Reducing human body heating and temperature rises due to inductively-powered implantable medical devices

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Abstract. The use of inductive power transfer (IPT) for medical implants can cause heating and temperature rises in human tissue. Increasing the load resistance away from the optimal value for maximising link efficiency has been shown to reduce the specific absorption rate (SAR) and temperature rises in a human body model in electromagnetic and thermal simulations. However, these quantities are generally difficult to measure as they are internal to the body, so experiments with a phantom should be used to verify the theory and simulations. This paper presents results of an experiment, in which 10W is delivered to a receiver immersed in salt water with two different load resistances. The temperature rise of the salt water was lower with the use of the increased load. Simulations of the experimental setup demonstrate good agreement with experimental results.

1. Introduction
Inductive power transfer (IPT) systems for high-power medical implants such as left ventricular assist devices (LVADs) can cause heating in the human body due to magnetic fields inducing eddy currents in biological tissue and heat dissipation in the implanted receiver coil. Many groups have considered human body heating when using IPT for LVADs. Tissue heating effects were explored in [1], but their simulation studies did not include the implanted receiver coil responsible for localised heating. In [2], the authors identified coil dissipation and losses in the power electronics as the main heating sources; however, they did not consider losses from human exposure to magnetic fields.

In previous work [3], it was shown with electromagnetic (EM) and thermal simulations that varying the load can reduce the heating and temperature rises in a human body model. This is beneficial for preventing excessive heating in the body that could lead to tissue damage, and allows safety regulations and standards to be met. These include the specific absorption rate (SAR) limits in the ICNIRP 1998 guidelines and the 2°C maximum temperature rise limit of an active implantable medical device in the EN 45502-1 standard for CE marking to comply with Directive 90/385/EEC. However, quantities such as SAR are generally difficult to measure directly, as they are internal to the body. Therefore, experiments can be performed on a phantom to ensure that the proposed design methods are valid. This paper presents results of an experiment showing a reduction in the temperature rise of the surrounding salt water.
solution in which the receiver is immersed, when the load resistance is changed. Simulations of the experimental setup are then carried out to verify the results and the theory.

2. Theory and design methods

IPT entails the transfer of power from a transmitter coil to a receiver coil using a high frequency magnetic field. To achieve maximum link efficiency, receiver resonance is used to cancel the receiver coil’s reactance and the optimal load is attached [4]. The maximum link efficiency can then be evaluated using a well-known expression [4].

The primary cause of SAR is the magnetic field through the conductive human tissue. The overall magnetic field at the centre of the receiver coil inside the human body can be considered as a combination of the magnetic field generated by the transmitter coil and the magnetic field due to current flowing through the receiver coil, which add in quadrature due to resonance [3]. The overall magnetic flux density for a pair of circular coils is given by Equation (1) in Figure 1, (where \( P_L \) is the useful load power, \( R_L \) is the load resistance, \( R_{RX} \) is the receiver coil’s resistance, \( N_{RX} \) is the number of turns of receiver coil, \( \omega \) is the frequency of operation multiplied by \( 2\pi \), \( a_{RX} \) is the radius of receiver coil) [3]. The plot in Figure 1 shows how the overall magnetic field, the field due to a 4-turn 10cm transmitter coil, and the field due to a 2-turn 5cm receiver coil (at the centre of the receiver coil) vary with the load resistance, where link efficiency is maximised with \( R_L = 3.87\,\Omega \) (assuming a constant received power of 15W, distance of 2cm and an operating frequency of 6.78MHz). It can be seen that overall magnetic field initially drops with load resistance, but then reaches a minimum (at 7.33Ω) before increasing again. Reducing the overall magnetic field should decrease SAR in the vicinity of the receiver, but it may possibly lead to additional heating in the tissue in other areas due to increased transmitter coil currents.

\[
B_{\text{overall}} = \sqrt{\frac{2P_L (R_{RX} + R_L)^2}{R_L \pi^2 \omega^2 N_{RX}^2 a_{RX}^2} + \frac{P_L \mu_0^2 N_{RX}^2}{R_L 2a_{RX}^2}} \tag{1}
\]

Figure 1. Variation of overall magnetic flux density (blue), flux density due to TX coil (yellow), and flux density due to RX coil (orange) with load resistance (at centre of RX coil)

Apart from heating caused by magnetic fields inducing eddy currents in the conductive human tissue, another source of heating is the power dissipation (\( P_{\text{diss}} \)) due to the current \( I_{RX} \) flowing in the receiver coil, which is calculated by \( P_{\text{diss}} = \frac{|I_{RX}|^2 R_{\text{coil}}}{2} = \left( \sqrt{\frac{2P_L}{\pi R_L}} \right)^2 R_{\text{coil}} = \frac{P_L R_{\text{coil}}}{R_L} \), where \( R_{\text{coil}} \) denotes the coil’s own skin effect and proximity effect loss resistances. The key to reducing heating of the surrounding tissue by dissipation in the receiver coil is to lower the receiver coil current. For a fixed received power, this can be achieved by increasing the load resistance \( R_L \).
Table 1. Impedance measurements of coils in the presence of salt water at 13.56 MHz

|       | $L_{TX}$ | $R_{TX}$ | $L_{RX}$ | $R_{RX}$ | $L_{TX,S}$ |
|-------|----------|----------|----------|----------|------------|
| Value  | 1.4134 μH | 856 mΩ  | 265 nH   | 418 mΩ   | 1.4068 μH  |

3. Wireless power experiments in salt water solution

The transmitter coil is a 4-turn circular spiral PCB coil with an outer diameter of 10 cm, a trace width of 4 mm, and a turn spacing of 4 mm (Figure 2). The receiver is a 2-turn circular spiral PCB coil, with an outer diameter of 5 cm, a trace width of 3 mm, and a turn spacing of 3 mm (Figure 3). These coils are chosen to have outer dimensions that are suitable for the wireless powering of an LVAD with a body-worn transmitter.

The distance between the transmitter coil and the receiver coil for the experiments is 4 cm. A 15 cm ($L$) × 9 cm ($W$) × 14 cm ($H$) plastic tank is used to hold the salt water solution, which is placed 2 cm above the transmitter coil. The receiver coil, coated in silicone conformal coating, is placed in the salt water 2 cm from the bottom of the tank. The solution is produced by adding 5.77 g of table salt to 337.5 cm³ of de-ionised water to achieve a conductivity of 2.512 S m⁻¹ [5]. This is four times that of muscle at 13.56 MHz to obtain more prominent temperature rises. An expanded polystyrene foam holder was constructed to insulate the salt water solution (Figure 4).

Impedance measurements of the coils in situ are performed with a Keysight E4990A Impedance Analyzer and 42941A Impedance Probe. These give the resistance $R_{TX}$ and inductance $L_{TX}$ of the transmitter coil and the resistance $R_{RX}$ and inductance $L_{RX}$ of the receiver coil, when loaded by the salt water solution. In addition, the inductance $L_{TX,S}$ of the transmitter coil when loaded with a shorted receiver coil in salt water is measured. The coil measurements at 13.56 MHz with salt water are summarised in Table 1. The coupling factor is given by $k = \sqrt{1 - \frac{L_{TX,S}}{L_{TX}}}$, which is 6.83% for this particular system.

The receiver coil PCB is connected to another PCB containing the series-resonant tuning capacitor and a current-driven half-wave Class D rectifier. The relationship between the AC load at the input of this rectifier ($R_L$) and output DC load ($R_{L,DC}$) is given by $R_{L,DC} = 0.5\pi^2 R_L$ [6]. The transmitter coil is driven by a load-independent Class EF inverter [7], which is able to sustain a constant output current over a range of load resistances without losing zero-voltage switching.

The purpose of the experiment is to show how increasing the load resistance can reduce the temperature rise of the salt water at the receiver, given a constant received power of 10 W.
Table 2 shows the link efficiency and coil currents when using the optimal load and increased load (1.5 times the optimal load). A K-type thermocouple will be used to measure the initial and final temperature on the surface at the centre of the receiver coil PCB, but is removed from the salt water solution during the operation of the system to prevent additional heating due to induced currents in the thermocouple.

|                  | $R_L$ (Ω) | $R_{L,DC}$ (Ω) | $\eta_{\text{link}}$ (%) | $I_{TX}$ (A) | $I_{RX}$ (A) |
|------------------|----------|----------------|--------------------------|--------------|--------------|
| Optimal load     | 2.55     | 12.6           | 72.8%                    | 2.30         | 2.80         |
| Increased load   | 3.82     | 18.9           | 71.3%                    | 2.69         | 2.29         |

Figure 5 is a scatter plot of the experimental results, showing the temperature rise of the salt water solution for each run. For each type of load (optimal and increased), twelve 20-minute runs were performed. The mean temperature rises for the original and increased loads are 4.48°C and 3.59°C, respectively. Therefore, the increase in the temperature rise when switching from the original load to the increased load is 0.89°C.

Figure 4. Polystyrene foam holder insulating the salt water solution in the tank

Figure 5. Experimental results: temperature rise of salt water solution

4. Simulation verification

Figure 6 shows the system modelled in CST Studio Suite. The EM simulations of the system are performed with an optimal load of 3.07Ω and an increased load of 4.60Ω (assuming always a received power of 10W). Table 3 shows the power dissipation in the transmitter coil, the receiver coil and the salt water block.

Table 3. Power dissipation in the system

|                  | $P_{TX}$ (W) | $P_{RX}$ (W) | $P_{salt}$ (W) |
|------------------|--------------|--------------|---------------|
| Optimal load     | 2.30         | 2.80         | 0.80          |
| Increased load   | 2.69         | 2.29         | 0.78          |

After performing the EM simulations, thermal simulations are carried out to obtain temperature rises, assuming an ambient and initial temperature of 20°C. Figures 7 and 8 show the cross-sectional temperature distribution through the centre of the receiver coil and the salt water block, 20 minutes after the start of the simulation, for the optimal load and increased load, respectively. For the optimal load, temperature on the surface at the centre of the receiver coil rose by 4.45°C to 24.45°C, whilst for the increased load, it rose by 3.48°C to 23.48°C. The difference in temperature rise between the optimal load and the increased load is 0.97°C. Note that the temperatures in the upper corners of the salt water box have increased due to the higher voltages and currents on the transmitter coil when the increased load is used.
Table 3. Power dissipation in coils and salt water with both optimal and increased load

| Component      | Optimal load | Increased load |
|----------------|--------------|----------------|
| Transmitter coil | 0.58 W       | 0.80 W         |
| Receiver coil  | 0.36 W       | 0.24 W         |
| Salt water block | 2.2 W       | 2.3 W         |

5. Conclusion

This paper presented an experiment comprising an IPT system with a receiver immersed in a salt water solution. It has been demonstrated that changing the load resistance away from the optimal value for maximising link efficiency can reduce heating and temperature rises in the surrounding medium. The results are in good agreement with EM and thermal simulations of the experimental system. Therefore, varying the load is a viable method of reducing heating when using IPT for medical implants, which can allow systems to comply with safety standards.

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