Design and Analysis of a Wireless Power Transmission System with Magnetic Coupling Resonance in the Weak-coupling Region

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Abstract: The magnetically coupled resonant wireless power transfer (MCR-WPT) system for implantable medical devices has gradually become one of the research hotspots. The transmission characteristics of the system in weak-coupling regions are of research significance. The mathematical model of the system's transmission characteristics is established by equivalent circuit theory. Several key factors that affect the transmission characteristics of the system are obtained, to include the self-inductance of the coil, the mutual inductance between the coils, and the load resistance. A magnetically coupled resonant wireless power transfer system is designed. A method of coil design is proposed. The transmission efficiency of the system in the weak-coupling region is improved by optimizing the design of the coil. Simulation and experimental results exhibited good consistency. The results show that the transmission performance of the system in the weak-coupling area is improved. This work provides guidance and practical significance for the research of the MCR-WPT system with implantable medical equipment.

Keywords: Wireless power transfer, frequency splitting, weak-coupling region, coil design

1 Introduction

Inspired by the work of Nikola Tesla, wireless power transfer (WPT) technology has been widely used in electric vehicles, portable electronic devices, industrial manufacturing, and implantable medical devices. As first proposed by MIT [1] in 2007, magnetically coupled resonant (MCR) WPT technology is much more convenient than the traditional inductive coupling technology. MCR-WPT has a high transmission efficiency over a long effective transmission distance, and has triggered a great deal of research interest [2-3].

Implantable medical devices are often implanted in deep tissues [4]. Zhang et al. [5] focused on in-depth analysis and simulation verification of brain-implanted wireless energy output systems. Tang et al. [6] developed a device that can be embedded in the interior of the garment for the artificial heart or left ventricle-assisted use of the firing structure of the segmented coil. Wei et al. [7] studied the wireless energy transfer system for application in the deep tissue of the human body, solving the problem of power transmission and the influence of coil size on stimulation depth and focus. In the implantable medical wireless power transmission system, Xing Li has designed the resonant coil to achieve 60 MW transmission at 3 mm, and its transmission efficiency has reached 92.6%.

For the design of MCR-WPT systems, power transfer efficiency is one of the most important factors to consider when evaluating the transfer performance of the system. To enhance the power transfer efficiency, previous work typically focused on factors such as impedance matching [8], frequency splitting [9-10], and relay resonators [11], among others. In addition, in contrast to traditional inductive WPT, optimization approaches could be used to achieve the maximum power transfer efficiency of the MCR-WPT system at a specific distance. In the case of fixed loads, mutual inductance, as a critical factor to optimize the efficiency of WPT systems, needed more in-depth analysis [12]. The main parameters that affect the mutual inductance included the number of turns, radius, material, and the relative spatial position between the two coils. In the implantable WPT system, the use of Litz wire can reduce the skin effect to
improve the coil quality factor \(^{[13]}\). In the MCR-WPT system, the transmission efficiency tends to increase first, and then decrease, when the axial distance between the sending coil and receiving coil increases.

To eliminate the frequency splitting phenomenon, Ref. \([14]\) proposed a method of adjusting the load resistance. Due to the maximum efficiency point in the MCR-WPT system, the frequency splitting is not the operating frequency point in the strong coupling region. In addition, the size of the receiving coil is limited in the containment of the implantable medical devices. It is an urgent problem to optimize the transmission characteristics of the system through the design of the coil. This study presents a coil design method that not only improves the transmission efficiency of a system which worked at the weak-coupling region, but also meets the need for the coil miniaturization. The correctness of the optimization method is verified by simulation and experiment, and an MCR-WPT system is designed. To improve the receiving performance of the system in the weak-coupling region, the voltage doubler rectifier is selected for the receiver design. Finally, the rectification circuit is tested.

This paper is organized as follows. In Section 2, the MCR-WPT system is analyzed by the equivalent circuit theory and the transmission characteristics of the system are investigated. The mathematical model of the resonance coil is established in Section 3. An MCR-WPT system is then designed in Section 4. The analytical model is validated by simulation and experiment in Section 5. Finally, conclusions are drawn in Section 6.

## 2 Analysis of the WPT system

### 2.1 Description of the WPT System

The structure diagram of the MCR-WPT system is shown in Fig. 1. The inverter circuit converts input direct current (DC) to high-frequency alternating current (AC). Under the action of the high-frequency alternating current, the transmitting coil will produce an alternating magnetic field. By adding several matching resonating capacitors at the transmitter and receiver, the resonant coils will operate at the resonant frequency point. The power is then exchanged between these coils. Finally, the rectifier circuit at the receiving end converts the received alternating current to the direct current to supply the load.

![Fig. 1 The structure diagram of the MCR-WPT system](image)

### 2.2 Modeling and analysis of the transmission characteristics

For implantable medical devices, the MCR-WPT system has a higher transmission efficiency and greater transmission distance than the inductive power transfer, making it easier to transfer power through deeper biological tissue to the receiving device. This study analyzes the four-coil MCR-WPT system, because it could reduce the influence of the transmitting circuit and the receiving circuit at the sending coil and the receiving coil \([15-16]\).

Fig. 2 shows the equivalent circuit model of the MCR-WPT system. It consists of a power coil, a sending coil, a receiving coil, and a load coil. \(L_i\) is the inductance of the \(i\)-th coil; \(C_i\) is the corresponding resonating capacitor; \(R_i\) is the parasitic resistance of each coil (where \(i = 1, 2, 3, 4\)); and \(R_L\) is the load resistance. The coefficients \(M_{12}\), \(M_{23}\), and \(M_{34}\) are the dominant coupling coefficient, while \(M_{13}\), \(M_{14}\), and \(M_{24}\) are negligible here.

![Fig. 2 Equivalent circuit of the basic MCR WPT system that has four coils](image)

By applying circuit theory to this system, according to Fig. 2, we can obtain

\[
\begin{align*}
I_1(R_s + R_1 + j\omega L_1 + \frac{1}{j\omega C_1}) + j\omega l_1M_{12} &= U_s \\
I_2(R_2 + j\omega L_2 + \frac{1}{j\omega C_2}) + j\omega (I_1M_{12} - I_2M_{23}) &= 0 \\
I_3(R_3 + j\omega L_3 + \frac{1}{j\omega C_3}) + j\omega (I_2M_{23} - I_3M_{34}) &= 0 \\
I_4(R_4 + R_4 + j\omega L_4 + \frac{1}{j\omega C_4}) + j\omega l_3M_{34} &= 0
\end{align*}
\]
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then

\[
k_i = M_i / \sqrt{L_i L_j}
\]

where \(0 < k_i < 1; i = 1, 2, 3, 4; j = 1, 2, 3, 4; i \neq j\).

In MCR-WPT systems, the scattering (S) parameters are used to analyze the forward gain \[15\].

\[S_{21}\] is defined as the ratio of output voltage to input voltage. The transmission efficiency \(\eta\) of the system can also be expressed as

\[
\eta_{21} = 2 \frac{U_L}{U_s} \left( \frac{R_s}{R_L} \right)^{1/2}
\]

Using formula (1), the voltage across the load resistor can be obtained

\[
U_L = \frac{io^k L_1 k_2 L_2 L_L L_3 L_4}{\sqrt{k_2 L_2 L_L L_3 L_4} \sqrt{k_2 L_2 L_L L_3 L_4} \sqrt{k_2 L_2 L_L L_3 L_4} \sqrt{k_2 L_2 L_L L_3 L_4} \sqrt{k_2 L_2 L_L L_3 L_4} \sqrt{k_2 L_2 L_L L_3 L_4}}
\]

where

\[
\begin{align*}
Z_1 &= R_1 + R_s + j\omega L_1 + \frac{1}{j\omega C_1} \\
Z_2 &= R_2 + j\omega L_2 + \frac{1}{j\omega C_2} \\
Z_3 &= R_3 + j\omega L_3 + \frac{1}{j\omega C_3} \\
Z_4 &= R_4 + j\omega L_4 + \frac{1}{j\omega C_4}
\end{align*}
\]

### 2.3 Transmission characteristic analysis

Owing to formula (4), it is seen that the transmission efficiency varies with the changes of coupling coefficient. From formula (2), the coupling coefficient is closely related to the mutual inductance. The numerical simulation is performed to analyze the effect of the coupling coefficient on the transfer characteristics of the MCR-WPT system. To simplify the analysis, assuming that other parameters are invariable, only the \(k_{23}\) and the resonant frequency changed. The parameter descriptions of the MCR WPT system are listed in Tab. 1.

| Parameter Description | Value |
|-----------------------|-------|
| Source internal resistance \(R_s/\Omega\) | 50 |
| Load resistance \(R_L/\Omega\) | 50 |
| Coil resistance \(R_1, R_2, R_3, R_4/\Omega\) | 0.073 95 |
| Coil self-inductance \((L_1, L_2, L_3, L_4)\mu H\) | 14.074 |
| Resonance capacitance \((C_1, C_2, C_3, C_4)/pF\) | 1.799 92 |
| Coupling coefficient \(k_{12}, k_{34}\) | 0.223 9 |
| Resonant angular frequency \(\omega\) | 2 000 000 \(\pi\) |

Fig. 3 demonstrates the change of the power transfer efficiency with changes to \(k_{23}\) and the resonant frequency. The optimal transmission efficiency can be obtained by changing the values of the coupling coefficient. In general, for a typical four-coil MCR-WPT system, the transmission distance is often referred to as the relative distance between the sending coil and the receiving coil. From Fig. 3, one can see that, with the change of distance, the system worked in the over-coupling region, at the critical coupling point, and in the weak-coupling region, respectively.

### 3 Analytical model of coil

In Section 2, several key factors that affect the transmission characteristics of the system are obtained. They include self-inductance, mutual inductance, frequency, and load resistance. Most of these parameters are related to the coil. In this section, the coil model is established.

#### 3.1 Self-inductance

If the magnetic field is generated by a current loop, the magnetic flux of a defined area is proportional to the current in the loop in a linear isotropic medium (formula 6).

\[
\Psi_L = LI
\]

where \(L\) is the self-inductance; \(\Psi_L\) is the self-inductance flux; and \(I\) is the current. Regardless of the current passed through the conductor and the flux linkage, the inductance is only related to the size,
shape, and material of the coil. The self-inductance calculation for a multi-turn coil is divided into internal self-inductance and external self-inductance. The calculation of the coil's self-inductance is the sum of the internal self-inductance and the external self-inductance (formula 7).

\[ L = L_{\text{internal}} + L_{\text{external}} \]  

For the self-inductance of a single-turn coil, assume that the radius of the circular coil is \( r \) and the wire radius of the conductor is \( a \). Then, the single-turn coil self-inductance formula is as follows

\[ L = \mu_0 r \left[ \ln \left( \frac{8r}{a} \right) - 2 \right] \]  

where \( \mu_0 = 4\pi \times 10^{-7} \) (N·A²) is the vacuum permeability. Thus, for a multi-turn coil, the coil self-inductance formula is as follows

\[ L = \sum_{i=1}^{n} L_{\text{internal}} + \sum_{i=1}^{n-1} \sum_{j=i+1}^{n} M_{ij} (i \neq j) \]  

where \( n \) is the number of turns of the coil and \( M_{ij} \) is the mutual inductance between each turn of the coil.

3.2 Mutual inductance

Fig. 4 shows two single-turn coils. Assuming that the position of coil1 is constant, only the relative spatial position of coil2 relative to the coil1 is changed. Take the geometric center \( O \) of coil1 as the origin of the global coordinate system and the plane of coil1 as the global coordinate system \( x_1Oy_1 \) plane. Take the geometric center \( O' \) of coil2 as the origin of the local coordinate system and the plane of coil2 as the global coordinate system \( x_2O'y_2 \) plane. The mutual inductance between the two coils can be obtained by solving Neumann’s formula as follows

\[ M_{12} = \frac{\mu_0 r_1 r_2}{4\pi} \int_0^{2\pi} \int_{-\infty}^{\infty} \sin \theta \sin \phi \cos \alpha + \cos \theta \cos \phi \ \phi \ d\phi \ d\theta \]  

where \( r_1 \) and \( r_2 \) are the radii of coil1 and coil2, respectively.

3.3 Coil resistance

When the coil operates at a high-frequency state, the loss resistance primarily includes two parts: the ohmic loss resistance \( R_0 \) and the radiation loss resistance \( R_r \). In the working frequency band of an MCR WPT system, the radiation loss is generally much less than the ohmic loss. Therefore, this study only considers the ohmic loss of the coil. It can be expressed as

\[ R_0 = \frac{\mu_0 \omega l}{4\pi \sigma a} \]  

where \( \omega \) is the operating angle frequency; \( \mu_0 \) is the vacuum permeability; \( \sigma \) is the conductivity; \( l \) is the total length of the coil; and \( a \) is the wire radius.

4 System design

The transdermal wireless power transfer system
can be divided into an external transmission circuit and an internal receiving circuit, as shown in Fig. 1. The external transmitting part includes a DC power supply, an inverter circuit, and a transmitting coil. The receiving part includes a receiving coil, a rectifier circuit, and a load.

4.1 Design of the transmitting circuit

Because the theoretical energy conversion rate of the Class-E power amplifier is 100%, and it has a simple structure, the Class-E power amplifier is selected for the transmitter inverter circuit. The circuit structure of the Class-E power amplifier is shown in Fig. 5.

![Fig. 5 Circuit structure of the Class-E power amplifier](image)

As shown in Fig. 5, the topology of the Class-E power amplifier includes the DC input power $V_{DD}$, a choke inductor $L_{RFC}$, a shunt capacitor $C_d$, a switch $S$, the $L_0-C_0$ series resonant filter circuit, and load $R_L$. The key parameters are as follows: $L_0 = 46.8 \, \mu\text{H}$, $C_d$ is 1008 pF, $C_0$ is 784 pF, and $R_L$ is 50 $\Omega$. The choke inductor acts as a current stabilizer; it has a high AC impedance and only allows the DC component in $V_{DD}$ to pass through.

The switch $S$ is periodically turned on and off at a frequency in the $RF$ input range. The series resonant filter circuit operates at the resonant frequency, such that the fundamental frequency signal is transmitted to the load. The role of the resonant filter circuit is to keep the output signal sinusoidal.

$$\omega = \frac{1}{\sqrt{L_1C_1}} = \frac{1}{\sqrt{L_2C_2}} = \frac{1}{\sqrt{L_3C_3}} = \frac{1}{\sqrt{L_4C_4}} = \frac{1}{\sqrt{L_0C_0}}$$ (14)

$$f = \frac{\omega}{2\pi}$$ (15)

$$f = \frac{1}{2\pi \sqrt{\frac{L_0C_0}{2 \times 3.14 \times 14.07 \times 10^{-3} \times 1.8 \times 10^{-12}}} = 10^6 \text{ Hz} = 1 \text{ MHz}}$$ (16)

By selecting a high-$Q$ coil in the transmitting circuit to make the transmitting circuit have a higher $Q$ value, and by adjusting the values of the matching network components to make the filter circuit and the voltage coil circuit both resonate at a high operating frequency, the coil self-inductance value $L_i=14.07 \, \mu\text{H}$ ($i=1, 2, 3, 4$), and the resonance capacitance $C_i=1.8 \, \text{pF}$ ($i=1, 2, 3, 4$). According to the above formulas (14-16), the resonant frequency of the four-coil system is calculated to be 1 MHz. The output current $i_0$ through the resonant filter and the transmitting circuit approximates a sinusoidal output current to drive the voltage coil for energy transfer.

4.2 Design of the resonant coil

Due to their particular application in the field of implantable medical devices, resonant coils of a wireless power transfer system require special consideration. For the receiving coils, the coil size is limited due to the size of the implanted device. In the wireless energy transmission system, there are three primary types of winding methods, as mentioned in Section 3. They are shown in Fig. 6.

![Fig. 6 Three types of coils](image)

(a) Helical coil  (b) Planar spiral coil  (c) Multilayer planar spiral coil

First, assume that these three types of coils have a number of turns of 20, an inner diameter of 21.22 mm, and a wire diameter of 1.10 mm. The volumes of these three types of coils are 3 538.02 mm$^3$, 8 856.57 mm$^3$, and 3 112.17 mm$^3$, respectively (the number of layers of the multilayer planar spiral coil is two). Therefore, under the condition of certain coil parameters, the volume of the multi-layer planar spiral coil is the smallest, and the double-layer spiral coil is selected as the receiving coil in this study.

Second, assuming that the outer diameter of these three types of coils is 21.22 mm and the diameter of the wire is 1.10 mm, then the number of turns of the helical coil is two, the number of turns of the planar spiral coil is ten, and the multilayer planar spiral coil is ten turns of two layers each. These three types of coils are used as the transmitting coil, and the
receiving coil is the double-layer planar spiral coil.

Comparing these three cases, the change of the system transmission efficiency is shown in Fig. 7 which is obtained by finite element simulation characteristics when the transmitting coil is a single-layer planar spiral coil. Therefore, a single-layer planar spiral coil is selected as the transmitting coil in this study.

![Fig. 7 Changes in transmission efficiency of different types of transmission coil](image)

We now analyze and compare three different types of coils under the same number of turns. Assume that the number of turns of the three coils is 20, and the inner diameter of the coil is 10.22 mm, the diameter of the coil wire is 1.10 mm, and the wire material is copper. The three coils are compared from the three aspects of coil volume, radial area, and thickness, as listed in Tab. 2. From Tab. 2, it can be seen that the multi-layer planar spiral coil occupies the smallest volume under the same number of turns. In the case of the same number of turns, because the volume of the solenoid coil is much larger than that of the planar spiral coil and the double-layer planar spiral coil, it is not the optimum choice for the type of the coil at the receiving end. The type of coil for the receiving coil in the body is selected as the multilayer planar spiral coil.

| Tab. 2 Size parameters of three types of coils |
|-----------------------------------------------|
|                                | Flat spiral coil | Solenoid coil | Double layer spiral coil | Radial area φ/mm² | Flat spiral coil | Solenoid coil | Double layer spiral coil | Coil thickness/mm |
|-----------------------------------------------|
|                                | 3 538.02        | 8 856.57      | 3 112.17            | 3 261.38          | 402.57          | 1 414.62      | 1 10.0        |

4.3 Design of receiving rectifier circuit

As the implant depth increases, the distance between the transmit coil and the receive coil also increases, making it easier for the system to work in weakly coupled areas. The power received by the device becomes smaller and the received voltage also becomes smaller. This study used a voltage doubler rectifier circuit as the receiver rectifier circuit. Although the voltage doubler rectifier circuit reduces the ability of the circuit to generate current, the circuit's voltage output capability is improved and, therefore, does not affect the functionality of the implantable medical device.

The working principle of the voltage doubler rectifier circuit is shown in Fig. 8. It is assumed that, during the positive half cycle of the AC voltage, $D_1$ conducts, $D_2$ is cut off, and $C_1$ is charged through $D_1$. In the negative half cycle, $D_2$ conducts, $D_1$ is cut off, and the voltage charges $C_2$ via $D_2$. The parameters $C_1$ and $C_2$ have the same resistance value of 50 μF, and are charged in the positive and negative periods of the received voltage, respectively. $C_1$ and $C_2$ are then discharged into the load after the voltage is superimposed, made the output voltage of the system more stable. In theory, the voltage doubler rectifier output voltage can reach twice the input voltage, but by the loss of the diode and other devices in the circuit,
the output voltage often cannot be doubled. Under different loads, the voltage doubler output voltage amplitude will have some variability.

5 Simulation and experiment

5.1 Finite element simulation and experimental measurement

The relationship between the mutual inductance and the coupling coefficient can be seen from formula (2). The self-inductance of the coil is basically determined when the coil parameter is determined, and the coupling coefficient varies with the mutual inductance. Therefore, the coupling coefficient between the coils can be indirectly obtained by obtaining the mutual inductance between the coils. Suppose there are two multi-layer planar spiral coils with the same parameters. Their parameters are: two layers of ten turns per layer, the inner radius of the coil is 10.22 mm, the material of the wire is copper, and the wire diameter is 1.10 mm. Thus, the mutual inductance between them can be obtained by formula (12), finite element simulation and experiment. Therefore, the coupling coefficient can be obtained indirectly. Fig. 9 is the finite element simulation model. Fig. 10 is the experimental measurement platform. Fig. 11 shows the variation of the coupling coefficient between two multi-layer planar spiral coils with the axis offset.

From Fig. 11 we can see that the maximum relative error between theoretical and simulated values is 1.5%. The maximum relative error between theoretical and experimental values is 2.6%. The maximum relative error between the simulated and experimental values is 1.7%. In the simulation using ANSYS software, different mesh subdivisions bring discrete errors, which is the main source of the relative error. In addition, this study reduces the random error existing in the experiment by taking multiple measurements. As can be seen from Fig. 11, the relative errors are all below 5%, within the allowable error range of the project. The results show that the coupling coefficient theory is correct. The coupling coefficient decreases with the increase of the axial offset. At the same time, it can be seen from Fig. 12 that, as the lateral displacement \(d\) increases, the mutual inductance also gradually decreases.
5.2 Effect of coil axial offset on efficiency

In a magnetically coupled resonant system, due to the resonant coupling between the coils, the energy transmission efficiency decreases as the axial offset increases. However, within a certain range of offset, the reduction of efficiency is not very large. Fig. 13 is a diagram showing changes in transmission efficiency caused by changing the axial shift degree of a single coil in the case where the positions of the axes of the other coils are not changed. Fig. 14 shows the effect of the axial offset on the efficiency when the transmitting coil and the receiving coil are simultaneously offset from the power coil and the load coil. In the experiments of Fig. 13 and Fig. 14, the relative axial center distances of the coils $L_1$ and $L_2$, to the coils $L_3$ and $L_4$ are 10 mm, and the axial distance between the coils $L_2$ and $L_3$ is 50 mm.

The wireless power transfer experiment platform is shown in Fig. 15. In the experiment, the working frequency of the system is 1 MHz. The parameters of the power coil, the load coil, and the receiving coil are kept consistent, and they are all double-layer planar spiral coils. The load resistance is 50 Ω after selecting the appropriate matching tunable capacitor to make the coil operate at the resonant frequency. In the experiment, the axial distance between the transmitting coil and the receiving coil changes in 5 mm steps, and the peak-to-peak voltage across the load is measured. The experimental results are shown in Fig. 16.

As can be seen from Fig. 16, the peak-to-peak load voltage first increases, then decreases as the axial distance increases, and reaches a maximum at 30 mm. As the distance continues to increase, the voltage across the load begins to decrease, indicating that the system has experienced frequency splitting. When the sending coil is a single-layer planar spiral coil, the higher output voltage can be received in the weak-coupling region load. We can know that, when the inner diameter and the number of turns of the single-layer planar spiral coil are kept consistent with the double-layer planar spiral sending coil, choosing a single-layer planar spiral coil enhances the wireless energy transmission of the system, and improves the transmission performance in the weak-coupling region.

5.3 Distance experiment

Under the condition that the receiving coil is fixed as a double-layer planar spiral coil, the transmitting coil used for the wireless energy transmission experiment is a single-layer planar spiral coil and a double-layer planar spiral coil, respectively.
5.4 Receiving circuit experiment

To verify the operation of the voltage doubler rectifier circuit in the wireless power transfer system, the receiver circuit is measured experimentally. The voltage waveforms on the load coil and the output voltage waveform of the voltage doubler rectifier circuit are measured using an oscilloscope. The data obtained from the oscilloscope are exported and processed by MATLAB software. The test waveform of the receiving circuit is shown in Fig. 17.

Fig. 17 The test waveform of the receiving circuit

In Fig. 17, the orange line is the output voltage waveform of voltage doubler rectifier. The green curve is the voltage waveform of the load coil. It can be seen from Fig. 17 that the output voltage of the voltage doubler rectifier circuit is approximately twice the effective value of the AC voltage on the load coil. The output voltage is stable, which enhances the transmission performance of the system in the weak-coupling region.

6 Conclusions

In this study, the theoretical model of the magnetically coupled resonant wireless power transfer system was established. The transmission characteristics of the system were analyzed, and the key factors affecting the system were obtained. The cause of frequency splitting was explained. A key research result was the determination of the transmission characteristics of the magnetically coupled resonant wireless power transfer system in the weak-coupling region. Aimed at the special application field of implantable medical devices, a coil design method was proposed to improve the transmission characteristics of the system in the weak-coupling region. The percutaneous wireless power transfer system was designed. To solve the problems of low energy transmission and low output voltage amplitude under weak energy reception, the voltage doubler rectifier circuit was selected as the receiver rectification circuit. The results demonstrated that the system designed in this study can improve the transmission characteristics of the system in the weak-coupling region.

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