# An Efficient Wireless Power Transfer System for An Implantable Deep Brain Stimulation

1st Sevilay Cetin
*Department of Biomedical Engineering*
*(Pamukkale University)*
Denizli, Turkey
scetin@pau.edu.tr
0000-0002-9747-4821

2nd Veli Yenil
*Department of Electrical-Electronics Engineering*
*(Pamukkale University)*
Denizli, Turkey
veliyenil@pau.edu.tr
0000-0002-0257-5305

3rd Umit Akin Dere
Faculty of Medicine
*Department of Neurosurgery*
*(Pamukkale University)*
Denizli, Turkey
udere@pau.edu.tr
0000-0002-6678-6224

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***Abstract*—This paper presents an efficient wireless power transfer (WPT) system for implantable deep brain stimulation (DBS) devices. A head-mounted DBS device is taken into account in the design procedure of the presented WPT system. A three coil system is designed to provide high efficiency power transfer. The LCC compensation network which has constant voltage characteristic is used in the transmitter side and two receiver coils using series compensation are located in the receiver side, into the human tissue. Finally, performance of the proposed WPT system is tested by a 3D electromagnetic simulation work at a 10 mm distance. Based on the simulation results, 1.19 V DC output voltage is regulated while the DC output current is 24 mA. The power transfer efficiency is achieved as 4.63% at full load condition.**

***Keywords—wireless power transfer, deep brain stimulation (DBS), efficient power transfer, voltage regulation.***

## I. Introduction

The brain implantable medical devices with stimulation function are gained attention for *in vivo* monitoring, diagnosis, treatment and personal healthcare [1], [2]. The stimulation function of these devices can introduce determined amount of charge for the neuronal tissue by providing electrical signals in a predefined period. Thus, brain implantable devices provide therapy for neurological diseases [3]-[5]. In addition, some experimental studies have been also published about traumatic brain injury patients [6]. Deep brain stimulation (DBS) is a popular method used in different diseases caused by deficiency of neurotransmitter or neural cell dysfunction [7]. It is used in the treatment of degeneration of neurons, which can cause neuron death resulting in different diseases such as Parkinson, Alzheimer, tremor, dystonia, epilepsy or obsessive-compulsive disease [8]-[10].

The implanted DBS devices can operate with a battery surgically introduced into the human body. At present, patients with Parkinson disease (PD) need operation in a determined period, varying depend on function and usage frequency of DBS, to replace their battery [11], [12]. For instance, nowadays, mostly implemented conventional open-loop DBS requires more energy, causing short surgery period, compared to the adaptive closed-loop DBS, automatically adjusting stimulation parameters [13].

In the conventional DBS mounting, chest implanted primary batteries, which have long interconnect across the neck, can cause mechanical failure due to neck movement [11], [14]. Thus, nowadays, studies on head-mounted or skull mounted DBS systems are gaining importance [15]-[17]. The development of head-mounted DBS systems is advantageous as it also allows single stage surgery operation without anesthesia [18]. Therefore, in this work, a head-mounted DBS device with limited size is taken into consideration.

Recently, the demand for wirelessly powered medical devices has been increased to extend the surgery period of patients. The wirelessly powered DBS can reduce the infection risks and patient discomfort resulting in surgery operation. However, size limitation for the receiver unit causes difficult design the wireless battery charger for DBS device [12]. The size of the receiver coil should be as small as possible to provide comfort of patients and reduce their suffering. On the other hand, enough energy should be provided by receiver to charge the battery of DBS device.

In order to increase the power transfer efficiency, four-coil systems are usually implemented for wirelessly powered DBS device [11], [12]. In [11], an inner dual-layer printed spiral coil (PSC) and outer helical coil structures are designed in the proposed WPT system. In addition, T-type impedance matching network is used to reduce the reflection from the load to the source and increase the power transfer efficiency. However, T-type impedance matching network and outer helical coil implementation result in complex design procedure. In [12], a design optimization procedure is presented to charge the battery which can be implanted into patients' skull. In the optimization work, sizes of coils and capacitances are evaluated. However, the outer diameter of the receiver coil is still large.

To provide high efficiency power transfer, the compensation network also plays important role in the design of the WPT systems. Series-series and series-parallel topologies are usually used in WPT systems of the implantable medical devices [19], [20]. In these basic topologies, the maximum power point is limited by coupling coils parameters. The hybrid compensation topologies can improve the power transfer efficiency of the basic compensation topologies [21]-[23]. The implementation of the hybrid topologies in WPT system of pacemakers are given in [24], [25]. In these studies, operation frequency is 300 kHz and the receiver coil size is large enough to cover pacemaker case. According to our knowledge, hybrid compensation network has not been applied to the WPT system of DBS devices.

This paper presents the efficient WPT system for head-mounted DBS device with rechargeable battery. A three-coil system is designed to provide high coupling between the transmitter and receiver side. In the receiver side, secondary and load coils are designed, as small as possible, to prevent patients suffering. To prevent loading effect, LCC compensation network is applied to the transmitter side while series compensation is applied for the other two coils in the receiver side of the WPT system. The voltage regulation and efficiency performance of the proposed system are tested by ANSYS software providing 3D electromagnetic simulation. According to obtained results, the output voltage is regulated at nominal and light load conditions. An efficient power is also transferred to the load.

## II. DESIGN PROCEDURE OF THE PROPOSED WPT SYSTEM FOR DBS

The concept design of the proposed WPT system is given in Fig. 1. In the receiver side, secondary and load coils are implanted under skin and DBS is mounted to the skull. In the clinical applications, biomaterial with poor conductivity is needed between the tissues and coils including resonant capacitors and rectifier diodes, in the receiver side, to prevent unwanted biological responses. $V_{in}$ represents the sinusoidal power source. $L_1$ is the transmitter coil, $L_2$ is the secondary coil and $L_3$ represents the load coil. $k_{12}$ is the coupling coefficient between $L_1$ and $L_2$ while $k_{23}$ is between $L_2$ and $L_3$, $k_{13}$ is between $L_1$ and $L_3$.

The induced AC voltage across the load coil is converted to DC voltage by the rectifier. Then, generated DC voltage feeds the DBS device through the power management unit including boost converter. The DBS device produces the exciting signals, which are applied to the stimulation area via leads. The implemented leads are significantly shorter compared to leads of the chest mounted DBS device. Therefore, they are more favorable for high density DBS and less invasive [14], [26].

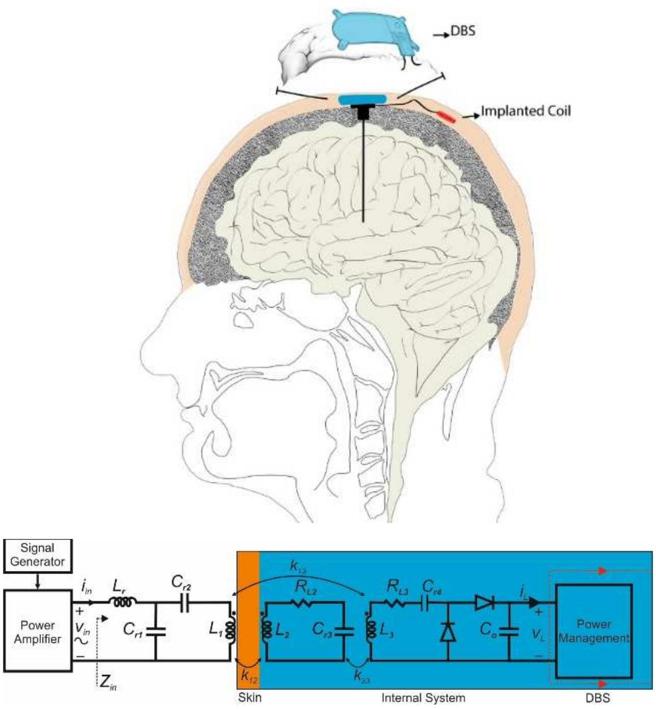


Fig. 1. The concept design of the proposed WPT system for a head-mounted wirelessly powered DBS device.

### A. Three Coil Modelling of the Proposed WPT System

In the design procedure of the proposed WPT system, transmitter, secondary and load coils are modeled in ANSYS 3D magnetic simulation software. The magnetic model of 3D coil configurations of the proposed WPT system are given in Fig. 2. In the simulation model, it is assumed that transmitter coil is in the air, outside of the skin, while the secondary and load coils in the receiver side are beneath the head skin. The distance between the transmitter coil and the secondary coil is determined as 10 mm to represent fat and skin tissue thickness. The fat and skin tissues are implemented with their magnetic properties in the modeling work. The load coil is located very close to the secondary coil. The load coil is placed one side of FR4 while the secondary coil is placed to the other side of the FR4. Thus, they are aligned facing each other. The secondary and load coils to be implanted are designed with 5 mm outer diameter. In case of any misalignment, transmitter coil is designed with outer diameter of 24 mm, larger compared to secondary and load coils.

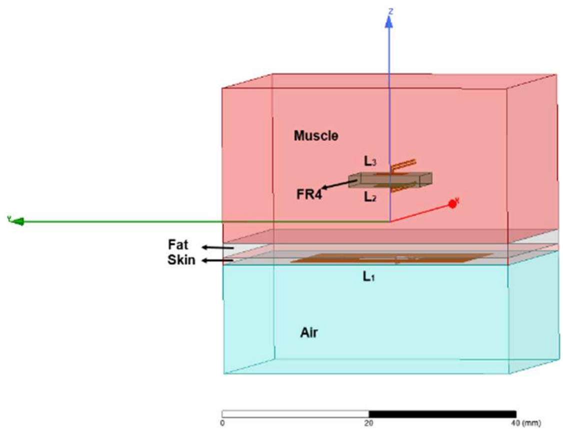


Fig. 2. The 3D magnetic modelling of transmitter, secondary and load coils in ANSYS software.

When 3D magnetic coil modeling is properly built, the inductance values of each coil and coupling coefficient between them can be extracted by excitation current applied to the coils. The inductance values and coupling coefficient between transmitter and receiver coils are obtained as given in Table I.

TABLE I. THE EXTRACTED INDUCTANCE VALUES BY 3D MAGNETIC MODELING.

| Coil Parameters | Values |
|---|---|
| L1 | 3.85 µH |
| L2 | 69.3 nH |
| L3 | 68 nH |
| k12 | 0.0142 |
| k13 | 0.0103 |

### B. Compensation Network Design

The AC equivalent circuit model of the proposed WPT system is shown in Fig. 3. At the transmitter side, an LCC compensation network, composed of $L_r$, $C_{r1}$ and $C_{r2}$, is used to compensate the reactive power while the series compensators, $C_{r3}$ and $C_{r4}$, are used for both secondary and load coils in the receiver side. $R_{L2}$ and $R_{L3}$ represent the resistance of $L_2$ and $L_3$ coils. $R_L$ represents the equivalent AC resistance of the load and the rectifier.

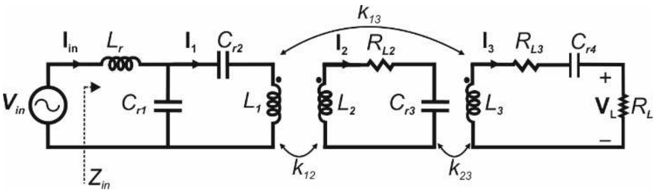


Fig. 3 The equivalent circuit model of the proposed WPT system.

The values of the compensation capacitors are determined taking into consideration the operation at resonant frequency. Then, the value of the $L_r$ inductance is determined. If resonant frequency of all resonators are identical, following statement can be written

$$\omega_o = \omega_1 = \omega_2 = \omega_3 . \tag{1}$$

In the compensation network, the power absorbed by the load is defined as

$$P_L = I_{3-RMS}{}^2 R_L . \tag{2}$$

The input power supplied by input source $V_{in}$ can be defined as

$$P_{in} = V_{in-RMS} I_{in-RMS} . \cos \theta_{in} . \tag{3}$$

Where $\theta_{in}$ is the phase angle of the input impedance $Z_{in}$ and it can be written as

$$\cos \theta_{in} = \frac{\mathrm{Re}|Z_{in}|}{|Z_{in}|} . \tag{4}$$

The input impedance of the compensation network can be extracted by

$$Z_{in} = \frac{\mathbf{V_{in}}}{\mathbf{I_{in}}} . \tag{5}$$

Finally, the power transfer efficiency of the proposed WPT system for DBS can be given as

$$\eta = \frac{P_L}{P_{in}} . \tag{6}$$

## III. Simulation Results

The performance of the proposed WPT system was tested with ANSYS software providing 3D electromagnetic simulation. The magnetic modeled three coils are used in the electromagnetic simulation. At the transmitter side, $L_r$ inductance is determined as 91.7 nH while $C_{r1}$ is 1.5 nF and $C_{r2}$ is 36.65 pF. At the receiver side, $C_{r3}$ and $C_{r4}$ are determined as identical and 1.98 nF. The operation of the WPT system is tested at 13.56 MHz.

The waveforms of the regulated DC output voltage and the load coil current are shown in Fig. 4 and Fig. 5. The output voltage regulation of the proposed system is tested at two different load conditions being full and light enough. As shown in Fig. 4, 24 mA DC output current and 1.19 V DC output voltage are regulated. At the light load condition, 1.25 V is regulated while the output current is 12.5 mA, as shown in Fig. 5. According to the obtained results, the output voltage of the rectifier is acceptable almost constant with little difference.

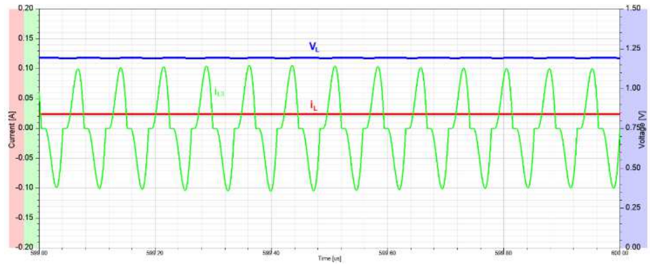


Fig. 4. The waveforms of the regulated output and the load coil current in the proposed WPT system. $V_L$=1.19 V, $I_L$=24 mA.

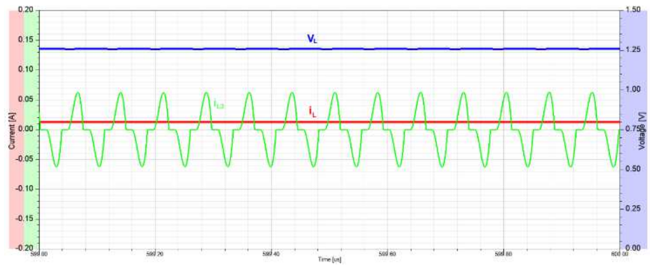


Fig.5. The waveforms of the regulated output and the load coil current in the proposed WPT system. $V_L$=1.25 V, $I_L$=12.5 mA.

The power transfer efficiency of the proposed WPT system was also tested for the same power levels used for output voltage regulation. At 24 mA output current level, the power transfer efficiency is obtained as 4.63% while it is 2.9% at 12.5 mA output current level.

## IV. Conclusion

In this work, an efficient WPT system for head mounted and wirelessly powered DBS is presented. A three coil system is used in the proposed WPT system. An LCC compensation at the transmitter side and series compensation for the secondary and load coils were used. The output voltage and power transfer efficiency of the proposed system were tested by ANSYS software proving 3D electromagnetic simulation. According to obtained results, the output voltage of the system is regulated at full and light load conditions. The DC output voltage is regulated around 1.2 V while the output current is 24 mA and 12.5 mA. The maximum power transfer efficiency was extracted as 4.63% at 24 mA output current level.


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