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# Limits for increasing the WPT distance in AIMDs

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## Abstract

Wireless Power Transfer (WPT) can be used in Active Implantable Medical Devices (AIMDs) to reduce battery size and/or extend durability. Typically, the power transmitter (Tx) is placed over the patient's skin during the recharge process. In this paper, we address the main challenges to overcome when trying to extend the distance between the Tx and the implant. An experimental example is presented which can deliver 5 mW at 30 cm. The specific absorption rate (SAR) was within the regulatory limits according to the evaluation performed with sim4life software by Zürich MedTech.

## 1 Introduction

Traditionally, primary batteries are used to power Active Implantable Medical Devices (AIMDs). However, the power that can be stored in a battery, considering the limited available space, is not enough in many applications, e.g. neurostimulators. In those cases, secondary batteries can be used [1], coupled with inductive Wireless Power Transfer (WPT) to recharge the battery or, even, power the device. This has enabled new applications and improvements in its size and battery lifetime.

A basic diagram of an inductive WPT system is presented in Fig. 1. A Transmitter (Tx) circuit (Tx-circuit) generates a sinusoidal current in the Tx coil, $I_{\text{Tx}}$, creating an alternating magnetic field, $H$, which induces a voltage in the receiver (Rx) coil. The Rx circuit (Rx-circuit) is powered from that induced voltage.

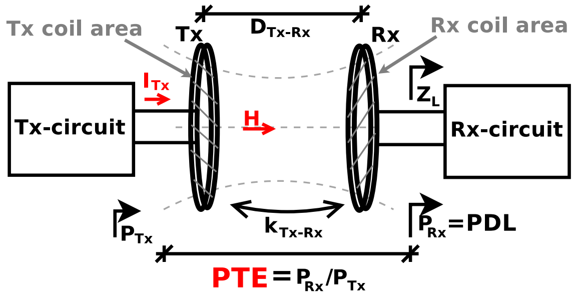



**Figure 1.** Block diagram of inductive power transmission.

The main parameters that define the WPT link are: 1) The distance between Tx and Rx, $D_{\text{Tx-Rx}}$, 2) The Tx and Rx areas 3) The Power Delivered to the Load (PDL) and 4) The Power Transmission Efficiency (PTE). In AIMDs, the Rx size is very limited, therefore the smaller the Rx coil, the better. Regarding distance, most of the current commercially available AIMDs are recharged resting the transmitter over the patient's skin, with a maximum transfer distance of a few cms. However, if the link supports larger distances, this would have two significant benefits. First, in traditional rechargeable AIMDs, it would make the recharge process to be more comfortable for the patient, as in the case study presented later in this paper. Second, it would allow to recharge smaller implants placed deeper inside the body, which is a interesting goal to be able to act closer to the target organ.

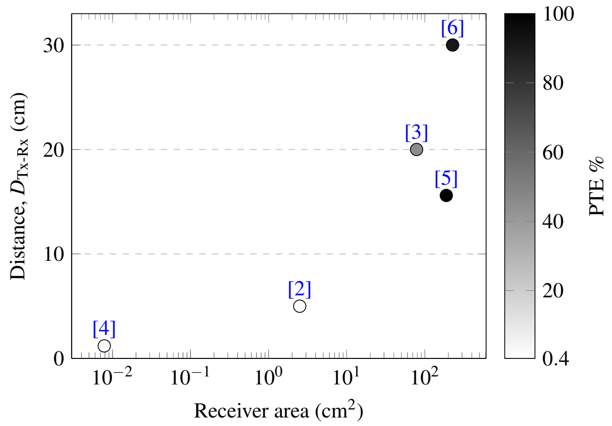


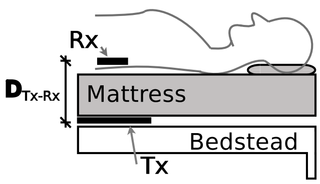

**Figure 2.** 2-coil Link state-of-the-art. Distance, Rx coil size and Link efficiency PTE.

In Fig. 2, the Tx to Rx distance and PTE achieved as a function of the Rx coil area is presented for relevant works in the state-of-the-art. As can be seen, there is a lack of papers that reach long distances with small Rx. In this paper, we address the challenges to overcome to work in that up-left deserted area of Fig. 2 which is of interest for AIMDs.

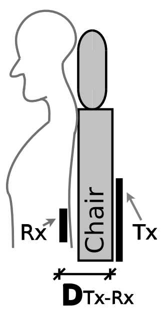
The rest of the paper is organized as follows. First, in Section 2, the case study in this paper is further described. Then in Section 3, the different limits that prevent us from achieving long distances with small Rxs are discussed. Finally, a practical example of implementation is presented in Section 4 followed by the conclusions.

## 2 Description of case study

A charging system for AIMDs that do not require placing the Tx over the patient skin, would enable more comfortable and non-invasive recharging schemes, as the ones represented in Fig. 3. For instance, the AIMDs, or wearable devices, could be recharged while the patient is sleeping by placing a Tx under the mattress, or while sitting at work, by placing the Tx behind the chair backrest. Therefore, in this scheme, a power transmission distance of, approximately, 30 cm is required. In this approach, the AIMDs would receive power for several hours during the day, on a daily or almost daily basis, which makes the link less restrictive in terms of PDL. Certainly, the PDL could be lower for moments due to angular misalignment and distance variations. Nevertheless, if, for instance, it is considered that the AIMD (or wearable device) is receiving 5 mW during 5 hours each day, considering this energy and a 100% efficient Rx, then the AIMD can consume, during the whole day, an average power of approximately 1 mW. This power level is in the order of what is required for various AIMD and wearable devices such as hearing aid, analog cochlear processor, and body-area monitoring [1].




**(a)** Power transmission under the mattress




**(b)** Power transmission behind chair backrest.

**Figure 3.** Representation of case study

## 3 Limits for increasing the distance in inductive powering for AIMDs

In this section, the three main issues that limit the distance range in inductive powering are qualitatively addressed, which are: 1) low PTE, Section. 3.1. 2) High output power in the transmitter driver in order to provide the required transmitter current, $I_{Tx}$, Section 3.2. 3) Safety and electromagnetic compatibility (EMC) limits, Section 3.3. These three limitations were highlighted in red in Fig. 1. This analysis is quantified in the example presented in Section 4.

### 3.1 Power transmission efficiency

Assuming the optimum load condition, e.i. optimum $Z_L$ in Fig. 1, the PTE can be calculated as [7]

$$\mathrm{PTE} = \frac{k_{\text{Tx-Rx}}{}^2 Q_{\text{Tx}} Q_{\text{Rx}}}{\left(\sqrt{k_{\text{Tx-Rx}}{}^2 Q_{\text{Tx}} Q_{\text{Rx}} + 1} + 1\right)^2}, \quad (1)$$

where $Q_{\text{Tx}}$ and $Q_{\text{Rx}}$ are the Tx and Rx coils quality factors, respectively, and $k_{\text{Tx-Rx}}$ is the coupling coefficient between coils. The greater $Q_{\text{Tx}}$, $Q_{\text{Rx}}$ or $k_{\text{Tx-Rx}}$, the higher the PTE.

The coupling coefficient could be estimated as [8]

$$k_{\text{Tx-Rx}} = \frac{r_{\text{Tx}}{}^2 . r_{\text{Rx}}{}^2}{\sqrt{r_{\text{Tx}} . r_{\text{Rx}}}\left(\sqrt{D_{\text{Tx-Rx}}{}^2 + r_{Tx}{}^2}\right)^3}, \quad (2)$$

where $r_{\text{Tx}}$ and $r_{\text{Rx}}$ are the Tx and Rx coils radius respectively, and it is assumed that $r_{\text{Tx}} > r_{\text{Rx}}$. Therefore, the larger the distance, the lower the coupling coefficient, and the lower the PTE. Additionally, the smaller the Rx, as desired in AIMDs, also the lower the coupling coefficient and the PTE.

To increase the low coupling coefficient, and thus improve the PTE, the optimum Tx radius, $r_{\text{Tx}}$, should be selected. This optimum value can be obtained derivating (2) as follow

$$\frac{\partial k_{\text{Tx-Rx}}}{\partial r_{\text{Tx}}} = 0 \Leftrightarrow r_{\text{Tx}} = D_{\text{Tx-Rx}}. \quad (3)$$

The coupling coefficient is maximized if the Tx radius equals the distance between coils. This means that if a $D_{\text{Tx-Rx}} = 30$ cm is desired, the optimum transmitter diameter is 60 cm which may be too large in certain use cases. Additionally, the larger the coil, the greater its self-inductance, $L_{\text{Tx}}$, which may bring the self-resonance of the coil, $f_{self\text{-}res}$, underneath the desired operating frequency ($f_{self\text{-}res} = \frac{1}{2\pi\sqrt{L_{\text{Tx}} C}}$, where $C$ is the parasitic capacitance). If that is the case, the parasitic capacitance dominates over the coil and WPT is extremely inefficient. This issue is numerically exemplified in the practical example of implementation presented in Section 4.

Summarizing, although some measures can be taken to alleviate it, low PTE is expected when the distance between the Tx and Rx is pushed to the limit. This means that most of the power consumed is transformed into heat in the Tx coil instead of reaching the load. As a consequence, to meet the required PDL $= P_{\text{Tx}} \times$ PTE, the transmitter output power, $P_{\text{Tx}}$, should be increased. Increase the Tx output power brings the two following problems addressed in Sections 3.2 and 3.3.

### 3.2 Transmitter power amplifier

In inductive powering, usually class-D or class-E amplifiers are used to drive the Tx coil. When its output power is increased, the following practical issues arise. 1) Heat dissipation. 2) Hazardous voltages and/or currents, which have to be withstood by the circuit components, and the hazardous voltages need to be correctly insulated from the user. 3) Size and weight increase: to bear the heat dissipation and insulation. 4) Cost increase. All these issues set practical limits to the Tx power.

### 3.3 Safety and EMC

The electromagnetic fields generated in the Tx coil and the overall link arrangement, must be such that the system is safe from the electromagnetic exposure point of view for the patient and other people that may be in the proximity of the link. The system must also comply with EMC limits in order to not disturb other systems and be approved by the applicable regulatory authority. Both limits (EMC and safety) vary slightly between different countries (in the case of safety see [9, 10]).

Regarding EMC limits (see e.g. FCC CFR 47.18.305), we will consider operation in the ISM band, in particular at 13.56 MHz, therefore emissions in the frequency band (which will be the desired one in WPT) are not limited. There is a limitation of the out of band field (25 uV/m at 300 m) but compliance with this will depend on the harmonic filtering in the Tx power amplifier, whose design is out of the scope of this paper.

Regarding safety, the Specific Absorption Rate (SAR) limits that will be taken as reference here are [9, 10]: Local SAR (averaged on a 1 g cube of tissue) $< 1.6$ W/kg Whole Body SAR $< 0.08$ W/kg considering the more restrictive case of the general public.

In the next section we will consider the impact of these limits in our study case.

## 4 Practical example of implementation

In this section, an example of implementation is presented, which is capable of transmitting 5 mW at 30 cm, taking into account the challenges previously discussed in Section 3.

The selected carrier frequency is 13.56 MHz. Many WPT systems are designed at this frequency since it allows the implementation of high-quality factors coils with small sizes, which are required in AIMD applications. In addition, this frequency is within the Industrial Scientific and Medical (ISM) band, which simplifies EMC compliance.

The Tx coil radius is the optimum one obtained from (3), $r_{\text{Tx}} = D_{\text{Tx-Rx}} = 30$ cm, and it has only one turn. This Tx has a self-inductance of $L_{T\text{x}} = 2.2$ $\mu$H, thus the required resonance capacitor is $C_{T\text{x}} = 62.6$ pF, to resonate at 13.56 MHz $=1/(2\pi\sqrt{L_{T\text{x}}C_{T\text{x}}})$. If a turn is added (thus having two turns) the self-inductance is approximately multiplied by four to $L_{T\text{x}} = 8.8$ $\mu$H, as it is proportional to the number of turns squared, and the required resonant capacitor is divided by four to 15.65 pF. Such a small capacitor is in the order of magnitude of the parasitic capacitance of the coil due to turn to turn and connections capacitance. Therefore, if this large coil (30 cm radius) has more than one turn, its self-resonance would be below the desired operating frequency, making the WPT link extremely inefficient as discussed in Section 3.1.

**Table 1.** Parameters of the practical example.

| Parameters | | Value |
|---|---|---|
| Tx coil Diameter, ⌀ 60 cm | $L_{T\text{x}}$ | 2.2 $\mu$H |
| | $Q_{T\text{x}}$ | 71 |
| Rx coil □ 25 × 25 mm | $L_{R\text{x}}$ | 877 nH |
| | $Q_{R\text{x}}$ | 34 |
| Distance, $D_{\text{Tx-Rx}}$ | | 30 cm |
| Coupling coefficient, $k_{T\text{x-}R\text{x}}$ | | 0.0012 |
| Carrier frequency, $f_S$ | | 13.56 MHz |
| Load power $P_L$ | | 5 mW (5.3 $V$@5.6 $k\Omega$) |
| Peak Tx current $I_{T\text{x}}$ | | 1.96 A |
| Tx power $P_{T\text{x}}$ | | 5.56 W |
| PTE = $P_L/P_{T\text{x}}$ | | 0.09 % |

The Rx coil is a commercial coil from Pulse/Larsen, model 7001, which is a flexible 5 turns 25x25 mm coil.

The implemented system, using the described Tx and Rx, is shown in Fig. 4 and further described in the caption. The main parameters of the system are presented in Table 1. This initial evaluation only measured operation in air.

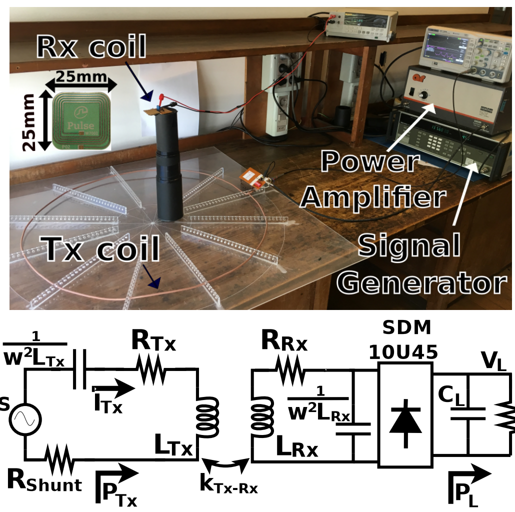



**Figure 4.** Measurement setup and circuit schematic. The $R_{T\text{x}}$ and $R_{R\text{x}}$ are the equivalent series resistance of Tx and Rx coils respectivelly. The $R_{shunt} = 2.5$ $\Omega$ was included to measure the Tx current. The Rx has a full-wave rectifier using SDM10U45 schottky diodes.

As mentioned in Section 3.1, the low PTE, 0.09 % in this case, makes it necessary to consume 5.56 W in the Tx to receive the required 5 mW. This power consumption is feasible if the Tx is powered from the power grid, which is possible in the case study of this paper, Section 2. In this example, we used a power amplifier from AR RF/Microwave instrumentation as the design of the Tx driver is out of the scope, but examples of implementations exist in the state-of-the-art at this frequency and even with higher output power levels [11]. Due to the relatively large current required in the Tx, $I_{T\text{x}} = 2$ $A$ (peak), the Tx resonant capacitors, $C_{T\text{x}}$, have to withstand around 400 V (ac peak), which should be taken into account.

Regarding safety limitations, the SAR was simulated using sim4life [12]. Although the user is supposed to be, approximately, at 30 cm far from the Tx coil, the worst-case situation of the patient (or any person) laying over the Tx coil (1 cm distance due to Tx case) was considered. The SAR in that situation, with a Tx current of 2 A is presented in Fig. 5. This simulation was done using the heterogeneous Duke virtual population model from sim4life and the SAR was calculated as the average over 1 g of tissue. Even in that worst-case situation, the maximum SAR was 1.13 W/kg which is below the more restrictive (general public) limit, presented in Section 3.3, of 1.6 W/kg. Regarding the whole body limitation, we obtain 0.035 W/kg which is also below its limit of 0.08 W/kg. Additionally, a shielding could be implemented, e.g., behind the chair backrest, see Fig. 3b, avoiding unnecessary magnetic field in the surroundings.

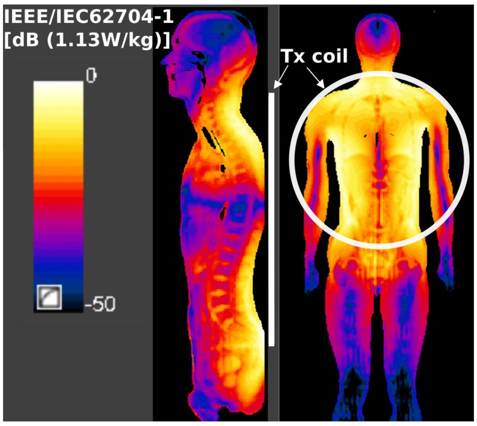



**Figure 5.** SAR simulation using Sim4life [12], Tx current $I_{Tx} = 2$ A and distance $D_{Tx\text{-}Rx} = 30$ cm

## 5 Conclusion

The main challenges that arise when extending the distance between power Tx and AIMDs were discussed. The very low PTE, due to the low coupling coefficient, forces to increase the Tx output power complexing its design and, eventually, reaching the safety limitations (SAR). A simple practical example of implementation was experimentally verified. It was shown that it is possible to deliver 5 mW to a 25 mm by 25 mm reception coil at 30 cm of distance. The simulation evaluation of this design predicts that it respects the SAR safety limits without any special measure to prevent humans to be very close to the Tx coil.

## 6 Acknowledgements

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## References

[1] A. P. Chandrakasan, N. Verma, and D. C. Daly, “Ultralow-power electronics for biomedical applications,” *Annu. Rev. Biomed. Eng.*, vol. 10, pp. 247–274, 2008.

[2] R. F. Xue, K. W. Cheng, and M. Je, “High-efficiency wireless power transfer for biomedical implants by optimal resonant load transformation,” *IEEE Trans. Circuits Syst. I*, vol. 60, no. 4, pp. 867–874, April 2013.

[3] L. Chen, S. Liu, Y. C. Zhou, and T. J. Cui, “An optimizable circuit structure for high-efficiency wireless power transfer,” *IEEE Trans. Ind. Electron.*, vol. 60, no. 1, pp. 339–349, Jan 2013.

[4] A. Ibrahim and M. Kiani, “A figure-of-merit for design and optimization of inductive power transmission links for millimeter-sized biomedical implants,” *IEEE Trans. Biomed. Circuits Syst.*, vol. 10, no. 6, pp. 1100–1111, 2016.

[5] H. Kim, C. Song, D. H. Kim, D. H. Jung, I. M. Kim, Y. I. Kim, J. Kim, S. Ahn, and J. Kim, “Coil design and measurements of automotive magnetic resonant wireless charging system for high-efficiency and low magnetic field leakage,” *IEEE Trans. Microw. Theory Techn.*, vol. 64, no. 2, pp. 383–400, Feb 2016.

[6] S. H. Lee and R. D. Lorenz, “Development and validation of model for 95%-efficiency 220-W wireless power transfer over a 30-cm air gap,” *IEEE Trans. Ind. Appl.*, vol. 47, no. 6, pp. 2495–2504, Nov. 2011.

[7] P. Pérez-Nicoli and F. Silveira, “Maximum efficiency tracking in inductive power transmission using both matching networks and adjustable ac–dc converters,” *IEEE Transactions on Microwave Theory and Techniques*, vol. 66, no. 7, pp. 3452–3462, 2018.

[8] K. Finkenzeller, *RFID handbook*. John wiley & sons.

[9] H. M. Madjar, “Human radio frequency exposure limits: An update of reference levels in europe, usa, canada, china, japan and korea,” in *2016 International Symposium on Electromagnetic Compatibility-EMC EUROPE*. IEEE, 2016, pp. 467–473.

[10] A. Christ, M. Douglas, J. Nadakuduti, and N. Kuster, “Assessing human exposure to electromagnetic fields from wireless power transmission systems,” *Proceedings of the IEEE*, vol. 101, no. 6, pp. 1482–1493, 2013.

[11] W. Chen, R. Chinga, S. Yoshida, J. Lin, C. Chen, and W. Lo, “A 25.6 w 13.56 mhz wireless power transfer system with a 94% efficiency gan class-e power amplifier,” in *2012 IEEE/MTT-S International Microwave Symposium Digest*. IEEE, 2012, pp. 1–3.

[12] Sim4life by zmt. [Online]. Available: www.zurichmedtech.com