Conformal Device for Wireless Powering in Biomedical Application

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***Abstract*—The wireless powering efficiency of the Strongly Coupled Magnetic Resonance (SCMR) method has been reported for both homogenous and non-homogenous interfaces. Here, a new Conformal (i.e., planar) SCMR (CSCMR) method is presented that achieves high efficiency and uses significantly smaller volume than conventional SCMR. Also, the performance of traditional resonant inductive coupling, SCMR and CSCMR are compared for wireless powering of devices that are implanted in the human body.**

I. INTRODUCTION

Implantable medical devices(IMDs) play an important role in diagnosis and treatment of disease due to their capability to monitor, stimulate and regulate vital internal organs. Also, certain IMDs can communicate report data to an external host. IMDs have found applications in a wide range of areas, including pacemakers, physiological monitoring devices, pain relief devices, cochlear hearing implants, functional electrical simulators (FES), left ventricular assist devices (LVAD) [1], artificial hearts, bladder-pressure monitoring devices [2] and neuro-stimulators [3].

The power required by IMDs depends on the application and the typical range is from a few microwatts [4] to hundreds of milliwatts [4]. Therefore, such restrictions require that wireless powering systems for IMDs must exhibit high power transmission efficiency in order to transfer substantial amount of power without using excessive transmitting power that will generate high intensity EM fields on the human body [5] and increase SAR.Resonant inductive coupling and SCMR are currently the two most popular wireless powering techniques for IMDs. Resonant inductive coupling involves near-field wireless transfer of electrical energy between two devices that are tuned to resonate at the same frequency [6]. On the other hand, SCMR employs resonators to transmit power wirelessly and efficiently over mid-range distances [7]. SCMR has also been applied to in-vitro and in- vivo experiments [8], [5].

II. CSCMR FOR BIOMEDICAL IMPLANTS

In this section, a novel design of a CSCMR system suitable for wireless powering of biomedical devices is presented. The performance of the proposed CSCMR system is compared to the performance of traditional resonant inductive coupling and SCMR. Here, all three wireless powering systems are designed based on self-resonating bifilar spirals, as shown inFig. 1. Self-resonating TX and RX elements (i.e., helices and spirals) are preferred for SCMR systems because external capacitors lower the Q-factor and in-turn the efficiency of SCMR systems. In fact, helices and spirals have been used to design and power SCMR systems for IMDs [7], [10]. Bifilar resonators are used here because they can be excited at their center thereby facilitating a direct comparison of the performance of the three wireless powering systems shown in Fig. 1. The basic geometrical parameters of a bifilar spiral are *N, W, S, T* and d_{out}, which are the number of turns, cross-sectional width, spacing between turns, thickness of the trace material, and the outermost side length of the spiral, respectively (see Fig. 2).Next, the three systems are briefly described and compared.bifilar spirals) that are fed at their center. For both methods the source loop is connected to the power source and the load loop is connected to a 50 Ohm load. All systems are designed to operate at 39 MHz.

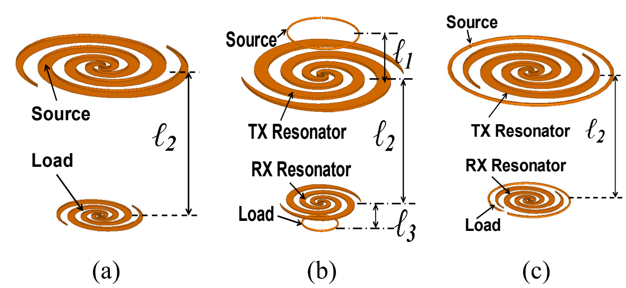


Fig. 1.Models of three wireless powering systems. (a) resonant inductive coupling, (b) SCMR, and (c)conformal SCMR (CSCMR).

The three systems of Fig. 1 use the same two bifilar spirals, one in air (external) and one in tissue (implanted). The implanted spirals are embedded at a depth of 2.6 cm inside muscle tissue, as shown in Fig. 3(a). The geometrical parameters of the external TX bifilar spiral are as follows:d_{out} = 98 mm, N = 8, T =0.5 mm,W =1.52 mm,and$S$ =1.52 mm. The geometrical parameters of the implantedRX bifilar spiral are as follows: d_{out} = 30 mm, N = 9, T = 0.3 mm, W = 0.48 mm,and$S$ = 0.48 mm. The distances in Fig. 5are as follows:ℓ_1 = 5mm, ℓ_2 = 50mm andℓ_3 = 4 mm. The SCMR system shown in Fig. 2(b) also uses: (a) a source loop with radius of 30mm and cross-sectional of 1 mm, and (b) a load loop with radius of 15mm and cross-sectional of 0.5 mm. The CSCMR system shown in Fig. 2(c) also uses: (a) a source loop with radius of 55mm and cross-sectional of 1 mm, and (b) a load loop with radius of 17mm and cross-sectional of 0.1 mm. All elements in Fig. 4 are assumed to be made from copper (i.e., σ= 5.8 x10^7S/m). In order to achieve miniaturization of the SCMR and CSCMR implanted RX resonator bifilar spirals, were embedded in a dielectric with permittivity value of 10.2.

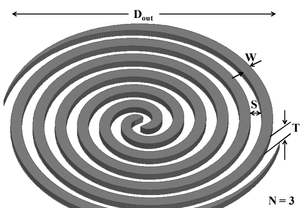
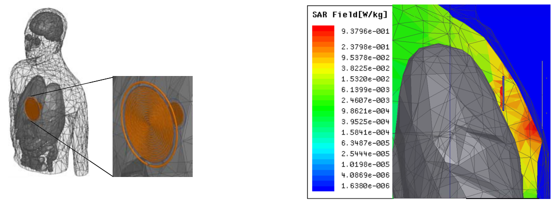


Fig. 2. Bifilar spiral geometry.

The performance of the three wireless powering systems of Fig. 1 is compared through simulations. All simulations are performed using Ansoft HFSS and the top part of a 3D human body model in which the properties of the various tissue types are specified. For example, Fig. 3(a) shows the part of the human phantom included in our simulations with the CSCMR system of Fig. 1(c).First, the conformal SCMR model is simulated and optimized in Ansoft HFSS and Nexxim with a 1W input power. An output power of 385 mW was achieved at $\ell 2$ = 50 mm. In order to compare the three techniques for the same power delivered to the load, the input power of the resonant inductive coupling and the SCMR systems is adjusted so that an output power of 385 mW is delivered to the load. The performance of the three methods is compared in Table I, where the input power, maximum H-field intensity, maximum SAR, and wireless powering efficiency are reported. As expected, it is clearly seen that SCMR and CSCMR significantly outperform resonant inductive coupling. Specifically, SCMR and CSCMR achieved an efficiency of 36.9% and 38.5%, respectively, versus resonant inductive coupling that achieved an efficiency of only 0.46%.

Therefore, the resonant inductive coupling must utilize significantly greater amount of input power (i.e., 97 Watts) to achieve the same output power with SCMR and CSCMR. hence, SCMR and CSCMR exhibit substantially lower levels of SAR(approximately 35 times lower maximum SAR) and magnetic field intensity, and safer methods for wireless powering of implanted or wearable devices.TheSAR distribution for the CSCMR model is shown in Fig. 3(b). In addition to achieving large efficiency and reducing SAR levels, CSCMR exhibits a compact size because it uses planar transmitter and receiver configurations thereby minimizing the volume required. Minimization of volume is particularly important for IMDs and biomedical applications. Therefore, CSCMR is very well suited for wireless powering of wearable and implantable devices. A comparison of the dimensions, volume and height of reported SCMR systems for biomedical applications is shown in Table II.



(a) (b)

Figure. 3. CSCMR in the human body. (a) Geometry the human body, and (b) SAR distribution .

TABLE I. COMPARISON OF DIFFERENT WIRELESS POWERING SYSTEMS FOR OUTPUT POWER OF 385 MWATTS..

Coupling Type	Field Parameters and Efficiency			
	Input Power(W)	*Max H-field (A/m)*	*Max SAR (W/kg)*	*Efficiency (%)*
Resonant Coupling	97	50	35	0.46
SCMR	1.3	5.6	0.98	36.9
CSCMR	1	3.4	0.91	38.5

TABLE II. COMPARISON OF DIFFERENT SCMR SYSTESFOR IMD APPLICATIONS.

Papers	*Dimension (r, H) (mm)*	*Volume (mm)3*	*Volum e (mm)*	*Freq. of operation (MHz)*
[8]	(85,50)	1,134,900	50	6.9
[7]	(11, 2.5)	950.3	10	0.7
SCMR	(15, 4)	2,827.4	4	39
CSCMR	(15, 0.35)	247.4	0.35	39

III. CONCLUSION

This paper introduced a new conformal SCMR method, that achieves high efficiency and minimizes the volume of wireless powering systems. Itoutperformed resonant inductive coupling and performed similarly to SCMR in terms of the SAR levels for the same power delivered. It is also planar and more suitable for planar and flexible implantable or wearable.

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