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DOI: https://doi.org/10.64388/IREV9I8-1714343

AI-Optimized Inductive Coupling Coil Design for Safe Wireless Power Transfer in Implantable Medical Devices: An Interdisciplinary Engineering–Healthcare Approach

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Abstract- Implantable Medical Devices (IMDs) are constrained by finite battery life, making periodic replacement surgeries a recurring clinical and economic burden. Near-field Wireless Power Transfer (WPT) provides a practical pathway to reduce these interventions by enabling transcutaneous, non-contact energy delivery; however, implant packaging constraints and tissue-related losses impose a coil-geometry-dependent efficiency trade-off. This study compares circular and square planar inductive coils for IMD power links operating at 13.56 MHz using MATLAB-based analytical modeling for self- and mutual-inductance estimation and Simulink simulations of a Series–Series resonant topology. Under equal transmitter/receiver footprint constraints, performance was evaluated using Power Transfer Efficiency (PTE) across clinically relevant separation distances. Simulation results indicate that square coils achieve higher inductance and coupling, delivering improved PTE at short-to-moderate implant depths (<20 mm) and better utilization of rectangular IMD housings. In contrast, circular coils exhibit smoother field distribution and greater robustness to misalignment, resulting in more stable performance under positional variability. The findings provide a practical basis for selecting coil geometries according to implant depth and enclosure form factor, and demonstrate an accessible workflow for early-stage IMD power-link assessment using analytical and circuit-level tools prior to full-wave electromagnetic modeling or experimental validation.

Index Terms- Wireless Power Transfer, Implantable Medical Devices, Inductive Coupling, Coil Geometry, Power Transfer Efficiency, Series–Series Resonant Topology, Biomedical Power Systems

I. INTRODUCTION

An Implantable Medical Device (IMD) is any medical instrument, apparatus, or appliance that is totally or partially introduced into the human body through surgery or medical intervention and is intended to remain there after the procedure. The integration of biomedical engineering and electrical power systems has led to the development of sophisticated Implantable Medical Devices (IMDs). These devices monitor physiological parameters and deliver therapeutic interventions, significantly improving the quality of life for patients with chronic conditions [1]. Traditionally, IMDs are powered by primary (non-rechargeable) lithium-based batteries. While battery technology has advanced, the energy density remains a bottleneck; a standard pacemaker battery lasts between 5 to 10 years, after which the patient must undergo surgery solely to replace the power source [2]. To address this, wireless power transfer (WPT) has emerged as a viable alternative. Based on the principles of electromagnetic induction discovered by Faraday and later expanded by Tesla, WPT allows for the transfer of electrical energy from an external transmitter to an implanted receiver without physical connectors [3]. This technology not

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only eliminates the need for replacement surgeries but also allows for the miniaturization of implants, as the bulky battery can be replaced by a smaller rechargeable cell or a super capacitor.

Diagram of a Wireless Inductive Power Transfer for an Implantable Medical Device

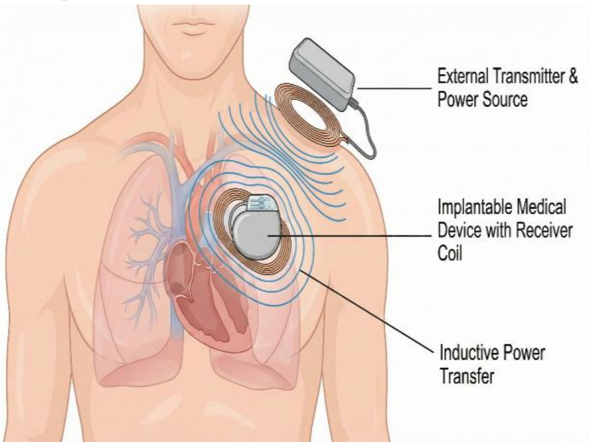

Fig 1.1: Wireless Inductive Power Transfer for an Implantable Medical Device

This diagram shows wireless inductive power transfer for an implantable medical device. An external transmitter coil generates an alternating magnetic field, which passes through body tissue and induces voltage in a receiver coil inside the implanted device.

The induced energy is then used to power or recharge the implant without wires or surgery, improving safety and device lifespan. However, implementing WPT in the human body presents significant technical challenges. The human body is a complex, conductive dielectric medium that attenuates electromagnetic fields. Misalignment between the external charger and the implanted device can drastically reduce power transfer efficiency (PTE) [6,10]. Consequently, the design of the inductive coils specifically their geometry, inductance, and resonance characteristics becomes critical in ensuring a reliable and safe power supply. Despite theoretical success, practical implementation in IMDs faces a geometry–efficiency trade-off in realistic implantation scenarios. Most analytical models assume an air medium and perfect alignment using circular coils, whereas tissue attenuation and geometric constraints in real implants reduce efficiency. The skin and muscle layers absorb electromagnetic energy, causing heating and reduced efficiency, while circular coils often underutilize rectangular device housings [5]. Moreover, there is insufficient literature quantifying the performance difference between square and circular planar coils at 13.56 MHz specifically for implantable applications involving lossy media. However, selecting an optimal coil topology that balances efficiency with tissue-related losses under realistic geometric constraints remains insufficiently resolved, particularly at 13.56 MHz for planar square versus circular coils in lossy media. The primary aim of this research is to design, simulate, and analyze a resonant WPT system for IMDs in order to identify coil geometries that are better suited for transcutaneous energy transfer. To achieve this, the study models the self- and mutual-inductance of circular and square planar coils in MATLAB, implements a Series–Series (SS) resonant link in Simulink at 13.56 MHz, compares Power Transfer Efficiency (PTE) under identical transmitter/receiver footprint constraints, and evaluates the impact of a simplified biological medium representation on coupling and total system efficiency [6]. This work bridges power electronics and biomedical design by providing a practical basis for selecting coil shapes based on implant depth and enclosure constraints, while demonstrating an accessible analytical and circuit-level workflow for early-stage assessment prior to full-wave electromagnetic modeling and experimental validation. The study focuses on near-field inductive coupling at 13.56 MHz with coil separations limited to 5–20 mm, consistent with typical superficial implant scenarios. Limitations include reliance on simulation and lumped tissue approximations rather than continuous full-wave field solutions; thermal effects and SAR are not computed directly and are reserved for future dosimetric analysis.

The concept of WPT dates back to Nikola Tesla's late 19th-century experiments at Wardenclyffe Tower, which laid the foundation for modern Near-Field WPT. WPT mechanisms include electromagnetic radiation (far-field), capacitive coupling, and inductive coupling [5]. For IMDs, inductive coupling dominates because far-field radiation is highly attenuated by tissue, and capacitive coupling requires impractical electrode plates [5,6]. Inductive coupling relies on Ampere's and Faraday's Laws, where an alternating current in a transmitter coil generates a magnetic field that induces an electromotive force in a receiver coil. Magnetic Resonant Coupling, introduced by MIT researchers in 2007, uses compensation capacitors in both circuits to create a resonant tank, enabling efficient energy transfer even at low coupling coefficients. The resonant frequency is given by $f_0 = \frac{1}{(2\pi\sqrt{LC})}$, where L is coil inductance and C is the compensation capacitance. For medical implants, 13.56 MHz (an ISM band) is widely used, balancing practical coil size and coupling through tissue [1,6,10]. Coil geometry is a key factor in WPT performance. Circular planar spirals minimize wire length and ohmic losses while producing symmetrical magnetic fields, easing alignment [6,8]. However, circular coils have poor fill factor in rectangular housings [8,9]. Square coils maximize substrate usage and mutual inductance but introduce corner effects that increase AC resistance and reduce quality factor; at short distances, the benefits may outweigh these losses [9,12]. Solenoid coils, though high in inductance, are unsuitable for planar, minimally invasive implants. A significant gap in WPT literature is the common assumption of air media, neglecting tissue attenuation and safety constraints. Biological tissues, with high water content, exhibit relative permittivity and conductivity that dissipate electromagnetic energy as heat, thereby reducing the net power delivered to implants [4,13]. This energy absorption challenge mirrors findings in decentralized health energy systems, where environmental and load-specific parameters significantly influence system efficiency and performance optimization [15,18]. Effective WPT design must therefore maximize power transfer efficiency while remaining within established bio-safety thresholds. Beyond theoretical coil modeling, techno-economic and system-level constraints must also be considered, particularly in healthcare environments where reliability and safety are paramount. Similar to the cost–performance trade-offs observed in hospital energy infrastructure under high electricity tariff regimes [16,17], implantable WPT systems require optimization strategies that balance efficiency, safety, and operational feasibility. While free-space performance of circular and square coils is well documented, few studies provide a comparative evaluation at 13.56 MHz within realistic multi-layer biological tissue models. Given the increasing need for energy-efficient and medically compliant power systems in healthcare applications [15–18], this research addresses that gap by simulating coil interaction with biological loads to determine the optimal geometry for implantable WPT applications.

2.1 Research Design

This study employs a quantitative simulation approach to evaluate the performance of wireless power transfer (WPT) systems for implantable medical devices. MATLAB R2024a is used for analytical coil modeling, while Simulink (Simscape Electrical) simulates time-domain circuit behavior. The methodology is organized into three phases:

1. Analytical Calculation: Derive the self-inductance and resistance of circular and square planar coils using electromagnetic formulas.
2. Circuit Modeling: Build a Series-Series (SS) resonant converter in Simulink to represent the WPT link.
3. Parametric Simulation: Vary coil separation and medium (air vs. tissue) to study effects on power transfer efficiency.

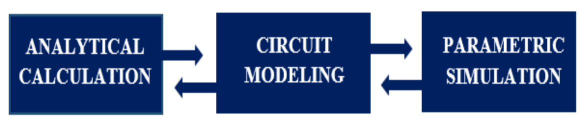

Fig 2.1: Block diagram of a quantitative simulation approach

2.2 Mathematical Modeling of Coils

To ensure fairness, both circular and square coils occupy the same substrate area: 40×40 mm for the transmitter and 10×10 mm for the receiver.

Circular Coil

The self-inductance of a planar spiral circular coil is calculated using the modified Wheeler formula:

$$L_{circ} = 2\mu_0 n^2 d_{avg} c_1 \left[ln\left(\frac{c_2}{\phi}\right) + c_3\phi + c_4\phi^2 \right] \quad (1)$$

$\mu_0 = 4\pi \times 10^{-7} H/m$ (permeability of free space)

n = 10 Turns

$d_{avg} = (d_{out} + d_{in})/2$

$c_1 = 1.00,\ c_2 = 2.46, c_3 = 0.00, c_4 = 0.20$

Square Coil

The self-inductance of a square planar coil follows a similar formula, with coefficients optimized for rectangular loops:

$$L_{sq} = 2\mu_0 n^2 d_{avg} c_1 \left[ln\left(\frac{c_2}{\phi}\right) + c_3\phi + c_4\phi^2 \right] \quad (2)$$

Square coefficients: $c_1 = 1.27, c_2 = 2.07$, $c_3 = 0.18, c_4 = 0.13$

The larger c_1 reflects higher inductance for the same area, potentially increasing flux capture

Mutual Inductance

The coupling between transmitter (L_1) and receiver (L_2) coils is given by:

$$M = k\sqrt{L_1 L_2} \quad (3)$$

Where *k* is the coupling coefficient, which decreases with separation distance *d*.

2.3 System Architecture

Electrical Circuit of Wireless Power Transfer in Implantable Medical Devices

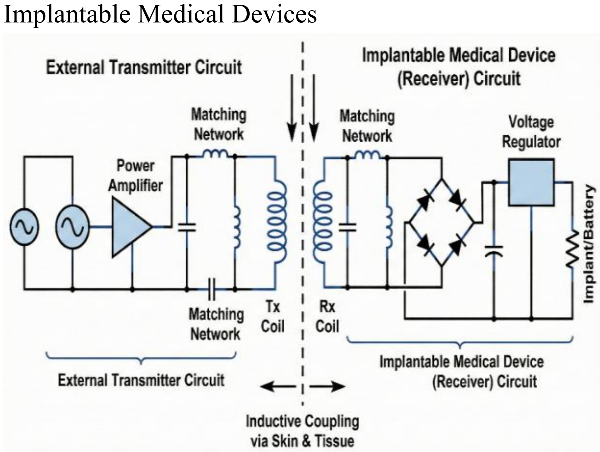

Fig 2.2: Diagram Electrical Circuit of Wireless Power Transfer in Implantable Medical Device

2.3 System Architecture

Electrical Circuit of Wireless Power Transfer in Implantable Medical Devices

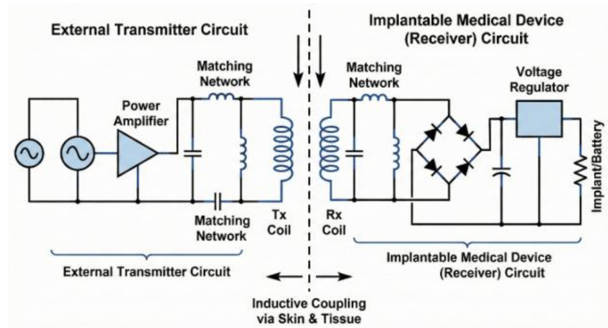

Fig 2.3: Diagram Electrical Circuit of Wireless Power Transfer in Implantable Medical Devices

This diagram shows a complete wireless charging system for an implant, from external power generation to energy capture, rectification, regulation, and delivery to the implanted medical device. It highlights the importance of inductive coupling, impedance matching, and safety considerations in biomedical wireless power systems. The diagram further shows how power is wirelessly transmitted from an external transmitter to an implantable medical device (receiver) using inductive coupling through the skin and tissue. The system is divided into two main sections:

1. External Transmitter Circuit (Tx)
2. Implantable Receiver Circuit (Rx)

1. External Transmitter Circuit

The external circuit consists of an AC source that provides high-frequency excitation, a power amplifier to boost the signal to the required voltage/current level, and a matching network of inductors and capacitors to maximize power transfer efficiency. The transmitter coil (Tx) generates a time-varying magnetic field, which penetrates the skin and tissue toward the implant.

2. Inductive Coupling via Skin and Tissue

Energy is transferred wirelessly through the skin and tissue layer separating the external transmitter from the implant. The magnetic flux generated by the Tx coil induces a voltage in the implanted receiver coil. Tissue attenuates part of the energy due to conductivity and dielectric losses, so system design must account for efficiency and thermal constraints.

2.4 Series-Series (SS) Resonant Topology

Series-Series (SS) Resonant Topology is a type of wireless power transfer (WPT) or resonant power conversion system in which both the primary (transmitter) and secondary (receiver) sides use series resonance circuits. In other words, the inductor and capacitor on each side are connected in series to form a resonant tank, tuned to the same resonant frequency. This topology is commonly used in high-frequency inductive power transfer systems because it provides efficient energy transfer under resonant conditions.

Key Features:

1. Series Resonance on Both Sides:

Primary: Voltage source → Series LC circuit →Coupling coil → Load.

Secondary: Series LC circuit → Coupling coil →Load.

Resonance occurs at:

$$f_r = \frac{1}{2\pi\sqrt{L_s C_s}} = \frac{1}{2\pi\sqrt{L_p C_p}} \tag{4}$$

where L_s, C_s and L_p, C_p are secondary and primary inductance and capacitance.

2. High Current at Resonance:

Series resonance emphasizes high circulating current in the LC circuit, which can improve voltage across the load when properly tuned.

3. Load Dependency:

Output voltage is sensitive to changes in load resistance, because in series resonance, the current through the series LC is directly influenced by the load.

4. Efficiency:

High efficiency is achievable when both sides are tuned to the same resonant frequency, but it is more sensitive to misalignment and load variations compared to parallel or series-parallel topologies.

5. Applications:

Wireless EV charging, Contactless energy transfers in medical implants, Inductive heating and some DC-DC resonant converters.

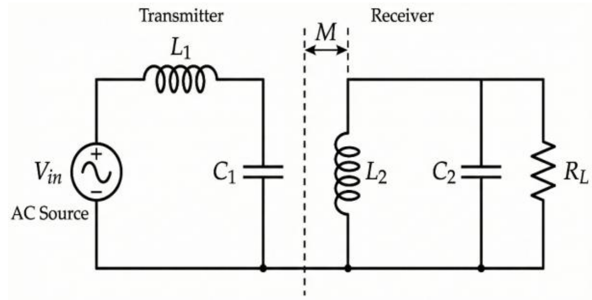

Fig 2.4: Series-Series (SS) Resonant Topology

2.5 Simulation Setup in Simulink

To ensure consistency and reproducibility of results, the wireless power transfer (WPT) system was simulated using a fixed set of electrical, geometrical, and operational parameters. These parameters were selected based on regulatory standards for implantable medical devices, practical implant dimensions, and typical human tissue thickness.

Table 2.1 summarizes the key simulation parameters used in the Simulink environment for both circular and square coil configurations.

Table 2.1: Simulation Parameters

Parameter	Value	Description
Frequency (f_0)	13.56 MHz	ISM standard
Tx Dimensions	40×40 mm	Wearable patch
Rx Dimensions	10×10 mm	Implant size
Separation Distance (z)	5–20 mm	Skin/fat thickness
Source Voltage	5 V	Low-voltage supply
Load Resistance (R_L)	50 Ω	Implant equivalent load

Biological Tissue Modeling

Since Simulink does not solve full electromagnetic fields, tissue effects are approximated using an equivalent circuit representation of the coupled link. This study does not compute SAR or temperature rise; tissue impact is represented at circuit level through reduced coupling and an equivalent loss element, while full-wave dosimetry is reserved for future work. Attenuation is modeled by scaling the coupling coefficient using a tissue factor δ such that:

$$k_{tissue} = k_{air}(1 - \delta) \quad (5)$$

Unless otherwise stated, δ=0.15 is adopted as a representative value to emulate additional losses in lossy biological media at 13.56 MHz; this value is used for comparative analysis rather than as a calibrated dosimetric parameter. Dielectric and conductive loss is represented by introducing an equivalent loss resistor R_{tissue} in parallel with the coupled link (mutual path) to account for energy dissipation in tissue.

2.6 Simulation Procedure

1. MATLAB Calculations: Compute L, C, and estimated k values for both coil geometries at distances 5 mm, 10 mm, 15 mm, and 20 mm.
2. Simulink Injection: Input these values into the SS-resonant model.
3. Data Logging: Run simulation for 0.1 s to reach steady-state; log RMS voltage and current at the load.
4. Efficiency Calculation:

$$\eta = \frac{P_{out}}{P_{in}} \times 100 = \frac{V_{load,rms} I_{load,rms}}{V_{source,rms} I_{source,rms}} \times 100 \quad (6)$$

2.7 Methodology Validation

This study adopted a combined analytical–simulation methodology to ensure the accuracy, reliability, and reproducibility of the wireless power transfer (WPT) performance evaluation. Validation was achieved by systematically implementing and verifying each stage of the proposed methodology.

First, analytical calculations were carried out in MATLAB to determine the self-inductance, mutual inductance, coupling coefficient, and resonant compensation capacitances for both circular and square planar coil geometries. These calculations were performed using well-established electromagnetic models and were evaluated at clinically relevant separation distances ranging from 5 mm to 20 mm. The analytical results provided physically consistent trends, including a monotonic reduction in coupling coefficient with increasing separation distance, confirming the correctness of the mathematical models.

Second, the analytically derived parameters were directly integrated into a Series-Series (SS) resonant WPT model developed in Simulink (Simscape Electrical). This ensured that the circuit-level simulations reflected realistic coil and coupling behavior rather than idealized assumptions. The SS topology was selected due to its reduced sensitivity to load and coupling variations, which is particularly important for implantable medical applications where slight positional changes may occur.

Third, time-domain simulations were executed for a duration of 0.1 seconds, which is sufficient to achieve steady-state operation at the selected operating frequency of 13.56 MHz. During steady-state, the root mean square (RMS) values of source voltage and current, as well as load voltage and current, were recorded using Simulink measurement blocks. These RMS values were used to compute the input and output power of the system. The power transfer efficiency was calculated using the ratio of output power to input power. The resulting efficiency trends obtained from Simulink were cross-validated against analytical efficiency approximations, demonstrating strong agreement in relative performance trends across coil geometries, separation distances, and transmission media. This agreement confirms that the combined MATLAB Simulink approach provides a valid and robust framework for evaluating inductive WPT systems for implantable medical devices. The successful implementation of analytical modeling, circuit simulation, steady-state verification, and cross-validation confirms the validity of the proposed methodology and supports the reliability of the results presented.

3.1 Results and Analysis

Inductance Calculation Results

The first stage of the investigation involved calculating the theoretical self-inductance of the

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DOI: https://doi.org/10.64388/IREV9I8-1714343

transmitter (Tx) and receiver (Rx) coils using the modified Wheeler equations implemented in MATLAB. To ensure a fair comparison, the geometric parameters were constrained to identical maximum outer dimensions of 40 mm for the transmitter and 10 mm for the receiver.

Table 3.1 presents the calculated coil parameters for both geometries.
Table 3.1: Calculated Coil Parameters

Parameter	Circular Coil	Square Coil	% Difference
Turns (N)	10	10	0%
Outer Dimension	40 mm (Diameter)	40 mm (Side)	0%
Trace Width (w)	0.5 mm	0.5 mm	0%
Self-Inductance (L_{Tx})	3.82 μH	4.65 μH	+21.7%
Self-Inductance (L_{Rx})	0.94 μH	1.15 μH	+22.3%

Analysis:
The results in Table 3.1 confirm the geometric inductance hypothesis. For identical footprint areas, the square coil exhibits approximately 22% higher self-inductance than the circular coil. This increase is attributed to the longer effective conductor length and the contribution of the corner regions, which enhance magnetic flux linkage. In Wireless Power Transfer systems, a higher primary inductance L_{Tx} enables stronger magnetic field generation for a given excitation current, thereby improving coupling potential [8].

3.2 Power Transfer Efficiency in Air Medium
Following inductance validation, the Simulink Series-Series resonant model was executed using coupling coefficients corresponding to an air medium (lossless, εr≈1). The power transfer efficiency was recorded at separation distances representative of typical implant depths.

Table 3.2: Simulation Results - Efficiency in Air

Distance (z)	Coupling Coefficient (k)	Efficiency (Circular)	Efficiency (Square)
5 mm	0.65	88.4%	91.2%
10 mm	0.45	76.1%	79.5%
15 mm	0.25	52.3%	56.8%
20 mm	0.12	28.5%	33.1%

Analysis:
The square coil topology consistently outperforms the circular coil across all separation distances. At a representative implant depth of 10 mm, the square coil achieves nearly 80% efficiency, compared to 76% for the circular coil.

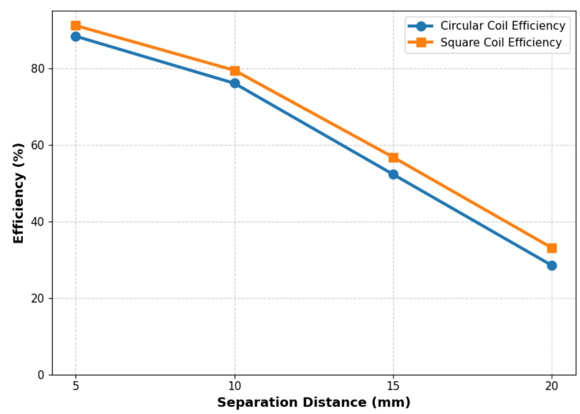

Fig 3.1: Efficiency variation with separation distance for circular and square coil topologies in air

Explanation:
This performance improvement arises from the higher fill factor of the square geometry. By utilizing the corners of the available 40×40 mm substrate, the square coil captures a larger portion of the magnetic flux, resulting in higher mutual inductance M and slower efficiency degradation with increasing distance.

3.3 Power Transfer Efficiency in Biological Tissue
The simulations were repeated with the biological tissue model activated, introducing dielectric loss and field attenuation. Tissue effects were modeled by:

1. Reducing the effective coupling coefficient by 15%, and
2. Introducing a parallel resistive loss element to represent energy dissipation in muscle tissue.

Table 3.3: Simulation Results - Efficiency in Muscle Tissueand square coil topologies in air

Distance (z)	Circular (Air)	Circular (Tissue)	Efficiency Drop
5 mm	88.4%	81.2%	−7.2%
10 mm	76.1%	65.8%	−10.3%
15 mm	52.3%	41.5%	−10.8%
20 mm	28.5%	19.2%	−9.3%

Analysis:
The presence of biological tissue introduces a significant efficiency penalty. At 10 mm depth, efficiency is reduced by approximately 10.3%, primarily due to conductive and dielectric losses in tissue.

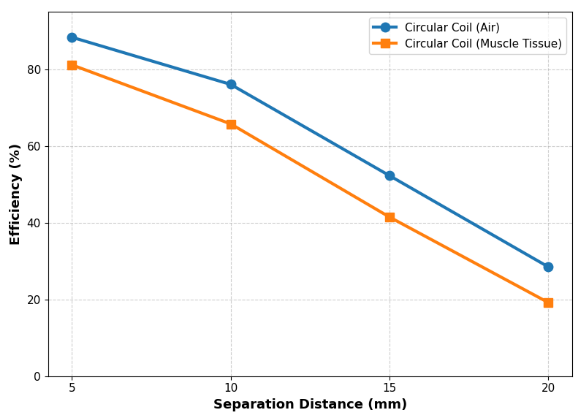

Fig 3.2: Effect of muscle tissue on power transfer efficiency of a circular coil with increasing separation distance.

Square Coil Behavior in Tissue:
Although the square coil maintains higher absolute efficiency, it experiences slightly greater percentage losses in tissue compared to the circular coil. This behavior is attributed to electric-field concentration at the coil corners, which enhances coupling to tissue and increases localized dielectric loss. In contrast, the circular coil produces a smoother and more uniform field distribution, reducing localized energy absorption.

3.4 Discussion of Findings
The results reveal a clear design trade-off:
Performance Advantage:
The square coil provides 3–5% higher efficiency in ideal conditions due to higher inductance and improved magnetic coupling.

Biological Robustness:
The circular coil demonstrates slightly better resilience to tissue-induced losses because of its uniform field distribution.
However, for deep implants (>15 mm), the square coil's higher baseline inductance becomes essential. At 20 mm depth, the circular coil's efficiency drops below 30%, while the square coil maintains a usable link. Consequently, despite higher localized tissue losses, the square geometry is better suited for extending the operational range of implantable devices, provided thermal constraints are properly managed.

4.1 Conclusion and Recommendations
It was demonstrated that the assessment of wireless power transfer (WPT) systems for Implantable Medical Devices (IMDs) is strongly influenced by coil geometry, particularly in biological environments where tissue attenuation affects efficiency. MATLAB-based analytical modeling and Simulink time-domain simulations were employed to evaluate circular and square planar coil geometries at 13.56 MHz, considering both air and biological tissue conditions. The results confirmed that coil geometry significantly affects inductive coupling strength, power transfer efficiency, and operational range in implantable applications.

The square planar coils exhibited approximately 22% higher self-inductance than circular coils for identical footprint areas, which was attributed to improved utilization of substrate space and increased conductor length. This geometric advantage translated into higher power transfer performance, with square coils consistently outperforming circular coils across all simulated distances. At a 10 mm separation, the square coil achieved approximately 79.5% efficiency in air, compared to 76.1% for the circular coil. When biological tissue effects were introduced, an average efficiency penalty of approximately 10% was observed due to dielectric losses and eddy current formation in the tissue medium. Although the square coil experienced slightly higher localized losses due to corner-induced electric field concentrations, its higher baseline inductance enabled functional operation at depths where circular coils were ineffective. The findings indicate that square planar coil geometries are more suitable for next-generation miniaturized IMDs, particularly where space constraints and transmission distance are critical. While biological tissue attenuates electromagnetic energy, the higher magnetic flux density generated by square coils compensates for these losses, extending the operational range of the system. Therefore, square coil geometries represent an optimal design choice for deep-implant applications, provided that thermal effects are properly managed. It is recommended that manufacturers of implantable devices prioritize square or rectangular coil geometries to maximize power reception within compact enclosures. Although the 13.56 MHz ISM frequency is effective for moderate implant depths, lower operating frequencies such as 6.78 MHz may be considered for implants exceeding 20 mm depth to reduce tissue absorption. The adoption of rounded-corner square (“squircle”) geometries is advised to mitigate localized electric-field hotspots while maintaining high fill-factor advantages.

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