Wireless Power Transfer to a Visual Prosthesis

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Wireless Power Transfer to a Visual Prosthesis:
100 mW at 6.78 MHz

Abstract—An inductive wireless power transfer system for a neural implant is presented. The prototype can transfer 100 mW at 6.78 MHz transcutaneously for $0.08 \leq k \leq 0.31$, which corresponds to a distance of 8 to 15 mm, and a lateral misalignment of 0 to 15 mm. The achieved efficiency ranges from 55 to 77 %, depending on the relative distance between the coils. The transmitter coil outer diameter is 55 mm, the receiver coil outside diameter is 35 mm. The inverter is class-E, using a GaN MOSFET, and the receiver is class-DE.

Index Terms—wireless power transfer, biomedical implant, bioelectronics, class-E inverter, class-DE rectifier, transcutaneous power transfer.

I. INTRODUCTION

Over 43 million people are blind, and this number is expected to reach 61 million by the year 2050 [1]. Less than 10 % of the blind population is born blind. In the rest, the blindness is caused by disease, accident, and/or aging, implying that the visual areas of the brain are still functioning in most cases. Functional vision could one day be restored using a neuroprosthesis [2], which is the goal of the NESTOR project [3].

The electronic part of the brain implant that is developed in the NESTOR project will be subcutaneous, i.e. located under the skin, outside the skull. Data as well as power have to be transferred to and from the implant wirelessly [4]. It is expected that the implant will need at most 100 mW. This is significantly higher than the power levels commonly encountered in biomedical implants in the human head [5], where wireless power transfer (WPT) occurs transcutaneously (through the skin).

In this work, the development of a WPT system delivering the required power (at most 100 mW) to the visual prosthesis is discussed. The transmitter will be wearable, and the receiver will be implanted; the system should thus be small. As it will be battery-powered, it should be efficient.

II. INVERTER AND RECTIFIER SELECTION

Inverters and rectifiers can be divided into classes, that give a broad indication of the characteristics, such as efficiency, component count, and design complexity.

Inverters of classes A to C use transistors in their active region, which is beneficial for achieving low harmonic distortion, but also introduces significant losses, since the voltage drop over the switch is significant. These classes are thus not optimal for use in efficient power inverters. Classes D and above use transistors as switches, which reduces the losses significantly, as the voltage drop over the switches is low. The switches are loaded with higher-order passive circuits to obtain an output waveform with low harmonic distortion. The transmitter (Tx) coil doubles as a part of the filter in this case.

A class-D inverter is very robust to changes in the load and coupling factor $k$. It operates on a duty cycle $D = 50\%$, making it simple to control. However, it uses a high-side switch, which means that it requires additional level shifting circuitry to drive the gate. Furthermore, the parasitic capacitor inside each switch is charged and discharged each cycle, causing losses. It is challenging to use a class-D inverter at MHz frequencies, as these losses increase with frequency.

The part count of the class-E inverter is lower than class-D. It uses zero-voltage switching (ZVS) to eliminate the losses associated with the (dis)charging of the parasitic capacitors in the switches, resulting in a higher efficiency. It can operate at a range of duty cycles, including $D = 50\%$. It is less robust to changes in $k$, but as we will see in this paper, the class-E inverter can be designed to be sufficiently robust for the broad range for $k$ that we encounter in this application.

The efficiency of class-F is a few percent higher than class-E [6]. Additional LC-filters result in the output waveform being less distorted, and in less voltage stress on the switch. However, the part count is higher, and the duty cycle for optimum operation is lower [6], which can be harder to generate in practice.

In view of the size, efficiency, and design complexity requirements, the class-E inverter was chosen.

In terms of losses, the rectifier can be analyzed in the same way as the inverter. Just as the class-D inverter, the class-D rectifier suffers from losses associated with parasitic capacitances. The efficiency of the class-E rectifier is again higher than class-D. However, its output voltage is significantly lower.

A class-DE rectifier combines the advantages of both: the output voltage of the class-D rectifier, together with the efficiency of class-E. Therefore, the class-DE rectifier was chosen. Rectifiers of higher class exist, which might add a few percent efficiency, but they add components to the receiver side, which is not desired, as the receiver should be kept small.
III. **System realization**

The class-E inverter and class-DE rectifier are designed with the aid of [7] and [8], the details of which are the subject of a follow-up paper. The load is assumed to consume 100 mW at 7 V, corresponding to a load of 490 Ω. The switch was chosen to be a GaN MOSFET, as its resistance and parasitics allow for a design with low losses. The system operates at 6.78 MHz, for this lies in an industrial, scientific and medical (ISM) frequency band.

The implanted receiver (Rx) coil has a diameter of 35 mm and a wire diameter of 0.25 mm. It is closely wound and has 8 turns. Increasing this number further would bring the self-resonance frequency (SRF) too close to 6.78 MHz. From a clinical point of view, this coil can still be implanted reasonably well*. The Tx coil has an outer diameter of 55 mm, to ensure sufficient robustness against coil misalignment. Its wire diameter is 1 mm. The FastHenry field solver [9] was used to calculate the self and mutual inductance of the coils. It was found that, when increasing the number of turns from 5 to 10 in steps, the increase in mutual coupling outweighs the increase in equivalent series resistance (ESR), which results in a higher maximum efficiency. Therefore, the Tx coil was chosen to have 10 turns.

The simulated and measured coupling factor $k$ is between 0.12 and 0.30 for the separation distance between the coils ranging from 8 to 15 mm, and the lateral misalignment ranging from 0 to 15 mm. That range is assumed to be sufficient to cover the practical situation. The system was designed to have maximum power transfer efficiency at $k = 0.18$.

IV. **Results**

The designed prototype was manufactured. The total printed circuit board (PCB) area of the inverter prototype measures 21.0 $\times$ 16.0 mm$^2$, including a micro-USB connector for power, and the rectifier measures 7.4 $\times$ 6.4 mm$^2$. Figure 1 shows the measured output voltage versus the coupling factor. Results for different separation distances as well as misalignment all translate to a different coupling factor. As can be seen, the output voltage ranges from 7.3 to 10.5 V. Figure 2 shows that the achieved efficiency ranges from 55 to 77 %. The range for which the output power is at least 100 mW is $0.08 \leq k \leq 0.31$.

![Fig. 1: Measured output voltage vs. $k$.](image1)

![Fig. 2: Measured efficiency vs. $k$.](image2)

*This was discussed with a neurosurgeon.

V. **Conclusion**

A WPT system that can transfer 100 mW to a biomedical brain implant transectunately was designed, manufactured, and validated. The system consists of a class-E inverter, class-DE rectifier, two coils, and a controller.

The transmitter measures 21 $\times$ 16 mm$^2$, the rectifier measures 7.6 $\times$ 6.4 mm$^2$. The Tx coil has an outer diameter of 55 mm, and is wound with 10 turns of 1 mm diameter wire. The Rx coil consists of 8 turns of 0.25 mm diameter copper wire, wound closely with an diameter of 35 mm.

The required output power is achieved for $0.08 \leq k \leq 0.31$, which corresponds to a separation distance of 8 to 15 mm, and a misalignment from 0 to 15 mm, which is deemed suitable for the application. The efficiency ranges from 55 to 77 %, depending on the relative position between the coils.

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