Multi-coil approach to reduce electromagnetic energy absorption for wirelessly powered implants

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Near-field inductive coupling is a commonly used technique for wireless power transfer (WPT) in biomedical implants. Owing to the close proximity of the implant coil(s) with the tissue (~ 1 mm) and high current (~ 100 – 300 mA) in the magnetic coil(s), a significant induced electric field can be generated for the operating frequency (1–20 MHz). In this Letter, a multi-coil-based WPT technique is proposed to selectively control the currents in the external and implant coils to reduce the specific absorption rate (SAR). A three-coil WPT system, that can achieve 26% reduction in peak 1-g SAR and 15% reduction in peak 10-g SAR, as compared to a two-coil WPT system with the same dimensions, is implemented and used to demonstrate the effectiveness of the proposed approach. To achieve the seamless design for the external and implant electronics, the multi-coil system achieves the same voltage gain and bandwidth as the two-coil design with 46% improvement in the power transfer efficiency.

1. Introduction: Wireless power transfer (WPT) has proven to be a viable and necessary technology to power implantable electronics [1–3]. To accommodate the advances in prosthetic systems [2, 3], the power requirement of the implanted device can vary hundreds of milli-watts [2–5]. To ease the mobility of the patient, most of these devices use body-worn batteries as a power source. This requires an efficient wireless powered system to improve the battery life. Traditionally, a two-coil based inductive link is commonly used to design a wireless power system [2, 4, 5]. Such systems require a power amplifier to drive a resonating driver (external) coil and cause a time-varying magnetic field. The load (implant) coil is connected to the implant electronics. The power transfer efficiency (PTE) of the two-coil-based WPT depends on the magnetic coupling and loaded Q -(quality) factor of the magnetic coils.

For many prosthesis devices such as an epiretinal prosthesis [4, 5], the driver coil (3–5 cm) is much larger than the implant coil (0.5–2 cm) to achieve sufficient magnetic coupling over a long operating distance (0.5–1.8 cm) between the coils. Moreover, the driver and implant coils are in close proximity of the tissue, which can cause a significant induced electric field and eddy currents in tissues. The induced electric field is directly proportional to the currents in the magnetic coils. Thus, to design a safe electromagnetic WPT system, currents in the magnetic coils may need to be reduced. Traditionally, a two-coil WPT system provides only few parameters (Q -factors and coupling k , Fig. 1) to tune the currents in the magnetic coils, which need to be optimised to achieve sufficient PTE ($>30\%$) under design constraints and operating conditions [4, 5].

Recently, a multi-coil-based wireless powered system was proposed to achieve high wireless PTE between external and implant electronics [6, 7]. The multi-coil WPT system can achieve more than twice the PTE as compared to the two-coil design under the same size restrictions and operating conditions [7]. Multi-coil WPT systems utilise intermediate coils to improve the PTE and result in a high number of design parameters (k_1 – k_3 , Q_d , Q_t and Q_l , Fig. 2) that can be used to control the current in each magnetic coil. In this Letter, a multi-coil-based WPT system is demonstrated to achieve the same system performance (voltage gain and bandwidth) as a two-coil WPT design under the same size restrictions and operating conditions. It is shown that without any change in the driver or implant electronics, this new multi-coil WPT system

can replace the traditional two-coil system requiring the same dimensions. The current in each coil can be selectively tuned to reduce the induced electric field in tissue. Formulation and demonstration of a multi-coil WPT system, which can reduce the specific absorption rate (SAR) in telemetry systems, are the main contributions of this work.

2. Safety aspects: SAR is a standard quantity used to provide a measure of the electromagnetic energy deposited in the conductive tissue (1). Absorbed energy can be a key contributor of thermal [8] or non-thermal effect [9] in tissue. At low frequencies, induced electric field is linearly proportional to the current in the magnetic coil [8].

$$\text{SAR} = \frac{\sigma(r)|E_{\text{rms}}(r)|^2}{\rho(r)} \quad (1)$$

where at location r , $\sigma(r)$ is the tissue electrical conductivity of the

Figure 1 Magnetically coupled two-coil WPT system with coupling k (Fig. 1a); schematic of WPT system with load coil (L_l) in series with load resistance R_L (Fig. 1b); schematic of the WPT system with load coil (L_l) in parallel with load resistance R_L (Fig. 1c); effect of load coil inductance L_l on the efficiency η and Q_{ll} of the two-coil WPT system (Fig. 1d and Table 1)

Figure 2 Block diagram (Fig. 1a); schematics of three-coil (multi-coil)-based WPT system showing the coupling k_{1-3} between each coil (Fig. 1b)

tissue, $E_{rms}(r)$ is the RMS peak induced electric field and $\rho(r)$ is the tissue density in kg/m^3 .

SAR increases with the conductivity of the tissue and current in the magnetic coils. Therefore; the current of the coil near high conductivity tissue needs to be minimised to keep the absorbed power in tissue well within the safety standards. In the following Sections, an analytical formulation is presented to identify the key parameters that can be utilised to tune the current in the individual coils for the traditional two-coil WPT system and multi-coil systems.

3. Currents in the two-coil WPT system: Two-coil WPT system is a traditional technique for transferring energy from the external source to the implant coil. Fig. 1a shows the basic block diagram of this system. Current in the driver coil (2) and load coil (3) can be calculated based on the coupling k , resistance $R_d = R_{\text{driver}} + R_{\text{CD}}$, (where R_{CD} is self-resistance of the driver coil), and Q -factors Q_d (driver coil) and Q_l (load coil). R_d and R_l are the effective series resistances connected to the driver and load coils, respectively

$$I_d = \frac{1}{R_d(1 + k^2 Q_d Q_l)} V_1 \quad (2)$$

$$I_l = -j \frac{k \sqrt{Q_d Q_l}}{\sqrt{R_d R_l} (1 + k^2 Q_d Q_l)} V_1 \quad (3)$$

The load resistance R_L depends on the current and the voltage requirements of the implant electronics. Depending on the value of R_L and load inductance L_L , there are two popular topologies (series and parallel) to connect R_L with the resonating load coil. In a series load configuration (Fig. 1b), the Q -factor of the load coil is inversely proportional to R_l ($Q_l = \omega L_L / R_l$, $\omega = 2\pi f_{\text{res}}$), where $R_l = R_L + R_{\text{CL}}$ (R_{CL} is self-resistance of the load coil). From (4), it can be seen that the required driver current increases with the reduction in coupling k . Owing to the small inductance of the implant coil (below 10 μH), only a small load resistance (below 100 Ω) can achieve a moderate Q_l of 3–6 for f_{oper} below 10 MHz

$$\left| \frac{I_d}{I_{\text{load}}} \right| = \frac{R_l}{\omega M} = \frac{R_l}{\omega k \sqrt{L_d L_L}}; \quad \left| \frac{I_l}{I_{\text{load}}} \right| = 1 \quad (4)$$

where M is the mutual inductance between the driver and the load

coil, and L_d and L_L are the self-inductance of the driver coil and load coil, respectively.

For a large load resistance R_L (200 Ω –10 k Ω), a parallel load configuration utilises an impedance transform mechanism to achieve smaller effective resistance $R_l = R_L / Q_{\parallel}^2 + R_{\text{CL}}$ (where $Q_{\parallel} = R_L / \omega L_L$). The current of the load coil I_l gets divided between the resonating capacitor (C_L) and R_L (Fig. 1c, (7)). For a fixed R_L , increasing the load coil inductance L_L reduces $Q_{\parallel}$ and increase R_l , limiting the value of Q_l . The PTE for the two-coil WPT system increases monotonically with Q factors (Q_d and Q_l) and coupling k between the coils (5). Therefore L_L needs to be adjusted to achieve a moderate $Q_{\parallel}$ (below 15) to reduce the current division and sufficient Q_l to achieve PTE above 30% ((5) and (6), Fig. 1d)

$$\eta = \frac{k^2 Q_d Q_l}{1 + k^2 Q_d Q_l} \quad (5)$$

Table 1 System specifications

Parameters	Two-coil		Three-coil		
	Driver	Load	Driver	Tx	Load
D_{out} , cm	4.0	1.5	1.2	4.0	1.5
N_{coil}	12	12	4	12	15
L_{coil} , μH	5.68	2.58	0.39	5.68	4.2
R_{coil} , Ω	0.4	0.22	0.1	0.4	0.45
R_{driver} , Ω	8	—	8	—	—
R_{load} , Ω	—	450	—	—	450
$Q_{\text{(loaded)}}$	8.5	12.7	0.6	178	7.95
Litz [10]	100/44	3/22/48	100/44	100/44	33/48
d , mm	15		15		
f_{res} , MHz	2		2		
load type	parallel		parallel		
coupling	$k = 0.072$		$k_1 = 0.33, k_2 = 0.072, k_3 = 0.04$

Figure 3 Head model for epiretinal prosthesis with external coil position (Fig. 1a); simulation model with $1 \times 1 \times 1$ mm resolution including tissue heterogeneity (Fig. 1b); positions of implant and external coil with respect to eye (cross-section model) (Fig. 1c); conductivities of different tissues types at 2 MHz [11] (Fig. 1d)

Table 2 Coil current

Coil	Driver I_d , mA	Transmitter I_t , mA	Load I_l , mA	$R_L I_{load}$, mA
two-coil	229.5		310.6	22.32
three-coil	156.2	282.35	190.9	22.24

$$Q_1 = \frac{\omega L_L}{(\omega L_L)^2 / R_L + R_{CL}} \quad (6)$$

$$\left| \frac{I_d}{I_{load}} \right| \approx \frac{Q_1 R_L}{\omega M}; \quad \left| \frac{I_l}{I_{load}} \right| \approx Q_{||} \quad (7)$$

where M is the mutual inductance between the driver and the load coil; and L_d and L_L are the self-inductance of the driver coil and load coil, respectively.

For most biomedical implants, R_L lies in a higher value range (200 Ω –10 k Ω) [1–3], making a parallel load configuration as a practical topology to achieve sufficient PTE above 30% [4, 5]. However, care needs to be taken in designing a load coil to achieve trade-off between the efficiency and current division (Fig. 1d). In the following Sections, our focus is to reduce the current division without losing the PTE of the wireless power link.

4. Currents in a multi-coil WPT system: Recently, we proposed a multi-coil wireless power system for the biomedical applications [7]. The proposed design utilises intermediate transmitter and(or) receiver coils to decouple the effect of the source and(or) load resistances and achieve high Q factor external and(or) implant coils. Fig. 2a shows the block diagram of the three-coil system. At resonance, (8) can be used to calculate the current in each coil

$$\begin{bmatrix} I_d \\ I_t \\ I_l \end{bmatrix} = \begin{bmatrix} R_d & j\omega M_{dt} & j\omega M_{dl} \\ j\omega M_{dt} & R_t & j\omega M_{tl} \\ j\omega M_{dl} & j\omega M_{tl} & R_l \end{bmatrix}^{-1} \begin{bmatrix} V_1 \\ 0 \\ 0 \end{bmatrix} \quad (8)$$

where M_{dt} , M_{tl} and M_{dl} are the mutual inductances between the driver and transmitter coil, the transmitter and load coil and the driver and load coil, respectively.

In multi-coil design, M_{dl} is many orders smaller than M_{tl} , allowing us to ignore M_{dl} in the current calculation [7]. To generate the unit current in the load resistance, the current ratio of the driver, transmitter and load (implant) coils can be calculated as shown in the following equations

$$\left| \frac{I_d}{I_{load}} \right| = \sqrt{\frac{R_l}{R_d k_1 k_2}} \frac{1 + k_2^2 Q_t Q_l}{\sqrt{Q_d Q_t} \sqrt{Q_t Q_l}} \quad (9)$$

Figure 4 Cross-section view of induced electric field and distribution of SAR

a Induced E-field due to two-coil WTP system

b Induced E-field due to three-coil WTP system

c SAR two-coil WTP system

d SAR three-coil WTP system

Coil currents are based on Table 2

$$\left| \frac{I_t}{I_{\text{load}}} \right| \simeq \frac{Q_{\parallel} R_t}{\omega M_{t\parallel}}; \quad \left| \frac{I_t}{I_{\text{load}}} \right| \simeq Q_{\parallel} \quad (10)$$

where k_1 , k_2 and k_3 are the coupling between the driver coil and transmitter coil, the transmitter and load coil, and the driver and load coil, respectively, and Q_d , Q_t and Q_l are the Q -factors of the driver, transmitter and load coil, respectively.

Fig. 2b shows the schematic of the three-coil-based WPT system. The load resistance is connected to the load coil in parallel to achieve sufficient Q factor and to sustain PTE above 30%. However, multi-coil WPT can reduce $Q_{\parallel}$ significantly compared to its two-coil equivalent and the effect of the low Q -factor load coil Q_l on the PTE is compensated using a high Q -factor transmitter coil [7].

5. Simulation model and methods: To model a practical wireless powered implant, an epiretinal prosthesis is taken as a design example [4, 8]. The traditional design uses two-coil-based WPT system [3–5]. Thus, to improve the PTE and to reduce the absorbed energy in the tissue, a three-coil-based WPT system is selected. Table 1 shows the electrical and mechanical properties of two-coil and three-coil equivalent systems. For both designs, the physical dimensions of the external coil and implant coil are identical to ensure fair comparison (Fig. 3c). However, the inductance of the load coil in the two-coil system is optimised to achieve $\eta > 30\%$ and $Q_{\parallel} < 15$ (Fig. 1d). For the three-coil WPT system, $Q_{\parallel}$ is reduced by 40% compared with its two-coil equivalent and its effect on the system's PTE is compensated by a high Q -transmitter coil ($Q_t \sim 178$). The driver coil is fed with a voltage source with amplitude 3 V and operating frequency of 2 MHz. Implant electronics is modelled as a resistive load of 450 Ω , which requires a load current of 22.3 mA to generate 10 V across implant [2]. To achieve maximum coupling, the external coil is positioned parallel to the surface of the head (Fig. 3c) and wireless power is transferred over the operating distance of 15 mm. In both designs, a driver resistance R_{driver} of 8 Ω is used to emulate the practical power-amplifier output resistance [12].

For our simulation, we took the heterogeneous tissue model of the human head [13] with $1 \times 1 \times 1$ mm resolution. For the frequency range of 1–10 MHz, the conductivity of most tissues (skin, sclera, retina and muscle) vary between 0.01–1.5 S/m [11]. Fig. 3d shows the conductivity map for different tissues at the operating frequency of 2 MHz [11]. As the implant coil is in close proximity to the high conductive vitreous humour (1.5 S/m), the implant coil contributes highly to the absorbed energy.

Figs. 3a–c show the head model, 3-D simulation model and cross-section of the simulation model, showing the external and implant coils. Fig. 3c shows the position of the external coil parallel to the head surface and implant coil next to the eye sclera. For both designs, an external coil with diameter 40 mm and an implant coil with diameter 15 mm are used (Table 1). The equivalent three-coil system consists of external coils (driver and transmitter coil) and an implant coil as shown in Fig. 2b. For the two-coil WPT system and its equivalent three-coil WPT system, the induced electric field is calculated using the impedance method [14] for current carrying coils to determine the absorbed power in the tissue.

6. Results and comparison: For the two-coil system, (2) and (3) can be used to calculate the currents in the driver coil, the load coil and the load resistance. Similarly, for the three-coil system, current in each coil can be calculated based on (9) and (10). Table 2 shows the current in each coil and load resistance for the two-coil and three-coil WPT systems. Table 2 also shows that the current in the implant coil is significantly reduced as compared to the two-coil design to generate the same load current of ~ 22.3 mA.

Figs. 4a and b show the cross-section view of the induced electric field owing to the two-coil and the three-coil WPT system,

Figure 5 PTE of two-coil and three-coil system (Fig. 1a); voltage-gain of two-coil and three-coil system (Fig. 1b)

respectively. For the two-coil system, because of the large current in the implant coil (Table 2), the induced field near the implant coil is maximum (Fig. 4a). For tissue density of 1000 kg/m³, the peak 1-g SAR and peak 10-g SAR are calculated based on the FCC (Federal Communications Commission) [15] and IEC (International Electrotechnical Commission) [16] standards for SAR, respectively. Figs. 4c and d show the distribution of SAR owing to the two-coil and three-coil WPT system, respectively.

To evaluate the performance of the new three-coil WPT system, the system performance is characterised based on the PTE, the voltage gain and the bandwidth [7]. Fig. 5a shows that the proposed three-coil WPT system can achieve significant PTE improvement over two-coil design as expected from the multi-coil design approach [7]. The new three-coil design achieves the same voltage gain over frequency as the initial two-coil design without change in driver electronics, implant electronics or system dimensions (Fig. 5b). Therefore it can be seamlessly incorporated into existing two-coil WPT systems.

Table 3 System performance

	1-g SAR, mW/kg	10-g SAR, mW/kg	Efficiency, %	Voltage gain	Bandwidth, kHz
two-coil	48.2	15.1	32.5	3.348	209
three-coil	35.7	12.8	47.5	3.336	207
variation	−26%	−15.2%	+46%	<1%	<1%

Table 3 shows the significant reduction in absorbed electromagnetic energy by using three-coil WPT over a traditional two-coil design.

7. Conclusion: Two-coil-based WPT systems only have a few factors (k , Q_d and Q_L) to control the current in each coil. However, multi-coil WPT systems utilise multiple coils to improve the PTE and result in a high number of design parameters (k_{1-3} , Q_d , Q_t and Q_l) that can be used to control current in each magnetic coil. For the proposed design example, a reduction of 26% in peak 1-g SAR and a reduction of 15% in peak 10-g peak are achieved with the improvement of 46% in PTE. A two-coil WPT system and its three-coil equivalent are demonstrated to achieve the same voltage gain and frequency bandwidth over the same operating distance. Although the presented design example used the typical dimensions for biomedical WPT system, the design approach is valid for any near field WPT system to reduce the absorbed electromagnetic field in tissue.

8 References

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