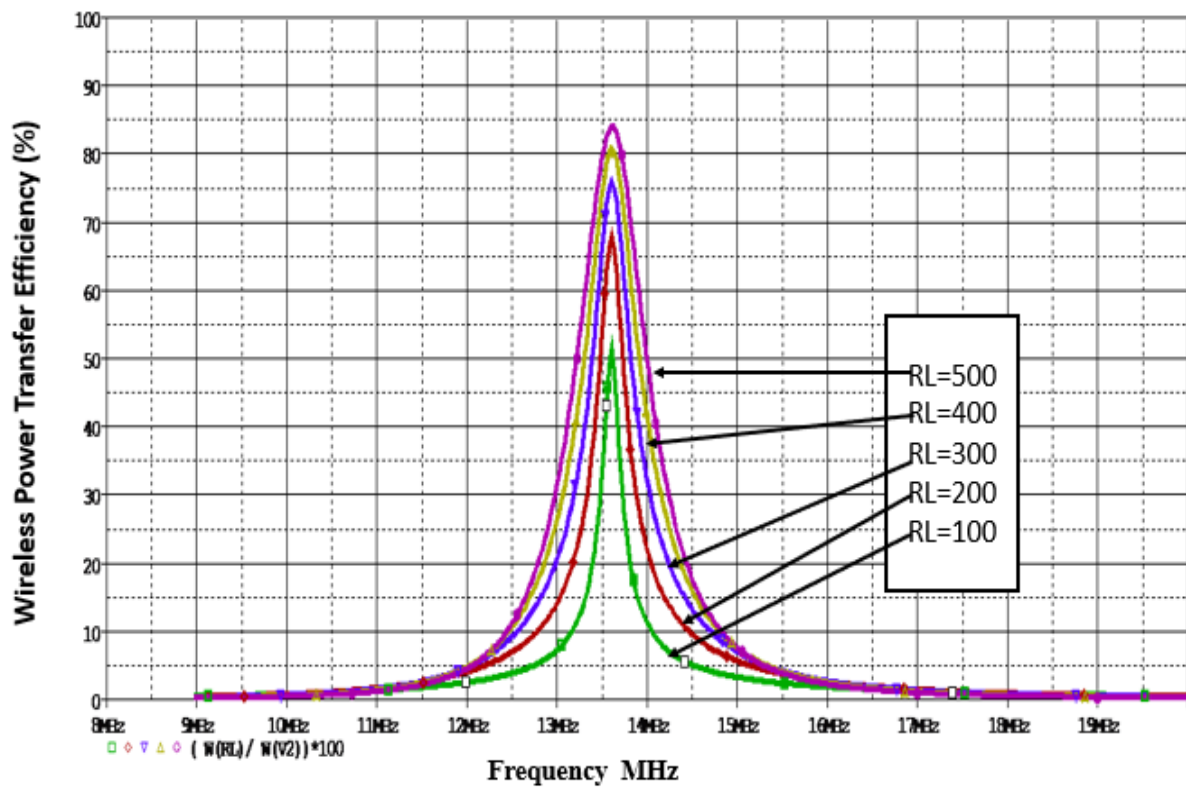


Figure III-3: Energy transfer efficiency as a function of frequency for different values of load resistance R_L

The results show that the energy transfer efficiency also increases proportionally with the load resistance, varying between 80% and 91%. It can be observed that the received output energy reaches its maximum at a frequency of approximately 13.56 MHz (resonant frequency).

The maximum output energy can be transferred if the circuit is tuned to this frequency to make it more efficient.

III-2-4- Effect of K on the output power of an inductive link circuit

With the simulation circuit parameters (Figure III-1), the system's resonant frequency is 13.56 MHz the coupling coefficient varies between 0.05 and 0.2 in 0.025 increments, and the load resistance between 100 Ω and 700 Ω is also varied in 200 increments. For different values of the coupling coefficient (K) and load resistance (RL), AC sweep was used with the PSpice simulation circuit.

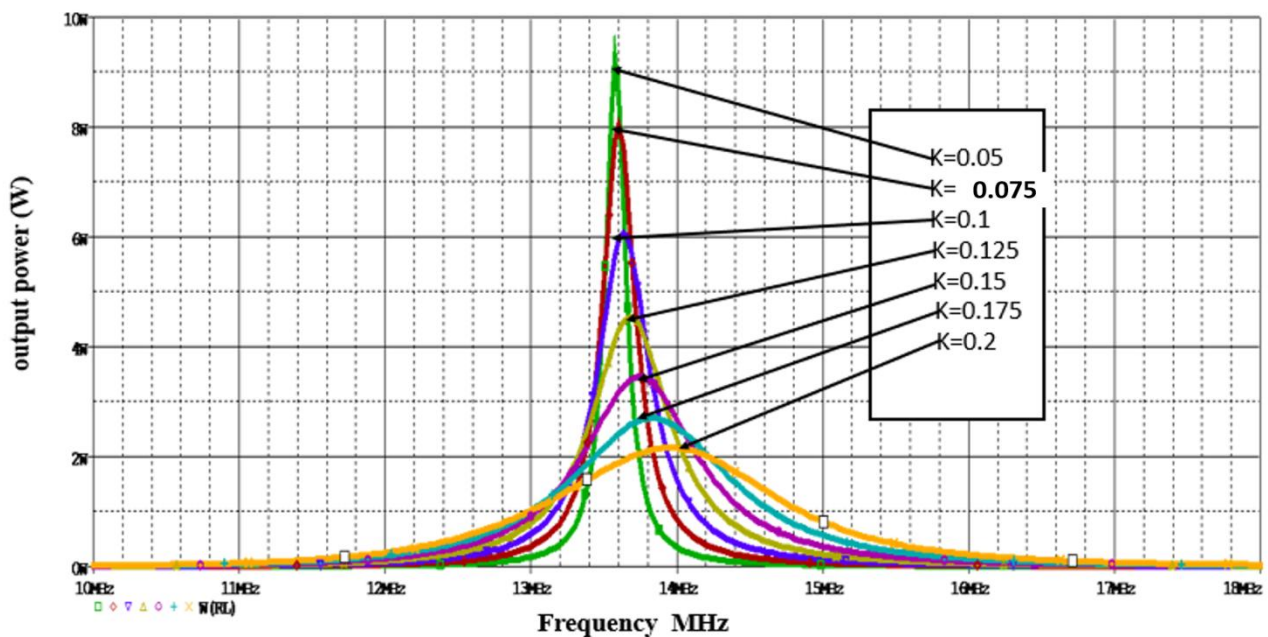


Figure III-4 illustrates the variation of the system's output power as a function of frequency for different coupling coefficient values

Note that at circuit resonance ($f = 13.56$ MHz), the output power is maximum. Therefore, as the coupling coefficient (k) decreases, the circuit achieves a higher output power.

III-2-5- Effect of RL on the output power of an inductive link circuit

Figure III-5 illustrates the variation of the system's output power as a function of frequency for different values of the load resistance. At circuit resonance ($f = 13.56$ MHz), the output power is maximum. Thus, as the load resistance RL decreases, the circuit achieves a higher output power.

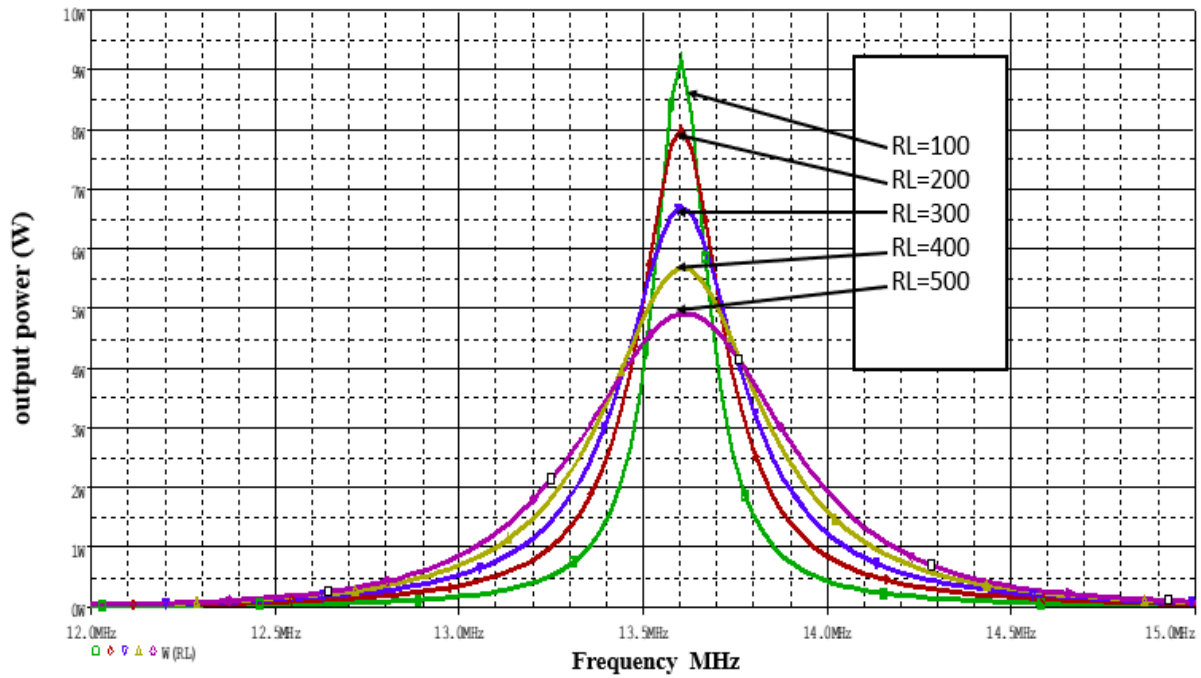


Figure III-5: Output power as a function of frequency for different values of load resistance R_L .

III-3 Coupling Coefficient as a Function of Distance

Figure III.6 illustrates the variation of the coupling coefficient (k) as a function of the separation distance between the coils. A continuous and smooth decay is observed as the distance increases. This behavior is due to the spatial dispersion of the magnetic field, where the magnetic link between the transmitter and the receiver becomes weaker at longer distances

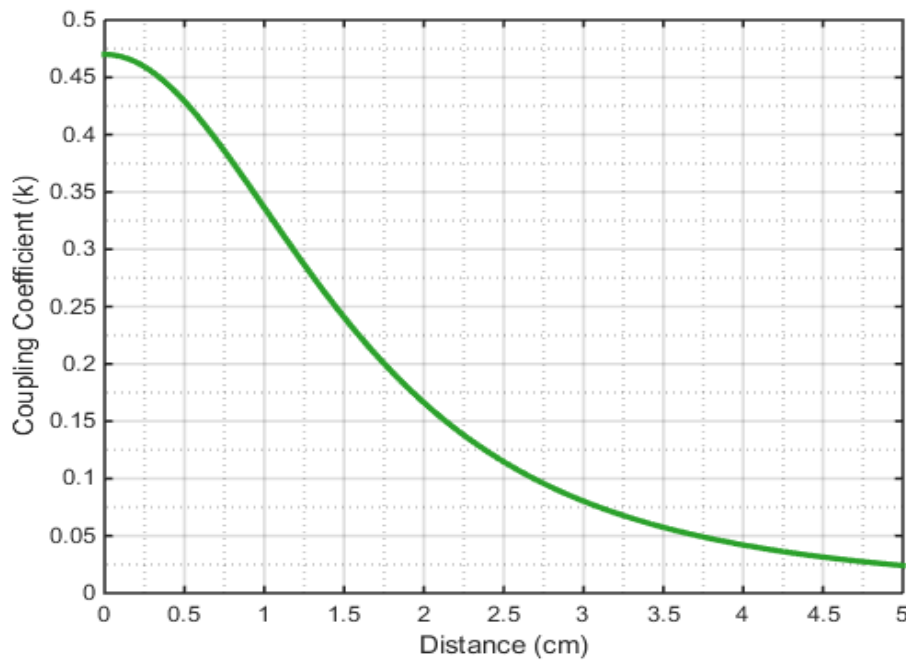


Figure III-6: Coupling Coefficient as a Function of Distance

III-4 The Mutual Inductance as a Function of Distance

Figure III-7 shows that the mutual inductance (M) decreases rapidly as the distance between the two coils increases. This happens because the magnetic flux becomes weaker and highly dispersed with distance, which reduces the magnetic link between the external transmitter and the implantable receiver

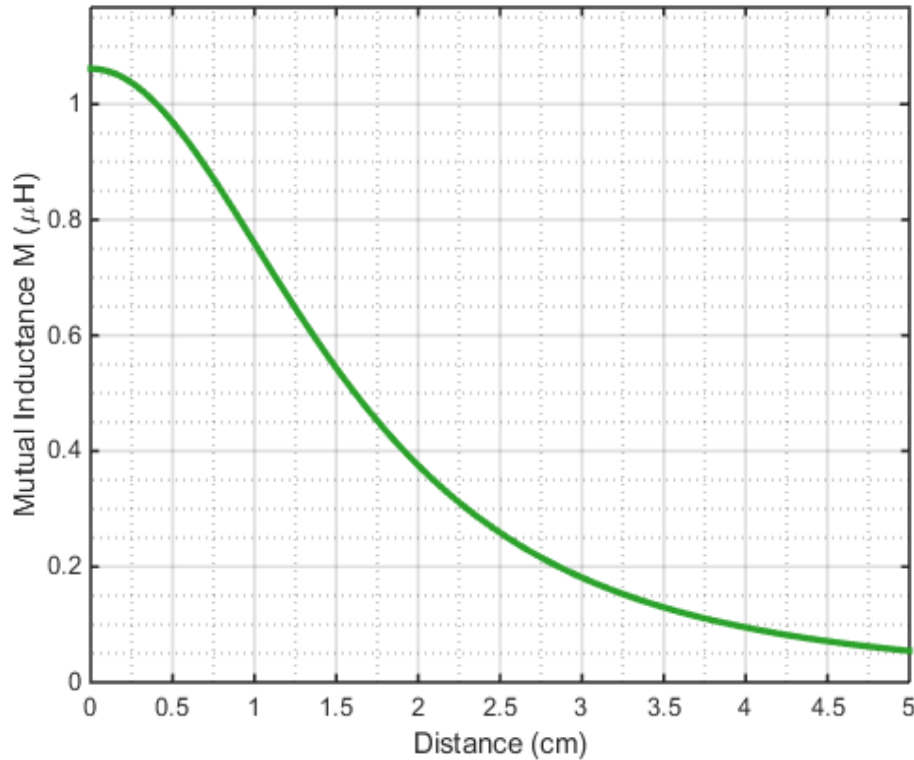


Figure III-7: The mutual inductance as a function of distance

III-5 Experimental Validation

III-5-1 Experimental testbench setup and instrumentation

Our experimental testbench is illustrated in the figures below. The practical setup consists of a GW INSTEK AFG-2225 arbitrary function generator (1 μHz - 25 MHz) operating as the high-frequency AC source for the primary transmitting circuit. For measurements, a digital multimeter and a RLC meter were used to record the output voltage and coil inductances, respectively. The output waveforms were monitored and captured using a GW INSTEK GDS-1102-U digital storage oscilloscope featuring a 100 MHz bandwidth. The electronic circuit, including various resistors and tuning capacitors, was assembled on two breadboards, with an LED serving as the terminal electrical load

III-5-2 First test: Primary and secondary solenoid coils

This first test investigates the wireless power transfer performance using identical, non-miniaturized solenoid-solenoid (solo-solo) coils for both the primary (transmitter) and secondary (receiver) sides. Both coils feature a radius of 4.5 cm to establish a baseline for system performance prior to size reduction



Figure III-8. WPT prototype using solenoid coils

III-5-3 Experimental validation through biological tissue obstacle

To validate the practical feasibility and biological safety of the developed system, an experimental test was conducted using a human hand as a direct obstacle between the coils. The obtained results, as illustrated in Figure III.9, demonstrate the system's ability to transfer power effectively through biological tissue.



Figure III-9: WPT System Testing with A Human Hand Obstacle

III-5-4 Second test: Miniaturized solenoid coils

For the second test, the experiment was conducted using miniaturized solenoid coils to evaluate the system's performance at a smaller scale. In this configuration, the primary transmitting coil was designed with a radius of 4.5 cm, while the secondary receiving coil featured a further reduced radius of 1.5 cm.

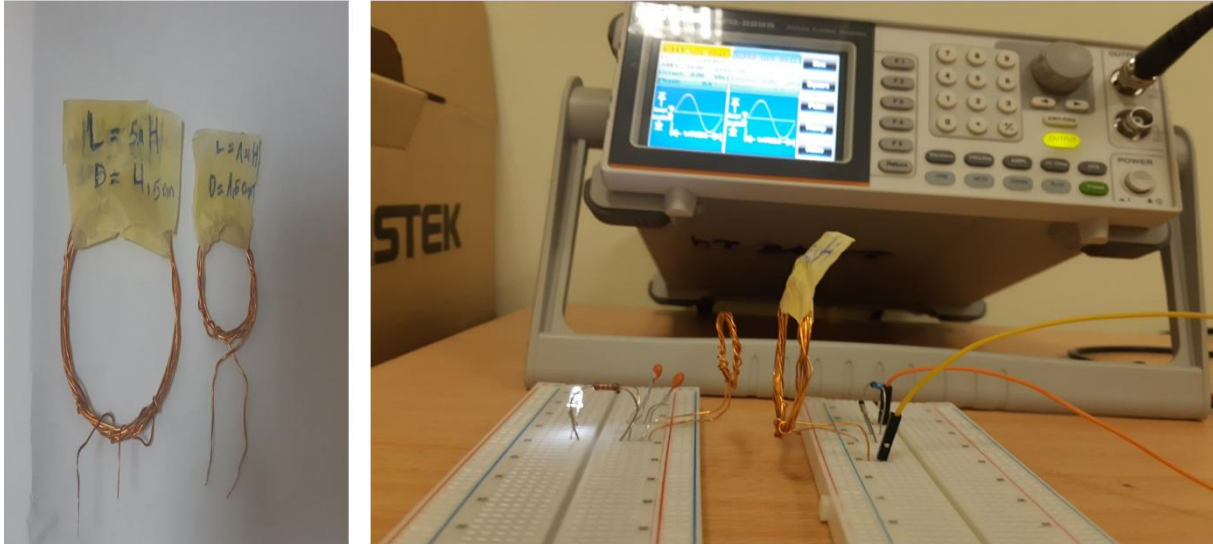


Figure III-10: WPT prototype using solenoid coils

III-5-5 The effect of distance on system performance

In this step, we measured the output voltage and current while changing the distance between the transmitter and receiver coils. This test allows us to see how the distance affects the power transfer efficiency and to find the best operating range for our system.

Table III-2. Measured output voltage and current at various transmission distances

Distance(cm)	0	0.5	1	1.5	2	2.5	3	3.5	4	4.5
Voltage(V)	3.5	2.5	1.7	1.25	0.95	0.8	0.75	0.5	0.48	0
Current(mA)	4	3.4	1.25	0.75	0.25	0.11	0.03	0.0025	0	0

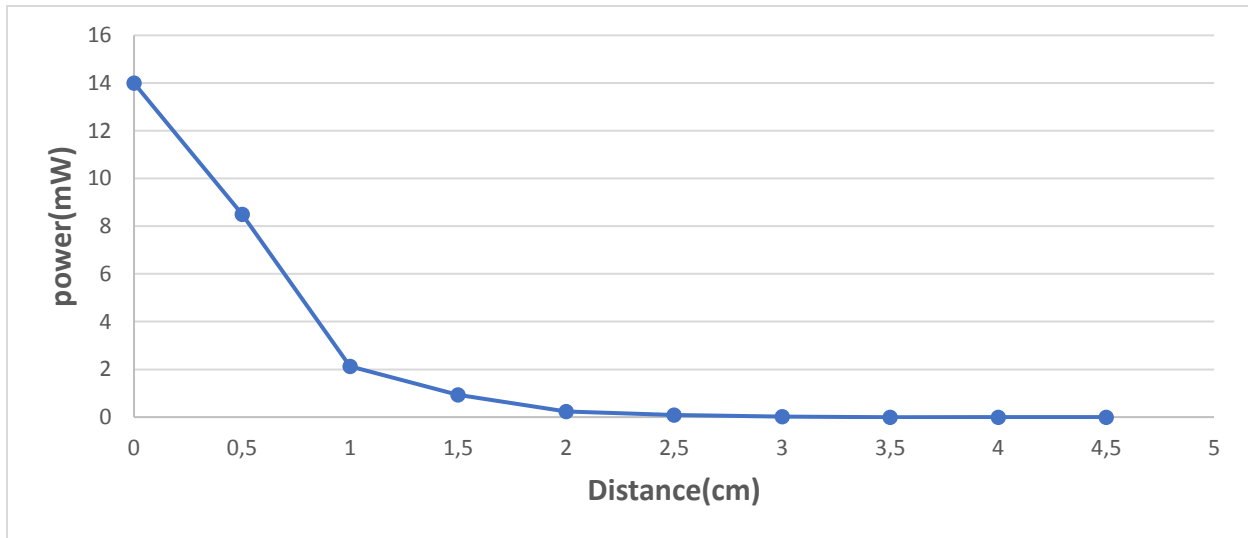


Figure III-11: Power as a function of distance (solenoid coils)

III-5-6 Third test: planar coils configuration

In this third test, planar coils were implemented. The primary transmitting coil was designed with an outer radius of 4.5 cm, while the miniaturized secondary receiving coil featured an outer radius of 1.5 cm.

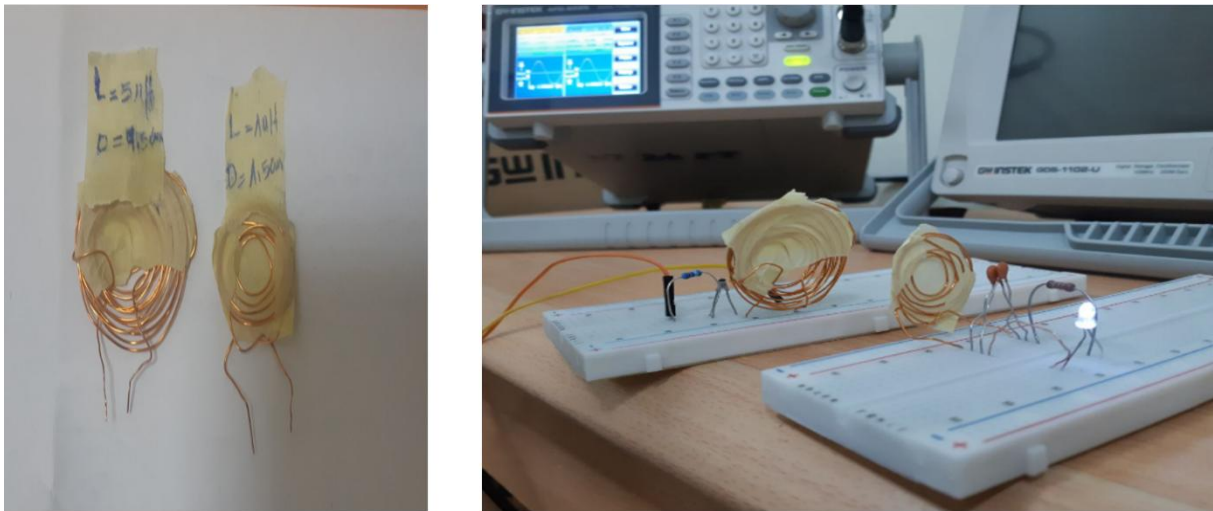


Figure III-12: WPT prototype using planar coils

Table III-3. Measured output voltage and current at various transmission distances

Distance(cm)	0	0.5	1	1.5	2	2.5	3	3.5	4	4.5
Voltage(v)	3.5	3.25	2.4	1.5	1	0.65	0.35	0.2	0	0
current (mA)	18	15	9	4.4	2.14	0.52	0.4	0.0015	0	0

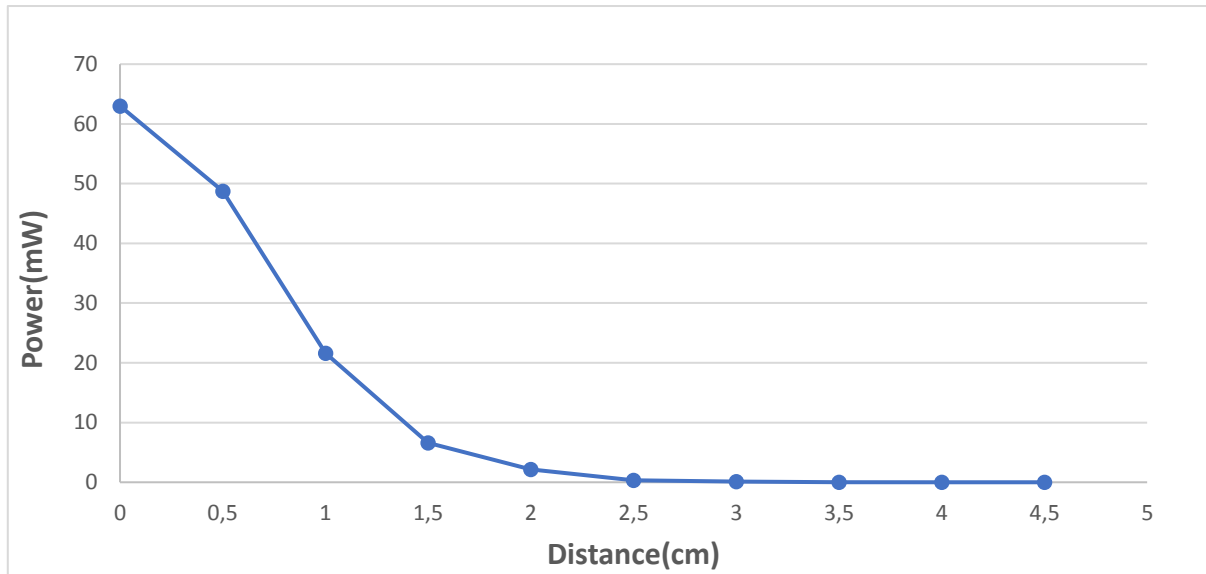


Figure III-13: Power as a function of distance (planar coils)

III-5-7 Fourth test: Hybrid coils configuration (Planar-Solenoid)

In this fourth test, a hybrid coils configuration was implemented to evaluate the system's performance. For this setup, a planar coil with an outer radius of 4.5 cm was used as the transmitter, while a miniaturized solenoid coil with a radius of 1.5 cm was utilized as the receiver.

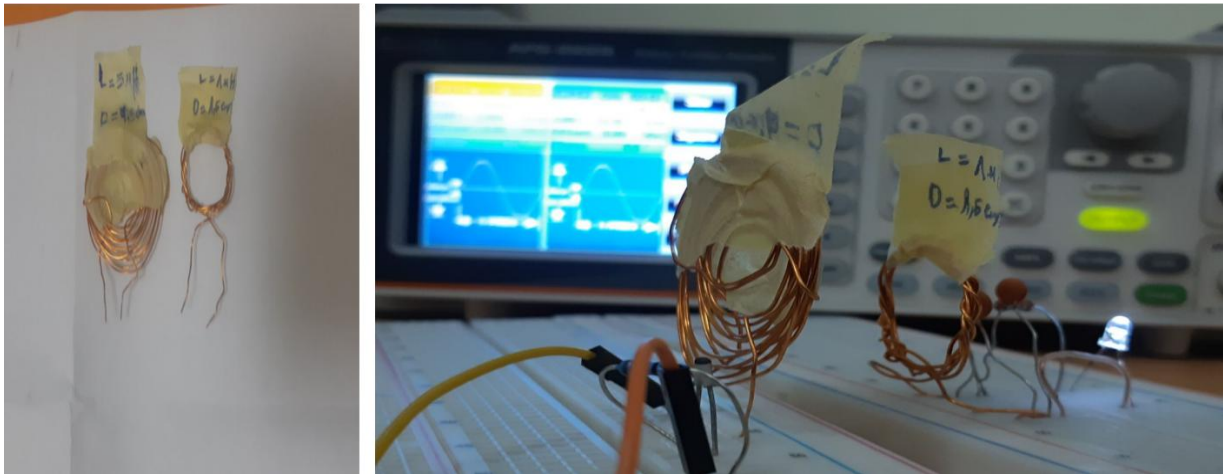


Figure III-14: WPT Prototype Using Hybrid Coils (Planar-Solenoid)

Table III-4. Measured output voltage and current at various transmission distances

Distance(cm)	0	0.5	1	1.5	2	2.5	3	3.5	4	4.5
Voltage(v)	3.75	3.5	2.6	1.5	1	0.48	0	0	0	0
Current (mA)	19.5	17.3	9.22	2.6	0.82	0.0005	0	0	0	0

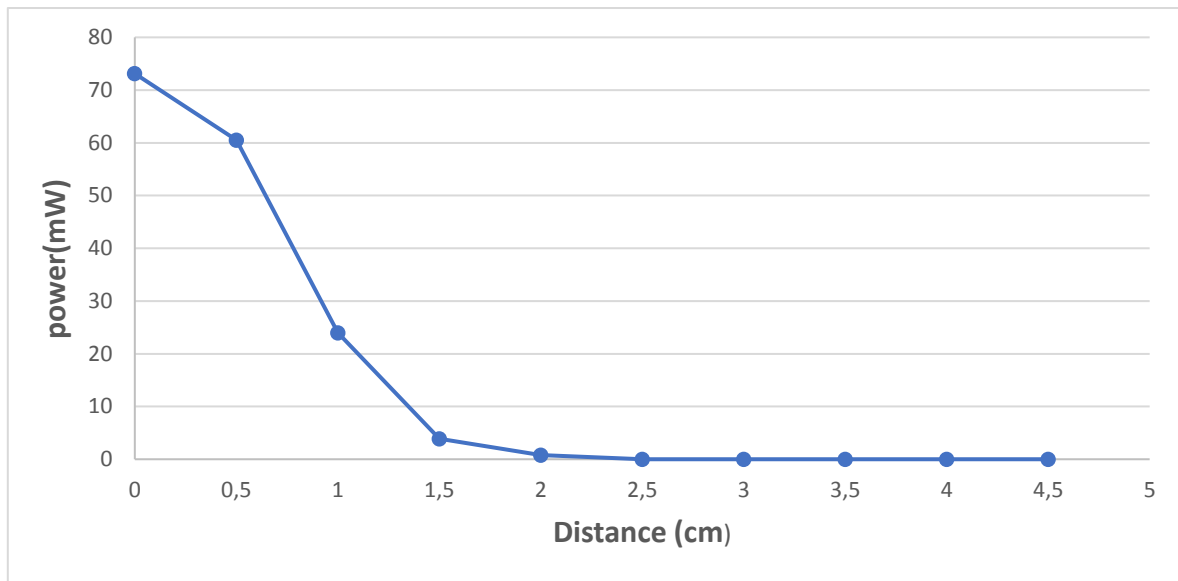


Figure III-15: Power as a function of distance (Planar-Solenoid coils)

III-5-8 Fifth Test: Hybrid coils configuration (Solenoid- Planar)

In this fifth test, a reversed hybrid configuration was examined using a 4.5 cm radius solenoid transmitter and a miniaturized 1.5 cm outer radius planar receiver. This setup allows us to evaluate the system's performance under altered coupling conditions

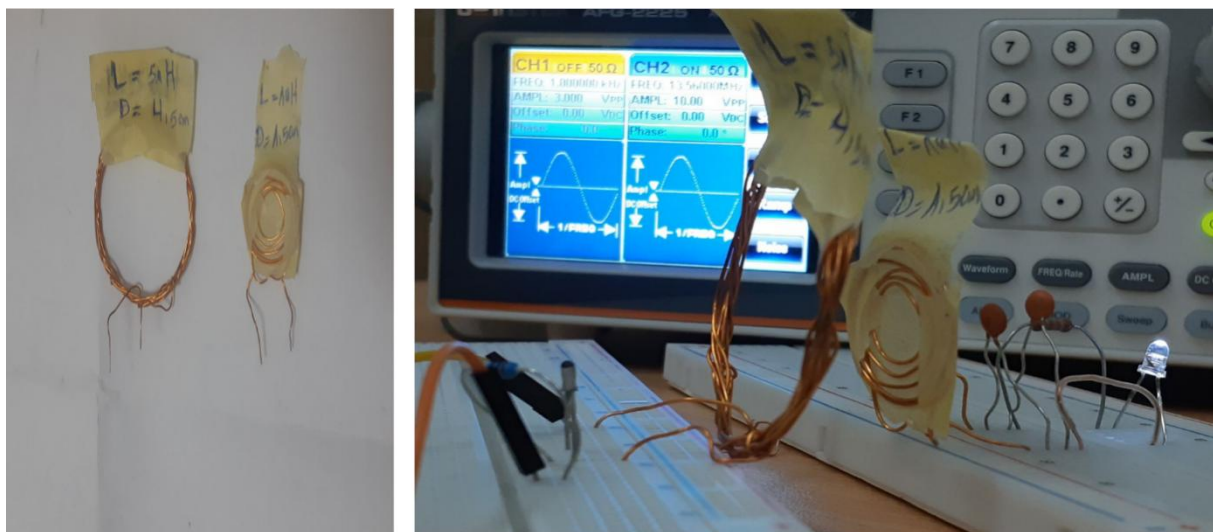


Figure III-16: WPT prototype using Hybrid coils (Solenoid- Planar).

Table III-5. Measured output voltage and current at various transmission distances (Solenoid- Planar)

Distance(cm)	0	0.5	1	1.5	2	2.5	3	3.5	4	4.5
voltage(v)	3.75	3.5	2.6	1.5	1	0.48	0	0	0	0
current (mA)	19.5	17.3	9.22	2.6	0.82	0.0005	0	0	0	0

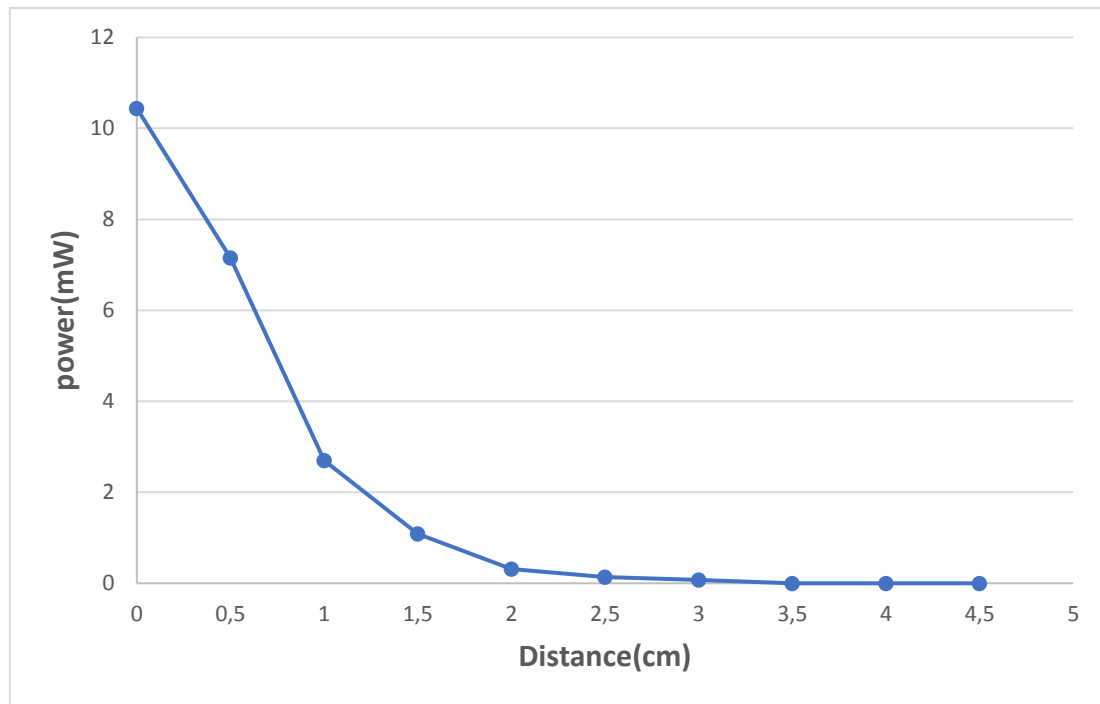


Figure III-17. Power as a function of distance (Solenoid -Planar coils)

III-6 Comparative Analysis of Power Curves

III-6-1 Planar Vs Solenoid Configurations

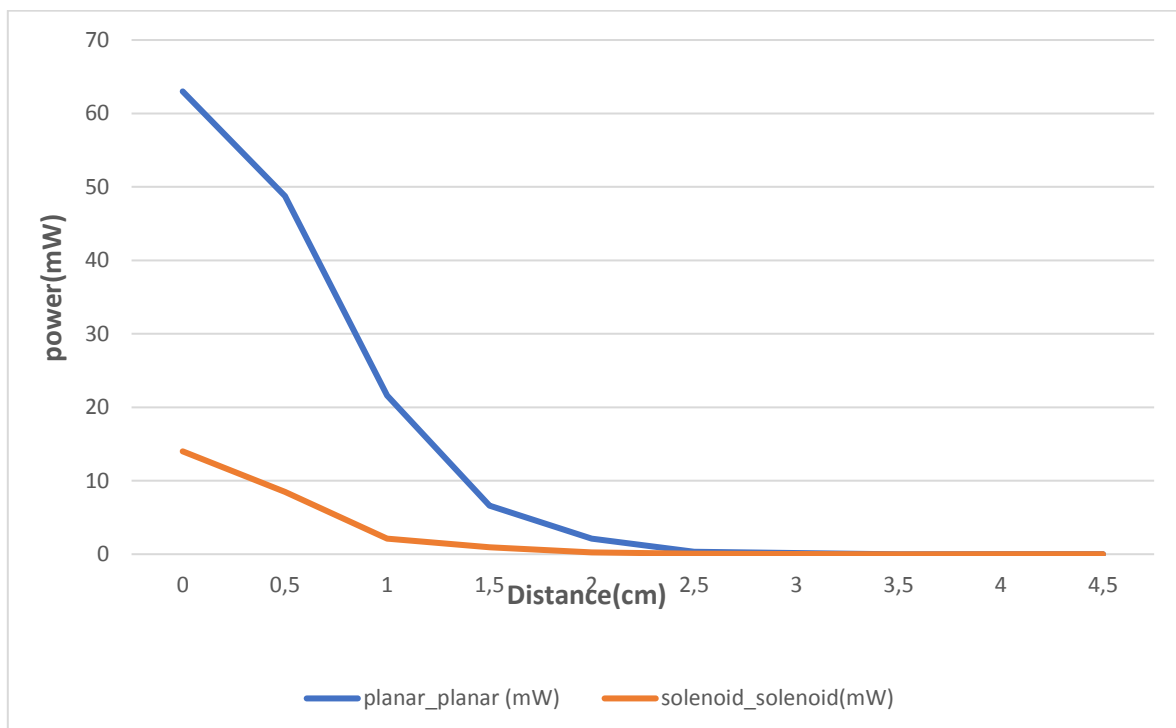


Figure III-18. Planar-Planar Vs. Solenoid-Solenoid Power Comparison

III-6-2 Planar-planar vs planar- Solenoid configurations

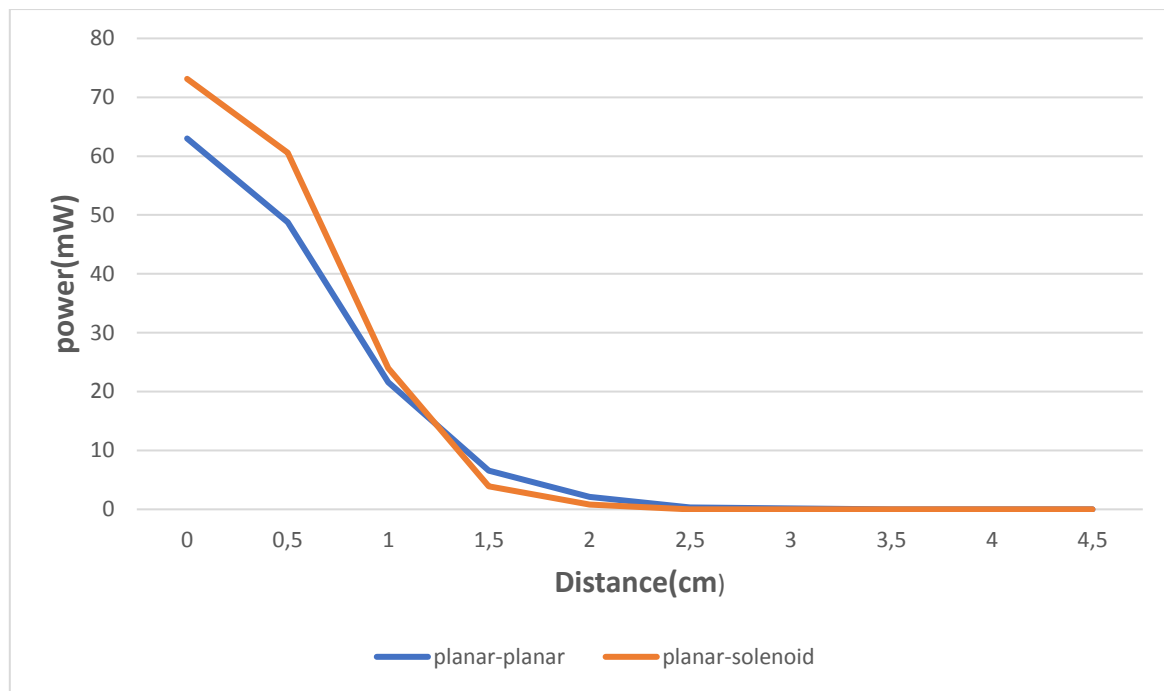


Figure IV-17. Planar-planar vs. planar-solenoid power comparison

IV-12. Discussion:

It is observed throughout the trials that each experiment exhibits its own unique characteristics. In this work, two main comparisons were conducted: the first between the strictly planar configuration (Planar-Planar) and the strictly cylindrical configuration (Solenoid-Solenoid), and the second between the planar configuration (Planar-Planar) and the hybrid configuration (Planar-Solenoid). The experimental results clearly demonstrate the superiority of the hybrid (Planar-Solenoid) topology, which achieved a peak power of 73.125 mW at zero distance. Furthermore, it maintained its advantage at a critical distance of 1 cm, delivering a power of 23.972 mW compared to only 21.6 mW for the planar-planar setup, making it the most efficient choice for implantable medical systems.

This outstanding performance is directly attributed to the precise selection of the coils' dimensions and diameters. A flat planar transmitter with a large outer diameter of 4.5 cm was utilized to maximize the magnetic field distribution and ensure sufficient tissue penetration. Conversely, a miniaturized solenoid receiver with a compact diameter of 1.5 cm was used to tightly concentrate the magnetic flux lines and minimize flux leakage within the implant, all without increasing the surgical incision size. On the other hand, while larger solenoid coils offer a more attractive transmission range, their bulky dimensions prevent them from being integrated inside the human body. In general, the output power decreases rapidly with distance,

proving an inversely proportional relationship and highlighting the necessity of stabilizing the coupling at short ranges (between 0 and 1.5 cm) to ensure the thermal safety of the medical implant.

III-7 Conclusion

In this chapter, the electrical optimization and experimental validation of the wireless power transfer (WPT) system were investigated. The simulation results demonstrated that the power transfer efficiency strongly depends on the coupling coefficient and the load resistance, with maximum efficiency achieved when the system operates at its resonant frequency of 13.56 MHz. An increase in the coupling coefficient and load resistance was found to improve the overall energy transfer efficiency.

Experimental tests were then carried out using different coil configurations, including solenoid, planar, and hybrid structures. The obtained results confirmed the ability of the proposed system to transfer power efficiently over short distances and through biological obstacles, demonstrating its suitability for biomedical applications. Among the tested configurations, the planar–planar structure exhibited the best overall performance due to its compact geometry, stable power transfer characteristics, and compatibility with miniaturized implantable medical devices.

Overall, the agreement between simulation and experimental results validates the effectiveness of the proposed resonant inductive link and highlights the planar–planar configuration as the most promising solution for wireless powering of implantable biomedical devices.

Bibliographical references for Chapter III

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GENERAL
CONCLUSION

This thesis presented the design, analysis, simulation, and experimental validation of a wireless power transfer (WPT) system intended for powering implantable biomedical devices through resonant inductive coupling. Initially, an overview of implantable medical devices and wireless power transfer technologies was provided, highlighting the advantages of WPT as a safe and reliable alternative to conventional wired power delivery methods. Particular attention was given to the safety requirements associated with human exposure to electromagnetic fields.

A theoretical and analytical study of a series–parallel resonant inductive link operating at 13.56 MHz was then conducted. The influence of key electrical parameters, including the coupling coefficient (K), load resistance (R_L), operating frequency, and transmission distance, was thoroughly investigated. Simulation results obtained using PSpice demonstrated that the power transfer efficiency increases with both the coupling coefficient and the load resistance, reaching values above 90% under optimal resonance conditions. The analysis also showed that increasing the separation distance between the coils leads to a reduction in both the coupling coefficient and mutual inductance, which consequently decreases the transferred power and overall efficiency.

To validate the theoretical and simulation results, an experimental testbench was developed and several coil configurations were evaluated, including solenoid, planar, and hybrid structures. Additional experiments were carried out using a human hand as a biological obstacle, confirming the capability of the proposed system to transfer power through biological tissues and demonstrating its suitability for biomedical applications.

The experimental results revealed that the hybrid planar–solenoid configuration provides the best overall performance among the investigated topologies. This configuration achieved a maximum transferred power of approximately 73.1 mW at zero distance and maintained about 24 mW at a transmission distance of 1 cm, which is particularly relevant for implantable medical devices. Furthermore, the hybrid structure combines the compactness of planar coils with the strong magnetic coupling provided by solenoid coils, making it highly suitable for miniaturized implant applications.

Overall, the good agreement between analytical, simulation, and experimental results confirms the effectiveness of resonant inductive wireless power transfer as a promising solution for powering implantable biomedical devices. The proposed system offers an efficient, safe, and reliable method of energy delivery without physical connections and provides a solid

foundation for future developments aimed at increasing transmission distance, improving efficiency, and further miniaturizing implantable systems.