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On the Design of an Efficient Inductive Wireless Power Transfer for Passive Neurostimulation Systems

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Abstract— **In this paper, a minimally invasive wireless powered electronic lens (e-lens) with passive electrodes is presented for an ocular electrical stimulation. Previous research has focused on the differentiation property of the induction phenomenon and half wave rectifiers. However, these approaches are generally application specific, non efficient, suitable for low current, and deliver monophasic current stimulation. Existing rectifier-based techniques can lead to safety concerns as the offset voltage could change unpredictably. A new wireless power transfer circuit is presented for the design of an efficient system to wirelessly deliver charge-balanced biphasic waveforms through passive electrodes for transcorneal electrical stimulation. The absence of active components allows the development of a flexible e-lens system for therapeutic electrical stimulation of the eye.**

I. INTRODUCTION

The scientific innovations behind retinal prosthetic systems [1]–[3], spinal cord pain management [4], and deep brain stimulation (DBS) for Parkinson's disease [5] are often supported by wireless power transfer (WPT) techniques. WPT systems can help achieve wireless powering of the implanted device [6]–[8], and enhance user experience and safety. There are several solutions to wireless powering of the implanted electrodes based on the source of power used: ultrasonic, infrared, and microwave sources of power can be utilized to meet the power demands of the implanted systems [9]–[11]. Inductive power transfer is also a possible solution for power delivery as it is generally safe [12] and easy to design [13]. Inductive power delivery systems generally consist of a transmit and a receive coil. The transmit coil is connected to a voltage/current source while the receiving coil is located on or inside the body to deliver power to an body-worn or implanted device [13]–[17]. In the literature, two types of inductive wireless power delivery systems for biomedical electrical stimulation applications are found: active and passive power delivery systems. Active power delivery systems [18]–[23] utilize an implanted battery/capacitor that stores the energy received from the receiving coil, often to drive a CMOS chip that delivers the regulated current/voltage waveforms to stimulate the tissue using electrodes. Passive wireless power delivery to implanted electrodes [9]–[11], [24]–[26] is also possible where the storage system is avoided and the stimulating waveforms are either transmitted directly from the transmit coil or an additional circuitry at the receiver converts the received signal into a stimulating voltage/current waveform. This technique simplifies the design process by avoiding the use of complex CMOS ICs for the purpose of generating stimulation waveforms inside the body. In this work, we propose a novel passive wireless electrode power delivery system for injecting an asymmetric charge-balanced biphasic waveform through electrodes.

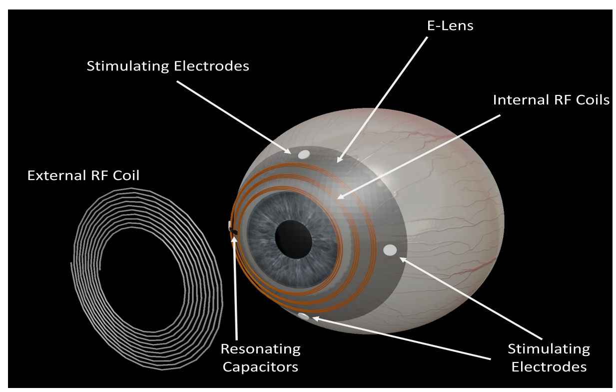

Fig. 1. A representative schematic of the transmit and receive coils used to passively power the electrodes in contact with sclera, for an electronic lens (e-lens) for the electrical stimulation of eye.

Electrochemistry at the interface between electrodes and tissue play a significant role on the system performance. Thus, the receiver coil must be carefully designed to receive the appropriate waveforms. The typical design and safety parameters of an electrochemical system such as reversible reactions and charge balance [27]–[29], waveform characteristics (pulse width/amplitude/duty cycle) [30]–[32], and charge/charge density per phase [33], [34] need to be controlled through appropriate design of the transmit/receive coils. Although the efficiency of the system is lowered by the low transmission frequency, several techniques such as modulation, rectification and filtering can be utilized to improve the performance.

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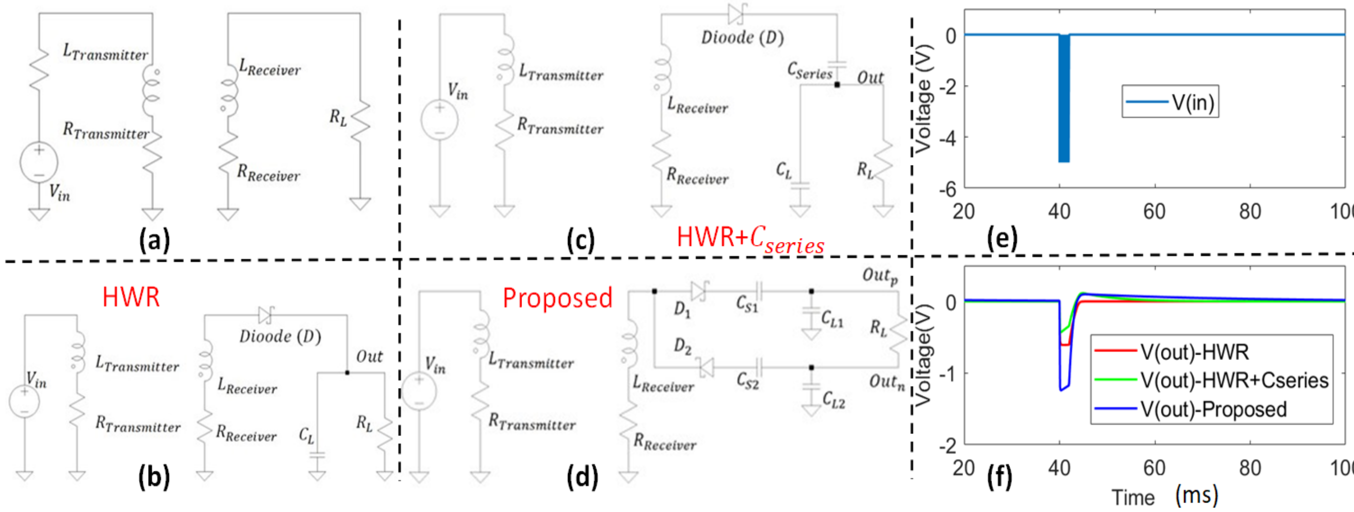

Fig. 2. There are four schematics in this figure. Only rectifier-based systems of Fig 2b, 2c and 2d are compared. Schematic of Fig. 2a is not part of the stimulation study as it needs triangular input current to induce rectangular pulse at the receiver. Transmission of triangular pulse is inefficient. (A) Schematic of a traditional inductive wireless power transfer system (Section II A). (B) Schematic of Half Wave Rectifier (HWR) based wireless power transfer system (Section II B). (C) Schematic of Half Wave Rectifier with series capacitor ($HWR_{C-Series}$) based wireless power transfer system (Section II C). (D) Schematic of the proposed wireless power transfer system (Section II D). (E) Input voltage waveform to compare systems of Fig. 2b, 2c and 2d. (F) Output/Load voltage waveform to compare systems of Fig. 2b, 2c and 2d. Parameters of the three systems are given in Table I. The three circuits are compared under similar input, load, transmitter and receiver conditions. Thus, higher load output voltage (V(out)) translates to higher system efficiency.

In this work, we will first compare different commonly utilized solutions to the design of passive wireless electrodes to deliver asymmetric short cathodic-first charge-balanced biphasic pulses. We will then introduce a new high-efficiency, charge-balanced, passive wireless stimulation system and compare its measured performance with other solutions.

II. Techniques for Wireless Power Delivery to Passive Electrodes

Half wave rectifier (HWR)-based solutions that takes advantage of the rectification, modulation, and filtering principles to deliver power to load characterized by large resistance are discussed first. A new circuit that can deliver high performance asymmetric charge-balanced biphasic waveforms is then proposed. Thus, the goal of this work is to compare commonly used passive wireless electrode systems and propose a new circuit and its benefits over the previous system designs. A proof of concept implementation of the proposed system is presented, along with measured results.

A. *Charge Balanced Power delivery using induction*

A traditional inductive wireless power transfer system consists of two coupled planar spiral coils. An embodiment of the transmitter and receiver coils for electrical stimulation of the eye is shown in Fig. 1. The transmit coil is located in front of the eye and could be part of a system embedded on eyeglasses. The receive coil is positioned instead on the sclera and can be embedded into a soft contact lens. Two stimulating electrodes are located on the two ends of the receive coil shown in Fig. 1. The transmit and receive coils are modeled as an inductor in series with a resistor, as shown in Fig. 2a.

A charge-balanced asymmetric biphasic waveform can be delivered to the implanted receiver coil without using any matching components on the load side [24]. This can be achieved using the WPT circuit shown in Fig. 2a. The induced voltage at the receiver can be expressed as a function of the input current simply as $V_{Receiver} = -L_{12}\frac{\partial i_{transmitter}}{\partial t}$. This induced voltage can deliver current to passive electrodes connected to the receive coil. Thus, a suitable triangular input current waveform with a different rising and falling edge time periods can deliver an asymmetric biphasic waveform to wireless passive electrodes (load) connected to the receiver inductive coil. The drawback of this technique is that the transmitted signal contains low frequency components, and is therefore inefficient for therapeutic applications using electrical stimulation with pulse widths in the order of milliseconds. Further, the waveform of the current in the receive coil is very sensitive to impedance variations as the voltage-current relation is affected by the presence of a capacitance in the receive loop.

B. *Half Wave Rectifier (HWR)*

A wireless system to deliver a monophasic stimulation waveform is shown in Fig. 2b. A modulated pulse shown in Fig. 2e is used at the transmit coil. The transmitter voltage source generates a signal at the frequency of 250 kHz for a duration of 1 ms. The transmitter voltage can be rectified to retain only one polarity of the voltage (with the negative polarity shown in Fig. 2e) to avoid designing electronics for both positive and negative voltage swings. Further, rectification may increase the harmonic frequency content which may enhance the power delivery.

As an example of implementation, we have considered a 60-turn receive coil, made of AWG 36 copper wire. The

Quality factor of such coil is low, with a typical value of 3 for the parameters considered in this example. This leads to a receive coil parasitic resistance of 10 Ohms. Given the low quality factor, the performance cannot be enhanced using parallel resonance. However, the power delivered can be increased by increasing the number of turns of the receive coil. Using a smaller number of turns of the receive coil leads to a lower induced voltage at the receiver and therefore requires an increased input voltage and current.

For this example, the simulated receiver pulse voltage across the load resistor is shown in Fig. 2f. Simulation parameters are given in Table I. The received pulse is monophasic as the diode conducts for only half of the cycle. Use of Schottky diodes improves the performance due to their reduced threshold [25]. The HWR solutions can be used to deliver small charge without safety concerns [25]. However, in order to meet the goals of this work, the monophasic output of the HWR based system must be converted to a charge-balanced output. This can be done using a series capacitor [31], [32], [35], [36] as discussed in the next section.

C. HWR with series Capacitor ($HWR_{C-Series}$)

A charge-balanced waveform in an HWR system can be achieved using a series capacitor, as illustrated in Fig. 2c. The capacitor provides a reverse conduction path for the current after a cathodic stimulation (it modifies the biasing of the Schottky diode accordingly) and thus provides a charge-balanced pulse. The simulated output voltage across the load for the same input voltage provided in Fig. 2e is shown in Fig. 2f. The value of the peak-to-peak voltage of this circuit is reduced compared to the peak-to-peak voltage of the HWR monophasic output shown in Fig. 2f. Thus, placement of a series capacitor may provide a charge-balance output at the cost of reducing the power delivered to the load. The amplitude of the output voltage for such systems can be improved by adopting the circuit proposed here in Section II.D and shown in Fig. 2d.

D. Dual-Capacitance System for Efficient Wireless Power Delivery to Passive Electrodes

The output load resistor is connected between the two nodes named Out_n and Out_p (Fig. 2d). In this implemented circuit, we utilize additional load capacitors C_{L1} and C_{L2} to increase the output load voltage. Conventional voltage doubler and rectifier circuits provide monophasic output; the proposed circuit can be considered a charge-balanced version of the voltage doublers. The working principle of the circuit can be explained as follows: i) D_1 is forward biased and D_2 is reverse biased when the voltage at the anodic end of the Schottky diode D_1 is positive; ii) A high frequency current flows from D_1 to C_{L1} via C_{S1} and enter the C_{L2} via the ground; iii) The current completes the closed loop path by flowing through C_{S2} and D_2 in a reverse direction; this causes the node Out_n to be at a negative potential compared to the node Out_p. Thus, the circuit provides a higher negative reference for the output load voltage compared to the circuit in Fig. 2c. The circuit will provide a charge-balanced output to the load as the voltage at the receiver coil (the node connected to D_1 and D_2) will be pure AC. While the proposed circuit results in a similar charge-balanced output voltage as the $HWR_{C-Series}$ circuit, the efficiency is not compromised under similar transmitter, receiver, input and load conditions as shown in Table I.

TABLE I

PARAMETERS USED IN SIMULATION OF 3 SYSTEMS SHOWN IN FIG. 2

Simulation and System	Parameters and Values
These parameters and values are common to all three simulations of schematics Fig 2(b),(c) and (d).	L_{Tx} = 1μ H,L_{Rx} = 10μ H,R_{Tx} = 0.5 Ω, R_{Rx} = 10 Ω, R_L = 10k Ω , K_{Tx-Rx} = 0.1, All Diodes: 1N5819
HWR circuit in Fig.2(b)	C_L=0.2 μ F
HWR+C_{series} in Fig.2(c)	C_L=0.2 μ F C_{Series}=0.5 μ F
Proposed circuit in Fig.2(d)	C_{L1} = C_{L2} =0.2 μ F, C_{S1} = C_{S2} = 5 μ F

III. OFFSET PERFORMANCE AND ADVANTAGES OF THE PROPOSED CIRCUIT OVER HWR AND $HWR_{C-Series}$

The offset voltage at the tissue-electrode interface can develop and vary due to change in concentrations or imbalanced equilibrium potential in cathodic and anodic reactions [27]. Thus, the effective load seen by the receiver coil connected to passive electrodes during a stimulation can include an offset voltage, interface capacitance and resistance which constitute the electrochemical load. The generation and variation of offset voltage depends on the type of circuits used. The simulation results in Fig. 2 only considered resistive load to study circuit behavior and efficiency and hence do not show offset voltage. However, when used to stimulate biological tissue, these circuits result in different offset voltage. In this section, we discuss the variation in offset voltage of the circuits shown in Fig. 2b, c and d.

The HWR rectifier shown in Fig. 2b delivers monophasic waveform to the resistive load. Thus, in the presence of electrochemical effects, the HWR system leads to the development of an offset voltage [25]. The HWR system, when used for low current stimulation application, achieves a charge-balanced output due to the electrode polarization [25]; however, when HWR is used for high current stimulation, it can lead to a large offset voltage.

Unlike the HWR circuit, the $HWR_{C-Series}$ circuit shown in Fig. 2c can deliver a charge-balanced current. The choice of the series capacitor determines the design of the anodic pulse to achieve a zero net charge. However, the presence of the series capacitor leads to the reduced voltage drop across the load resistor, as shown in Fig. 2f, which reduces the efficiency of the circuit.

The circuit proposed in this paper, shown in Fig 2d, can deliver a charge-balanced current without reducing the voltage drop across the load resistor. The circuit also prevents any automatic offset voltage generations as it has two diodes connected in an opposite direction in series with the load.

The diode connection helps maintain a constant voltage across the load. Any increments in the offset voltage at node 'out_p'(Fig. 2d) leads to the reduced conduction of the diode D1 and increased conduction of the diode D2 to restore the voltage. Similarly, it restores any increase/decrease of the voltage at the node 'out_n' (Fig. 2d). This active control of the node voltage provides a more stable offset voltage at the nodes 'out_p' and 'out_n' (Fig. 2d). The better controlled offset voltage of the proposed circuit enables the implementation of high current passive electrode stimulation systems.

IV. Result and Conclusion

The proposed charge-balanced circuit of Fig. 2d and HWR circuit of Fig. 2b were experimentally implemented for performance comparison. The first observation is that the proposed circuit shows a smaller offset voltage over a 2-hour period of stimulation compared to the HWR system, which is desirable. Second, the V(out) amplitude of the proposed circuit is higher than that of the HWR circuit. The experiment was conducted on a sample biological tissue to prove the concept. Tungsten electrodes of diameter 200 um were used for the stimulation and ground in the experiments.

The values of the physical and electrical properties of the coils are given in Table II. The output voltage of the two systems measured across the stimulating and ground electrodes in the tissue is shown in Fig. 3. The offset created by the HWR system is increased to about -1.4V after the 2 hours of stimulation while the proposed circuit results in a zero offset. When the offset voltage of the HWR system is subtracted from the measurement data to compare the amplitudes, the AC potential difference observed in the proposed circuit is 25% higher than the HWR circuit.

TABLE II

Electrical and Physical Properties of the Coils Used

Parameters	Tx.Coil	Rx.Coil
Outer Diameter(mm)	70	26
Inner Diameter(mm)	28	14
AWG	20	32
Number of Turns	3	40
Inductance(μH)	1	22
Resistance(Ω)	0.9	2

We should note that the rectifiers are nonlinear circuits. The resistances offered by diodes depend on bias amplitude, frequency etc. Thus, the amplitude gain offered by the proposed circuit over the HWR circuit depends on the load, signal level, distance of the receiver coil from the transmitter coil, and other factors.

In conclusion, in this work we have proposed a new circuit for the wireless power delivery to passive electrodes. The proposed circuit can perform better than the traditional HWR in terms of efficiency, charge balance and offset voltage. Using a tissue simulating material, we have also demonstrated that the proposed circuit will not lead to an offset voltage over a long period (2 hours) of stimulation. This advantage over HWR gives a possibility of implementing the proposed circuit in medical implants which usually require hours of stimulation. Although in this specific study we used an asymmetric short cathodic-first biphasic waveform as it was shown to reduce the subthreshold of retinal ganglion cells in epiretinal electrical stimulation [37], future experiments with our multi-scale computational platform will be focused on devising better stimulation strategies, including the stimulus waveform for selective activation of retinal neurons [38]–[40].

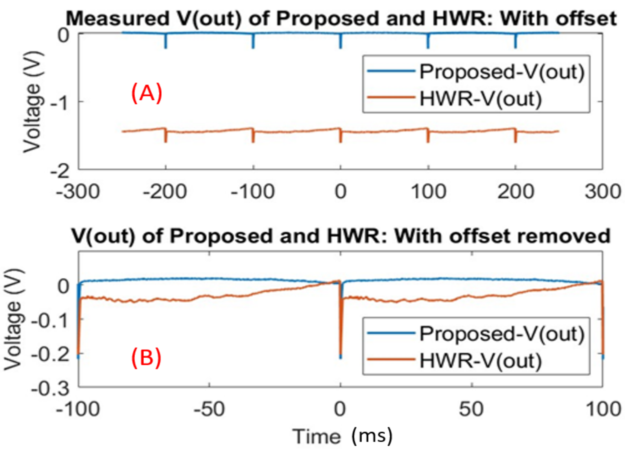

Fig. 3. (A) Measurement of output voltages of the proposed circuit and the HWR circuit in a biological-simulant tissue after 2 hours of stimulation. It can be observed that the offset offered by the HWR is close to -1.5 V while the offset of the proposed circuit is almost negligible. Both circuits started with zero offset conditions. (B) To compare the amplitude of the output voltages of the HWR and the proposed circuit, offset voltage was removed from the recorded measurement results. The amplitude of the voltage of the proposed circuit is higher than that of the HWR.

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