Design and Implementation of Wireless Low-Power Transfer for Medical Implant Devices

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Abstract. Wireless power transfer (WPT) in medical implanted devices (MIDs) has received significant interest from both academic and the medical industry. These systems have suffered from battery-life that must be charged or replaced. Also, some implant devices are large, leading them to be uncomfortable. In addition, the device may interact with internal tissues, which may lead to reactions that affect the patient. This paper aims to produce a small MID operated by WPT to transmit vital signs (i.e., temperature) to an external station to ensure that the device does not affect the patient’s body. The proposed system used a flat spiral coil as a transmitter and a multi-layer copper wire coil as the receiver coil. The transmitter circuit was implanted inside a rabbit’s body. The temperature of the rabbit was sent using the nRF24L01 transceiver to the external monitoring station. The system reached an efficiency and power of 23.37% and 1.98 W respectively on 50 Ω load resistors. The proposed system was acceptable due to the small size of the coil, which provides a sufficient reception at a distance of 3 cm, exceeding the required power to operate the MID (i.e., 73 mW) and send data correctly.

Keywords: electromagnetic field; inductive coupling; medical implant device; near-field; nRF24L01; wireless power transfer

1. Introduction

Wireless power transfer (WPT) has recently found its way into the development of the nerve- and muscle-simulating system [1, 2], heart pumps [3], cochlear aids [4], retinal implants [5], infusion pumps [6], pacemakers [7], brain pacemakers [8] and other implantable devices. At the beginning of the 21st century, some medical implanted device (MID) applications, such as the nerve- and muscle-simulating system, heart pumps, cochlear aids, pacemakers, and other implantable devices, operated using batteries, despite their heavyweight quality, limited lifetime and chemical side effects, as well as the inconvenience of charging cables and the long amount of time required for the charging process. Therefore, MIDs needed to eliminate the need for batteries through WPT.

In addition, the demand for WPT systems in MIDs increased due to the ability to emulate internal organs and transfer internal vital signs to external receiver stations. MIDs have several advantages, such as being lightweight, small and more comfortable to implant directly within the tissues or organs of patients. The diagnostic and treatment effects are also more obvious. Researchers found new ways to power and observe the implant devices, such as through vibration [9], thermal and coupling links methods [10]. Most implanted devices are powered using inductive coupling links, which are used to emulate biological signals and allow for continuous observation in real-time [11]. They can be used long term and allow the patient to behave freely. In general, the MID simulator contains an external part and an internal part. The external part is fixed outside the patient and transfers power inductively to the
internal part, which is located inside the patient and receives data from it. Less power consumption, high data rate, visibility, and small size are major factors that can impact the efficiency of the complete MID and allow the patient to feel relaxed. The power is reflected and absorbed by the human tissue, which is neglected if the operating frequency is less than 20MHz [12].

The main aims of the proposed system are to transfer the power from the stationary to the implant device inside the rabbit, then the implant device transmits the temperature of the rabbit to the external station. Then, related works, formulas, schematics and works that were done will be shown. Also, in the upcoming sections, the results and safety consideration of the proposed system will be shown. The contribution of the paper can be summarized as follows:

1. The implantable system was supplied by energy-based on magnetic resonant coupling (MRC) WPT.
2. The size of the implantable device was miniaturized as much as possible.
3. The transfer distance between the transmitter and the implantable device was increased to 30 mm.
4. The temperature of the rabbit was sensed and sent outside its body to a data acquisitions system.
5. The proposed system was compared with previous research based on the following parameters: type of coupling, distance, operation frequency, power and efficiency.

2. Related Work

Kumar et al. [13] designed and implemented high-quality factor coils in a four-coil system to improve maximum power efficiency in which the four-coil system could be optimized to compare experimental results. In addition, this system was compared with a two-coil system, and it was found that the proposed system was better than the two-coil system when the transmitter and receiver coil radii were 64 mm and 22 mm, respectively. Moreover, this system was tested with a variable frequency between 100 kHz and 4 MHz the experimental result showed that the optimum frequency was 700 kHz because the input power and output power were 0.25 W and 0.2 W, respectively, with an air gap of 20 mm between the transmitter and the receiver. Qi et al. [14] designed a novel mat-based WPT for biomedical devices with moving targets using seven coils in the transmitter connected in a series and arranged in a hexagonal shape to make the magnetic field nearly flat for MIDs inside an animal. They used ANSYS HFSS software to simulate the coils and generated the signal using an Agilent (33250A) generator. They reached an average of 1.3 V at a frequency of 26.6 MHz at a distance of 8 cm. When a single animal is studied, power can be sent to a cell where the animal is located. This requires a set of driver coils that can be separately activated and a tracking system to determine the position of the animal.

Xue et al. [15] presented a high-efficiency WPT technique for MIDs, such as those used for neural recordings. They provided analysis and an intuitive physical explanation of the limitation of conventional resonant coupling. The authors adopted printed spiral coils (PSCs) and used parameters such as resistance, power, Q-factor, and voltage when tuning. They measured performance using a network analyser (Agilent E5071B). The authors measured the efficiency at a distance of 1-5 cm; it achieved 0.16% power efficiency, at an operation frequency of 13.65 MHz and output power of 100 mW. This antenna can be used in many applications that require only a small amount of power. Miao et al. [16]optimized a WPT for intraocular implantable devices. The authors developed three designs for the coils, which were tested at different frequencies between 1–10 MHz, with a change in the number of coil turns. The authors improved the WPT design by matching the network and taking the frequency as a variable coil system parameter for both the transmitter and the receiver. The coupling efficiency was approximately 19.1% at 10 MHz.

Kim et al. [17] proposed a WPT system for small hearing aids. This system was designed using an electromagnetic simulator (Maxwell 2D) and worked to recharge the batteries of small medical devices. Two different size and shape transmitter and receiver coils were used. They used a bowl-shaped coil as a transmitter coil and a box-shaped coil as a receiver coil, as well as ferrite sheets, a Li-ion battery, Bluetooth Low and a function generator Agilent (33521A), in which the free position system for the WPT used the receiver coil to charge the battery with 30% efficiency, an output power of 100 mW and an operating frequency of 6.78 MHz. The proposed wireless charging for a small implanted device would use three coils: one in the transmitter and two as reserves. In addition, all calculations would be performed theoretically, and the proposed wireless charging system for hearing aids could be connected
to the LabVIEW program, which was employed in the analysis of the hearing aids signal that connected the external part of the hearing aid to the laptop.

Badr et al. [11] used WPT to operate MIDs for telemetry from small mice, in which mice-related behaviour data were used to study disease models. The authors used a class-E power amplifier, wireless measurement system (WMS), microcontroller and nRF24LE1-F16Q24. This design used a novel technique to collect real-time performance data from the implant and then correctly measured the power transfer of loosely coupled WPT systems using a cable-free (wireless) approach. WMS can measure corrected Voltage of rectifiers values from 0–63 V, with a 12-bit resolution at 29 Hz in the rodent MID. Furthermore, the WMS system was designed to reduce its interference with loosely coupled WPT signalling and conform to design requirements. WMS works as a useful and portable tool. Shon et al. [18] presented a WPT system to recharge the battery in the neural prostheses device using widely available commercial off-the-shelf components and tested the device in rabbit models. After covering the device with a PolyJet photopolymer (MED610) and polydimethylsiloxane (PDMS) (sylgard 184, DowCorning, Midland, MI, USA), it was used to collect data and power two single-pole, double-throw analogue switches (TS3A24159, Texas Instruments, Dallas, TX, USA). They used a probe (TCP0030A, Tektronix, Beaverton, OR, USA) and a ZL70102 (MICS) as data communication protocols. The implanted device succeeded in recording and stimulating the nerves and spine during the transmission of data to the external device. The overall operation power was reached to 2.5 W (5 V, 0.5 A) at a distance of 11 mm and an operating frequency of 110–205 kHz.

Seshadri et al. [19] designed a four-coil system and implemented it to charge the battery in a left ventricular assist device. Two- and four-coil systems were designed using magnetic software. They then compared the two- and four-coil systems; experimentally, the four-coil system produced better results than the two-coil system in the MIDs, with a distance between 1–4.5 cm and a frequency between 100–350 kHz with a loose coupling factor. The authors used a function generator as an oscillator and then tested the two systems based on variable frequency, misalignment and coupling factors. They found the best frequency was 350 kHz. Aldaoud et al. [20] designed a stent-based WPT to supply MIDs with power using the body as a power receiver and two systems, inductive and capacitive, as transmitters. The system was modelled using the statistical electromagnetic simulation software CST MICROWAVE STUDIO. They tested it on bovine muscle and lived sheep. The efficiencies reached 2.6% and 1% when they placed it at depths of 15 mm and 30 mm, respectively, in the muscle tissue with an input power of 53 mW. Both capacitive and inductive methods of power transfer were analysed, and it was found that these methods were suitable for WPT with a stent. The inductive link was more efficient.

3. Methodology

3.1. Proposed FSC with MLCWC System

In this section, the main work of the transmitter and receiver circuits were described as shown in Figure 1, where the stages of processing are shows from the power supply in the transmitter circuit to the load resistance receive circuit.

![Figure 1. The block diagram of the wireless power system.](image-url)
‘TX coil’. Cross-coupled feedback is supplied via the D1 and D2 diodes. R1 and R2 are the resistors for biasing the voltage on the MOSFET gate. When DC power is applied, current flows through both sides of the coil and to the MOSFETs’ drain. Simultaneously, the DC voltage emerges at both gates and begins to operate the MOSFETs. One MOSFET is invariably a little faster than the other and will switch to the ON position more rapidly. The current would continue to grow until the coil becomes saturated.

On the assumption that Q1 begins operating first, the voltages at the Q1 drain will be driven into the ground while the voltage in Q2 stimulation increases to its peak and then falls to the ground as a resonance tank is formed by the capacitors and the FSC through a half cycle. Afterwards, D1 will be biased forward with maximal voltage. It will then repeat the operation using Q2 and repeat the cycle that is causing the sine wave. Also, the Zener diodes and 10kΩ resistors have been added between the gates of the MOSFETs and the ground. The Zener diodes protect the MOSFETs from overvoltage that may occur and limit the voltage to 12 V, and the resistors prevent the MOSFET’s gate from latching-up and becoming stuck in one state. All the parameters values are shown in Table 1.

In the receiver circuit that demonstrated in (Figure 2, part B), multi-layer copper wire coil (MLCWC) with a diameter of 3.5 cm was used as a receiver circuit. This coil was tested with two different types of rectifier. The first rectifier used a commercial AC to DC converter type (T3168) that is commonly used in WPT, and the second rectifier was a full-wave rectifier using germanium diodes. Figure 2, part B illustrates the main circuits used in the WPT receiver section. The mechanism of power transfer in the MLCWC is based on the parallel to parallel topology as MRC. The FSC and MLCWC were set at the same frequency. After saturating the MLCWC by the current, the voltage passes to the AC to DC circuit and drives the DC voltage to the load, where the overall system power and efficiency were calculated. This circuit has also been printed as development board to minimize the size of the MID, and all components of this circuit are shown in Table 2.

![Diagram of the proposed flat spiral coil (FSC) and multi-layer copper wire coil (MLCWC) system.](image)

**Figure 2.** Proposed flat spiral coil (FSC) and multi-layer copper wire coil (MLCWC) system.

**Table 1.** Components of FSC WPT transmitter circuit.

| Name            | Symbol | Type                  | Value          |
|-----------------|--------|-----------------------|----------------|
| Inductors       | L1, L2 | CLF1006NIT            | 100μH          |
| MOSFET          | Q1, Q2 | IRFP260N              | 200V/50Amp.    |
| Diode           | D1, D2 | IN4008                | 200V/2Amp.     |
| Capacitor       | C1, C2 | CPP Capacitor         | 33μF           |
| Resistor        | R1, R2 | Ceramic Capacitor     | 470Ω/3Watt     |
| Transmitter coil| Tx coil| Copper Wire           | 10μH           |
| Zener diode     | D3, D4 | IN4742                | 12V            |
| Resistor        | R3, R4 | 10K 3W                | 10kΩ           |
### Table 2. Component of MLCWC receiver circuit.

| Name            | Symbol | Type                  | Value        |
|-----------------|--------|-----------------------|--------------|
| Diode           | Bridge | DB104S                | 200V/1Amp.   |
| Capacitor       | C4     | Film Capacitor        | 20 nF        |
| Resistor        | RL     | Ceramic Capacitor     | 50, 100, 150, 200 and 250 Ω |
| Receiver coil   | Rx coil| Cooper Wire           | 100 μH       |
| Capacitor       | C5     | 0603 CAP              | 100 μF       |

#### 3.2. Data Communication System

The data transmission mechanism used a low power consumption protocol. An nRF24L01 adaptive network technology (ANT) was used as both a transmitter and a receiver in two separate devices. The ANT technology works on a frequency of 2.4 GHz, and the data transmission rate is 2Mb/s, with a maximum of 125 nodes to the connection. The transmitter circuit was implanted inside a rabbit’s body, which contained the negative temperature coefficient (NTC) and nRF24L01, and the receiver circuit, which included Arduino Uno, OLED screen and nRF24L01, was placed externally. The data transmitter was connected to the implantable development board to transfer the data from within the body to the external station. The shield of the receiving device was made by hand in the laboratory on a Vero board, which facilitates the installation and lifting of the electronic devices. The overall communication system is shown in Figure 3. The implant device is shown on the top side in Figure 6a and the bottom side in Figure 6b.

Arduino IDE program was used in the current work to make and upload the working code for the ATmega328. A program library that handles the nRF24L01 transceiver communication was downloaded from the Arduino website[21] and used in this work. The channel and the pipe were set in the code to interface the two transceivers together. A simple code for the NTC was used to collect the temperature from the NTC, the average of 10 reads was taking with an interval of 100 milliseconds between each read to stabilize the temperature reads. Then, the temperature was sent throw the nRF24L01 to the external station. In the external station, the temperature either displayed on an OLED screen or collected on Parallax data acquisition for the Excel program in real-time.

![Figure 3. Block diagram of the communication system.](image)

#### 3.3. System Analysis

The inductance of the FSC is different in the induction formula as shown in Figure 4. This type of inductance can be expressed according to the following equation [22].

\[
L = \frac{R^2N^2}{8R+11w}
\]

where \( R \) is the radius of the coil, and \( N \) is the number of turns.

![Figure 4. Flat spiral coil.](image)
3.3.1. Mutual Inductance

The current in the primary coil will generate the voltage in the secondary coils as shown in Figure 5, which leads to the transfer of energy between them. This is self-induction and is used in many home appliances and engines. However, self-induction can be questionable in some cases; if two coils are placed close to each other, one must be aware of the position of the ground pole or horizontal placement at a certain angle to avoid induction between them. Mutual inductances in coils rely heavily on the divergence and angle between the two coils [23].

\[ M = \frac{\mu_0 \mu_r N_1 N_2 A}{l} \]  

(2)

In which \( \mu_0 \) is the permeability of free space, which equal 4.\( \pi \).10\(^{-7} \), \( \mu_r \) is the relative permeability of the iron core, \( N \) is in the number of turns in the coil, \( A \) is the cross-sectional area in m\(^2 \) and \( l \) is the length of the coil in meters.

![Figure 5. Mutual inductance between two coils.](image)

3.3.2. Coupling Coefficient

According to the laws of self-induction and mutual inductance, the coupling coefficient can be derived as a percentage of interlock, as described in equation (3). Additionally, this coefficient represents the amount of correlation between the first and second coils. It can be calculated in several ways, including by being derived from mutual inductance. According to [24], it can also be obtained from the physical parameters of the transmitter and receiver coils (such as the coil’s shape, the coil’s cross-sectional area and the distance between the two coils) as explained in equation 4.

\[ K = \frac{M}{\sqrt{L_T L_R}} \]  

(3)

In which \( L_T \) and \( L_R \) are the transmitter and receiver inductors, respectively. It can also be calculated using only the coil’s radius and the distance between the transmitter and receiver coils.

\[ K = \frac{a^2 b^2}{\sqrt{\pi d^2 (a^2 + d^2)^3}} \]  

(4)

In which \( a \) is the radius of the transmitter, \( b \) is the radius of the receiver, and \( d \) is the distance between them.

3.3.3. Basics of MRC

The WPT principle depends on the unification of frequencies between coils or antennas. Coupling between the transmitter and receiver coils is obtained after the voltage in the transmitter is stabilized, and the mutual induction state appears in a normal state; however, this coupling is weak. Therefore, it is necessary to add capacitors to both sides to obtain magnetic resonance between the coils, as shown in the resonance formula. This process is usually used in coils of different sizes in which the transmitter is large, and the receiver is small. In this situation, it is necessary to add capacitors to both sides to increase the magnetic field. Moreover, the impedance must be measured on both sides. If the pairing is to be as close as possible to the ideal efficiency, the impedance should be considered as it dramatically affects the bonding between the transmitter and the receiver (as is the case in all radio transmission modes).

\[ F_{self} = \frac{1}{2\pi\sqrt{L_T C_T}} = \frac{1}{2\pi\sqrt{L_R C_R}} \]  

(5)
In which $F_{sef}$ is resonance frequency, $L_T$ is transmitter coil inductance, $L_R$ is receiver coil inductance, and $C_T$ and $C_R$ are the transmitter and receiver capacitors, respectively. Using this law, the transmitter and the receiver can be set at the same frequency, as shown below, by changing the inductance or capacitance in one side or on both sides according to the calculations already measured by equation 5. Power dissipation and WPT efficiency can be calculated through several methods; one way is using Kirchhoff’s voltage and currents laws, in which the input power and the power dissipation in the load can be calculated and obtained in the lab. The output and input powers can be calculated by calculating the voltages and currents of the transmitter and the receiver based on Kirchhoff’s laws, as shown in equations 6 and 7.

$$p_L = \frac{V_L^2}{R_L} = I_L^2R_L = V_LI_L$$  \hspace{1cm} (6)

$$\eta = \frac{p_L}{p_{TD}}$$  \hspace{1cm} (7)

There are also many ways to calculate efficiency, including the calculation of losses in the receiving circuit as described in Equation (8) [25].

$$\eta (%) = \frac{p_L}{p_{TD}} = \frac{|V_{LT}|^2R_L}{|V_{LT}|^2R_L + |I_{LT}|^2(L_T + L_R + R_L) + |I_{LT}|^2R_L}$$  \hspace{1cm} (8)

In which $V_T$ is the source voltage from the oscillator in AC, $R_{LT}$ is the impedance in ohms for the oscillator circuit, $C_R$ and $L_T$ are the resonance tanks in the transmitter, $R_{LT}$ is internal resistance of the transmitter inductance, $I_T$ is the source current, $I_{LT}$ is current that flows in transmitter inductance, $M_{TR}$ is mutual inductance, $R_{LR}$ is the internal resistance of the receiver inductance, $L_R$ and $C_R$ are the resonance tanks of the receiver, $I_{LR}$ is the current flow in the receiver inductance, $V_L$ is the voltage induced in the load and $R_L$ is the load resistance.

### 3.4. Experiment configuration of FSC Transmitter and MLCWC Receiver

An FSC with an MLCWC experiment that's contains, a ZVS oscillator consisting of two MOSFET IRFB260Ns and two choke inductors of 100 μH and 3 Amp, two resistors of 330 Ω and 3 Amp, two Zener diodes of 12 volts and two resistors of 10 kΩ. The FSC transmitter coil was made by using a 1.45 mm copper wire that consists of 15 turns in the form of an FSC in which the external of 10.5 cm and internal diameters of 1 cm. It was connected in parallel with two of the 33 μF capacitors to act as a resonance tank. The MLCWC WPT receiver consisted of a 15-wind 0.54 mm copper wire with outer and inner diameters of 3.5 cm and 2.5 cm, respectively.

The WPT transmitter circuit was supplied with 12 V DC and 0.6 Amps from the DC generator (YIHUA PS-305D). In this experiment, a digital multi-meter (Pro’sKit MT-5211) was used to measure the currents, a digital multi-meter (Aswar DT-920N) was used to measure the DC voltage, and a storage oscilloscope type (UNI-T UTD2052CEX) was used to measure the AC voltage. This circuit was connected parallel to parallel and set at the same frequency by calculating the coil’s inductance and capacitor’s capacitance according to Eq. (5). The system parameters are shown in Table 3. Parallel to parallel topology is considered more efficient than other methods (e.g., series to series, series to parallel or parallel to series). Power and efficiency were measured by calculating the voltages and currents in the transmitting and receiving segments and using different load resistors in the receiving circuits. Two types of voltage converters were applied in the receiving part of this experiment.

### 3.4.1. Method of AC to DC Converter Using Model T3168

The first method used an AC to DC converter model (T3168), and the second circuit used a germanium diode as a full-wave bridge rectifier. In the experiment with an AC to DC converter model (T3168), this type was used to obtain a constant 5 V DC with a maximum current of 600 mA. Many resistors were used in the measured AC and DC voltages for load resistance with values 50, 100, 150, 200 and 250 Ω. The calculation was measured in 0.5-cm increments over distances ranging from 0–6 cm. The purpose of the experiment was to measure the system’s overall efficiency and compare it with other circuits to determine the most appropriate use for the MID. The final MLCWC circuit is shown in Figure 6a and b, and the experimental setup is shown in Figure 7.
3.4.2. Method of AC to DC Converter Using Germanium Diode

The second method used germanium diodes (1n60p). This type of diode is characterized by a low voltage drop of only 0.3 volts because that voltage is considered more efficient concerning currents <1 Amp as the voltage drop is less than that in conventional diodes (silicon diodes), which have a voltage drop of 0.7 volts. This experiment measured voltages and currents using the experimental setup of the previous experiment to compare it with other systems that were tested according to power and efficiency.

![MLCWC circuit](image1.png) ![MLCWC circuit](image2.png)

**Figure 6.** MLCWC circuit. (a) bottom side and b (topside).

![FSC with MLCWC](image3.png)

**Figure 7.** FSC with MLCWC during the experiment of WPT.

| Parameters                              | Value          |
|-----------------------------------------|----------------|
| Input voltage (E/Volt)                  | 12             |
| Operating frequency (f/KHz)             | 67             |
| Inductance of transmitter coil (µH)    | 10             |
| Inductance of receiver coil (µH)       | 30             |
| Compensating capacitor of transmitter coil (nF) | 33         |
| Compensating capacitor of receiver coil (nF) | 10         |
| Load resistance (RL/Ω)                  | 50, 100, 150, 200, and 250 |
| Air gap between coils(cm)              | 0-6            |
| Number of turns in the transmitter coil | 15             |
| Number of turns in the receiver coil    | 15             |
| Thickness of wire in the transmitter coil (mm) | 1.45         |
| Thickness of wire in the receiver coil  | 0.54           |
| Diameter of the coil in the transmitter coil (cm) | 10.5  |
| Diameter of the coil in the receiver coil (cm) | 3.5        |

4. Implantation Process

The process of implanting the development board under the rabbit’s skin in the chest under the left paw was performed by a veterinarian surgeon. After the surgery with the development, the electronic board was covered with acrylic material. The surgical steps are summarized below.

1. Make an alginate mould [26] after mixing it with water in a suitable container, and insert the development board (such as an implantable device) inside the mix in order to mould the shape of the surface device into the material of the alginate in order for it to serve as a template that is prepared...
for the process of covering the implantable device with acrylic material. After inserting the implantable device in the mix for 10 min, check to determine whether the material has hardened, and then extract the implantable device from the mould.

2. The inner surface of the alginate mould is covered with a greasy substance to prevent adhesion of the acrylic material to the mould.

3. Analyze the solid acrylic material [27] with a special liquid material of a specific percentage, and mix it until the material is ready.

4. Place a suitable amount of acrylic in the middle of the two-piece mould. Then, placed the MID inside the mould, and press the two pieces of the mould onto the MID. The acrylic will work as a cover for it. Maintain this pressure for an hour, and then open the mould to extract the MID that is covered.

5. Clean the surface of the cover to remove the fatty material, adjust its sides using an iron cooler and then leave it in the air for a period to dry.

6. Sterilize the surface of the MID and all the material that is used in operation. Anaesthetize the rabbit, and then begin the process of implanting the device, as shown in Figures 8a and b.

7. The device that is shown in Figure 8c is the implantation under the rabbit’s skin. After monitoring the rabbit for seven months, no side effects from the chemical aspect were observed. Figure 8d shows the overall final system during the measurement of the temperature readings obtained from the rabbit.

Figure 8. The operation of the implanting development board under the skin of the rabbit (a) during implant, (b) after the implant completed, (c) development board covered by acrylic and (d) while receiving data.

The hand shown in Figure 9 is used to receive vital signals or to stimulate the nervous system as needed. The development board provides an easy way to deal with the signals and send them because it contains a microcontroller and transceiver radio frequency (RF) that can be programmed according to what is needed with simple additions (such as sensors, actuators). However, it would be challenging to implant it in a patient because the device has not received a license for implantation in the human body and is still in experimental stages. In this study, the rabbit was chosen because a rabbit’s vital signals are close to those of humans, and the rabbit’s size is suitable for the device that was manufactured with the purpose of implantation without affecting the rabbit’s comfort. The device was planted under the rabbit’s skin and remained under surveillance for three months to date and is still easy to control.
5. Safety Consideration

One of the most critical parameters that should be given special attention is patient safety that is based on previous studies and global standards/guidelines. Specific absorption rate (SAR) and the amount of energy allowed are also described by these guidelines. The effects of biomedical materials used in the packaging of the MID were also studied. It has been shown that each of the parameters used in the current work does not affect the patient in an adverse way.

5.1. Frequency and Electromagnetic Safety

This section presents the most essential global standards for the RF, which set the limits of the usable frequency in the WPT. In addition to the RF and WPT frequency limits, global standards also present the effects of the magnetic field, the ability of the body to absorb the energy and allowable limits to which a body can be exposed. In this work, two different frequencies in WPT and data communication were studied to show the possibility of using both frequencies because these frequencies are within the permissible limits in terms of the frequency and energy to which a body can be exposed. The frequency of the device was 67 KHz. The communication frequency was 2.4 GHz, and the maximum power received by the implanted device did not exceed 2 W. There are two electromagnetic standards. One is the Standard for Safety Levels concerning Human Exposure to EMF [29] from the IEEE. The other is the Guidelines for Limiting Exposure to Time-Varying Electric, Magnetic and EMFs [30] from the International Commission on Non-Ionizing Radiation Protection (ICNIRP). Both IEEE standards and the ICNIRP guideline give the maximum permissible non-ionizing radiation exposure. The IEEE standard covers a high range of frequencies ranging from 1 Hz to 300 GHz. The ICNIRP guideline also covers a high range of frequencies in the range from 3 kHz to 300 GHz. Both ranges cover the usual operating frequency in the WPT, which ranges from 0.01–100 MHz the permitted exposure determines the maximum electrical field strength and the maximum magnetic flux density. Compared to conventional cable power systems, WPT suffers from safety issues. One of these is the risk of exposure to electromagnetic radiation. Exposure to the low-frequency magnetic and electric field causes absorbed energy to be ignored, but these two parameters produce a thermal effect on the body that can be calculated. However, exposure to the EMF with a frequency >100 kHz leads to effective absorbed energy and an increase in body temperature. Consequently, understanding and considering radiological safety issues are very important [31].

5.2. Specific Absorption Rate

The interaction of electromagnetic waves with organisms is very complicated. To describe the level of interaction of an organism with electromagnetic waves, the American researcher Schwan (in the 1860s) suggested defining Specific Absorption Rate (SAR). SAR is usually used to define the interaction of electromagnetic waves with humans and to determine the maximum level of electromagnetic radiation relevant to human safety [32]. It can be defined as the absorbed power per mass of tissue, and its unit is W/kg [32, 33]. SAR represents the time derivative of electromagnetic energy absorbed or consumed by a mass of 1 kg of human body weight. It can be calculated using Equation (9).

\[
SAR = \frac{d}{dt} \left[ \frac{dW}{dm} \right] = \frac{d}{dt} \left[ \frac{dW}{\rho \cdot dv} \right]
\]  

(9)

In which \( \rho \) represents the biological tissue conductivity, \( W \) represents radiated power, \( t \) represents time, \( m \) represents the biological tissue weight and \( v \) represents the biological tissue volume. SAR is usually calculated either for the entire body or for small sample size (usually 1 g or 10 g of tissue). SAR measures exposure to radio waves [34]. It is commonly used to measure absorbed energy during magnetic resonance imaging (MRI) scans, in which the measured value depends mainly on the geometry of the body that is exposed to RF energy, the body’s location and the RF source’s geometry. Therefore, the
test must be performed by each specific source and at the intended position of use [31]. In theory, the insulating properties of the different parts of human tissue are contradictory, resulting in uneven distribution of EMF in human tissues. The electromagnetic radiation energy levels absorbed by human tissues are closely related to the conductivity of the human organ in question. Most human tissues contain a lot of water, which leads to higher conductivities. SAR is different for different parts of the body. To understand the possible benefits and mechanisms of EMF and body interactions, the minimum SAR that can have a biological effect would be more valuable compared with the study of high-intensity fields. These studies can focus on the maximum effect of non-ionizing radiation. In addition, it is possible to reduce the complexity of the effective electromagnetic interaction in different cells by reducing energy exposure, thereby reducing at least the overall rise in cellular/body temperature. These parameters vary according to the situation being studied and depend on the physical and biological conditions of the intended target [29]. In this work, the energy did not exceed 2 W/kg, which indicates the work was within the safety limitation of SAR.

5.3. Biomedical Materials

One of the crucial properties that must be considered is the biomedical material’s effect on the body because of the chemical reaction that may occur between the implanted device and the patient. Many researchers have investigated this issue. In this work, a biomedical substance that has been authorized by the International Organization for Standardization (ISO 1567-type 2 class 1) was used [35], which is a certified world licensed material according to standards set by it for each type of MID, chemical materials or medical materials. The cold acrylic material, which is numbered according to the ISO, was already applied as a cover for the device that was implanted under the skin of the rabbit and remained under observation for three months. No side effects were detected. On the other hand, the tissue was built around this material.

6. Results and Discussion

After completing the FSC experiment with the MLCWC, the main parameters of the experiment were taken to calculate the experimental data, including the voltage peak to peak, DC voltage, transfer power and efficiency of the proposed system. Two methods were used to determine which is best at converting voltages from AC to DC. This proposed system was applied to operate subcutaneous MIDs. Adequate power must be provided at a suitable distance to run the implanted MID without affecting the patient’s comfort.

6.1. Result of AC to DC Converter Using Module T3168

The AC voltages of the transmitter and the receiver were measured for this proposed system (i.e., FSC with MLCWC when using AC to DC module T3168). The voltages were measured using the digital oscilloscope model (UNI-T UTD2052CEX), as seen in Figure 10a, which depicts the transmitted voltage when using 50 Ω load resistors and demonstrates that the signal is clear and does not suffer from any deformities. Although the received voltages were half-wave signals at a distance of 0 cm, the voltage stays in a half-wave signal configuration but smooths out at a distance of 3 cm. Then, it continues the rest of the distances in the same state, as can be seen in Figures 10c and d. The 50 Ω resistors achieved the best power and efficiency.
Figure 10. The measured AC voltage of (a) transmitter signal, (b) received signal at 0 cm, (c) received signal at 3 cm and (d) received signal at 5 cm.

For the T3168 AC to DC converter, the relationship between the AC voltages with the distance for different loads is illustrated in Figure 11a. As depicted, the AC voltages do not change significantly when the resistors change. Therefore, the AC voltage was calculated directly on either side of the receiver coil in the circuit when using resistors of 50, 100 Ω, 150, 200 and 250 Ω for distances between 0–5 cm. Then, the use of the AC to DC converter type (T3168) depicts when the AC voltage that is received has reached 60 volts in the best case and begins decreasing to 8 volts in the worst cases. Figure 11b shows the variance in DC voltages when using different resistors at different distances. As can be observed, the voltage increases as the values of the resistors increase, unlike the current that flows in an opposite relation with the resistance according to the readings of the current, which were measured and which conform to the ohms law. When the voltage has exceeded 3 volts at a distance of 3 cm, it is possible to operate the MID at this distance. In addition, this model (i.e., T3168) demonstrated a voltage exceeding 5 volts. An increase in temperature was observed when the voltages were increased by more than 5 volts. Therefore, this model is suitable when using different resistors at a distance between 1–3 cm and when the voltage exceeds 3 volts, which is sufficient to operate the sensor, wireless protocol, and microcontroller in the MID.

Figure 11. Relationship for different load over distance for (a) voltage peak-to-peak and (b) DC-voltage.

The output power of the proposed system was calculated according to Equation (6). The received power was calculated for five resistors (i.e., 50, 100, 150, 200 and 250 Ω). As shown in Figure 12a, the power in the lower resistors is better than the power in the larger resistors because the current is higher in the smaller resistors. The 50 Ω resistors reached 0.7 W of power, which is more than that achieved in the other resistors. Despite the reduced power in this proposed system, there is sufficient power to operate the MID up to a distance of 3 cm; when the power supply is above 100 mW, it is enough to operate MIDs. The efficiency of the proposed system reached 9.3% for the 50 Ω resistor, as seen in Figure 12b. However, the efficiency of the proposed system did not exceed 3% in large resistors (e.g., 200 and 250 Ω), likely as a result of the DC voltage specified in the receiver circuit (i.e., T3168) and the difference in size between the transmitter coil and the receiver coil. The measurements were calculated up to 5 cm because 5 cm is a greater distance when the system can receive the voltage and the current.
Moreover, the relationships between the transfer power and the transfer efficiency were determined in the form of a 3D stereogram. Figures 13a, b, c, d, and e demonstrate the third relationship for the resistor of 50, 100, 150, 200 and 250 Ω, respectively. These figures clearly illustrate the holographic connections and accurately depict any defects. As shown in Figure 13, efficiency decreased over several intervals, especially in the first 2 cm. These figures also showed that a decrease in the first measures was commonly followed by an increase in the second measures. This may have happened due to the effect of the high voltage being clipped in the receiving circuit, or it may have occurred due to the difference in impedance matching between the transmitter and the receiver coils. This will cause a lack of coupling between the circuits, creating a ripple in the rise and fall in the signals received. These cases are reflected in the shape of power and efficiency. The voltage gradually stabilizes when the voltage that was clipped is lower and then passes a stable DC voltage that is less effective in the circuit. The relationships were calculated for five different resistors in this method; the colours in the figures represent the amount of energy received and the efficiency. The colour of the nut in the 3D figures indicates high efficiency: orange indicates a slightly reduced efficiency, yellow refers to medium efficiency, and blue indicates poor efficiency.

6.2. Method of AC to DC Converter Using Germanium Diode

The second converter in the same proposed system was used with the same coil for the transmitter and the receiver but was changed in the receiver circuit. The germanium diode was used in the full-wave rectifier converter and was found to be better than the previous module (i.e., T3168) because germanium diodes are more flexible regarding DC voltage. Moreover, the distance of the proposed converter was more than the previous model by 1 cm. The AC voltage in both the transmitter and receiver of the FSC with the MLCWC when using a germanium diode was measured in a digital oscilloscope. As depicted
in Figures 14a, b, c and d, a clear sinusoidal signal appears. Additionally, the signal does not suffer from any distortion in either the transmitter signal or the receiver signals. The signal in the transmitter side is fixed at 78 volt-AC, as shown in Figure 14a. The signal received when using the 50 Ω resistors, however, reached 52 volt-AC (see Figure 14b) in its first reading at 0 cm. At a distance of 3 cm, it reached 14.2 volt-AC (see Figure 14c), and at the distance of 6 cm, it reached 3.56 volt-AC (see Figure 14d).

![Figure 14](image)

Figure 14. The measured AC voltage of (a) transmitter signal (b) received signal at 0 cm (c) received signal at 3 cm (d) received signal at 6 cm.

The received AC voltages were in the form of a sinusoidal wave, as opposed to the previous model, in which half waves were used. Figure 15a depicts the peak-to-peak voltage of five different resistors. The voltage of the large resistors was found to be better than the voltage in the smaller resistors because the mutual index of the larger resistance in this circuit is better than the smaller ones, and the current of the larger resistance was lower than the small resistance. In addition, the relationship between the DC voltages over a given distance with different loads was measured (see Figure 15b). Although the DC voltage was better than in the previously proposed converter (i.e., T3168), the proposed germanium rectifier demonstrated the best voltage in the 250 Ω resistor, which reached 13 volts. However, the current flow in the 50 Ω is larger than in the other resistors.

![Figure 15](image)

Figure 15. Relationship for different load over distance for (a) voltage peak-to-peak and (b) DC-voltage.

The output power of the FSC with the MLCWC when using germanium was also calculated. The power of five resistors over a given distance was measured, as shown in Figure 16a. Maximum power was achieved by the 50 Ω resistor, while the least amount of power was achieved by the 250 Ω resistor. The 50 Ω resistor reached 1.7 W, whereas the 250 Ω resistor only reached 0.7 W. The force reduction in the large resistors occurs because of the increased difference in impedance matching between the transmitter
and receiver circuits, resulting in a lack of coupling between the circuits, causing a ripple in the rise and fall of incoming signals as explained previous literature \([36-40]\). These situations are reflected in the form of power. Moreover, the efficiency of this proposed system was measured for five resistors (see Figure 16b).

The WPT efficiency of the proposed system of the FSC with the MLCWC when using germanium diodes as a full-wave rectifier reached 23.37\% when using the 50 Ω resistor and 10\% when using the 250 Ω resistor. The transfer efficiency was considered to exhibit adequate levels of efficiency for the operation of the MIDs utilizing this type of coil (i.e., MLCWC). The efficiency of this system reached 1\% at a distance of 3 cm and is, therefore, a good and efficient machine to operate the MIDs. If the efficiency reaches more than 1\% in the receiving party, it will be sufficient to run the implanted device with no more power than 0.1 W. Moreover, the efficiency gradually decreased as distance increased, but this is normal in near-field WPT techniques.

![Figure 16](https://example.com/fig16.png)

**Figure 16.** Relationship for the different load over distance for (a) transfer power and (b) transfer efficiency.

A 3D graph of this system over five different loads is presented in Figures 17a, b, c, d, and e. The 3D graph illustrates the relationship between efficiency, power, and distance. This relation demonstrates the advantages of the system and reveals which resistor is better based on the measured parameters. These figures also reflect the characteristics of the power and the stability of received power at different distances. As shown in Figure 17a, the load resistor is used as a 50 Ω resistor that works properly as the descent is gradual. While the 150 Ω resistor exhibits a slight decrease in the signal received at a distance of 1 cm, the rest of the resistors are stable.

![Figure 17](https://example.com/fig17.png)

**Figure 17.** The relationship between the power and efficiency with distance for different loads (a) 50 Ω, (b) 100 Ω, (c) 150 Ω, (d) 200 Ω and (e) 250 Ω.

### 6.3. Characteristics of the MID

After implanting the manufactured MID, data were collected on the temperature of the rabbit’s body using the NTC sensor. The data were sent from the body and received by an external device using a...
small and low power consumption transceiver (i.e., nRF24L01). The collected data were measured for 100 sec when the device was operated by WPT. The readings were close to the rabbit’s actual temperature of 39 degrees Celsius. The received temperature was displayed in the DAQ in real-time. In addition, the period between the measurements was 1 sec, as shown in Figure 18a. Ten measurements were taken every 1 sec, and the total was averaged to obtain a more accurate measurement. The current consumption for the MID was also measured according to the current that operates every component of the MID (i.e., temperature sensors, microcontroller, and wireless protocol). The current that operates the microcontroller was 9 mA, and the current that runs the temperature sensor was 0.2 mA, while the current of the wireless protocol was 13 mA. The current that was sufficient to run the proposed MID reached 22.2 mA. The power to run the MID was achieved 72 mW when the device sent data and 30 mW when the device measured the data of the sensor. In addition, the size of the MID (development board) was measured at 3.4 x 2.5 x 0.5 cm³ and weighed 7 gm. The MID was then covered with acrylic material to avoid interactions with the rabbit tissue. Thus, the weight of the MID increased to 10 grams, and its size increased to 3.5 x 2.6 x 0.6 cm³ the electronic piece was implanted inside the rabbit’s body during surgery.

The rabbit was placed under general anaesthesia by a veterinarian at the veterinary clinic. The surgery took place on 3 September 2018, and the device is still implanted in the body of the rabbit. No side effects have been noticed since the surgery. Moreover, the rabbit has been moving and eating naturally, and we were able to measure its temperature when it approached the WPT circuit, as measured in Figure 18a for 100 sec (samples) (on 7 Sep. 2018). Figure 18b presents the temperature measurement (on 13 Dec. 2018) for 80 sec (samples).

Figure 18. The measurement of temperature from the rabbit (a) on 7 Sep. 2018 and (b) on 13 Dec. 2018.

6.4. Comparison of the Proposed System Result with Previous Works

The proposed system of the FSC with the MLCWC, which used a small coil, was compared with similar previous works depending on transfer efficiency, transfer power, distance, techniques, and frequency. As presented in Table 4, it was found to perform better than previous works in terms of transmitter distance and efficiency. The efficiency of the system was 23.37%. The proposed system also enabled the operation of the MID at a distance of 3 cm. In addition, the proposed system could operate devices with less power at a maximum distance of 6 cm.

| References | Type | Distance (cm) | Frequency (MHz) | Power (W) | Efficiency (%) |
|------------|------|---------------|----------------|----------|----------------|
| [11]       | MRC  | >9.5          | 2.057          | 0.85     | 1.5            |
| [20]       | IC   | 1.5           | 4              | 0.53     | 2.25           |
| [41]       | IC   | <3            | 1500           | 15.7     | 0.5            |
| [42]       | MRC  | 1             | 13.56          | 110      | 11.7           |
| [43]       | MRC  | 0.5           | 10             | 0.2      | 15.2           |
| [44]       | IC   | 0.5           | 7.15           | 0.029    | 4.1            |
| MTCWC      | MRC  | 0            | 1.68@ 0 cm     | 23.37%   |                |
| (This work)|      | 3             | 0.067          | 0.104@ 3 cm |                |
|            |      | 6             | 0.008@ 6 cm    | 0.012@ 6 cm |                |
7. Conclusions
In this paper, the FSC with the MLCWC system was tested for their ability to transfer power wirelessly in MID application. In this system, an FSC with a diameter of 10.5 cm was used as a transmitter coil, and an MLCWC with a diameter of 3.5 cm was used as the receiver coil. This system was tested on five resistors (50 Ω, 100 Ω, 150 Ω, 200 Ω and 250 Ω) with two types of AC to DC converters (i.e., T3168 and germanium diode rectifier). Maximum efficiency and output power were achieved when using a germanium diode rectifier with a resistor of 50 Ω. The proposed system reached an efficiency of 23.37% with an output power of 1.7 watts. Meanwhile, the efficiency reached 10% with an output power of 0.7 watts when using a T3168 AC to DC converter with the same load resistor. The germanium diode rectifier shows better transfer power, transfer efficiency and transfer distance than the conventional module (i.e., T3168). In addition, it has been shown that the received capacity is inversely related to the distance. Moreover, the small resistors give higher efficiencies than they do in large resistors. A PCB was manufactured as a WPT receiver circuit with the ability to replace the receiving coil in the future, hence reducing the size of the circuit to be implanted. The MLCWC circuit was used to be implanted in the body of the rabbit due to its acceptable size, and the efficiency of this system was acceptable based on a comparison with previous works. As shown this works were better than previous works in many parameters like the power consumption, transfer distance and received power. In addition, the data were transmitted and received correctly, and the temperature of the rabbit was measured with significant accuracy compared to the degree of the natural temperature of the rabbit. The encapsulating materials and used frequencies also did not show any side effects on the rabbit. Moreover, this current work is very important because it has been applied in practice since most of the previous research was limited to design only. In the future, the MID can be improved by adding other sensors to the device.

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