High-Efficiency Wireless Power Transfer for mm-Size Biomedical Implants

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***Abstract*— In this study, a wireless power transfer system for miniaturized biomedical implants has been optimized for a 1mm-diameter receiver coil. This coil is wire wound around a 7.5 mm-long ferrite core and achieves a 3.56 µH inductance. For a distance of 10 mm between the industrial primary coil and the handmade secondary coil, measurements show a coupling factor of 3% and power transmission efficiencies of 21.6%, 20.8% and 20.5% respectively for a load power consumption of 1 mW, 500 µW and 250 µW, at 1cm distance and at 1 MHz, surpassing previously-reported comparable links.**

***Index Terms*—Biomedical implant, ferrite, wireless power transfer**

I. Introduction

In future medecine, the continuous monitoring of patients from hospitals to homes, by biomedical implants, could solve numerous health care issues as discussed in [1-3]. Wireless power transfer (WPT) is already known as a convenient principle to power such implants: It is minimaly invasive thanks to the absence of battery and allows the implant to be operated on demand or to remain permanently into the body. However, the transmitted power is limited by the size of the implant and by the distance between the two inductively-coupled coils [4, 5]. Solutions to improve WPT efficiency involve impedance matching between the primary and the secondary [6, 7], adding a phase locked loop controller to track the best operating frequency [8, 9], using ferrite cores at the primary and/or at the secondary coils [3, 10, 11], optimizing the shape of the coils [12, 13], or using multiple coils at the receiver [13]. Transfer frequencies ranging from MF (0.3-3 MHz) to HF (3-30 MHz), VHF (30-300 MHz) and even UHF (0.3-3 GHz) have been advocated. The present study is focused on the performances that can be reached for mm-size implants thanks to the use of a handmade, wire-wound, ferrite-core secondary tiny-coil with optimal impedance matching. Our target case of study consists in a miniaturized sensing capsule to be implanted at a few mm depth in the body (e.g., in eye or skin), to be powered and read by a cm-size external reader with minimal power consumption for increased overall autonomy.

II. System design

A. Topology

Fig. 1 shows a schematic view of the WPT system. On one side, the primary coil features an inductance L_s and a resistance R_{Ls}, and is placed in series with capacitor C_{ss} to limit the reactive power to be provided by the source V_s. Power is collected by a small-diameter secondary coil, with inductance L_l and resistance R_{ll}, linked to the primary coil by a coupling factor *k*. This factor can be computed according to (1), where V_{Ls} is the signal amplitude applied to the primary coil and V_{Ll} is the one measured at the terminals of the secondary coil, assuming that the resonance frequency of the coils interacting with the measurement setup is far from the frequency of the applied signal.

$$\mathrm{k} = \frac{\mathrm{V}_{Ls}}{\mathrm{V}_{Ll}}\sqrt{\frac{\mathrm{L_l}}{\mathrm{L_s}}} \tag{1}$$

As the receiver diameter is small, the coupling factor is also small and the efficiency η of the power transfered to the load R_l is limited. However, η could be increased using a resonant compensation at the receiver [4, 5]. In this design, series and parallel capacitors, C_{sl} and C_{pl}, are added to easily achieve the impedance matching [14].

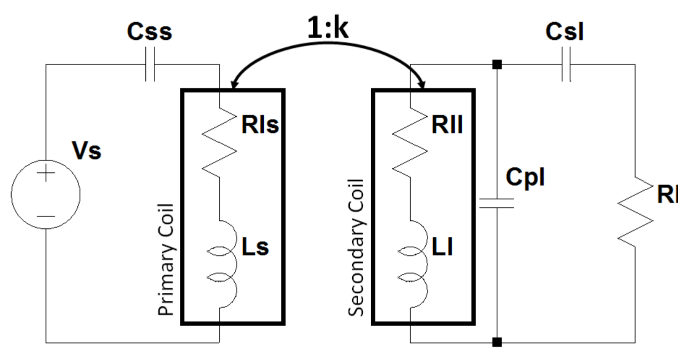

Fig. 1 Schematic view of the wireless power transfer (WPT) system.

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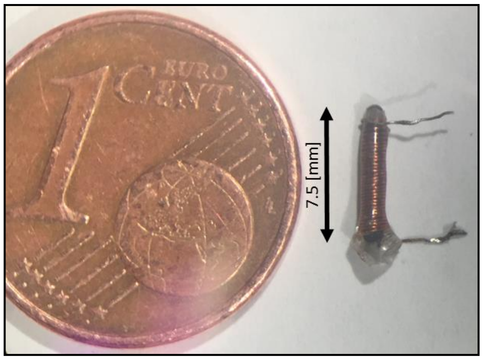

Fig. 2 View of the micro-coil used for power transfer.

B. Receiving coil

A 0.75 mm-diameter, 7.5 mm-length NiZn ferrite rod from *Fair-Rite Products*, Corp., Wallkill, NY, US, with a relative permeability μ_r of 35 is choosen as core for the tiny-coil and 31 turns of 100 µm-diameter coated-wire are manually wound around the rod in a single layer. An optical UV-curing adhesive is then added to fix the wire and to make the coil robust. Fig. 2 shows the resulting 1 mm-diameter, 7.5 mm-length handmade tiny-coil.

C. Operating frequency and passive components

The two variables of interest in the system are the efficiency η and the RMS supply voltage V_s to obtain a RMS voltage V_l on the load R_l . These variables are computed as

$$\eta = \frac{P_l}{P_s} = \frac{re(V_l I_l^*)}{re(V_s I_s^*)} = \frac{V_s^2 re(Y_{sl} R_l Y_{sl}^*)}{V_s^2 re(Y_t^*)} = \frac{re(Y_{sl} R_l Y_{sl}^*)}{re(Y_t^*)} \quad (2)$$

$$V_s = V_l \sqrt{\frac{1}{re(Y_{sl} R_l^2 Y_{sl}^*)}} \quad (3)$$

where Y_t is the total admittance of the system seen by the source, Y_{sl} is the transadmittance between the source and the load, P_s the active power provided by the source and P_l the active power dissipated by the load. A genetic algorithm is then used to select the values of the capacitors and the operating frequency that produce the best η with V_s as low as possible for specific coils, k and R_l. Fig. 3 shows the *Pareto front* given by the algorithm.

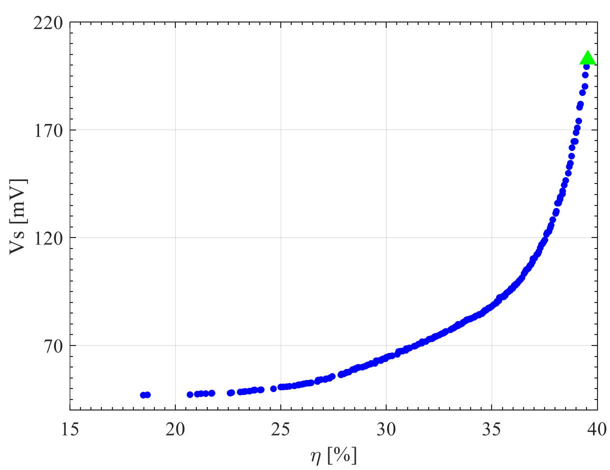

Fig. 3 Pareto Front obtained by the genetic algorithm for a 560 Ω load at the secondary that dissipates 1 mW of power, with the best efficiency solution (▲).

III. Experimental Results

A wireless charging coil (Model 760308102210, from *Würth Electronics*, Inc., Niedernhall, Germany) has been chosen as primary coil. It is a 3.7 cm-diameter pancake coil on a sheet of ferrite. The secondary handmade tiny-coil has been characterized using an E5061B ENA vector network analyser. Its measured inductance is 3.56 µH and its resistance varies from 200 mΩ at dc to 8.45 Ω at 14 MHz.

The cables used to measure the coupling factor have a capacitance of 120 pF leading to a resonance with the measurement setup for the secondary coil around 8 MHz. The primary coil is connected to an Agilent 33120 that generates V_s and a Tektronix TDS7104 scope is used to measure the voltage amplitudes V_{Ls} and V_{Ll}. The secondary coil is placed at a distance d perpendicular to the primary coil. Fig. 4 shows that for d = 10 mm, at 1.5 MHz, a k of 3% is reached thanks to the ferrite while k is only of 0.1% in the case of a similary-sized secondary coil without ferrite.

The WPT prototype is optimized for a load of 560 Ω and a distance between the primary and the secondary of 10 mm corresponding to our target study case. The capacitors determined by the algorithm are 3.52 nF, 6.7 nF and 940 pF respectively for C_{ss}, C_{pl} and C_{sl} and an operating frequency of

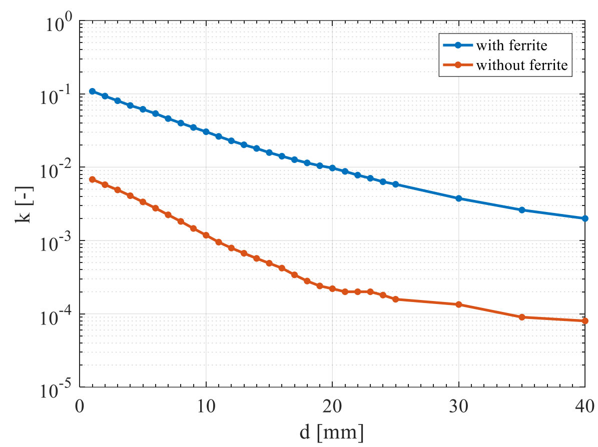

Fig. 4 View of the coils coupling factor k varying with the distance between the coils d for a secondary µ-coil with and without ferrite core.

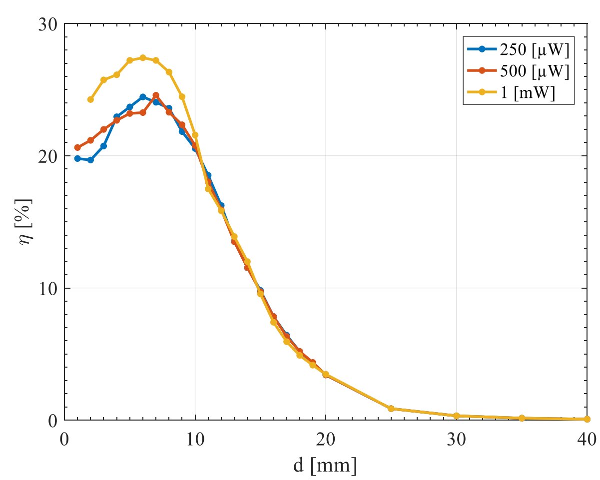

Fig. 5 View of the WPT efficiency η for a 560 Ω load at 3 power levels on the load.

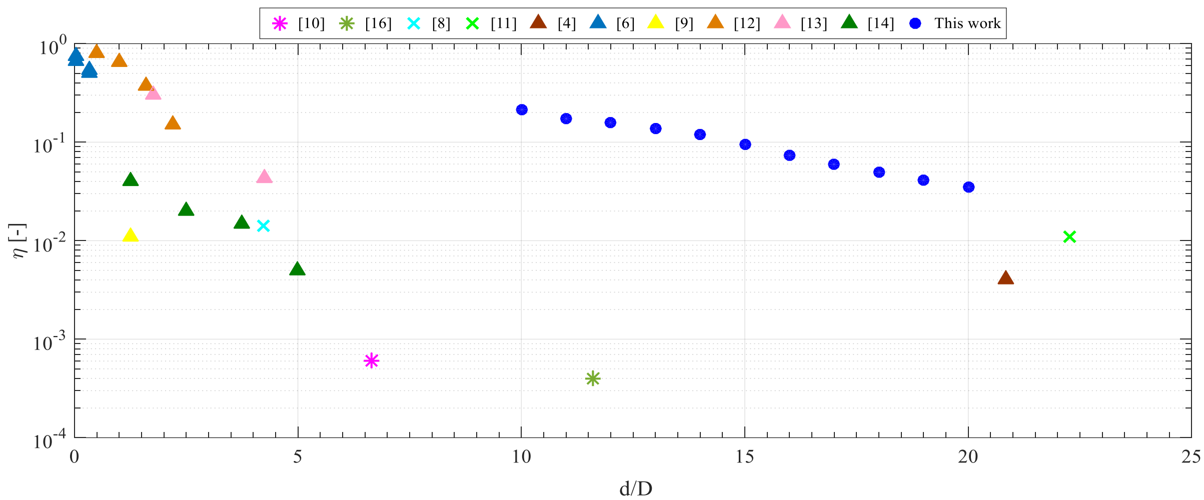

Fig. 2 Comparison between this work and the literature in terms of efficiency η compared to the ratio of the distance between the coils and the receiver diameter *d/D*. Shapes of the data are relative to the study operating frequencies range: (*) UHF, (x) VHF, (▲) HF, (●) MF

1.008 MHz. The prototype has been tested with three different power levels transmitted to the load, i.e. 1 mW, 500 µW and 250 µW, and with a distance between the coils, varying from 1mm to 40 mm. Fig. 5 shows the result of the measurements. At 10 mm, efficiencies of 21.6%, 20.8% and 20.5% are reached respectively for 1 mW, 500 µW and 250 µW. We have verified that inserting a 1 cm-thick slice of ham (to mimic the effect of tissues) has no significant influence at such low frequency. Reduction of efficiency for lower distances is due to the change of mutual and self-inductance of the coils that depend on the distance between the two ferrite cores. Fig. 6 shows a comparison between this work and the literature in terms of efficiency related to the distance between the coils and the diameter *D* of the receiving coil. At similar *d/D* ratio, the efficiencies reached in this study are observed to lie one order of magnitude higher than [3], two orders greater than [15] and similar to [5, 11, 12] but a 10x higher *d/D* ratio. It is also visible that MF and HF frequencies appear to lead to the highest known efficiencies for the study case we consider, when compared with much higher frequencies targeted in other works.

IV. Conclusion

A resonant wireless power transfer system for miniaturized biomedical injectable capsule implants, with ferrite-core tiny-coil receiver, has been designed, optimized and prototyped. The tiny-coil achieves an inductance of 3.65 µH for a 1mm diameter and a 7.5 mm length leading to a coupling factor of 3% at 1cm of the transmitting coil. With this *k*, the optimized resonant design allows to provide power levels of 1 mW, 500 µW and 250 µW with respective efficiencies of 21.6%, 20.8% and 20.5%, to a 560 Ω load at 1 MHz. Comparison with literature shows a greater interest in the MF-HF frequencies and the use of ferrite as core for actually using implanted power-receiving coils with mm-size diameter.

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