Mid-range Wireless Power Transfer with Segmented Coil Transmitters for Implantable Heart Pumps

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***Abstract*- In wireless power transfer systems, the transmitting coil dimensions can substantially affect the transmission range and alignment sensitivity. We found that a transmitting coil with larger inner and outer diameter has a wider transmission range and lower alignment sensitivity. Thus, we developed a larger coil (24×30 cm^2) designed to be embedded in the back of a vest to power DC pumps for artificial hearts or LVADs. To significantly reduce the required voltage, the coil was divided into 8 segments with resonant capacitors. The coil was operated at 6.78 MHz and evaluated with a 5.3-cm diameter receiving coil. A circuit model for the energy coupling coils was developed to predict the output power and efficiency. Having a coil separation of 7.7 cm, the output power and efficiency of the energy coupling coils are higher than 48 W and 80%, respectively. The system was experimentally tested with a DC pump, demonstrating that the proposed coil segmentation technique can significantly reduce the transmitter voltage to a safe level (~10 V_{rms}).**

I. Introduction

Differences in the dimensions of energy transmitting coils in wireless power transfer systems can lead to substantial differences in transmission range and coil alignment sensitivity. In transcutaneous transformers, which have been investigated for powering total artificial hearts (TAHs) and left ventricular assist devices (LVADs) for decades, the transmitting coil diameter is typically less than 12 cm, resulting in a transmission range limited to 20 mm, and an allowed lateral misalignment of approximately 10 mm [1],[2]. Thus, precise coil alignment is required; otherwise, the energy efficiency will be diminished. Since the receiving coil must be implanted under the patient's skin and the implantable heart pump is located relatively deep in the body, long wires are needed to connect the receiving coil to the implanted device. This arrangement substantially increases the surgical time, complexity, and thus cost. The wires also create reliability issues, particularly when the patient moves vigorously.

In mid-range wireless power transfer systems, relatively large transmitting coils (e.g. 30 cm) can power implantable devices located virtually anywhere in the body without precise alignment [3]-[9]. However, the required excitation voltage over the transmitting coil is much higher than what is needed for the transcutaneous transformer. For example, in an application that powered a 0.35-W capsule endoscope, an excitation more than 3-kV was required [3],[4]. In applications with higher power consumption, this voltage will be even more demanding. This high voltage requirement is obviously a serious concern in terms of patient safety and manufacturing cost due to the need for bulky electrical insulation and high-voltage electronic components. Moreover, the system energy efficiency can be drastically reduced because of the excessive dielectric power loss under a high voltage stress [7].

A novel low-operating-voltage, mid-range wireless power transfer method was reported previously [6]-[8], in which the larger transmitting coil was divided into multiple segments using resonant capacitors. The induced coil segment voltage is canceled by the capacitor voltage at the resonant frequency, thus the overall voltage can be reduced to a safe level. In this paper, we demonstrate the use of this method to a mid-range transmitting coil designed for powering a heart pump.

II. *H*-Field Distributions of Transmitting Coils

A. *Conventional Transcutaneous Transformer*

In applications of implantable TAH and LVAD, the transmitting coil of a transcutaneous transformer typically ranges from 5 to 12 cm. Fig. 1 shows a representative 5-cm transmitting coil (Coil 1). The magnetic field intensity of the coil in the axial direction, H_z, can be deduced by summing the H_z-field generated by each winding given by Equation (1) [10].

$$H_z = \frac{I}{2\pi}\frac{1}{\sqrt{(a+x)^2+z^2}}\left[K(k)+\frac{a^2-x^2-z^2}{(a-x)^2+z^2}E(k)\right] \tag{1}$$

where $K(k)$ and $E(k)$ are complete elliptic integrals of the *1st* and *2nd* kind, a is the coil radius, z and x are the distances from the coil center along the coil's axial and radial axes respectively, I is the winding current, and $k^2=4ax/[(a+x)^2+z^2]$.

Finite-element-analysis (FEA) was also used to simulate the H_z-field. Both the calculated and simulated H_z-field were plotted along the axial and radial directions (Fig. 2). The coil excitation was set to 1 A, i.e. 10 A-turn. Along the z-axis, at z=2 cm, H_z is reduced by 81.7%. Along the x-direction at z=2 cm, comparing with H_z at x=0 cm, H_z drops by 50% at x=1.7 cm. For this reason, the separation between the transmitting and receiving coils is usually less than 2 cm, and the lateral misalignment tolerance is less than 1 cm.

B. *Larger Transmitting Coils*

Recent reports have demonstrated that larger transmitting coils (∅≥30-cm) can transfer energy deep in the body [3]-[9]. In heart pump applications, sustaining continuous energy transfer is of importance, the preferable configuration is to put the larger transmitting coil in parallel to the patient's body, e.g. by embedding it in a vest, chair back or mattress [11]. In this configuration the receiving coil for an implanted heart pump could be located more than 7 cm from the transmitting coil.

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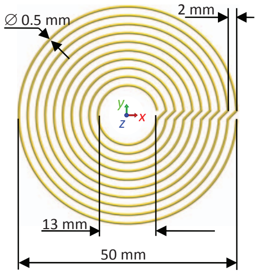

Fig. 1. Dimensions of Coil 1.

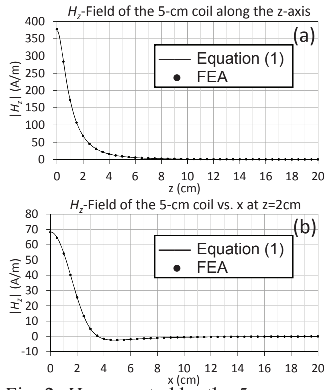

Fig. 2. H_z generated by the 5-cm transmitting coil (a) along the z-axis, and (b) versus x at z=2 cm.

To investigate the applicable range of a larger transmitting coil placed in parallel to the body, we analyzed the H_z-field distribution of four larger transmitting coils (Coils 2-5) with same outer diameter of 30 cm but different inner diameters (Fig. 3). Fig. 4(a) shows H_z along the z-axis of the four coils with an excitation of 10 A-turn. The results show that H_z with a larger inner diameter decreases more slowly than a smaller inner diameter, and slower still than the 5-cm transmitting coil. The lateral misalignment sensitivities of the 30-cm coils were investigated by plotting H_z versus x at z=10 cm (Fig. 4 (b)). The x-range, where H_z was within 50% of H_z at x=0, of the coil with the largest inner diameter (Coil 5) was 37% larger than that of Coil 2 and 8 times larger than that of the 5-cm coil.

III. The Proposed Transmitting Coil

Based on the results in the previous section, transmitting coils with larger inner and outer diameters can transmit energy deeper in the body with lower alignment sensitivity. Thus, we propose a larger transmitting coil (Fig. 5(a)) designed to power an implanted heart pump that can be embedded in the back of a vest. The energy transfer system operates at the 6.78 MHz industrial, scientific and medical (ISM) band. The coil input impedance at the no-load condition is about 116 Ω. In higher power applications (e.g. I_t = 5 A_{rms}), the required voltage is 579 V_{rms}. Such a high voltage requirement over the transmitting coil is obviously a major concern in terms of patient safety and manufacturing cost.

Recently, we demonstrated that dividing an energy coupling coil into multiple segments using resonant capacitors can significantly reduce the coil voltage to a safe level [6]-[8]. Here, we apply this technique to divide the proposed coil into 8 segments (Fig. 5(b)) to reduce the coil voltage. The coil was tuned to 6.78 MHz with 1.623 nF resonant capacitors.

IV. Output Power and Energy Efficiency

The receiving coil used in the energy coupling system is shown in Fig. 6. The receiving coil was tuned to 6.78 MHz with a capacitor network (C_1–C_3) shown in Fig. 7. The capacitors' equivalent-series-resistances (ESRs) are denoted by R_1-R_3. The self-inductance and winding resistance of the transmitting and receiving coils are represented by L_t, R_t, L_r, and R_r, respectively. The capacitor C_t denotes the resultant capacitance of the resonant capacitors used to segment the transmitting coil. The mutual inductance between the transmitting and receiving coils is denoted by L_m.

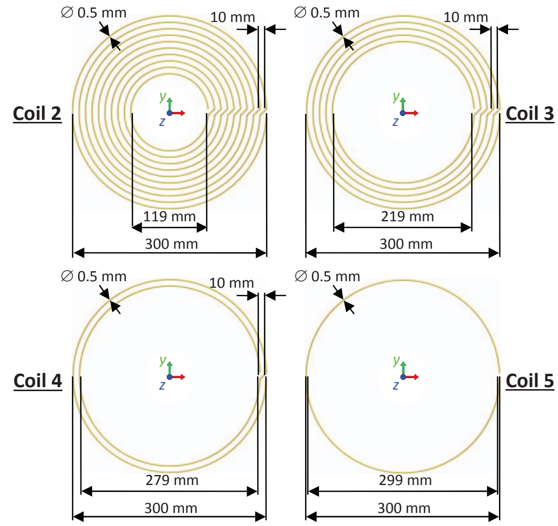

Fig. 3. Dimensions of Coils 2–5.

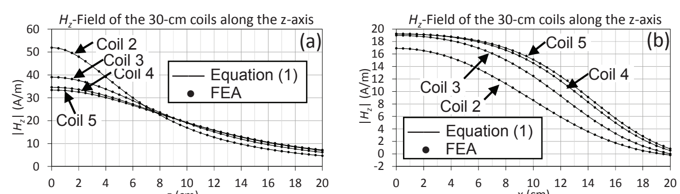

Fig. 4. H-field generated by the 30-cm energy transmitting coils (a) along the z-axis, and (b) versus x at z = 10 cm.

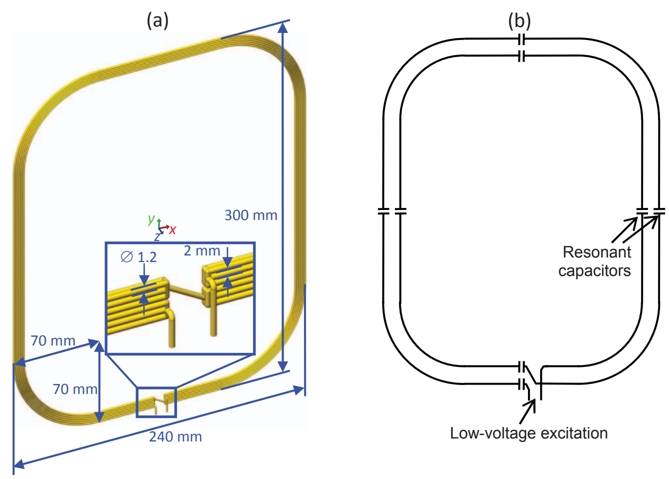

Fig. 5. (a) A 3-D drawing of the proposed transmitting coil, (b) a circuit schematic for the segmented coil.

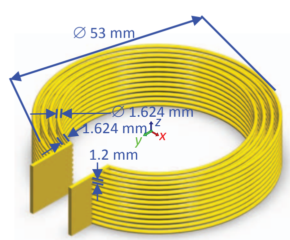

Fig. 6. The energy receiving coil.

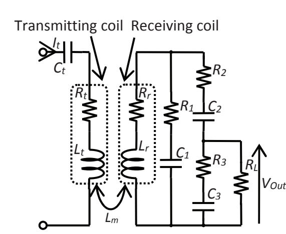

Fig. 7. Equivalent circuit model of the energy coupling circuit.

The output power and efficiency were calculated and measured in different locations along the *x*-, *y*-, and *z*-directions with an excitation of 1 A_{rms} and a load resistance of 16 Ω (Fig. 8). Both calculated and measured results show that the proposed energy coupling coils have a much wider range than conventional transcutaneous transformers. The power efficiency of the energy coupling coils is more than 80% even when the separation is 7.7 cm. The output power and efficiency were calculated and measured with different excitation current levels (Fig. 9) when the load resistance was set to 16.8 Ω. The maximum output power at z= 7.7 and 10.7 cm were 48.2 W and 38.3 W, respectively. We could not investigate higher power levels, since we reached the maximum output power capability of the RF power amplifier adopted for driving the transmitting coil.

The mid-range wireless energy coupling coils were verified with a simplified circulatory model driven by a 24-V DC pump (Fig. 10). The pump, which was used to represent an LVAD actuator, propelled the water flow cycling throughout the circulatory model. An additional load of 152 Ω was connected in parallel to the DC pump to represent the power consumption of auxiliary circuits, such as motor drive, control and communication. The measured DC load power was 19.7 W, and the transmitting voltage and current were 8.57 V_{rms} and 4.23 A_{rms}, respectively. These results verify that, even in high-power applications, the high excitation voltage necessary for overcoming the back emf induced by a larger transmitting coil is not necessary with the segmented coils developed in this study.

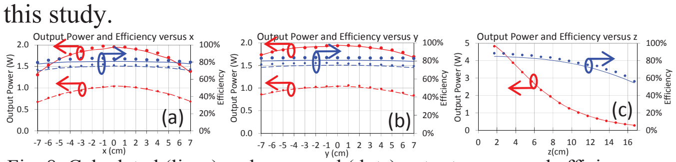

Fig. 8. Calculated (lines) and measured (dots) output power and efficiency (a) versus x when $y = 0$, and (b) versus y when $x = 0$, (c) along the z-axis. In (a) and (b), $z = 7.7$ cm (solid lines) and 10.7 cm (dashed lines).

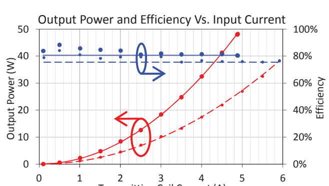

Fig. 9. Calculated (lines) and measured (dots) output power and efficiency versus transmitting coil current when $x = y = 0$, $z = 7.7$ cm (solid lines) and 10.7 cm (dashed lines).

V. Conclusions

We demonstrated that wireless power transfer using a segmented coil transmitter can drive relatively high power applications, such as heart pumps, while maintaining a safe voltage. It was found that transmitting coils with larger inner and outer diameter can provide both a wider transmission range and lower alignment sensitivity. In the system with a 24×30 cm^2 transmitting coil and a 5.3 cm receiving coil, the output power and efficiency were measured to be higher than 48 W and 80%, respectively, when the coil separation is 7.7 cm. Experimental validation has also been carried out in LVAD model with a 24-V DC pump, which was used to propel fluid through a circulatory model. The experimental results verified our hypothesis that the coil segmentation technique can achieve mid-range wireless power transfer without requiring high-voltage excitation.

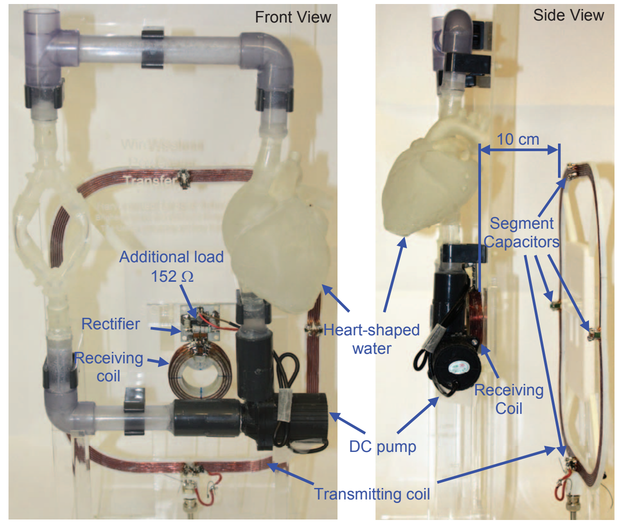

Fig. 10. Front and side views of the wirelessly powered circulatory model.

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