

Design of a Magnetically Coupled Resonant Implantable Medical Device Wireless Power Transmission System

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Abstract. This paper systematically describes the design and optimisation strategy of wireless charging technology for implantable medical devices. Starting from the theoretical basis of magnetically coupled resonant wireless power transmission (MCR-WPT), an efficient transmission circuit is designed and the resonant circuit characteristics are analysed. Meanwhile, the effects of the outer diameter, the number of turns of the transmitting coil and the turns spacing on the magnetic flux density and the AC internal resistance are obtained through the principle analysis. The coil parameters are adjusted through simulation experiments to improve the transmission efficiency. Then, the safety and stability of the device are ensured by temperature-power double closed-loop control. Then the closed-loop phase-shift control strategy is introduced to ensure that the system can output power stably in complex environments. In this paper, a low-power, safe and high-efficiency wireless energy transmission system is built, which provides a comprehensive and in-depth technical reference for the development of wireless power supply technology for implantable medical devices.

Keywords: Magnetically coupled resonant wireless energy transmission, implantable medical devices, dual closed-loop control, phase-shift control.

1. Introduction

With the rapid advancement of medical technology, implantable medical devices have become a key innovation in improving the quality of life and treatment outcomes for patients. They are able to penetrate deep into the body to continuously monitor health conditions, administer treatments or assist in the rehabilitation process. However, a central challenge in ensuring the long-term, stable operation of these devices is the continuous supply of energy.

Traditional implantable medical devices rely on built-in batteries or wired connections for power, but batteries have a limited lifespan, and once they run out of power, the device will not function properly. Due to the lack of battery life, patients may need to undergo multiple surgeries to replace the battery, which not only increases patient pain, but also brings additional surgical risks and financial burdens. At the same time, traditional implantable medical devices may have potential risks in terms of electrical and mechanical safety, such as wire interface infections, accidental electric shocks, and other safety issues. These problems may endanger patients' lives. In summary, traditional implantable medical devices have obvious drawbacks in terms of battery life and replacement, safety and maintenance costs. These drawbacks limit the application and development of traditional implantable medical devices, prompting people to continuously explore new technologies and methods to improve these problems. Wireless energy transmission technology provides a new solution for implantable medical devices, which is expected to solve the battery life and replacement problems and improve the safety and reliability of the devices.

Wireless energy transmission technology uses electromagnetic fields, microwaves or lasers as a transmission medium to safely and stably transmit electrical energy from an external power source to an internal device without physical contact. The application of this technology not only fundamentally solves the problem of power supply for implantable medical devices, but also greatly reduces the inconvenience and potential risk for patients who need to frequently replace batteries or maintain wired connections. Currently, the main applications of wireless energy transmission technology in medical systems include artificial organs, cochlear implants, pacemakers, micro-robots in medical

institutions, hearing aids [1-5]. However, the wireless energy transmission method is mainly magnetic coupling inductive, for magnetic coupling inductive wireless charging, it requires accurate coil position alignment, while the transmission distance is relatively short. Therefore, it is difficult to achieve inductive charging in the existing way, this paper adopts the magnetic coupling resonant wireless charging to achieve a longer transmission distance and low precision requirements for the device parking position.

For this reason, this paper focuses on the magnetic coupling resonance-based wireless energy transmission technology, aiming to build a low-power, safe, and high-efficiency wireless energy transmission system with good robustness for the application of implantable medical devices.

2. Analysis of the Principles of Wireless Energy Transmission

2.1. Basic Structure of the System

As shown in Figure. 1, the implantable medical device wireless energy transmission system consists of an extracorporeal electrical energy transmitter and an intracorporeal receiver-receiver to form an energy transmission path. The DC source generates high frequency AC power through the inverter circuit, which drives the transmitting coil to produce an alternating magnetic field, generating current in the receiving coil and thus realising non-contact energy transfer. The induced current is processed by the rectifier and filter circuit and converted into stable DC power, which is supplied to the implantable medical device. The system utilises the principle of electromagnetic induction to achieve the mutual conversion of electrical and magnetic energy, ensuring the stable and safe operation of the device in the body.

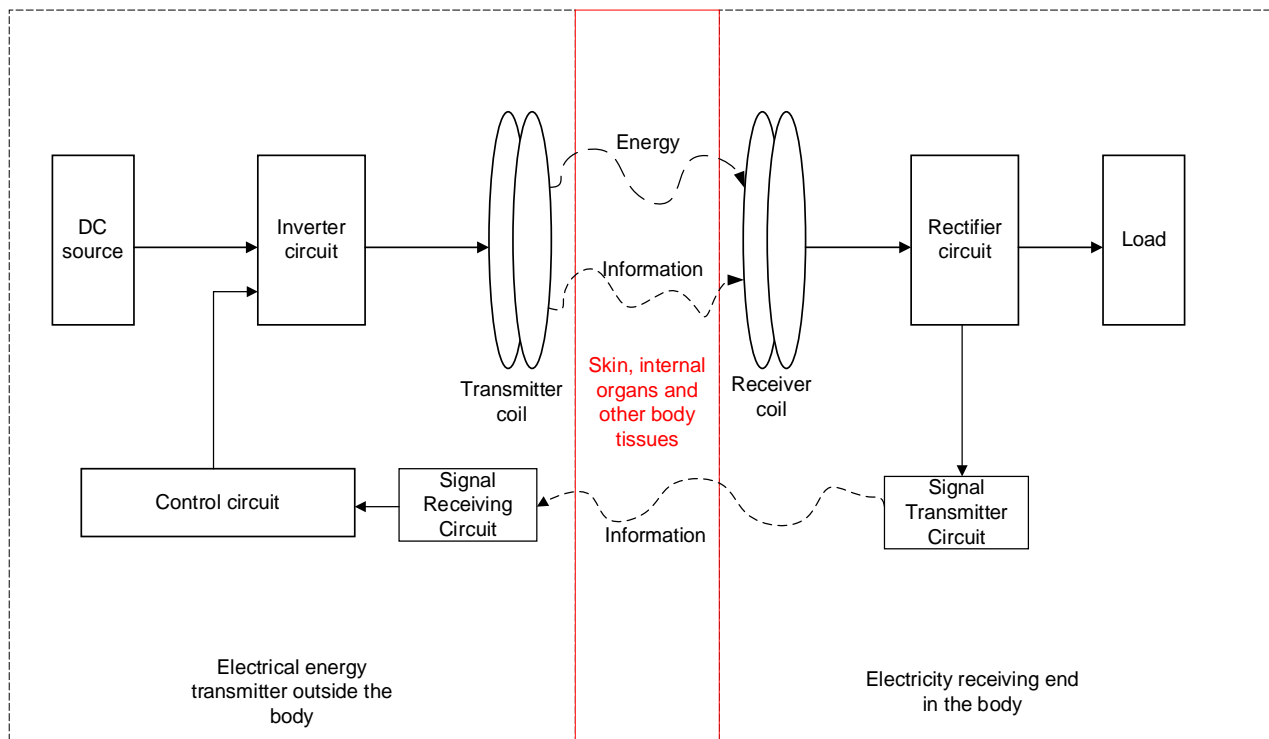


Fig. 1 Wireless energy transmission system for implantable medical devices

2.2. Resonant Circuit Analysis

The phenomenon of resonance is based on the interaction between the intrinsic frequency of the vibrating system and the frequency of the external excitation. When the external excitation frequency matches the intrinsic frequency of the system, the vibration system will resonate, absorbing more energy and increasing the system amplitude. Resonance has the characteristics of high amplitude gain, phase invariance, and maximum power transfer, which have a significant impact on system

performance. Magnetic resonance coupled wireless power transmission (MCR-WPT) is a technology that uses an electromagnetic system with a specific operating frequency to transmit electrical energy over long distances without the need for a physical connection. When the operating frequency of the transmitter (primary side) and the receiver (secondary side) of the system are matched, the two will enter a strong coupling state. At this time, the power provided by the transmitter side can be efficiently transmitted to the receiver side, thus completing the wireless transmission of power.

2.2.1. Series Resonant Circuits

In a series resonant circuit, a frequency point f_0 occurs when the inductive reactance value of the circuit is equal to the capacitive reactance value, at which point it is obtained that $X_L = X_C$. The basic circuit diagram for achieving resonance in series is shown in Figure. 2. Corresponding resistance: R , inductance: L and capacitance: C . The resonance is provided by the AC power supply as the energy source.

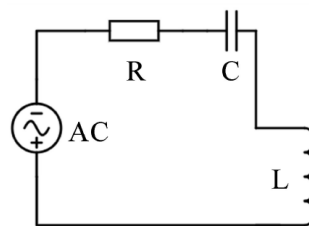


Fig. 2 Series resonant circuit

The frequency of the AC current is f_0 , the corresponding angular frequency is ω_0 , and the circuit impedance is:

$$Z = R + j\omega L + \frac{1}{j\omega C} \quad (1)$$

Simplified as:

$$Z = R + j\left(\omega L - \frac{1}{\omega C}\right) = R + jX \quad (2)$$

The voltage in the circuit is:

$$U_s = I_s \left[R + j\left(\omega L - \frac{1}{\omega C}\right) \right] \quad (3)$$

At resonance, the voltage is in phase with the current, when the imaginary part of the impedance is 0. It is known:

$$\text{Im}[Z] = X = \omega L - \frac{1}{\omega C} = 0 \quad (4)$$

Let the angular frequency at resonance be ω_0 , so that we have:

$$\omega_0 = \frac{1}{\sqrt{LC}} \quad (5)$$

From equation (5), it can be seen that the resonant frequency is independent. It is not affected by the magnitude of the external excitation source, but is strictly determined by the specific values of the circuit parameters L and C within the system. Therefore, once these circuit parameters are set precisely, the resonant frequency of the system is also determined.

The resonance phenomenon is initiated when the frequency of the applied AC signal is aligned with the intrinsic frequency of the system, also known as the resonance frequency. In this configuration, the system will attain a unique equilibrium wherein the reactive power generated by

the inductive elements is numerically equal but opposite in phase to that generated by the capacitive elements. This results in the complete cancellation of reactive power. At this point, the total impedance of the system is purely resistive, which facilitates the efficient transmission of electrical energy.

2.2.2. Circuit Compensation Network Analysis

The implementation of wireless charging technology can be classified into four basic topologies based on how the capacitors are accessed in the circuit: series-series, series-parallel, parallel-series, and parallel-parallel. The topologies in question are illustrated in Figure. 3. Each of the aforementioned topologies possesses distinctive circuit configuration characteristics that exert an influence on the performance of the wireless charging system, the energy transfer efficiency, and the overall stability of the system.

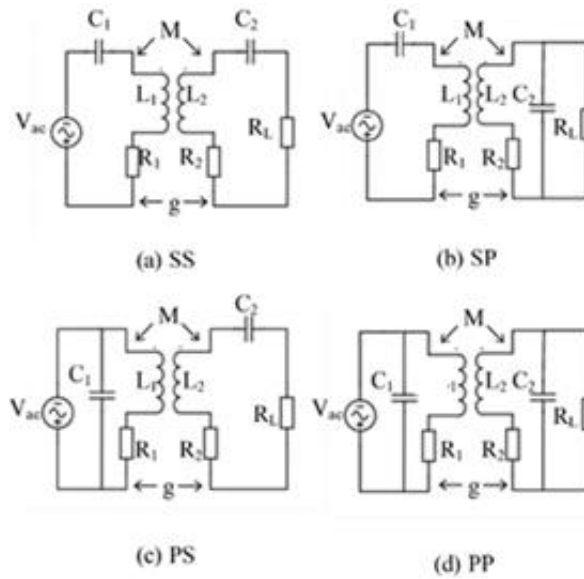


Fig. 3 Equivalent circuits of four capacitor compensation structures

The selection of the S-S type resonant compensation method (primary-side series and secondary-side series) as the object of study is mainly based on its relatively simple structure and high stability. In the S-S compensation network, the selection of the compensation capacitance depends only on the primary-side inductance L_1 and the operating frequency f of the system, and this independence enables the system to keep the compensation capacitance value stable and unchanged in the face of the coil position movement or load change. This feature is particularly important for application scenarios where charging a battery is concerned, as it ensures stability and reliability during the charging process. When the system operates at a frequency f and the capacitor C and the inductor L form a resonant circuit, their relationship satisfies the resonance condition:

$$f = \frac{1}{2\pi\sqrt{LC}} \quad (6)$$

The operating principle of the S-S circuit and the circuit parameters can be used to derive the formula for the compensation capacitance.

The primary winding side impedance is:

$$Z_s = R_1 + j\omega L_1 + \frac{1}{j\omega C_1} \quad (7)$$

At resonance, the inductive and capacitive reactances cancel each other out:

$$j\omega L_1 = \frac{-1}{j\omega C_1} \quad (8)$$

Therefore:

$$C_1 = \frac{1}{\omega_0^2 L_1} \quad (9)$$

Based on the same principle:

$$C_2 = \frac{1}{\omega_0^2 L_2} \quad (10)$$

Among them, R_L is the load, C_1 and C_2 are the primary and secondary compensation capacitors, ω is the system working angle frequency, and L_1 and L_2 are the primary and secondary coil self-inductance.

At this point, the parameters of the circuit can be designed as shown in Table 1:

Table 1. Circuit parameter values

Circuit parameters	Numerical value
System frequency f	100 kHz
Primary side coil self-inductance L_1	1.046 μ H
Side coil self-inductance L_2	0.757 μ H
Primary compensation capacitance C_1	24.198 nF
Secondary compensation capacitance C_2	99.183 nF
Primary coil equivalent resistance R_1	10 m Ω
Secondary coil equivalent resistance R_2	10 m Ω
Primary input voltage V_1	DC 240V
Secondary input voltage V_2	DC 0.12V
Mutual Inductance M	0.128 μ H
Load R_L	0.1 Ω

2.3. Design of the Inverter Circuit

In light of the aforementioned characteristics of the pacemaker, it is evident that the rectifier circuit must be designed in a manner that allows for the advantages of a stable output voltage, high efficiency, and a robust anti-interference capability. Consequently, this paper proposes the utilisation of a voltage-type full-bridge inverter circuit as the optimal solution. The voltage-type full-bridge inverter circuit is capable of mitigating the impact of power supply voltage fluctuations on the output voltage, enhancing the stability and reliability of the equipment while accommodating a range of diverse electrical loads, output current and voltage stability, and the inverter's robust load capacity. In addition, its output frequency is adjustable, adapting to a variety of operating frequency equipment. The full-bridge inverter is also able to realise sinusoidal waveform output with good output waveform quality, which can eliminate harmonic components of different frequencies and meet the working requirements of different equipment. Through the PWM (Pulse Width Modulation) technology, it can achieve precise control of the output voltage and current, including the amplitude, frequency and phase of the voltage, as well as the waveform and phase of the current. Therefore, voltage-type full-bridge inverter circuits are very suitable for use as the inverter part of wireless power transmission circuits.

2.4. Design of Rectifier Circuit

Similarly, the selection rules for rectifier circuits should be the same as those for inverter circuits, so bridge rectifier circuits are selected in this paper. Compared with other rectifier circuits, the bridge rectifier circuit has less ripple voltage in the output voltage, which makes the output DC voltage smoother. Since each diode is subjected to a lower maximum reverse voltage in the bridge rectifier circuit, this helps to improve the safety and stability of the circuit. In addition the power transformer

is fully utilised as the current is supplied to the load during both the positive and negative half-cycle, thus improving the efficiency of the rectifier circuit. Bridge rectifier circuits have a relatively simple structure, and with the rapid development and cost reduction of semiconductor devices, their cost advantage is more obvious. Bridge rectifier circuits are usually used in conjunction with filters, which can effectively filter out the AC component of the rectified voltage to obtain a more stable DC voltage.

In conclusion, this paper presents the construction of a wireless energy transmission circuit based on the magnetic resonant coupling principle, with a particular focus on its potential applications in implantable medical devices, as shown in Figure. 4.

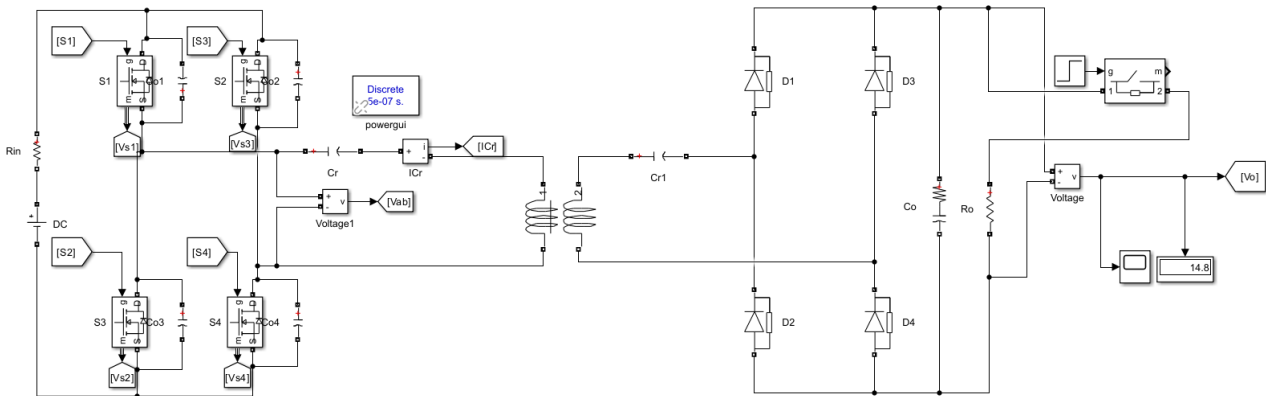


Fig. 4 Model of wireless energy transmission circuits

3. Parametric design of Magnetic Coupling Coils

3.1. Principle Analysis of Magnetically Coupled Coils

As shown in Figure. 5, for a magnetically coupled resonant coil with S-S topology, it can be equated to an RLC circuit in order to find a function about the system efficiency [6].

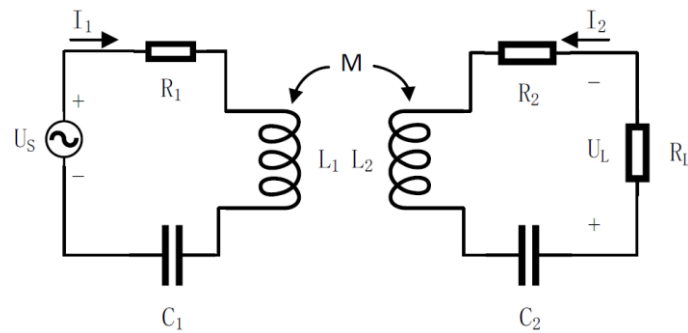


Fig. 5 Equivalent circuit of S-S type compensation structure

The KVL equations are given separately for the primary and secondary sides:

$$\begin{cases} \dot{U}_s = \left(R_1 + j\omega L_1 + \frac{1}{j\omega C_1} \right) \dot{I}_1 + j\omega M \dot{I}_2 \\ 0 = \left(R_2 + R_L + j\omega L_2 + \frac{1}{j\omega C_2} \right) \dot{I}_2 + j\omega M \dot{I}_1 \end{cases} \quad (11)$$

When the system is resonant:

$$\omega = \frac{1}{\sqrt{LC}} \quad (12)$$

So equation (11) reduces to:

$$\begin{cases} \dot{U}_s = R_1 \dot{I}_1 + j\omega M \dot{I}_2 \\ 0 = (R_2 + R_L) \dot{I}_2 + j\omega M \dot{I}_1 \end{cases} \quad (13)$$

Therefore, the transmission efficiency η is as follows:

$$|P_L| = \frac{|U_L|^2}{R_L} = I_2^2 R_L = \frac{(\omega M U_s)^2}{R_1^2 R_L \left(1 + \frac{R_2}{R_L} + \frac{\omega^2 M^2}{R_1 R_L}\right)^2} \quad (14)$$

$$\eta = \frac{\omega^2 M^2 R_L}{[R_1 (R_2 + R_L) + \omega^2 M^2] (R_2 + R_L)} \times 100\% \quad (15)$$

From equation (11), it can be observed that the efficiency of the transmission process is primarily influenced by the operating frequency, the mutual inductance between the coils, and the AC internal resistance of the coils, when the load impedance remains constant.

3.2. Mutual Inductance between Coils and Coil AC Internal Resistance Analysis

3.2.1. Mutual Inductance Analysis between Coils

Figure. 6 shows the most commonly used planar coils in wireless energy transmission systems, with the outer diameter R_{out} , the number of coil turns N , and the turn spacing S being the three important parameters in determining the performance of the coil. The mutual inductance between a multi-turn coil can be obtained by summing the mutual inductance between each turn of the coil [7]:

$$M = \sum_{i=1}^{N_1} \sum_{j=1}^{N_2} \frac{\mu_0}{4\pi} \frac{dl_1 dl_2}{l} \quad (16)$$

where μ_0 is the winding permeability, l is the spacing between coils, dl_1, dl_2 are the line elements, and L_1, L_2 are the single-turn line lengths.

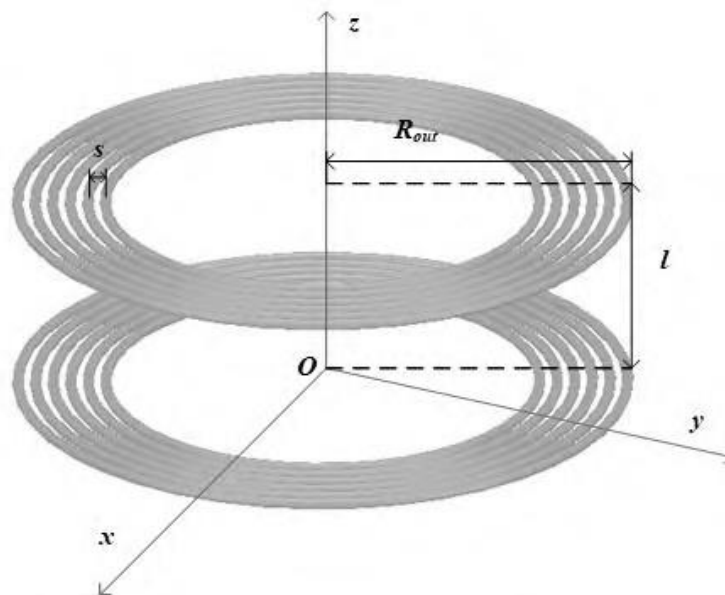


Fig. 6 Coaxial coil model

As the optimal solution of the above equations is more difficult to calculate, simulation experiments can be performed directly using Maxwell. With the receiving coil structure unchanged, the effect on the dynamic transmission performance is investigated when the transmitting coil structure is varied at the number of turns of 10, 15, and 25 turns, respectively [8], as shown in Figures. 7, Figures. 8, and Figures. 9.

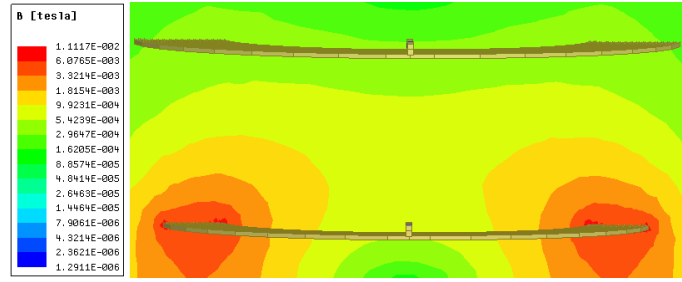


Fig. 7 $N=10$

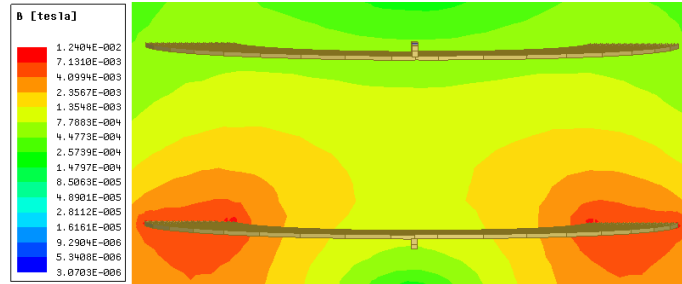


Fig. 8 $N=15$

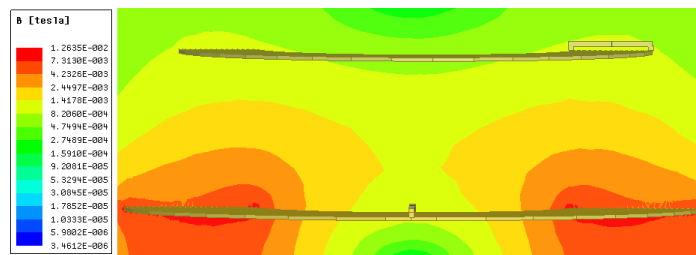


Fig. 9 $N=25$

With the increase in the winding turns, the flux density increases in both the axial and radial directions. It has been demonstrated that when the number of turns in the transmitting coil is less than that of the receiving coil, both the axial and radial flux densities reach a minimum. Conversely, if the number of turns in the transmitting coil exceeds that of the receiving coil, the asymmetric coil structure generates a greater flux density than the symmetric structure. This indicates that augmenting the number of turns in the transmitter coil is an efficacious method for augmenting the total magnetic flux traversing the receiver coil, thereby enhancing the mutual inductance between the two coils.

3.2.2. Analysis of AC Internal Resistance between Coils

The AC resistance of a coil is expressed as its intrinsic resistance plus the effects of the skin and neighbourhood effects of the current flowing through it:

$$R_{ac} = R_{prox} + R_{cond} \tag{17}$$

Where R_{cond} is the intrinsic resistance plus the skin effect of the current flow through the AC resistance, R_{prox} for the neighbourhood effect of the AC resistance. Skin effect and the neighbourhood effect in high-frequency AC coils often appear at the same time. Skin effect refers to the high-frequency current through the conductor, due to the role of electromagnetic induction, the current is mainly concentrated on the surface of the conductor flow, rather than uniformly distributed in the conductor's entire cross-section. The proximity effect, on the other hand, refers to the fact that when an alternating current passes through two neighbouring conductors, the alternating magnetic field generated between them interacts with each other, resulting in a more concentrated flow of current on the inner side of the two conductor surfaces opposite to each other. These two effects together affect the distribution of the current and the characteristics of the electromagnetic field [9]. The turn spacing S in a coil is the main factor that determines these two properties. Therefore we can explore the turn spacing S by simulation [8] as shown in Figures 10, Figures. 11 and Figures. 12.

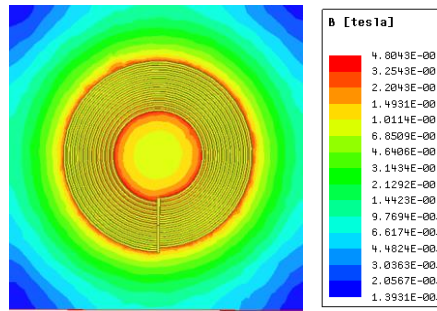


Fig. 10 $S=1\text{mm}$

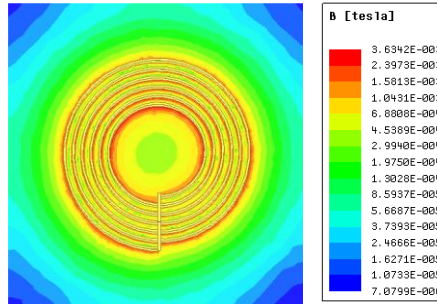


Fig. 11 $S=4\text{mm}$

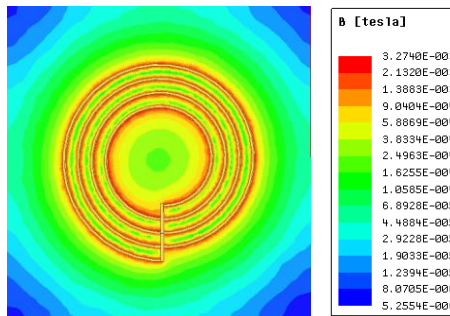


Fig. 12 $S=7\text{mm}$

It can be seen that under the condition of the established inner and outer diameters, the reduction of the distance between turns will directly lead to an increase in the number of turns, which in turn will improve the power and efficiency of current transmission. Therefore, for small coils with compact dimensions, the efficiency of energy transmission and power output can be improved more effectively by tightly arranged winding.

3.3. Coil Parameter Design

According to the previous section, a larger outer diameter and number of turns can effectively increase the mutual inductance effect. At the same time a tighter coil arrangement helps to increase the magnetic field strength at the centre of the coil. Therefore, the coil parameters are finally determined as shown in Table 2.

Table 2. Coil parameters

Coil parameters	Numerical value
Receiver coil inner diameter R_{S-in}	30 mm
Receiver coil outer diameter R_{S-out}	45 mm
Transmitter coil inner diameter R_{P-in}	30 mm
Transmitter coil outer diameter R_{P-out}	65 mm
Receiving coil turns N_s	15
Transmitter coil turns N_p	25
Turn spacing S	1 mm

4. Research on Closed-loop System Control Strategy

In practical applications, the distance between the coils and the load will change due to the consideration of the changes in the environment within the human body and the interference caused by external electromagnetic signals, so we need to introduce a closed-loop control system to ensure the power stability of electrical energy transmission. There are two main control strategies used in wireless energy transmission systems: primary edge control and secondary edge control. A distinctive feature of the vice-edge control strategy is that it does not rely on the wireless communication function, but directly regulates the charging power by adding a DC-DC converter at the receiving end (vice-edge). However, the disadvantage of this approach is that it may increase the cost, size, and weight of the receiver side, making the overall design complex and difficult to control. In contrast, the primary-side control strategy relies on wireless communication to transmit real-time information about the load back to the transmitter (primary side) and adjust the control accordingly. This approach requires additional communication devices and protocols, but it is more flexible in responding to load variations and achieves more efficient power delivery. For implantable medical devices, the weight and size need to be strictly controlled during design, so this paper chooses the primary phase shift control strategy based on PI algorithm to precisely regulate the voltage to achieve power control. Therefore, a dual closed-loop control of power and temperature is designed on the basis of power control to ensure the safety of the device [10].

4.1. Power and Temperature Double Closed-loop Control Principle

As shown in Figure. 13. The power closed loop is achieved by adjusting the phase of the H-bridge inverter voltage. Having less inertia and faster adjustment capability, it is more suitable as an inner loop control for fast and precise adjustment of the system state. Since temperature, as a physical quantity, changes as a cumulative effect of thermal power consumption in time, the rate of temperature change is relatively slow and the system response to it exhibits large inertia and significant hysteresis. Therefore, it is more appropriate to use the temperature control as an outer-loop control, so that the temperature can be regulated more smoothly and over a long period of time while ensuring the stability of the system. In the power control, $F_1(s)$ is the controller, which is used to change the input voltage of the coupled resonant unit $G_1(s)$ is the transfer function of the transmit resonant circuit. The received power P of the body coil is then the output of the coupling transfer function $G_2(s)$, where the coupling coefficient K is the disturbance term. The purpose of the closed-loop control is to ensure the power stability of the power transmission in the presence of disturbances.

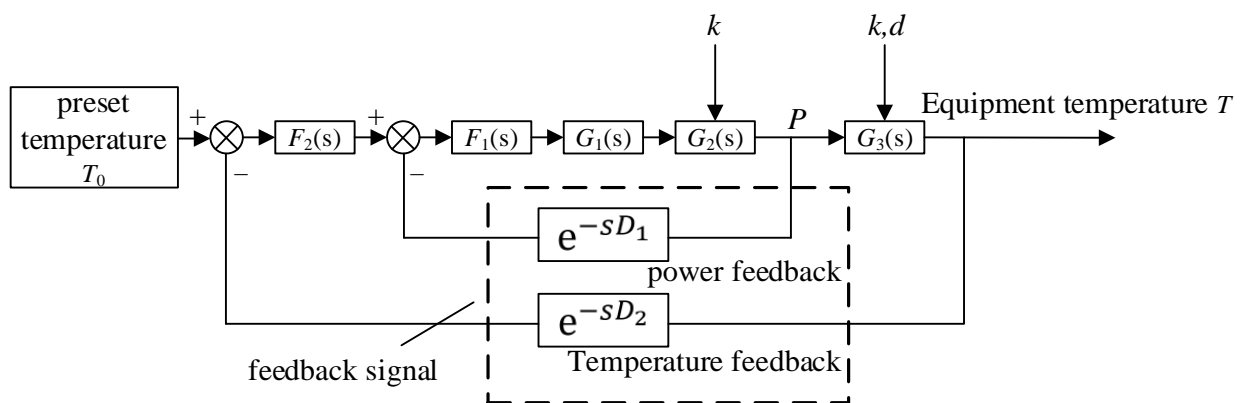


Fig. 13 Schematic diagram of double closed-loop control of power and temperature

In the temperature control system, $F_2(s)$ acts as an outer-loop controller that dynamically adjusts the power received by the in vivo coils based on the difference between the preset target temperature and the real-time device temperature T obtained via wireless communication. This adjustment process is intended to ensure that the system can be maintained within the desired temperature range. The heat producing part of the system is denoted by $G_3(s)$, and the power P received by the body coil is used as an input. This power first satisfies the battery charging requirements and the remaining power

is converted into heat, resulting in an increase in the temperature of the device. There are interference factors in the operating environment of the system. When the coupling coefficient k deteriorates, the coupling efficiency between the receiving coil and the transmitting coil decreases, which leads to a decrease in the energy conversion rate, and at the same time, the eddy current effect is enhanced, which in turn generates additional heat. On the other hand, a decrease in the heat dissipation coefficient d means that the diffusion of heat through the system or surrounding tissues slows down, further exacerbating the temperature rise.

Closed-loop temperature control is particularly important for implantable medical devices. By monitoring the temperature in real time and feeding it back to $F_2(s)$, the system is able to automatically adjust the charging power of the in vivo battery to cope with different coupling alignment conditions and heat dissipation environments. This adaptive adjustment mechanism ensures that the system can maintain safe and stable operation under various working conditions, effectively avoiding the risk of overheating.

4.2. Power Control Design

The phase-shift control technique is mainly used to regulate the input voltage by changing the conduction angle of the switching tubes in the full-bridge converter, i.e., adjusting their conduction time. This regulation is similar to adjusting the duty cycle of a DC/DC converter, both of which indirectly control the output of the system by affecting the input voltage of the coupled resonant unit. It allows the system to finely control the energy transfer by adjusting the on-time of the switching tubes without changing the switching frequency. This feature is particularly important because it avoids the problem of frequency detuning, which is common in MRC-WPT systems, and which can significantly reduce transmission efficiency and increase system losses.

The advantages of the phase-shift control technique are its flexibility and stability. Since the switching frequency remains constant, the system can more easily maintain a stable operating condition while reducing the frequency variations that affect the dynamic response speed or harmonic content. In addition, phase-shift control enables fast regulation of the system output, as changes in on-time can be made to respond quickly to changes in load or voltage.

4.2.1. PI Controller Parameterisation

The closed-loop control algorithm can ensure a higher accuracy of charging voltage, for this reason, this paper uses the phase shift control based on PI algorithm to regulate the charging voltage. In this paper, the Ziegler-Nichols parameter rectification method is used for the study, and the formula of discrete PID is listed first:

$$u(k) = K_p e(k) + \frac{K_p T}{K_i} \sum_{n=0}^k e(n) + \frac{K_p K_d}{T} (e(k) - e(k-1)) \quad (18)$$

$u(k)$ is our final output, $e(k)$ is the error at the current moment of our control system, $e(k-1)$ is the error at the previous moment, K_p is the proportionality coefficient, T is the control period, K_i is the constant of integration, and K_d is the constant of differentiation.

Based on this algorithm a MATLAB function Ziegler [11] can be written, from which the PI controller of the system can be designed directly. Eventually, K_p and K_i can be determined:

$$G_c = K_p + \frac{K_i}{s} \quad (19)$$

Where $K_p = 2.5$ and $K_i = 65$.

4.3. Power Control Simulink Simulation Analysis

The MCR-WPT system and control module shown in Figure. 14 are tested in Simulink.

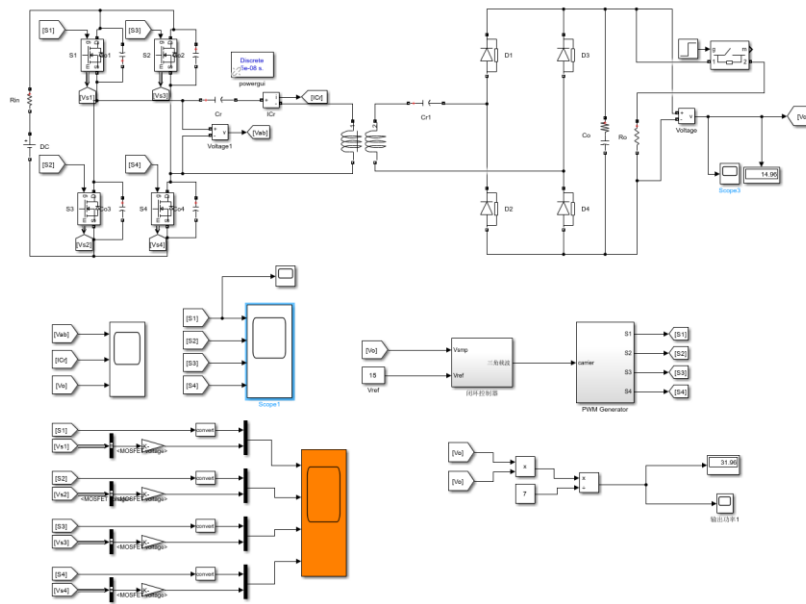


Fig. 14 Simulation of MCR-WPT system and control module

In order to verify the stability of the control system ground at $t=0.5s$, the load is added suddenly, the load is added from 0.1Ω to 0.2Ω , and the waveforms shown in Figure. 15 and Figure. 16 are obtained.



Fig. 15 Voltage change curve



Fig. 16 Power change curve

In $0-0.5s$, the load is 0.1Ω , in $0.5s$ to give a sudden increase in the load, it can be observed that the voltage and power in the interference fluctuations, but in $0.5s$ to return to the preset value of $0.12V$. It can be observed that the closed-loop PI controller has the potential to enhance the system's capacity to withstand interference and to reinforce its overall resilience.

5. Conclusion

The aim of this paper is to explore the innovative application of wireless charging technology in implantable medical devices to address the activity limitations and health risks associated with traditional power supply methods. The paper firstly describes the importance of implantable medical devices and the challenges they face in terms of energy supply, then analyses the advantages and necessity of wireless energy transmission technology and designs a robust, low-power, safe and high-efficiency wireless energy transmission system. Starting from the theoretical basis, this paper analyses in detail the circuit working principle of wireless energy transmission, including the series resonant circuit, circuit compensation network analysis, and the design of inverter and rectifier circuits. Through the theoretical analysis, it offers a robust theoretical basis for the development of the wireless energy transmission system. In terms of circuit design, the parameter design of the magnetically coupled coil is highlighted, including the analysis of mutual inductance and AC internal resistance between the coils, as well as the optimisation of the number of coil turns and turn spacing through simulation experiments to improve the transmission performance. These design optimisation measures aim to enhance the transmission efficiency of the system. In order to cope with the load variations and external disturbances that may occur in practical applications, the article also proposes a closed-loop system control strategy, including the principle of dual closed-loop control of power and temperature and its realisation method. By introducing the PI controller and phase shift control technology, the system is able to maintain stable power transmission under conditions such as sudden load changes, while controlling the temperature of the device within the safe range.

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