Optimal Design of Megahertz Wireless Power Transfer Systems for Biomedical Implants

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***Abstract*—Wireless power transfer (WPT) working at megahertz (MHz) is widely considered a promising technology for the mid-range transfer of low power. In the biomedical implantable WPT systems, the receiving coil is small. Meanwhile, in real applications, the required transfer distance is large. Thus, the coupling coefficient k is low. For the applications of large load, the low coupling coefficient k and large load R_L deteriorate the system efficiency largely. This paper proposes a optimal design method of MHz WPT systems for biomedical implants. A capacitive L-matching network is inserted in the conventional MHz Class E^2 WPT system to enlarge the reflected impedance of the receiving coil on the transmitting side, i.e., improve the power transfer capability and efficiency of the coupling coils. Then the input impedance of the matching network and efficiency of the proposed MHz WPT system are derived and serves as the basis of the proposed parameter design procedure. Based on the circuit improvement and analytical derivations, a numerical optimization design method is proposed to optimize the design parameters of the MHz WPT system. The final experiment verifies the feasibility of the design procedure. With loosely coupled coils (coupling coefficient k=0.035, distance of the coupling coils=1.5 cm, diameter of receiving coil=1.5 cm), the system efficiency can achieve 36.43% under a 0.5 W power transfer.**

***Keywords*—Megahertz wireless power transfer, system efficiency, matching network, optimal design.**

I. INTRODUCTION

Wireless power transfer (WPT) systems provide a convenient non-contacting charging method for devices that require electrical power, and there has been a growing demand for wireless charging in recent years. The magnetic resonance coupling working at megahertz (MHz) is being widely considered a promising technology for the mid-range transfer and low-power applications [1], [2]. It is because generally a higher operating frequency (such as 6.78 and 13.56 MHz) is desirable for a more compact and lighter WPT system with a longer transfer distance. Lots of research has been done on the design and optimization of WPT systems both at component and system levels, including the improvements on coupling coils [3]–[7], and power amplifier (PA) [8]–[11].

It is known that the soft-switching-based PAs are promising candidates to build high-efficiency MHz WPT systems, such as the Class E PA. The Class E PA was first introduced for high-frequency applications in [12]. It has been applied in MHz WPT systems thanks to its high efficiency and simple structure [8], [9], [13]. The Class E rectifier was first proposed for high-frequency DC-DC converter applications in 1988 [14]. Various Class E topologies were later developed, such as voltage-driven, current-driven, and full-wave ones.

Implantable biomedical devices, especially cardiac pacemakers and nerve stimulators, are playing more and more significant roles in curing many kinds of diseases [15]. However, energy depletion in the battery of the biomedical implantable devices eventually forces the patient to accept reimplantation of the device, adding extra ordeal to the patient and increasing the risk of surgery failure. Accordingly, an urgent solution is in vitro wireless energy supply that provides uninterrupted power supply for implantable devices. It has attracted the attention of professionals in medical and engineering fields [16].

However, the size of implants is much smaller compared with non-implantable devices. Therefore, the size of the receiving coil should be small, leading to the very weak coupling between transmitting and receiving coils and eventually the low system efficiency. What's more, in real applications for medical implants, the required transfer distance is large (usually more than 1 cm). In that case, the very small size and relative large transfer distance are the common challenges existing in the wireless power transfer system for biomedical implants.

In this paper, a capacitive L-matching network is added between the rectifier and the receiving coil. Basically, the newly added capacitive matching network will not enlarge the system size and involve more power loss. Based on the improved circuits, a system level design methodology is proposed to optimize the efficiency and power transfer of the WPT systems in the biomedical implant applications. For applications in real life, the changes in the distance and misalignment of the coupling coils are very common. The simulation and experiment results of system efficiencies under different distances and misalignments are also presented to demonstrate the change tendency of system efficiency under different k. This paper is organized as follows. Section II uses conventional design methodology and presents the system performance by Advanced Design System Software. Section III is the presentation of optimal design of biomedical implantable WPT systems using genetic algorithm. Section IV validates the results using simulation and experiments. Finally, section V draws the conclusions.

II. CONVENTIONAL DESIGN

Fig. 1 shows the circuit model of a typical MHz WPT system consisting of a Class E PA, coupling coils and a Class E rectifier. L_f is a RF (radio frequency) choke. C_S and C_0 are the shunt and series capacitors of the Class E PA. The coupling coils consist of the transmitting coil L_{tx} and the receiving coil L_{rx}. r_{tx} and r_{rx} are ESRs of L_{tx} and L_{rx}. C_{tx} and C_{rx} are the compensation capacitors. The Class E rectifier consists of a diode D_r, a parallel capacitor C_r, a filter capacitor C_f, and a filter inductor L_r. Here R_L is the final dc load. Z_{in} is the input impedance of the coupling coils and Z_m is the impedance seen by the receiving coil. V_{pa} is the input voltage of the WPT system.

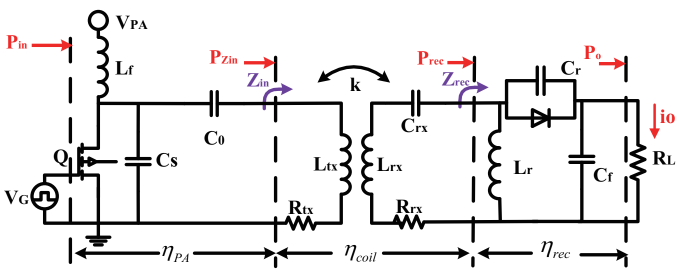

Fig. 1. Circuit model of the conventional MHz WPT system

In the conventional design, the compensation capacitors C_{rx} is determined to be exactly resonant with the receiving coil. Based on the input impedance of the coupling coils, Z_{in}, the shunt and series capacitors of the Class E PA, C_S and C_0, are designed to achieve the ZVS (zero-voltage switching) operation to maximize the PA efficiency as [17]

$$C_\mathrm{S} = \frac{0.1836}{\omega R_{Z_{in}}}, \tag{1}$$

$$X_0 = 1.1525 R_{Z_{in}}, \tag{2}$$

where X_0 is the pure reactance of C_0 and L_{tx}.

Based on the system configuration given in Fig. 1 and the constant parameters listed in Table I, the design parameters of the MHz WPT system using the conventional design are calculated as shown in Table II. Here, the coupling coefficient k=0.035. This k is used as the nominal coupling coefficient for both the conventional and design methodology. Note, in the real wireless charging applications of wearable and implanted devices, the receiving coil should have a small size and then a small self-inductance.

By using Advanced Design System, an electronic design automation software for RF, microwave, and high speed digital applications, the simulations on the conventional system are carried out and the results are given in Fig. 2. It can be seen that the system efficiency is quite low and the conventional design can not achieve the required power transfer for charging wearable and implanted devices when k varies from 0.007 to 0.089 that corresponds to the varying distance from 3.5 cm to 0.167 cm, respectively.

TABLE I
CONSTANT PARAMETERS IN THE SYSTEM

Parameters	Value	Parameters	Value
L_f	60 uH	r_{rx}	0.4 Ω
r_{L_f}	0.2 Ω	L_r	2.2 uH
L_{tx}	1.6 uH	r_{L_r}	0.1 Ω
C_{tx}	344.4 pF	r_{D_r}	0.3 Ω
r_{tx}	0.45 Ω	C_f	10 uF
L_{rx}	456 nH	R_L	50 Ω

TABLE II
CALCULATED RESULTS OF PARAMETERS

C_S	C_0	C_{rx}
8886.31 pF	347.25 pF	1208.41 pF

Fig. 3 gives the simulation results of the reflected resistance of the receiver, R_r. It can be seen that R_r is very small, especially at small k. It will lead to the high power loss on the ESR of the transmitting coil and then result in a very low system efficiency. Meanwhile, the voltage over the battery load is constant (usually 5 or 3.3 V). For the applications of low power (below 1 W), the load is large. Then the reflected impedance of the receiving coil on the transmitting side is small, which enlarges the power loss of the transmitting side, i.e., worsens the system efficiency.

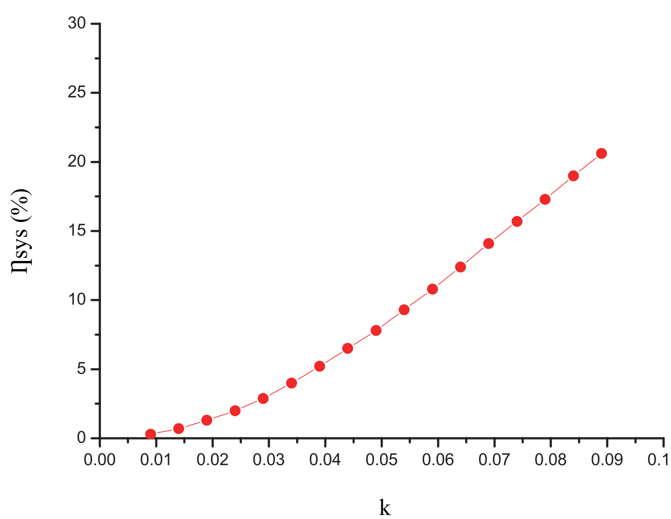

Fig. 2. Simulation results of system efficiencies under different coupling coefficients

As shown in Fig. 2, the simulation result of system efficiency under the nominal k is only 4.2%. As it is too small, it is unnecessary to do experiments.

Suppose the voltage over the battery load is 5 V. Therefore, the simulation and experiment results in this paper are obtained

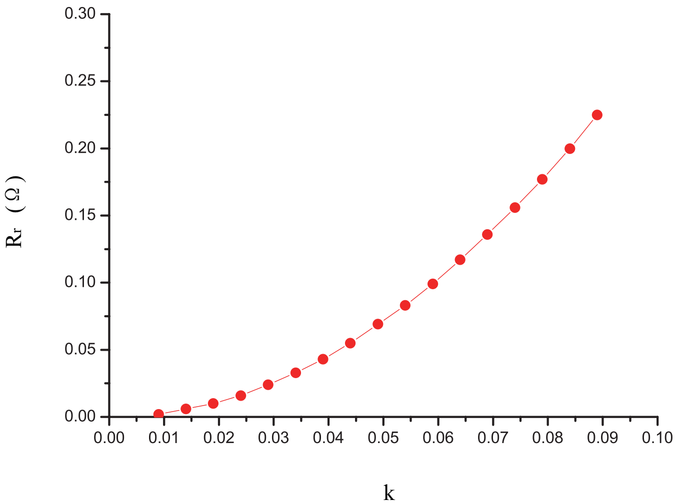

Fig. 3. Simulation results of the reflected resistance under different coupling coefficients

by keeping the current through the load constant, i.e., i_o equals 0.1 A [refer to Fig.1].

III. DESIGN METHODOLOGY

A. Proposed WPT System

Based on the aforementioned analysis, the system efficiency of the conventional design is low due to the small reflected impedance R_r. For applications that involve large R_L, the input impedance Z_{in} is small. Meanwhile, as the size of the receiving coil is restricted by the small size of biomedical implantable device, the coupling coefficient k and the self-inductance of receiving coil are small, which worsens the input impedance Z_{in}. Consequently, the power loss on the transmitting side is large and the efficiency of coupling coils will be small, which is a common problem for biomedical implantable WPT systems. It is known that, in series-series compensation WPT systems, R_r is inversely proportional to the impedance R_m. Here R_m is the real part of Z_m. In this case, decreasing R_m will improve the resistance R_r, finally leading to a lower power loss on the transmitting side. Meanwhile, the smaller R_m will adversely lead to a higher power loss on the ESR of the receiving coil. Thus, an L-matching network is added into the conventional system to transform R_m to an appropriate value to reduce the power loss of the coupling coils.

The system configuration is shown in Fig. 4. Here the Class E rectifier is equivalent to a series connected resistance and reactance, R_{rec} and X_{rec}. Since the matching network consists of two capacitors, basically it will not increase the size and power loss of the receiver. Based on the improved circuit of the WPT system, an optimization design procedure is proposed to optimize the capacitors of the matching network, receiving coil and Class E PA, C_{mns}, C_{mnp}, C_{rx}, C_S, C_0. In order to formulate the optimization problem, the system efficiency of the improved WPT system is analytically derived as follows.

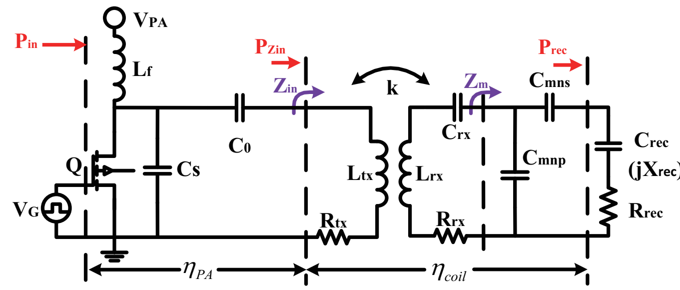

Fig. 4. The circuit model of designed topology

Based on the circuit configuration given in Fig. 4, the system efficiency can be expressed as

$$\eta_{sys} = \eta_{pa} * \eta_{coil} * \eta_{rec}. \tag{3}$$

where η_{pa}, η_{coil}, and η_{rec} are the efficiencies of the Class E PA, the coupling coils, and the rectifier. In this paper, the rectifier efficiency is assumed to be constant due to the fixed output power and dc load. Then, the efficiencies of PA and coupling coils are defined as follows

$$\eta_{pa} = \frac{P_{Zin}}{P_{in}}, \eta_{coil} = \frac{P_{rec}}{P_{Zin}}, \tag{4}$$

where P_{in} is the input power of the PA; $P_{Z_{in}}$ is the input power of the coupling coils; P_{rec} is the output power of the matching network. Based on the circuit model, the input impedance of matching network Z_m can be calculated as

$$Z_{\mathrm{m}} = R_m + jX_m, \tag{5}$$

where

$$R_m = \frac{Z_{rec}}{\omega^2 {C_{mnp}}^2 {Z_{rec}}^2 + (1 + \frac{C_{mnp}}{C_{mns}})^2}, \tag{6}$$

$$X_m = -\frac{\omega}{C_{mnp}} * \frac{{Z_{rec}}^2 + \frac{C_{mnp}+C_{mns}}{\omega^2 {C_{mns}}^2 C_{mnp}}}{\omega^2 {Z_{rec}}^2 + (\frac{1}{C_{mnp}} + \frac{1}{C_{mns}})^2}. \tag{7}$$

Then the reflected impedance of the receiving coil, Z_r, can be derived as

$$Z_r{=}R_r + jX_r, \tag{8}$$

where

$$R_r = \frac{\omega L_m (Z_m + r_{rx})[\omega(L_m - L_{rx}) + \frac{1}{\omega C_{rx}} + Z_m + r_{rx}]}{(Z_m + r_{rx})^2 {+} (\omega L_{rx} - \frac{1}{\omega C_{rx}})^2}, \tag{9}$$

$$\begin{aligned} X_r = \quad & -\omega L_m - \frac{\omega^2 L_m (\omega L_{rx} - \frac{1}{\omega C_{rx}})(L_m - L_{rx})}{(Z_m + r_{rx})^2 + (\omega L_{rx} - \frac{1}{\omega C_{rx}})^2} \\ - \quad & \omega L_m \frac{(\omega L_{rx} - \frac{1}{\omega C_{rx}})[\frac{1}{\omega C_{rx}} + Z_m + r_{rx}]}{(Z_m + r_{rx})^2 + (\omega L_{rx} - \frac{1}{\omega C_{rx}})^2}. \end{aligned} \tag{10}$$

Here L_m is the mutual inductance of the coupling coils. It can be seen from (9), the reflected resistance R_r is determined by the impedance seen by the receiving coil Z_m, and then determined by the combination of the matching network and the input impedance of the rectifier. Based on the derived reflected impedance Z_r, the input impedance of the coupling coils, Z_{in}, can be easily derived as

$$\begin{aligned} Z_{\text{in}} &= Z_r + r_{tx} = \\ & r_{tx} + j[-1/(\omega C_{tx}) + \omega(L_{tx} - L_m)] + \\ j\omega L_m & \frac{j[\omega(L_{rx} - L_m) - 1/(\omega C_{rx})] + Z_m + R_{rx}}{Z_m + R_{rx} + j[\omega L_{rx} - 1/(\omega C_{rx})]}. \end{aligned} \tag{11}$$

The efficiency of coupling coils can be further derived as

$$\eta_{coil} = \frac{\omega^2 {L_m}^2 R_m}{\omega^2 {L_m}^2 (R_m + r_{rx}) + r_{tx} f}, \tag{12}$$

where R_m is the real part of the input impedance Z_m and

$$f = (R_m + r_{rx})^2 + (X_m + \omega L_{rx} - \frac{1}{\omega C_{rx}})^2. \tag{13}$$

The PA consists of a DC power supply V_{PA}, a choke L_f, a switch S, a shunt capacitor C_S, and a series capacitor C_0.

The efficiency of the PA can be derived as

$$\eta_{PA} = \frac{g^2 R_{Z_{in}}}{2R_{dc} + 2r_{L_f}}, \tag{14}$$

where R_{dc} is the equivalent resistance PA shows to the DC power supply.

Here,

$$R_{dc} = \frac{\pi^2 - g(2\pi\cos\phi - 4\sin\phi)}{4\pi\omega C_S}, \tag{15}$$

$$g = \frac{2\pi\sin(\varphi + \phi) + 4\cos(\varphi + \phi)}{4\cos\phi\sin(\varphi + \phi) + \pi\cos\varphi}, \tag{16}$$

$$\phi = \arctan\frac{\frac{\pi^2}{2} - 4 - \pi\omega C_S(2R_{Z_{in}} + \pi X_0)}{\pi + \pi^2\omega C_S R_{Z_{in}} - 2\pi\omega C_S X_0}, \tag{17}$$

$$\varphi = \arctan\frac{X_0}{R_{Z_{in}}}, \tag{18}$$

where g and φ are the intermediate variables, ϕ is initial phase of i_{out}.

B. Optimization Design

C_S, C_0, C_{rx}, C_{mnp}, C_{mns} are the five design parameters of the biomedical implantable WPT system using L-matching network [refer to Fig. 4], namely X.

The feasible range of X is defined as:

$$\mathrm{X} = [C_S, C_0, C_{rx}, C_{mnp}, C_{mns}]_{1\times 5} \in (\mathrm{X}^{lower}, \mathrm{X}^{upper}), \tag{19}$$

where X^{lower} and X^{upper} are the lower and upper bounds of X, respectively. In real applications, due to different distances and misalignments between the coupling coils, the variation of the coupling coefficient k is common. Its nominal value (i.e., a target operating condition) is defined as k^{nom}. The variation range of k is defined as

$$k \in (k^{lower}, k^{upper}), \tag{20}$$

where k^{lower} and k^{upper} are the lower and upper bounds of k which are the predefined minimum and maximum values of k.

From (6), (12), (13), (14), (16), (17), the system efficiency can be expressed by a function of design parameters X, constant parameters Z_{con} and the variable k.

Here,

$$Z_{con} = [\omega, C_{tx}, L_{tx}, r_{tx}, L_{rx}, r_{rx}, r_{L_f}, r_{L_r}, r_{D_r}, R_L]_{1\times 10}. \tag{21}$$

The final design optimization problem is formulated as follows:

$$\max_{\mathrm{X}} \ \eta_{sys}(\mathrm{X}) \tag{22}$$

$$s.t. X \leq X^{upper}, \tag{23}$$

$$-X \leq -X^{lower}. \tag{24}$$

The purpose of the design procedure is to find an optimal set of design parameters, X_{opt} to obtain the highest achievable system efficiency under the constraints of design variables and the nominal k. Given the nature of the optimization problem in (22) - (24), it is appropriate to apply genetic algorithm (GA), a popular population-based heuristic approach, to find a global or at least near-to-global optimal solution.

IV. EXPERIMENTAL VERIFICATION

An implantable WPT system working at 6.78 MHz is built up for verification purpose. As shown in Fig. 5, this system includes a Class E PA, coupling coils, a Class E current-driven rectifier, and an electronic load. The diameters of the transmitting and the receiving coils are 7.2 cm, 1.5 cm, respectively. In this 6.78-MHz WPT system, a Schottky barrier diode (DFLS230L) and a MOSFET (SUD06N10) work as the rectifying diode D_r and the switch Q of the rectifier and Class E PA, respectively. The constant parameters in Table I are used for fair comparison.

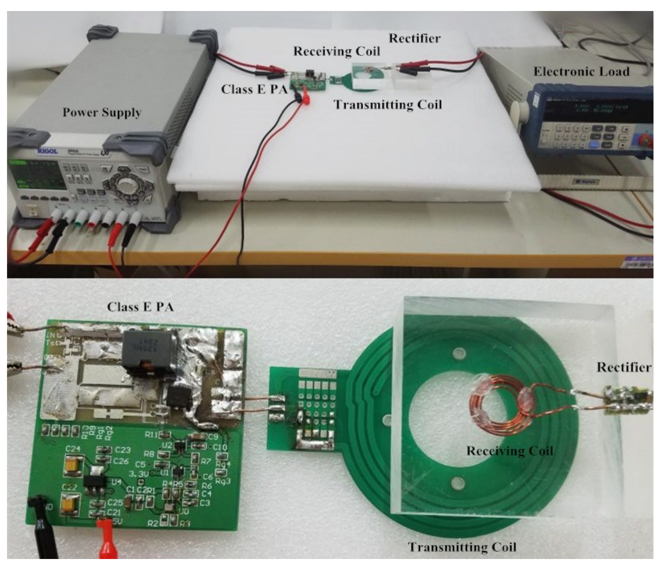

Fig. 5. The experimental biomedical implantable WPT system

Here the feasible ranges of the design parameters, $\mathrm{X} = [C_S, C_0, C_{rx}, C_{mnp}, C_{mns}]_{1\times 5} \in (\mathrm{X}^{lower}, \mathrm{X}^{upper})$, are

$$X^{lower} = [100\ pF, 100\ pF, 100\ pF, 100\ pF, 100\ pF], \tag{25}$$

$$X^{upper} = [1000\ pF, 1000\ pF, 5000\ pF, 5000\ pF, 5000\ pF]. \quad (26)$$

The ranges of C_S and C_0 are chosen as 100 pF-1000 pF according to a given input impedance Z_{in} [refer to (1) and (2)]. Based on the datasheet, the parasitic capacitor of the diode is about 30 pF. The calculated final C_S includes this parasitic capacitor. Following the design procedure in section III, the optimal design parameters, are

$$\mathrm{X}_{opt} = [850\ pF, 370\ pF, 2200\ pF, 2200\ pF, 3300\ pF]. \quad (27)$$

As shown in Fig. 6, as the distance becomes larger, the coupling coefficient k decreases. However, k doesn't change linearly with the misalignment between the coils. Among these four misalignments, when the misalignment=1.7 cm, k is the largest. Moreover, When the distance is small, k under zero misalignment is the smallest, but when the distance becomes larger, k under the largest misalignment is the worst.

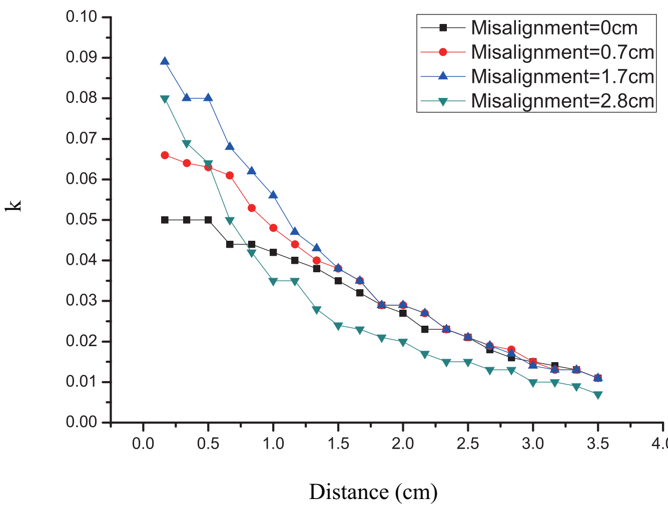

Fig. 6. Measured results of coupling coefficient k under different misalignments and distances

Fig. 7 shows the reflected resistance of the receiving coil on the transmitting side, R_r, under different k. It can be seen that the proposed methodology can increase R_r, i.e., decrease the power loss of the transmitting side.

Fig. 8 shows the simulation and experiment results of system efficiencies when k varies from 0.007 to 0.089 (i.e., distance and misalignment between coupling coils vary from 0.167 cm to 3.5 cm, 0 cm to 2.8 cm, respectively) [refer to Fig. 6]. Note that the quality factor of the receiving coil is 48.56.

It can be seen from the experiment results that the highest system efficiency is 66.67%. Meanwhile, the system efficiency is 36.43% for the nominal k=0.035, i.e., the distance between coupling coils is 1.5 cm and the misalignment is 0 cm. Besides, the simulation results of system efficiencies in conventional and proposed design are listed in Table III. This table shows the improvement of system efficiency by using the proposed methodology.

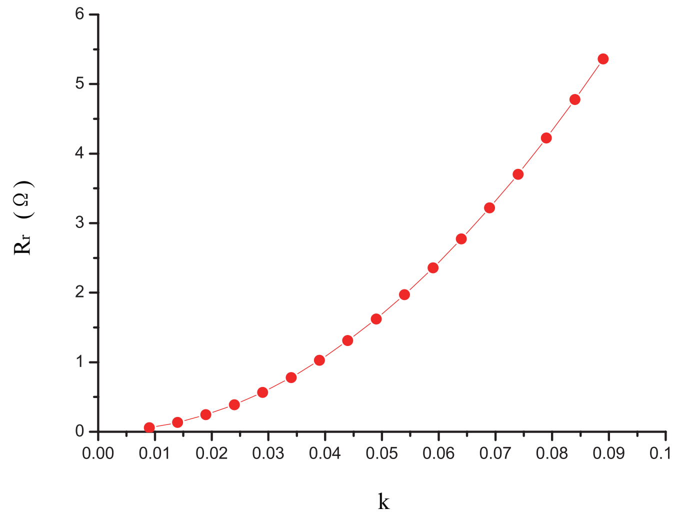

Fig. 7. Simulation results of the reflected resistance under different coupling coefficients

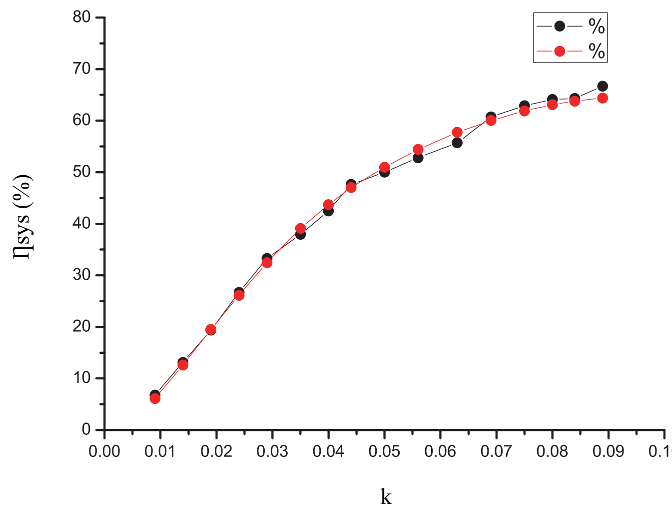

Fig. 8. Experiment and simulation results of system efficiencies under different coupling coefficients

TABLE III
SYSTEM EFFICIENCY PERFORMANCE

k	Conventional methodology	Proposed methodology
0.009	0.3%	6.1%
0.029	2.9%	32.5%
0.049	7.8%	50.4%
0.069	14.1%	60%
0.089	20.6%	64.4%

V. CONCLUSIONS

This paper discusses optimized parameter design for a 6.78-MHz biomedical implantable WPT system. The input impedance of the transmitting coil considering the matching network is accurately derived. Then this derived input impedance is used to guide the parameter design of the matching network, the coupling coils and the Class E PA. It is verified by experiment that the proposed L-matching network containing two capacitors can improve the system efficiency without causing much power loss. With loosely coupled coils due to the small size of the implants, the system efficiency can still reach 36.43% when the distance between coupling coils is 1.5 cm. Furthermore, this system can operate over the wide range of distance and misalignment of the coupling coils.

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