Circularly polarized RF coil for energy harvesting in clinical MRI

P. S. Seregin¹, O. I. Burmistrov¹, G. Solomakha¹, E.I. Kretov¹, N.A. Olekhno¹, A. Slobozhanyuk¹

¹Department of Physics and Engineering, ITMO University, 197101 Saint Petersburg, Russia
E-mail: pavel.seregin@metalab.ifmo.ru

Abstract. Radiofrequency (RF) harvesting is a promising technology for the wireless power supply of various in-bore devices used in magnetic resonance imaging. However, current technical solutions in this area are based on the conversion of linearly polarized RF fields, and thus their efficiency is limited, as they interact only with a fraction of circularly polarized RF fields. In the present work, we introduce and experimentally realize a novel harvesting setup allowing for converting circularly polarized RF fields to direct current.

1. Introduction
Magnetic resonance imaging (MRI) offers great perspectives in medical diagnostics. Nevertheless, some specific MRI tasks, such as abdominal imaging, require gating and, as a result, patient parameter tracking. For this, various in-bore devices are often used, including respiration sensors and heart rate monitors, which in most cases need an external power source [1, 2, 3]. However, powering these devices with cables can create imaging artifacts and decrease the comfort of patients [4]. As a result, various strategies of wireless power supply are actively developed. RF harvesting is one of the wireless supply technology. It relies on the conversion of RF field $B_1$ that is readily present in the MRI scanner, thus not requiring additional transmitting coils [1].

The use of RF harvesting in MRI poses several difficulties. For example, currents flowing in the harvesting circuit potentially can distort the magnetic fields $B_1$ and $B_0$ [6], which can result in the image quality decrease. Moreover, the majority of harvesting setups for MRI exploit only one linear polarization component of the transmit field [5]. However, the $B_1$ field is circularly polarized, and thus the efficiency of such devices is limited. In the present paper, we summarize these main challenges and aim to overcome some of them by proposing a harvesting setup capable of utilizing a circularly polarized field. Here, we realize and experimentally test the harvesting coil for clinical MRI and study the effects of its location within the scanner bore.

2. Drawbacks in linearly polarized harvesting for MRI applications
The main disadvantage of MRI RF harvesting is its low efficiency. Despite the significant peak power of the radio transmitter (10-25 kW), the average power that RF Harvesting could collect is about 100 mW. This is because a significant part of power should be dissipated in the human body to provide excitation of nuclei. On the one hand, it is possible to increase the power of the received energy by increasing the area of the harvesting coil. However, an increase in the
Figure 1. Design of the proposed circular field harvesting setup consisting of two coils, one in the shape of 0 (shown in red) and another one in the shape of 8 (shown in black). The setup includes two tuning capacitors and two rectifiers for each of the coils, along with a combiner. The grounding resistor emulates the load powered with the harvesting setup.

coil area can lead to distortion of the $B_1$ field and a rise in the specific absorption rate (SAR). Besides, DC after the rectifier may affect the $B_0$ field. Since it is a source of secondary constant magnetic fields.

Another factor that influences the power level of MRI RF harvesting is the pulse sequence used during the MRI examination. A powerful RF excitation pulse is present for a short time, usually no more than 10 ms, while the pause between the excitation pulses can be 100 ms.

3. Improving MRI harvesting efficiency with circular polarization conversion

To evaluate the impact of the pulse sequence, a series of experiments were performed with Siemens Espree 1.5 T MRI scanner. It was observed that the harvesting voltage for gradient-echo sequences (GRE) were small because of the low power examination (low flip angle (FA)) and only one pulse used for excitation. However, for some pulse sequences, such as the T2 turbo spin-echo (TSE) sequence, harvesting worked better due to the high-power examination (high flip angle), many pulses in the excitation phase, and low TR.

To get circularly polarized and, as a result, improve efficiency, we suggested a design with two coils with orthogonal $B_1$ fields. Design of coils are illustrated in Figure 1. Each coil receives the corresponding linear component of the circularly polarized $B_1$ field. Construction of each coil includes tuning capacitors to adjust the resonant frequency. The currents induced in the coils are rectified with the diodes and then combined together in the bridge-based combiner and go to the load. Its resistance was chosen to be 200 Ohms, which is equivalent to a DC load of a low noise amplifier (LNA) and a small RF transmitter to create a wireless communication channel. In experimental realization, we use the following coil parameters: length $L = 110$ mm, width $W = 67$ mm, and height $H = 32$ mm. Photo of the experimental realization is shown in Fig. 2(a).

It is essential to find an optimal position of the designed coil within the bore of the scanner to provide maximum energy harvesting and simultaneously eliminate the effect of the MRI RF harvesting setup on the $B_0$ and $B_1$ fields. To find the optimal location, the voltage dependence was evaluated, obtained with a harvesting setup as a function of the displacement of setup from the scanner isocenter. Since the region of the uniform RF field of excitation depends on the size of the body coil, a comparison was made between the Siemens Espree (red curve) and the Siemens Avanto (blue curve), which are equipped with different types of body coils. The output
voltage was measured using Owon XDS3202A 14-bit oscilloscope attached by a coaxial cable with multiple RF cable traps to prevent the cable effect. We measured the output voltage with a single excitation pulse, as RF harvesting should provide voltage to the local coil circuitry during each phase encoding step. To this end, we applied a manual frequency adjustment procedure employing three separate pulses. However, only the first one was used to measure the harvesting output. To estimate the region in which the harvesting setup maintains stable functionality, we select a 15% drop in the output voltage as criteria. Therefore, the realized harvesting coil can be freely moved within $\pm 20$ cm range for Espree setup and within $\pm 22.5$ cm range for Avanto setup without considering a change in the output voltage.

To predict the effect of the harvesting setup to $B_1$ distribution, an experimental study was performed. We investigated different locations of the harvesting coil to ensure that the harvesting setup does not impact the MRI functionality. We use the method estimation $B_1$ map with two GRE sequences [7]. The experiment was performed using Siemens large spherical phantom. It has a diameter of 17 cm and part number 04761065. The phantom was placed at the isocenter $Z = 0$ for Espree MRI scanner. Three $B_1$ map were measured for two cases: without harvesting setup(Fig. 3(a)), with a harvesting coil at the distance of 1 cm (Fig. 3(b)) and 3 cm (Fig. 3(c)) from the edge of the phantom. It was found that for the case when harvesting is located at a
distance of 1 cm, currents induced in the harvesting coil create local $B_1$ field inhomogeneity. However, if the distance between the harvesting coil and the phantom equals 3 cm, then the field again appears unperturbed. It is important to note that the given distance from the phantom depends on the size of the harvesting coil and the amplitude of the induced current.

4. Conclusion
This paper proposes a novel design of a RF harvesting setup based on two coils with an orthogonal magnetic field. Such a design is capable of converting a circularly polarized RF magnetic field to DC. Such design is promising for MRI applications due to the circular polarization of $B_1$ RF field in MRI scanners. As we have demonstrated with our experiments, the developed harvesting setup does not distort the $B_1$ field in the region of scan and can provide DC in a specific range of positions along MR-scanner isocenter.

The measurements demonstrated that it is possible to double voltage almost two times at the harvesting output with circular polarized RF coil. The reason for lower DC output could be in the losses in the voltage. The peak voltage during one excitation RF pulse reaches 100 V on the load 200 Ohm. Therefore the peak power is approximately 50 W. However, the duration of the exciting pulse is no longer than 5 mS, and the time repetition (TR) pulse time usually is 40-400 mS. The estimated average power at each step of the phase encoding cycle will be about 100-500 mW, and it depends on the selected pulse sequence. The proposed RF harvesting setup can be used to provide DC supply for cable-free receive arrays or wireless sensors.

Acknowledgments
This work was supported by the Russian Science Foundation (Project No. 21-79-30038).

References
[1] Nohava L, Ginefri JC, Willoquet G, Laistler E, Frass-Kriegl R. “Perspectives in Wireless Radio Frequency Coil Development for Magnetic Resonance Imaging” 2020 Frontiers in Physics 8, 11
[2] Mandal R, Babaria N, Cao J, Lu KH, Liu Z. ”MRI Powered and Triggered Current Stimulator for Concurrent Stimulation and MRI” 2019 bioRxiv, 715805
[3] Ganti A, Lin J, Wynn T, Ortiz T. ”Achieving electromagnetic compatibility of wireless power transfer antennas inside MRI system” 2019 Wireless Power Transfer 6, 138–153
[4] Shchelokova A, et al. ”Ceramic resonators for targeted clinical magnetic resonance imaging of the breast” 2020 Nature communications 11, 1–7
[5] Venkateswaran M, Kurpad K, Brown J E, Fain S, van der Weide D. ”Wireless Power Harvesting During MRI” 2020 2020 42nd Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC)
[6] Twieg M, de Rooij MA, Griswold MA. ”Active detuning of MRI receive coils with GaN FETs” 2015 IEEE Transactions on Microwave Theory and Techniques 63, 4169–4177
[7] Insko E, Bolinger L. ”Mapping of the radiofrequency field” 1993 Journal of Magnetic Resonance, Series A 103, 82–85