A Low-Operating-Voltage Wireless Intermediate-Range Scheme for Energy and Signal Transmission by Magnetic Coupling for Implantable Devices

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Abstract—The increasing sophistication and power demands of medical implantable devices have motivated a variety of research on wireless energy transmission, aiming to provide higher power capability, wider receiver position, and improved performance, including safety and efficiency. This paper demonstrates a low-operating-voltage wireless energy and signal transmission method to achieve these aims. We developed a 30-cm transmitting coil in a Helmholtz-coil configuration. Innovative coil segmentation and distributive resonance techniques are employed to substantially reduce the excitation voltage and the voltage over the coil. Magnetic field simulations show that the magnetic field generated by the scheme is uniform and applicable in a wide region bounded by the coils. The receiving coil has three turns and a diameter of 2 cm. The receiver can deliver 350 mW when the transmitting coil excitation is 1 A rms/6.68 V rms and the maximum voltage over the coil is 32 V rms, which is much lower than that required in the most up-to-date method (5 A/3.5 kV). This method can provide sufficient power to operate many kinds of medical implants, especially deep-seated and locomotive devices, such as capsule endoscopes. An operating frequency of 6.1 MHz is chosen. At such a high frequency, the receiving coil does not need ferromagnetic core so magnetic resonance imaging compatible implants can be accomplished.

Index Terms—Coil segmentation, implantable medical devices, low-voltage operation, wireless intermediate-range scheme for energy and signal transmission (WISEST).

I. INTRODUCTION

Research on wireless transcutaneous energy and signal transmission has been motivated by the growing energy requirements of the increasingly sophisticated medical implantable devices. Magnetic coupling is probably the most widely used method among various transcutaneous wireless energy transmission techniques. It has already been used in various applications such as implantable cardiac pacemakers, drug delivery systems [1], defibrillators [2], spinal cord stimulator [3], [4], and middle ear hearing devices [5]. The clinical need for advanced implants, for example, next-generation capsule endoscopes [6], [7] and artificial hearts [8], [9], has necessitated the demand for more power.

In transcutaneous magnetic coupling, the transmitting coil placed parallel to the skin surface is driven by a high-frequency current that generates magnetic field concentrated around the coil [10]–[13]. The receiving coil located under the skin converts the magnetic energy to electric energy to operate the implantable circuit. To maximize power transmission, the transmitting coil has to be well-aligned and close to the receiving coil [14]. A transcutaneous transformer could transfer 12–48 W of power for an artificial heart when the separation between the transmitter and receiver was 1–2 cm [8]. However, this method is only suitable for applications that the receiving coil is placed beneath and close to the skin with a fixed location. The separation between the transmitting and receiving coils is limited to 1–2 cm. In addition, the received voltage and power are very sensitive to the coil alignment error.

Wireless energy and signal transmission for a deep-seated implantable device located 15 cm from the skin was proposed in [15]. A single-turn current-carrying coil wrapped around the body is used to generate an evenly distributed magnetic field deeply inside the body. A 2-cm diameter air-core receiving coil at or around the center of the transmitting coil is used to receive the magnetic energy to operate an implantable ultrasonic pulser-receiver circuit without embedding a battery. Although the magnetic field around the transmitting coil center is even, the receiving coil position is limited to about 3 cm from the plane of the transmitting coil.

Wireless power and data transmission strategies that use a pair of 30-cm diameter Helmholtz coils to generate an evenly distributed magnetic field in a wider range deeply inside the body were proposed in [6] and [7]. These methods can provide sufficient energy to power up a capsule endoscope, which is locomotive and deeply inside the body, without the need of embedded batteries. While most commercial capsule endoscopes use two watch batteries, in which the stored energy is limited by their sizes, the operating time of such endoscopes is limited and the image resolution is not comparable with that provided by traditional endoscopes. The wireless energy transmission strategies were proven to transfer power over 300 mW to a locomotive capsule endoscope, in which three orthogonal coils were wrapped around a ferrite core to receive magnetic energy in all directions. This method can allow the endoscope to take higher resolution pictures as needed without the time limitation imposed by the embedded batteries. However, to transmit sufficient power to the capsule endoscope, the transmitting coil requires a driver that can...
produce an output as high as 5 A and 3.5 kV at 1 MHz [6]. Another study demonstrated by Xin et al. [7] also shows that the transmitting coil requires a high-voltage resonant driving circuit, in which the LC tank voltage is up to several kilovolts. Although the driving circuits require lower supply voltage, high voltage presents at both the driver output and also over the transmitting coil. Clearly, electric field shielding and high-voltage insulation for both the driver and the transmitting coil are required for the patient’s safety and to comply with safety regulations. The required shielding, electrical insulation, and high-voltage driver represent a large percentage of the overall manufacturing cost. Moreover, the electric field shielding would attenuate the magnetic field coupling to the implant.

This paper presents a novel low-operating-voltage wireless intermediate-range scheme for energy and signal transmission (WISEST), especially suitable for deep-seated or locomotive implantable devices, using magnetic coupling while traditional transcutaneous transformers can only transmit energy to the implant located around the center of the transmitting coil. The transmitting coil was driven by an ac current of 1.2 A_{rms} at a frequency of 5.7 MHz. The magnetic field intensity around the center of the transmitting coil was uniform, thus the voltage at the receiving coil was not very sensitive to the alignment error between the coils within a few centimeters.

In the analysis of the magnetic field pattern induced by the transmitting coil shown in Fig. 1, the following equations [22] describing the magnetic field intensity (H-field) in the radial, \( H_r \), and axial, \( H_x \), directions will be applied:

\[
H_r = \frac{I}{2\pi r \sqrt{(a+r)^2 + x^2}} \left[ -K(k) + \frac{a^2 + r^2 + x^2}{(a - r)^2 + x^2} E(k) \right] \quad (1)
\]

\[
H_x = \frac{I}{2\pi \sqrt{(a+r)^2 + x^2}} \left[ K(k) + \frac{a^2 - r^2 - x^2}{(a - r)^2 + x^2} E(k) \right] \quad (2)
\]

where \( K(k) \) and \( E(k) \) are the first and the second kinds of complete elliptic integrals, \( k^2 = 4ar/[\{(a + r)^2 + x^2\}] \), \( a \) is the coil radius, \( x \) and \( r \) are the distances from the coil center along the coil’s axial and radial axes, respectively, and \( I \) is the coil current.

Equations (1) and (2) show that the \( H \)-field induced by the current flowing through a circular filament is frequency independent in free space. However, in the actual situation, the coil has a finite thickness and wraps around the conductive human body. These factors will be considered in our magnetic field simulations using finite-element-analysis (FEA) with Ansoft Maxwell 3-D software to investigate the high-frequency effects of the conductivity and permittivity of human tissues on the magnetic energy absorption and energy transfer. At high frequencies, eddy current induced in the conductive human tissue can attenuate the magnetic field intensity. The operating frequency of the energy transmission system in this paper is chosen as 6.1 MHz (to be explained in Section VI). In the simulations, the field intensity is simulated at dc (without the eddy current effect), 6.1 and 10 MHz to investigate the eddy effects on the \( H \)-field attenuation at different frequencies. The conductivity and relative permittivity of some tissues in the human body at 6.1 and 10 MHz are listed in Table I. The parameters are computed on the basis of the 4-Cole-Cole Model described in [23] by using the RF_Tools MATLAB program developed by the Center for NMR Research at Penn State University [24].

Fig. 2 shows the calculated and simulated \( H \)-field of the 30-cm diameter transmitting coil excited by a 1-A current. The transmitting coil with a thickness of 0.2 mm wound around a cylinder (29.8 cm in diameter) filled with a material approximating the conductivity and permittivity of human tissue. According to Table I, because the small intestine has

![Fig. 1. Single transmitting coil carrying current I.](image)
TABLE I
CONDUCTIVITY AND RELATIVE PERMITTIVITY OF SOME TISSUES [23], [24]

<table>
<thead>
<tr>
<th>Tissues</th>
<th>Conductivity (S/m)</th>
<th>Relative Permittivity</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>6.1MHz</td>
<td>10MHz</td>
</tr>
<tr>
<td>Fat (Average Infiltrated)</td>
<td>0.04886</td>
<td>0.05261</td>
</tr>
<tr>
<td>Gall Bladder</td>
<td>0.9011</td>
<td>0.9026</td>
</tr>
<tr>
<td>Gall Bladder Bile</td>
<td>1.401</td>
<td>1.403</td>
</tr>
<tr>
<td>Heart</td>
<td>0.4634</td>
<td>0.5014</td>
</tr>
<tr>
<td>Kidney</td>
<td>0.4552</td>
<td>0.5081</td>
</tr>
<tr>
<td>Liver</td>
<td>0.2875</td>
<td>0.3167</td>
</tr>
<tr>
<td>Deflated Lung</td>
<td>0.4165</td>
<td>0.438</td>
</tr>
<tr>
<td>Muscle (Transverse)</td>
<td>0.598</td>
<td>0.6168</td>
</tr>
<tr>
<td>Small Intestine</td>
<td>1.272</td>
<td>1.345</td>
</tr>
<tr>
<td>Spleen</td>
<td>0.4302</td>
<td>0.5057</td>
</tr>
<tr>
<td>Stomach</td>
<td>0.7492</td>
<td>0.7844</td>
</tr>
</tbody>
</table>

In the Ansoft FEA program setting, the solution type was set as magnetic eddy current, and the percentage error was set to 1%. The simulation volume was bounded by a cylinder with a height of 80 cm and a diameter of 100 cm, and the coils were located at the middle of the cylindrical boundary.

From Fig. 2, the results calculated by (1) and (2) are consistent with the FEA simulation when the eddy current effect is not included. However, as the operating frequency increases, the $H$-field attenuation increases because of the eddy current effect in the body tissue. Maximum $H$-field attenuation occurs at the center of the coil, which is also the middle of the cylinder. The maximum $H$-field attenuation is 1.7% at 6.1 MHz and increases to 8.8% at 10 MHz.

From Fig. 2(a), the field variation along the coil’s radial direction is 5% within a region of 7.5-cm diameter around the coil center. However, outside this flat region, the field intensity increases rapidly by 50% at 9.8 cm and 100% at 11.6 cm from the coil center. Along the coil’s axis, the $H$-field drops 5% at 3 cm, and 25% at 6.8 cm from the coil center as shown in Fig. 2(b).

**III. MULTIPLE TRANSMITTING COILS**

Although energy and signal transmission to an implantable device can be achieved using a single transmitting coil, the position of the implant is limited to a few centimeters from the transmitting coil plane. For locomotive implants, such as capsule endoscope and endoscopic capsule robots [25], that need to move and perform procedures in different positions inside the body, a more even magnetic field pattern generated from the transmitting coil is required to power up such devices.

To generate a desired or evenly distributed magnetic field intensity in a wider applicable region, an additional transmitting coil with a distance from the original coil can be adopted as shown in Fig. 3. For instance, the Helmholtz coil, which comprised a pair of coils with the same diameter, separated by a coil radius, and placed in parallel and co-axially, is a well-known structure that can provide even magnetic field bounded by the coils. Other desired field patterns may be achieved with different coil parameters, such as number of coils, coil diameters, number of turns in each coil, coil separation, and inclination. In this paper, we use a 15-cm Helmholtz-coil-configured transmitting coil ($R_1 = R_2 = 15$ cm,
When \( R = d_1 + d_2 = 15 \text{ cm} \), the calculated \( H \)-field in free space is shown in Fig. 4. Similar to the single coil simulation, the \( H \)-field of the Helmholtz coil was simulated when the coil was wrapped around the conductive cylinder approximating the electrical properties of the small intestine at dc (without the eddy current effect), 6.1 and 10 MHz. The calculated results are also included in Fig. 4 for comparison. The calculated \( H \)-field is consistent with the simulated one when the eddy current effect is not included in the conductive tissue. When the operating frequency increases, the \( H \)-field attenuation increases due to the eddy current effect. The \( H \)-field attenuation at the midpoint between the centers of the coils is 3.2% at 6.1 MHz and increases to 11.5% at 10 MHz.

The two-coil configuration generates a more even \( H \)-field pattern in a wider applicable region than that generated by the single coil configuration. From Fig. 4(a), the variation of the \( H \)-field at 6.1 MHz is about 2.7% within a region of 15 cm diameter at the middle plane between the two coils. From Fig. 4(b), the \( H \)-field variation is 5.8% in the region bounded by the two coils along the coil axis. The simulated \( H \)-field on the \( r-x \) plane at 6.1 MHz is shown in Fig. 5. It shows that the field is even and the applicable area is wide enough for many fixed or locomotive medical implant applications.

IV. METHODS FOR REDUCING THE TRANSMITTING COIL VOLTAGE

In the design of a practical energy transmitter for medical implantable devices, the patient’s safety, overall cost, and power consumption are the prime factors to consider. The most up-to-date method for driving a transmitting coil for providing sufficient power of over 300 mW for next-generation capsule endoscopes requires a several kilovolts power converter [6], [7]. This high voltage not only presents at the output of the power converter, but also presents over the transmitting coil, which is designed to wrap around to the patient body. This high-voltage requirement could be an obstacle for the development of such an energy transmission technology being adopted in patients owing to safety and cost issues. Although electric shielding and high-voltage insulation may be used to reduce the risk caused by the high-voltage coils, the overall cost will be significantly increased and the magnetic energy coupling will be deteriorated.

In our prototype design, the transmitting coil diameter is 30 cm and each coil has three turns. The operating frequency is 6.1 MHz. The separation between the two coils is 15 cm. To obtain a power level of 350 mW from a three-turn 2-cm diameter air-core receiving coil, the required transmitting coil current is 1 A rms, or 6 A-turn of magneto-motive force (mmf). The measured inductance of the transmitting coil is 20 \( \mu \text{H} \). Neglecting the wire resistance of a few ohms, the transmitting coil impedance at 6.1 MHz is \( 2\pi f L = 767 \Omega \), thus the required driving voltage is \( 767 V_{\text{rms}} \) (or 2.17 kV pp). The driving voltage is proportional to the operating frequency, transmitting coil current, and inductance, which is determined by the coil size and the number of turns. Therefore, if the transmitting coil has a larger dimensions or number of turns, operates at a higher frequency, or carries a higher current, the required driving voltage would be increased proportionally. In this paper, we propose to significantly reduce both the required transmitting coil excitation voltage and the voltage over the whole coil by using a distributive resonance technique and dividing the coil into multiple segments.
Fig. 6. Coil divided into four smaller subcoils.

A. Dividing a Coil into Multiple Smaller Subcoils

The required driving voltage can be reduced by dividing the transmitting coil into multiple smaller subcoils. Fig. 6 shows an example of a coil divided into four identical subcoils. Since the currents flowing through each pair of wires that connecting the subcoil circumferences to the junctions are in opposite direction, the magnetic field generated by each current-carrying wire pair can be canceled each other theoretically. Thus, the voltage across these connecting wire pairs, which is mainly caused by the wire resistance rather than its inductance, is negligible compared with the coil segment voltage, \(V_{\text{seg}}\). The required voltage for driving each subcoil approximately equals to the required voltage for driving the whole coil divided by number of subcoils. The driving voltage for the transmitting coil can be further reduced by connecting a capacitor in series to each subcoil and operating the coil-capacitor circuit at its resonant frequency, at which the input impedance and thus the required driving voltage are minimized. The applied voltage is proportional to the resonant circuit’s equivalent series resistance (ESR), which is very small compared with the coil impedance without using the resonance and segmentation techniques, and the coil current.

Although dividing the coil into multiple subcoils can significantly reduce the driving voltage and is feasible for energy and signal transmission, it increases the conduction loss caused by the additional wires for connections. In addition, the tolerances in the dimensions of the subcoils will result in different subcoil inductance values, so each of the subcoil has to be fine tuned with different series capacitors individually. However, although all the subcoils can be fine tuned to the desired resonant frequency, the tolerance of the capacitor ESR, which is hard to control and usually not specified in the datasheet, will cause different subcoil currents, and therefore unbalanced field pattern will occur. A representative example of this unbalanced situation is demonstrated by the simulated magnetic field pattern on the coil plane shown in Fig. 7, in which the subcoil currents are 0.9, 1, and 1.1 A, respectively. As shown in the figure, the magnetic field intensity at a point 5 cm from the coil center in the 1.1-A subcoil is higher than that at the corresponding point in the 0.9-A subcoil by 69%. Apart from the unbalanced field pattern, the simulated field pattern also shows that this configuration generates undesired magnetic field around the connection wire pairs since the magnetic field generated by the unequal opposite currents in the wire pairs cannot be entirely canceled each other. This undesired field pattern may create stability and safety issues in the wireless energy transfer system.

Fig. 7. Simulated \(H\)-field induced by four smaller unbalanced subcoils.

B. Dividing a Coil by Multiple Series Capacitors

To avoid the disadvantages caused by the method mentioned in Section I, we propose a novel method to divide the transmitting coil into multiple segments by connecting low-loss high-frequency capacitors in series to the coil segments. Fig. 8 shows an example of this configuration with four segments in a single-turn coil. Compared with the multiple subcoil
configuration, the benefits of the series capacitor segmentation method are twofold. First, it does not need the additional bundles of wires for the connections, so its conduction loss is lower and it is easier to design and use, especially when the coil is designed to wrap around the patient’s body. Second, because all the coil segments and capacitors are connected in series, even there are tolerances in the arc shape, the series capacitance value, and the capacitor ESR, the currents flowing through all the coil segments are the same, so unbalanced magnetic field can be avoid.

The proposed method is different from the traditional designs [6], [7], in which the transmitting coil is driven by a resonant converter or a series capacitor is connected to the coil to form an LC resonant tank. In these traditional designs, although the system input voltage, that is, the dc supply voltage of the resonant converter or the driving voltage of the LC tank, can be very low, the excitation voltage and the voltage over the transmitting coil is still very high (several kilovolts). In the proposed capacitor-segmented coil, the segment voltage, which represents the maximum voltage over the whole coil, equals to the required voltage for the unsegmented coil divided by the number of segments. The coil segment voltage can be reduced proportionally by increasing the number of series capacitors and segments. When operating the circuit at the resonant frequency, if the series resistance is ignored, the voltage across each capacitor equals to the coil segment voltage, $V_{\text{seg}}$, but in opposite polarity, as shown in Fig. 8. The input impedance equals to the ESR of the resonant circuit, which is much smaller than the impedance of the coil without the series capacitors. Therefore, for a given transmitting coil current, the proposed segmentation and resonance techniques can significantly reduce both the required driving voltage and the voltage over the whole coil.

The transmitting coil in our design consists of two 3-turn coils connected in series. The coil diameter is 30 cm and the coil separation is 15 cm. The inductance of the transmitting coil is measured with an HP4194A Impedance Gain Phase Analyzer (Hewlett Packard, Santa Clara, CA) at 6 MHz. In our power transmission experiments, we divided the coil into 24 segments, that is, each turn is divided into four segments as shown in Fig. 9. With this configuration, the segment voltage ($32 \, \text{V}_\text{rms}$) is 24 times less than the voltage ($767 \, \text{V}_\text{rms}$) of the whole coil without using the segmentation technique. The required driving voltage is also significantly reduced by operating the coil at the resonant frequency. The operating frequency for the energy transmission is selected at around 6 MHz in our experiments [15]. The resonant frequency of the transmission network is given by

$$f = \frac{1}{2\pi \sqrt{L_{\text{TX}} C_{\text{TX}}}} \quad (3)$$

where $L_{\text{TX}}$ is the transmitting coil inductance and $C_{\text{TX}}$ is the required resultant capacitance of the series capacitors connecting to the coil segments.

The closest capacitance value available that achieves a resonant frequency at around 6 MHz is 820 pF. The resultant series capacitance is 820 pF/24 = 34.17 pF and the calculated resonant frequency is 6.088 MHz. The impedance of the transmitting coil network versus frequency was measured, calculated and is shown in Fig. 10. The calculated impedance is consistent with the measurement. The impedance is about 5 $\Omega$ at the resonant frequency. It is noted that the resonant frequency can be fine tuned to a desired value by connecting capacitors in parallel or series to any of the series capacitors though it may be more practical to put a parallel capacitor to one close to the coil input. Since the input impedance around the resonant frequency is also low, low-driving-voltage can be achieved by operating the transmitting coil around the resonant frequency.

V. ENERGY AND SIGNAL-RECEIVING CIRCUITS

A. Energy-Receiving Circuit

An air-core receiving coil with a diameter of 2 cm is used to convert the magnetic energy generated by the transmitting coil current to electrical energy to operate the potential implantable devices. The advantages of using an air-core receiving coil includes low cost, very high-power density, no frequency limitation due to magnetic cores, no magnetic loss, and ease of manufacturing [26], [27]. The receiving coil has three turns and is made of 26 AWG single-strand enameled wire. A resonance technique is used in the receiving circuit to
increase the energy receiving capability. Here, the three-turn receiving coil is used to demonstrate the proposed coil segmentation method and it has not been optimized. Its dimensions and number of turns should be tailored to fit the actual situation. In addition, more advanced tuning and control can be used to optimize the stability and maximize the power-receiving capability. A schematic of the power-receiving circuit is shown in Fig. 11. The functions of the capacitors $C_1$ and $C_2$ are twofold. First, they determine the resonant frequency of the receiving circuit which is given by

$$f_r = \frac{1}{2\pi \sqrt{L_r C_r}}$$

where $L_r$ is the inductance of the receiving coil and $C_r = (C_1 \cdot C_2)/(C_1 + C_2)$. Second, because they are connected in a potential divider configuration, the desired output voltage level can be chosen by an appropriate ratio of $C_1$ and $C_2$. The voltage across $C_2$ is given by

$$V_o = \frac{C_1}{C_1 + C_2} V_r$$

where $V_r$ is voltage across the receiving coil. High-frequency low-loss capacitors are used in the resonant network. In this paper, we select $C_1 = 2.18 \text{ nF}$ and $C_2 = 7.5 \text{ nF}$ to achieve a resonant frequency of 6.1 MHz and a voltage of about 6.6 V dc at the output of the rectifier connected in the Greinacher voltage doubler configuration. Schottky diodes are used in the rectifier to reduce the power dissipation due to the forward-voltage drop. It is noted that a half-wave rectifier, which is made of only one diode, cannot be used in this configuration since it loads the resonant capacitors with unbalanced current, that is, conducts in the positive half-cycle only. The net current flow of the resonant capacitors must be zero in each cycle to make the circuit operates properly. Thus, the presented rectifier or a full-wave rectifier, which conducts in both the positive and negative cycles, should be used.

### B. Power Measurement

The output power delivered from the receiving circuit was measured when the transmitting coil was driven by a current of 1 A using a radio frequency (RF) power amplifier (240L, Electronics & Innovation, Rochester, NY). The circuit and mechanical parameters in the testing setup are summarized in Table II. The operating frequency was 6.1 MHz and the measured transmitting coil voltage was 6.68 V rms. The transmitting coil voltage and current waveforms were measured and are shown in Fig. 12. The waveforms were captured with an oscilloscope (Tektronix, DPO 3034, Beaverton, OR) and the current was sensed by a current probe (Tektronix, TCP 312) with an amplifier (Tektronix, TCPA 300). The dc output power was measured with different load resistance and receiving coil locations.

<table>
<thead>
<tr>
<th>TABLE II</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>PARAMETERS OF THE TESTING SETUP</strong></td>
</tr>
<tr>
<td>Transmitting Carbon Dioxide</td>
</tr>
<tr>
<td>Wire Diameter</td>
</tr>
<tr>
<td>Coil Separation</td>
</tr>
<tr>
<td>Number of Turns per Coil</td>
</tr>
<tr>
<td>Number of Segments per Turn</td>
</tr>
<tr>
<td>Inductance</td>
</tr>
<tr>
<td>Series Capacitors</td>
</tr>
</tbody>
</table>

| Receiving Carbon Dioxide | |
| --- |
| Number of Turns | 3 |
| Wire Diameter | 0.405 mm |
| Inductance | 326 nH |

| Receiver Carbon Dioxide | 2.18 nF |
| --- |
| Resonant Capacitor C1 | 2.18 nF |
| Resonant Capacitor C2 | 7.5 nF |
| Rectifier Diode | NSR0340HT1G |

![Fig. 11. Schematic of the power- and signal-receiving circuit.](image1)

![Fig. 12. Measured transmitting coil voltage and current.](image2)

![Fig. 13. Measured receiving circuit output power versus load resistance when the transmitting coil current is 1 A and the receiving coil is located at the midpoint between the two transmitting coils’ centers and 1 cm from the center of one of the transmitting coils.](image3)
Fig. 14. Measured receiving circuit output power versus transmitting coil current when the receiving coil is located at the midpoint between the centers of the two transmitting coils.

Fig. 15. Measured modulated transmitting coil current and demodulated signal at the receiver output.

Fig. 13 shows the measured output power when the receiving coil is placed at the midpoint between the centers of the two transmitting coils and 1 cm axially from the center of a transmitting coil, respectively. From the measured results, the energy level at the midpoint is higher than that at 1 cm axially from the center of a transmitting coil. This variation can be predicted by the theoretical magnetic field plot in Fig. 4(b). The maximum output power delivered from the receiving circuit is about 350 mW. The output power level can be increased by increasing the transmitting coil current. Fig. 14 shows the maximum output power from the receiving circuit versus transmitting coil current. The output power is about 3 W when the transmitting coil current is increased to 3.1 A.

C. Signal Transmission

Signal transmission using the proposed transmitting coil and air-core receiving coil is demonstrated with on-off keying (OOK). The signal to be transmitted modulates the 6.1 MHz carrier frequency as shown in Fig. 15. The frequency of the modulation signal in this demonstration is 10 kHz. The demodulation circuit shown in Fig. 11 is composed of a diode-resistor-capacitor envelope detector and a buffer. The recovered waveform from the output of the demodulator is shown in Fig. 15. In this paper, we demonstrate the signal transmission using OOK, but other modulation techniques, such as frequency modulation, can be adopted depending upon the needs of the application.

VI. DISCUSSION

A. Transmitting Coil Configurations and Positions

This paper demonstrates a novel low-operating-voltage technique for transmitting energy and signal to medical implantable devices deep-seated in the body by magnetic coupling. The primary advantages of the proposed method are to significantly reduce the transmitting coil excitation voltage to less than 7 V_{rms}, and the voltage over the coil to less than 32 V_{rms} while the voltage over the coil using existing methods require several kilovolts [6], [7], and therefore improve safety, reduce overall manufacturing costs, and power consumption. Since the voltage over the whole coil and the potential difference between windings are substantially reduced, the dielectric loss caused by the displacement current flowing between windings through the coil former materials is naturally eliminated. Besides, since the magnetic field pattern generated by the transmitting coils is uniform and applicable in a wide region, this method can deliver stable power to deep-seated implantable or even locomotive devices.

In this paper, we used a Helmholtz-coil configured transmitting coil to demonstrate the low-operating-voltage coil segmentation method but the coil configuration has not been optimized for power transfer. First, in the testing setup, the transmitting coil was tightly wound, the separation between windings can be increased to reduce the ac resistance caused by proximity effect. Second, the coils’ geometry and number of turns can be varied to obtain the required magnetic field pattern and fit the patient’s body. For example, a more even field pattern can be obtained by adding one more coil with a larger diameter at the middle between the two coils, which is known as the Maxwell coil [28]. Third, to transmit maximum power to the implant in any orientation, the transmitting coils can be in the following configurations: 1) both coils wrapping around the patient’s body; 2) one in front and one on the back of the patient; 3) one on the left and one on the right of the patient; or 4) any combination of 1) to 3).

The wireless energy transmission scheme can be used to power up locomotive devices, such as capsule endoscopes. In these applications, a navigation system can be used to trace the implant position so that the position of the transmitting coil can be adjusted accordingly and the implantable device can be always around the midpoint between the centers of the transmitting coils to achieve maximum power transmission.

B. Transmitting Coil Excitation

The desired resonant frequency of the transmission network can be chosen flexibly by using appropriate capacitors in series to the transmitting coil segments. It was reported that [26] the energy transmission capability between air-core coils increases as frequency increases. On the other hand, the H-field attenuation by body tissues increases as the operating frequency increases. The FEA simulated results in Fig. 4 show that the attenuation increases from 3.2% to 11.5% when the
magnetic field frequency increases from 6.1 MHz to 10 MHz. In our demonstration, we select an operating frequency of 6.1 MHz, but it can be increased and optimized to maximize the energy transmission capability without causing excessive tissue energy absorption.

In our power transmission measurement, an mmf of 6 A-turns is required to deliver 350 mW of power at the receiver output with a three-turn and 2-cm receiving coil. The output power can be increased by increasing the mmf, that is, by increasing the current or number of turns of the transmitting coil, or by optimizing the dimensions and number of turns of the receiving coil. In this paper, both the transmitting and receiving coils are made of 26 AWG single-strand enameled wires. The energy efficiency can be improved using thicker wires or using bundles of wires to reduce the winding resistance. The efficiency can also be increased using a larger receiving coil when space is allowed.

The required maximum transmitting coil voltage in the proposed method is noticeably lower than that in the most recent studies [6], [7]. This means that the proposed method has a significantly lower specific absorption rate (SAR) in body tissue due to the electric field ($E$-field) emitted from the transmitting coils. If necessary, the $E$-field can be further reduced by increasing the number of segments and capacitors because the voltage division ratio is proportional to number of segments. Although using identical series capacitors evenly distributing around the transmitting coil are not necessarily, this configuration can minimize the voltage, and thus the $E$-field, at any point over the coil.

C. Energy-Receiving Circuit

Sophisticated and multifunctional implantable devices often require power supplies with multiple output voltage levels. Multiple linear voltage regulators may be used to deliver different voltage levels, but this configuration is not energy efficient, especially when the voltage difference between the regulator’s input and output is large. Energy-saving passive potential dividers constructed with low-loss high-frequency capacitors, $C_1$ and $C_2$, as shown in Fig. 11, can be used to step down the receiving coil voltage to voltage levels just higher than, for example, 0.5 V, the desired voltage. The desired voltage levels can be obtained by connecting low-drop-out (LDO) voltage regulators at the outputs of the rectifiers. If the implant requires a voltage level higher than the receiving coil voltage, a multistage low-loss diode-capacitor-based voltage multiplier proposed in [15] can be used to step up the coil voltage to the desired voltage level.

It is expected that the proposed method can provide sufficient power to many kinds of implants directly without embedded batteries. However, if the implanted device requires a higher current in a certain time interval, a temporary energy storage device, such as a supercapacitor or a miniaturized battery, can be used to store the energy and then discharge to the load at the desired current level. Because the $H$-field generated by the transmitting coil is even and applicable in a wide region, this method allows multiple implantable devices at different locations operating at the same time. The dimensions and number of turns of the 2-cm diameter air-core receiving coil used in our demonstration could be further optimized. The receiver output power depends on the coil dimensions and number of turns. Multistrand enameled wires (litz wires) can be used to reduce the ac resistance, and thus power loss, due to the skin effect at high frequency. The receiving coil can also be in other shapes to fit into the implanted device. Apart from capsule endoscopies, it is envisioned that the proposed method can deliver sufficient energy to operate other implantable devices deep in the body without embedded battery, such as implantable blood flow monitoring devices [29], [30], drug delivery systems [1], spinal cord stimulators [3], [4], middle ear hearing devices [5], and neuromuscular stimulator [31]. In these applications, the proposed method can transfer energy deep in the body and precise alignment is not required. Besides, it can further reduce the manufacturing cost by eliminating the need of the implantable medical grade battery and the cost of the replacement surgeries every several years due to the limited life of the battery.

VII. Conclusion

A low-operating-voltage WISEST for medical implantable devices is proposed. The operating principle of the coil segmentation and distributive resonance techniques for substantially reducing both the required transmitting coil excitation voltage and the voltage over the coil were demonstrated with experiments. With the proposed coil segmentation method, the excitation voltage for the transmitting coil is less than 7 $V_{rms}$ and the voltage over the coil is less than 32 $V_{rms}$, while the most up-to-date method requires a several kilovolts power converter to drive an unsegmented transmitting. We demonstrated that the proposed configuration can deliver an output power of 350 mW from a receiving circuit composed of a 2-cm diameter receiving coil, a simple resonant network, and a rectifier. The proposed method can significantly reduce the overall manufacturing cost, relieve safety concerns, and increase the power transmission capability. Since the receiving coil does not contain ferromagnetic material, MRI-compatible implants can be achieved. The proposed scheme generates evenly distributed magnetic field in a wide region bounded by the transmitting coils; it is especially suitable for deep-seated and locomotive implantable devices, such as capsule endoscopes and endoscopic capsule robots, without using the additional intermediate resonant coils that required in wireless power transfer system via strongly coupled magnetic resonance. The proposed low-operating-voltage coil segmentation method can also be used to reduce the extremely high voltage in the resonant coils in the applications that the coils are required.

REFERENCES

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