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Miniaturized Resonant Power Conversion for Implanted Medical Devices

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ABSTRACT The design of a millimetre-sized power converter is proposed which is based on coil coupling that could be integrated into the neural or cardiac implantable medical device (IMD) to provide isolated power or energy transmission by harvesting it from external transmitter device. A special step-up case of transformer coupling to a millimetre-sized receiver coil by a comparatively larger transmitter coil is examined. This paper is expected to increase research efforts to develop the battery-less IMD's with reduced size, low power, high efficiency, and improved reliability and feasibility. Based on our work, we believe that the inductive coupling link with low loss ferrite material is the suitable method to be used to power the batteryless devices. The converter produces the targeted output power of more than 400 μW with 1 mm^3 size of the coil at 2 MHz.

INDEX TERMS Energy harvesting, implantable medical devices, ferrite core, inductive coupling link, power efficiency.

I. INTRODUCTION

Implanted medical devices (IMD) nowadays have peak power consumption in the range of mW's. This is related to the physical size of electronic biomedical implants, which is an important objective in terms of realizable applications. The goal is that IMD devices, such as pacemakers, cochlear, neuro, cardio and brain implantable, to be made few microns in size to be injected in their respective locations within the body. However, one of the critical factors to such aggressive size reduction is the power supply, for which IMD devices need to consume ultra low power [1], [2], [12]. Due to this challenge, energy harvesting methods [1]–[4], [19] are attracting lot of research efforts especially in biomedical implants. The inductive coupling link on the other hand is possibly growing technology for biomedical applications in short power and data transfer. This technology uses magnetic coupling as the communication environment, which is common with radio frequency identification techniques [5], [6], [15], [16]. Most studies related to inductive links have discussed about frequencies in the range of 2 MHz to 13MHz [7], [8] in order to be within allowed limits to avoid tissue heating caused by power absorption within tissue. Generally continuous exposure levels are limited to hundreds of mW/cm2, although pulsed power levels can be much higher.

In fact using ultrasonic waves to intentionally destroy tissue (High Intensity Focused Ultrasound (HIFU)) has applications in tumor treatment [3], [4]. Example of existing implantable device [12] that uses power coupling to power itself, can be referred to [5], [14]–[16].

In fact in [17] a full biomedical interface chip consumes only 450nW. In the near future most of the implants will be scaled down to operating power of few μW's. It is reasonable to believe that some devices having predetermined functions may not require any data transfer and will only need to be powered wirelessly for performance, which will reduce the power consumption further.

Based on the link model analyzed, the focus is on the required power obtainable for a coil size in neural and cardio implant rather than the link efficiency.

The IMD system topology is composed of two coils, receiving coil located inside the human body (implant), and the transmitting coil located outside the body (transmitter). The topological connections in this system have four resonance possibilities: serial-to-parallel topology, parallel-to-serial topology, serial-to-serial topology and parallel-to-parallel topology, respectively. To ensure better power transfer efficiency of the inductive coupling link, both sides of the link are tuned at the same resonant frequency f_o.

The design and analysis of an inductive link requires simplified circuit representation of the inductive link to compute link response at the operating frequency. In this paper, the primary circuit (transmitter) has series resonant circuit for driving the transmitter coil, where the secondary circuit (receiver) is parallel LC circuit in for driving a non-linear rectifier load shown in Figure 1, which is discussed in Section II. Next step involves coil design, which plays a critical role in optimum power transmission.

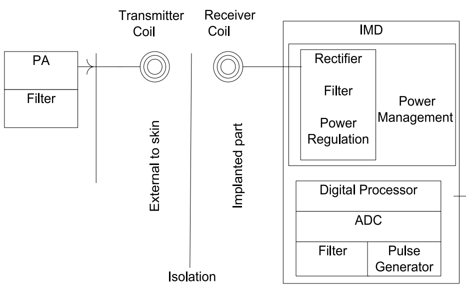

FIGURE 1. Electronics architecture: Resonant inductive-coils link provides both power and communication (both ways).

This includes optimization of coil geometry, coil dimensions, inductor losses, type of wire used and number of turns in windings to mention a few. The use of ferrite rods in inductive implants can improve power transfer efficiency, which is desirable due to low coupling in practical implants [6]. In this work an analytical model is verified experimentally showing improvement in that mutual inductance by inserting ferrite rods within the implanted device, however, the size should still remain within the specified limits. All these factors are comprehensively discussed in Section III and Section IV.

A. ELECTRONIC ARCHITECTURE FOR IMPLANTABLE DEVICE

The electronic circuitry in implant consists of a self-resonant receiving coil, a custom IMD chip, a Schottky diode, and electrodes connected at the output of the receiver circuit. On the transmitter side of this system a power amplifier is used, which enables the transmitter coil to generate magnetic field for the implanted receiver coil to generate current. The IMD chip derives direct-current (DC) power by rectifying, regulating and filtering the energy picked up by the receiving coil, to be encoded into a serial bit stream, which is decodedby a state machine inside Digital Processor within the IMD chip. The first data byte specifies an address, which is compared with the address specified by a hardwired read-only memory (ROM) in each IC. If they match, the subsequent data bytes are decoded to specify the desired operation. Stimulation operations require a pulse width and a pulse amplitude specification.

The implants draw very little power by inductive coupling between two coils that have a low coupling coefficient because of their physical separation and mismatch in size. This requires an intense magnetic field in calculated number of turns and size. The ferrite material with low loss and low initial permeabilities, μ_f having higher efficiencies is used to maximize the capture of energy from this coupling.

The amount of transmitted power from the large transmitter coil to the small receiver coil now only depends on μ_f and on the orientation and place of the small receiver coil in the magnetic field. The efficiency of inductive power transfer can be increased by using coupled coils working in resonance as shown in Figure 2. In this way the circuit will oscillate and store the energy that is inserted. In the ideal case all the energy is stored until it is transferred to the secondary circuit. In the non-ideal case some of the power is lost because of power dissipation in the primary circuit, mainly caused by the resistance of the coil.

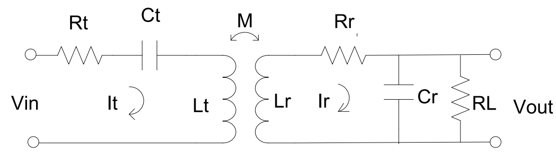

FIGURE 2. Representation of a coupled primary and secondary.

B. REGULATIONS

The maximum magnetic field, generated by the transmitter coil cannot be arbitrarily high and is limited by several restrictions:

1. Thermal limit: Tissue damage in humans can occur during exposure to high EM-levels because of the body's inability to cope with or dissipate the excessive heat that could be generated. Exposure to energy at power density levels above the 100 W/m2 can result in heating of the biological tissue [9], [10].

2. B-field: The maximum allowed magnetic field depends on the frequency [10], but we will assume that it is allowed to exceed this limit by a factor of 10, (as in the case of TMS, which has already FDA approval).

3. E-field: The maximum allowed electrical field depends on the frequency [10].

4. SAR-limit: The maximum allowed Specific Absorption Rate in the human brain is 1.6 W/kg [11].

C. OUTPUT POWER AND OUTPUT VOLTAGE ESTIMATION

A brief summary of the requirements for this topology to power the implantable medical device are described in the as:

- deliver 10-20 μW continuously, to power the system directly or, to recharge the existing battery continuously or 500 μW for 2 weeks of energy storage
- It needs to be safe, so not exceeding the regulated limits described above.
- It needs to generate appropriate voltages.
- The receiver coil needs to be as small as possible. The total volume of the power module needs to be in the millimetre range.

II. ANALYTICAL MODEL AND ANALYSIS

A. PERFORMANCE CHARACTERISTICS

An inductively coupled set of two coils which can be considered as transmitter and receiver of the circuit driven by an ac voltage source V_t which is sinusoidal in nature and having frequency f_o as shown in Figure 2. In this circuit, L_t represents the inductance of the transmitter coil and C_t is the capacitor in transmitter coil circuit whose value is chosen to be in resonance with L_t at the transmitter driving frequency, ω_o ($2\pi \mathrm{f_o}$). L_r and C_r are the inductor-capacitor pair that will be implanted and function as a receiver

$$\omega o=\frac{1}{\sqrt{Lt\ Ct}}=\frac{1}{\sqrt{Lr\ Cr}} \tag{1}$$

The resistance R_t, R_r represent the resistances in the inductor winding and other losses in the circuits for transmitter and receiver respectively. These losses comprise of radiated energy and absorption losses in both windings. The receiver voltage V_r is the open circuit voltage across the capacitor C_r. Other parameters considered are mutual inductance (M) and coupling coefficient (k) with respect to Lt and Lr, as proposed by [18]. $\mathrm{Z_t}$, $\mathrm{A_r}$ are the impedance and admittance in transmitter and receiver circuits respectively.

$$k=\frac{M}{\sqrt{L_t L}_r} \tag{2}$$

k factor equates the fraction of magnetic flux generated by the primary coil and flowing through the secondary coil and ranges from 0 to 1. This topology is usually used as a transformer with a coupling coefficient k very close to 1 for efficient power transfer. However it is impossible to achieve a coupling coefficient close to 1 when the distance between the coils is large. A parallel circuit at the receiver gets to gave a high output resistance, and therefore behaves like a current source in order to minimize the reduction of the resonator quality factor. The loop equation for this circuit is given by [6]:

$$\begin{bmatrix} V_t \\ 0 \end{bmatrix}=\begin{bmatrix} Z_t & -Mj\omega \\ -Mj\omega & Z_r \end{bmatrix}\begin{bmatrix} I_t \\ I_r \end{bmatrix} \tag{3}$$

The $X_L=\omega L$ and $X_C=1/\omega C$ be the magnitude of the inductive and capacitive reactance, which are equal at resonance respectively, cancelling each other. In this condition the impendence is having real part only. Now, using quality factor of the transmitter, Q_t and receiver circuit, Q_r, defined as the ratio of the magnitude of the inductive or capacitive reactance to the resistance in the circuit:

$$\mathrm{Q}=\frac{1}{R}\sqrt{\frac{L}{C}}=\frac{1}{R}\sqrt{X_L X_\mathrm{C}}=\frac{X_l}{R}=\frac{X_c}{R} \tag{4}$$

Now, assuming negligible losses, the efficiency as calculated in terms of coupling coefficient (k) and quality factors of the coils as given in [13] is:

$$\eta=\frac{k^2 Q_t Q_r}{1+k^2 Q_t Q_r} \tag{5}$$

The voltage generated at the receiving side inside the implantable device is modified from [14] as:

$$V_r=\frac{V_t}{k}\sqrt{\frac{L_r}{L_t}\frac{k^2 Q_t Q_r}{1+k^2 Q_t Q_r}} \tag{6}$$

This equation provides estimation of the receiver voltage V_r as a function of transmitter voltage V_t for coupled circuits with dependency on coupling ratio, having influence of the mutual inductance, the transmitter coil inductance, the quality factor of the transmitter circuit (Q_t), and the quality factor of the receiver circuit (Q_r). The complex power delivered to the at the receiver is

$$P_r=\left|\frac{wMV_t}{Z_t(w)Z_r(w)+w^2M^2}\right|^2 Real(Z_r(w)-R_r) \tag{7}$$

At resonance the reactive power of the inductor is equal to the reactive power of the capacitor. Therefore, the power delivered to the load is equal to the power dissipated in the resistor is:

$$P_r=\frac{|V_r|^2}{R} \tag{8}$$

The mutual inductance, M, can be estimated from the following analysis. Assume two coplanar coaxial coils with radii Y_t and Y_r (with $Y_t \gg Y_r$), and turns N_t and N_r, diameter $2Y_t$ and $2Y_r$ respectively. Coil diameter can be considered as twice the coil-radius, corresponding to a small diameter receiver coil. At the center of the coils, a current I_t through the transmitter coil produces a magnetic field and flux given by:

$$H_2=\frac{\mathrm{I_t N_t}}{2\mathrm{Y_t}} \tag{9}$$

$$B_2=\frac{\mu\mathrm{I_t N_t}}{2\mathrm{Y_t}} \tag{10}$$

with μ the permeability of the material between the two coils. The electromagnetic waves passing through the human body decline in intensity as they heat the tissues and are attenuated approximately as [16]:

$$I=I_o e^{(-2\alpha_\epsilon d_x)} \tag{11}$$

where I is the power per unit area transmitted through the tissue, I_o the intensity at the surface, d_x is depth of tissue penetration, and α_ϵ is the total attenuation factor including scattering and absorption:

$$v\alpha_\epsilon=\alpha_\epsilon\sqrt{f} \tag{12}$$

where f is the frequency of the electromagnetic wave, and $\alpha_\epsilon=5.10^{-3}\mathrm{s}^{1/2}\mathrm{m}^{-1}$ for soft tissue.

In this work, since the receiver coil is placed coplanar coaxial with repect to the transmitter coil, thus H is relatively uniform over the small area of the receiver coil, hence, the flux linkage is:

$$\Phi=\frac{\mu I_t N_t N_r}{2Y_t}\pi Y_r^2\cdot e^{(-2\alpha_\epsilon d_x)} \tag{13}$$

with $\mathrm{Y_r}$ the radii of the receiver coil and N_r the number of turns in the receiver coil. For the receiver coil, the voltage across the inductor $\mathrm{V_l}$ is:

$$V_l=-\frac{\partial\Phi}{\partial t}=\frac{N_t N_r \pi Y_r^2\mu}{2Y_t}\cdot e^{(-2\alpha_\epsilon d_x)}\cdot\frac{\partial I_t}{\partial t} \tag{14}$$

The expression for mutual inductance so for coaxial coplanar coils is:

$$M = \frac{\pi}{2} N_t N_r \mu \frac{Y_r^2}{Y_t} \cdot e^{(-2\alpha_\epsilon d_x)} \tag{15}$$

Coil dimensions simulated in this work have range of turns, thus for larger value of turns, the mutual inductance can be higher for the same self inductance. This higher mutual inductance results in a larger coupling coefficient k, which increases the efficiency.

In order to have highest magnetic field, the secondary coil needs to be placed at coplanar coaxial with respect to the primary coil. However, some portion of the initial energy is lost due to attenuation, and it results in a reduction of the local magnetic field. This reduction in the local magnetic field can be overcome by adding a ferrite core to the center of the receiver coil, which increases the self inductance, while the mutual inductance represented by including the factor μ_f into the equation 15. The magnitude of this factor depends on the relative length and cross section of the ferrite rod. The presence of this ferrite core increases the magnetic flux through the secondary coil by a factor μ_f as shown in eq. 17, which is between the 4 and 15 for millimetre sized ferrites [6]. For the calculations in this work the coil is modelled with value of $\mu_f = 10$.

The magnetic field, generated by the transmitter coil, at the position of the receiver coil is 14 μT (attenuation included).

The local magnetic field at the centre of the receiver coil can be increased to 140 μT, when a ferrite core ($\mu_f = 10$) is added to the receiver coil. This is above the limit [8]–[10], nevertheless, this is a very local increase of the magnetic field, and for this reason, it will be assumed that this is allowed, as long as the other limits are not exceeded. The electrical field and tissue heating are far below the limits.

Besides the strength of the magnetic field generated by the transmitter, the amount of transmitted power also depends on the receiver circuit and the design of the receiver coil. The receiver circuit needs to be in resonance with the transmitter frequency, for optimal coupling between the two coils. The receiver coil design depends on several things, like the shape/volume of the total device, the inductance, the resistance, the number of turns, the number of layers, and the diameter of the wire.

The inductance of the transmitter coil can be calculated with use of the Wheeler formula for round planar coils [16]:

$$L_t = \frac{31.3 \cdot \mu \cdot N_t^2 Y_t}{8} \tag{16}$$

and the inductance of the receiver coil can be calculated with use of the Wheeler formula for finite length solenoids [16]:

$$L_r = \frac{10\pi \mu \mu_f N_r 2 Y_r^2}{9 Y_r + 10 l_r} \tag{17}$$

with l_r the length of the receiver coil. The transmitter and receiver capacitance can then be calculated with use of equation 1 (the resonance frequency of the transmitter and receiver circuit needs to be equal to the input frequency).

The mutual inductance therefore scales linearly with the geometric mean of the radiuses of the coils if the relative distance is constant. By combining equations (2), (15), (16) and (17), it can be seen that the coupling coefficient is almost constant when both the coils and the distance between the coils are scaled by the same factor. This means that increasing the size of the coils can be used to increase the range of the power transfer.

The resistance of the coils used for calculating the resonant power at receiver can be calculated with use of:

$$R = \frac{\sigma N \pi Y}{\pi d^2} \tag{18}$$

With σ the specific resistance of the coil material, d the radius of the wire.

As discussed in Section II, we will take the transmission frequency of 2 MHz. This frequency is chosen to avoid significant attenuation at higher frequencies while keeping sufficient output voltage (V_r).

III. COIL DESIGN AND MEASUREMENTS

Several coil designs are possible with a proposed model of a receiver coil design is shown in Fig. 3, in this case it is a cylindrical multilayer solenoid, with a ferrite core or embedded electronics in the middle, and covered with PDMS (polydimethylsiloxane) coating for biocompatibility purpose.

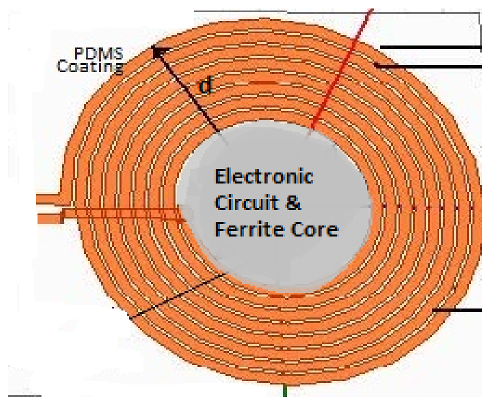

FIGURE 3. Electronics architecture: Resonant inductive-coils link provides both power and communication (both ways).

The variables as mentioned in equations representing the model above are: the number of turns, length of the coil, diameter of the coil, diameter of the wire, number of layers. For a particular requirement of the design, it is possible to calculate the amount of transmitted power and the voltage. Some possible coil designs, together with the transmitted power and voltage, are given in table 1. All these receiver coils have a total volume of around 1 mm^3, and there is no ferrite core in the middle. The volume in the middle of the receiver coil can then be used for the integrated electronics.

Nevertheless, the receiver coil designs with the best efficiency are a challenge, however through a hypodermic needle with receiver coil having a form factor suitable for injection through a hypodermic needle as shown in Fig. 4 is also investigated for power output, having total volume of $10 \times 1 \times 0.1 = 1$ mm^3, with wire having a cross section of 50×100 μm^2, and coil consisting of 5 layers, with a ferrite core in the middle ($\mu_f \approx 10$). To estimate the amount of transmitted power, we assumed it has the same energy flux as a ferrite core ($\mu_f \approx 10$) with a diameter of 2 mm.

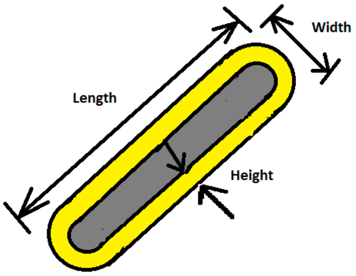

FIGURE 4. Receiver coil, which can be injected through a hypodermic needle. The coil has a total volume of 10 × 1 × 0.1 = 1 mm3. The coil has a ferrite core in the middle (μf ≈ 10).

TABLE 1. Transmitted power and output voltage for several receiver coil designs. All coils have a volume of 1 mm3 and a ferrite core in the middle (μf ≈ 10).

Wire diameter (mm)	Inner Coil diameter (mm)	Coil Length (mm)	No. of Turns	No. of Layers	Outer Coil diameter (mm)	Power (μW)	Voltage (V)
0.02	0.5	2.6	650	5	0.7	0.6	14
0.02	1	0.89	222	5	1.2	160	27
0.02	2	0.26	66	5	2.2	370	23
0.04	0.5	1.6	179	5	0.9	60	7.5
0.04	1	0.66	82	5	1.4	220	18
0.04	2	0.22	28	5	2.4	600	16
0.1	0.5	0.58	29	5	1.5	45	16
0.1	1	0.32	16	5	2	230	5.5
0.1	2	0.14	7	5	3	875	6.5

IV. EXPERIMENTAL RESULTS

Experiments on the above coil designs are executed to check the amount of harvested power delivered to the implanted chip. Experimental results showed that it is almost impossible to use a receiver coil of Fig. 3 with a volume of 1 mm^3 for required power conversion, without a ferrite core in the centre of the receiver coil ($\mu_f = 1$). Therefore, higher value of μ_f needs to be used.

However, the amount of transmitted power and the output voltage can be increased with a factor μ_f^2 by adding a ferrite core to the receiver coil. This means that we can multiply the amount of transmitted power and the voltage (table 2) with a factor of 100, when we assume that $\mu_f = 10$.

Table 1 shows the transmitted power and output voltage of some receiver coils with a ferrite core in the middle ($\mu_f = 10$), again with a total volume of 1 mm^3. In this case the volume needed for the embedded electronics is not taken into account.

TABLE 2. Transmitted power and output voltage for several receiver coil designs. All coils are injectable through a needle and have a ferrite core in the middle (μf ≈ 10).

Coil Length (mm)	Coil width (mm)	Coil Height (mm)	Number of Layers	Number of Turns	Power (μW)	Voltage (V)
5	0.75	0.1	5	5	42	0.35
5	0.75	0.3	5	15	140	2.47
5	0.75	0.5	5	25	230	5.5
5	1	0.1	5	5	180	1.0
5	1	0.3	5	15	500	6.9
5	1	0.5	5	25	800	17.2
5	1.5	0.1	5	5	590	2.45
5	1.5	0.3	5	15	2000	19.7
5	1.5	0.5	5	25	3000	48.6
10	0.75	0.1	5	5	85	0.58
10	0.75	0.3	5	15	270	3.8
10	0.75	0.5	5	25	450	9.0
10	1	0.1	5	5	400	1.58
10	1	0.3	5	15	990	12
10	1	0.5	5	25	1800	29.1
10	1.5	0.1	5	5	1400	4.0
10	1.5	0.3	5	15	3900	32.0
10	1.5	0.5	5	25	6500	80.3
15	0.75	0.1	5	5	1500	0.65
15	0.75	0.3	5	15	400	4.9
15	0.75	0.5	5	25	660	12.7
15	1	0.1	5	5	530	1.8
15	1	0.3	5	15	1800	15.1
15	1	0.5	5	25	280	36.7
15	1.5	0.1	5	5	2200	4.95
15	1.5	0.3	5	15	6000	41.4
15	1.5	0.5	5	25	9900	107

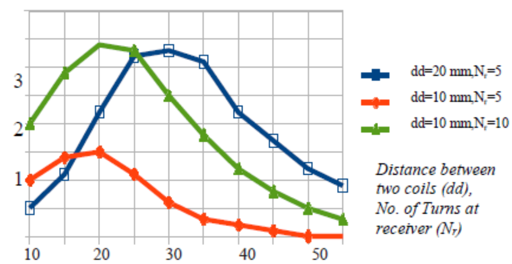

FIGURE 5. Receiver Voltage using a transmitter coil with 30 mm diameter.

The amount of transmitted power to a receiver coil with a volume of 1 mm^3, and a ferrite core in the middle ($\mu_f = 10$), fits the requirements, as can be seen in table 1.

The calculation shows that around 400 μW can be transmitted to this receiver coil with an output voltage of around 1.5 Volt, which is sufficient to power the IMD.

Also some other designs of the receiver coils that can be implanted through a hypodermic needle are proposed here.

They all have a shape similar to the receiver coil shown in Fig. 4, the same wire is used, only the dimensions of the total device are changed. The calculations (table 2) show that around 0.4 mW power can be transmitted to a receiver coil with a volume of 1 mm^3. This is enough to power the implant for three weeks with only 1 hour of recharging the battery.

For the comparison of different receiver coils, the same transmitter circuit was used for all verification by placing the receiver at around 3 cm from the same transmitter. The results are shown in figure 5.

V. DISCUSSION & CONCLUSION

Calculations have shown that enough power can be harvested to power up IMD at good power density to a receiver coil with a volume of 1 mm^3 with a ferrite core integrated in the middle. The configuration of the coil with ferrite can be optimised to allow straightforward injection through a needle as shown in this work. Calculations have shown that around the 400 μW can be transferred to this receiver coil with an output voltage of 1.5 Volt. All the safety limits are taken into account in the transmitter coil design. It was observed that the ferrites with the large length-to-diameter ratios gives better values in mutual inductance. The only limit which is exceeded is the limit for the maximum allowed B-field, but it is assumed that this is allowed. Further research is needed to optimise coil and circuit design for making it MRI compatible. To conclude, to inductively transfer power provides easy, robust and efficient method to deliver the needed power to the miniaturized implant.

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