# RF Excitation in 7 Tesla MRI Systems Using Monofilar Axial-Mode Helical Antenna

M. M. Ilić
School of Electrical Engineering
University of Belgrade
Belgrade, Serbia

A. A. Tonyushkin
Radiology Department
Massachusetts General Hospital
Harvard Medical School
Boston, Massachusetts, USA

N. Šekeljić, P. Athalye, B. M. Notaroš
Electrical & Computer Engineering Dept.
Colorado State University
Fort Collins, Colorado, USA
notaros@colostate.edu

***Abstract*—We present a novel RF exciter for traveling-wave magnetic resonance imaging (MRI) systems based on a monofilar axial-mode helical antenna. In specific, we present a bore-extended, subject-loaded monofilar helical antenna for 7-T (ultra-high magnetic field) MRI systems. By means of rigorous full-wave electromagnetic modeling, we confirm clear advantages of the novel exciter over the existing TW excitation methods, and its excellent performance in terms of circular polarization and uniformity of the RF magnetic field in simple phantoms inside MRI bores at 7 T.**

## I. Introduction

Magnetic resonance imaging (MRI) is an established medical diagnostic method and tool widely utilized to obtain high-resolution images of the internal structure of the body or its parts and organs. The physical foundation of MRI is the principle of nuclear magnetic resonance, whereby atom nuclei of the tissue that is imaged absorb and reemit applied radio-frequency (RF) radiation based on the resonant radian frequency with which the spin precesses in an external polarizing static (dc) magnetic field (magnetic flux density), $B_0$, the so-called Larmor frequency, $f_0$, that is proportional to $B_0$ [1]. The main polarizing static field $\mathbf{B}_0$ is generated by the main coil in the scanner, known as magnet, and its direction is longitudinal, along the MRI bore. An RF excitation magnetic field, $\mathbf{B}_1$, is applied in orthogonal direction to the main polarizing field, so in the transversal directions with respect to the MRI bore axis, to alter alignment of spins and induce an echo signal that is used in MRI. For maximum coupling between the RF field and the spins in the tissue, a rotating magnetic field with constant magnitude during rotation is desired, that is, the RF exciter needs to generate a circularly polarized (CP), and more precisely, right-handed CP RF magnetic field, usually denoted as $\mathbf{B}_1^+$.

## II. RF Coils for Traveling-Wave Ultra-High-Field MRI Systems

In 3-T clinical MRI scanners, the field $\mathbf{B}_1$ is generated by an RF exciter – the so-called RF coil – almost exclusively in the form of a birdcage coil. At ultra-high $B_0$ fields, RF is no longer purely concentrated in the near-field due to the shorter wavelengths, which negatively impacts the performance of traditional near-field whole-body coils. Therefore, the next-generation MRI systems at ultra-high magnetic fields ($B_0 > 3$ T) and ultra-high Larmor frequencies ($f_0 > 127.8$ MHz) utilize RF as traveling waves (TWs) in the bore and the images of subjects are generated (and received) by far-field coils, namely, by excitation probes that essentially operate as antennas, in place of the traditional quasi-static, near-field RF coils [2].

The generation and control of TW $\mathbf{B}_1$ RF fields inside a MRI bore and a phantom (or a subject under MRI imaging) is a new area of research. Significant work has been done on TW excitation using loops as antenna probes, e.g., [3], [4]. In addition, the state-of-the-art TW antennas as implemented at $B_0 \geq 7$ T include circular- or square-shaped patch antennas that excite linear or CP TW fields inside a scanner's bore [5]. This excitation of TW, however, if not aided by additional dielectrics or metamaterials, is highly localized, which results in rapid power dissipation in the body and quick attenuation with distance away from the antenna.

In this summary, we present a novel TW MRI exciter whose operation is based on the axial-mode helical antenna, which naturally provides a dominantly CP electromagnetic wave in the greater part of the domain encompassed by the helical coils. In specific, we present a bore-extended, subject-loaded monofilar helical antenna for 7-T MRI systems ($B_0 = 7$ T). The presented helical antenna exciter can be used for small animal, partial human body or full body imaging, provided that appropriate modifications (e.g., adding additional helical filaments and modifying the parameters of the turns, such as the size and pitch) are done.

## III. Bore-Extended, Subject-Loaded Monofilar Helical Antenna as Traveling-Wave MRI RF Exciter, Numerical Results and Discussion

We consider a cylindrical bore, shielded by a perfect electric conductor (PEC), and open at both ends (in essence a section of the open-ended circular waveguide), as shown in Fig. 1. The bore is loaded with a cylindrical phantom filled with saline solution. The operating frequency is $f_0 = 300$ MHz (Larmor frequency for a 7-T magnet, $B_0 = 7$ T). The right-handed CP RF magnetic field is often achieved using the loop exciters, where, however, the desired field in the phantom is achieved only in the close vicinity of the antenna and decays

978-1-4799-7815-1/15/$31.00 ©2015 IEEE 

quickly with the distance away from the antenna, while at the same time degrading the required unit axial ratio of the transversal CP magnetic field. To cope with this problem, we propose a monofilar axial-mode helical antenna exciter as shown in Fig. 1. The antenna is backed by a circular PEC plate located in the bore, centered on the bore axis ($z$-axis), and offset 1.6 cm from the bore opening.

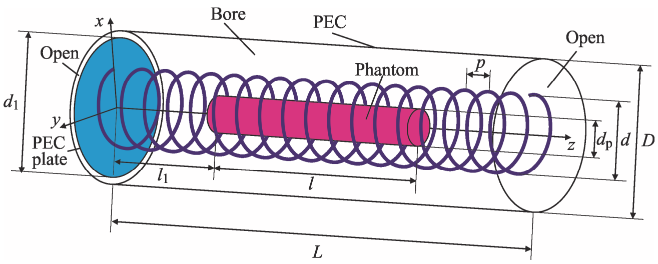


Fig. 1. 7 T MRI bore with a monofilar axial-mode helical antenna bore-extended exciter loaded with a simple phantom: saline solution $\varepsilon_r = 81$, $\sigma = 0.6\ \mathrm{S/m}$, $D = 60\ \mathrm{cm}$, $d = 32\ \mathrm{cm}$, $d_p = 15\ \mathrm{cm}$, $d_1 = 56\ \mathrm{cm}$, $L = 200\ \mathrm{cm}$, $l_1 = l/2 = 50\ \mathrm{cm}$, $p = 11.5\ \mathrm{cm}$, $f_0 = 300$ MHz ($B_0$ = 7 T).

The complete structure is next modeled using the finite element method (FEM), namely, commercial software ANSYS HFSS, and excited through a lumped port connecting the beginning of the first helical turn with the circular PEC plate. The incident power is set to $P_{inc} = 1\ \mathrm{W}$. Shown in Fig. 2 are the right-handed and left-handed CP components of the transversal rms magnetic fields, $|H_{rcp}|$ and $|H_{lcp}|$, respectively, in the phantom. We can conclude based on the figure that the novel helical exciter indeed produces excellent field uniformity throughout the whole length of the phantom with substantially higher right-handed CP component ($\mathbf{B}_1^+$ or $\mathbf{H}_1^+$) of the magnetic field.

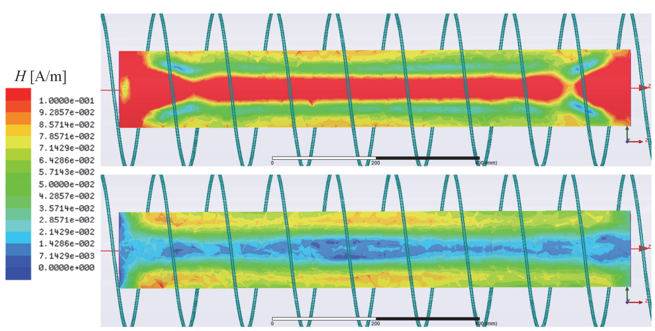


Fig. 2. Field maps of right-handed CP (top panel) and left-handed CP (bottom panel) components of the transverse RF rms magnetic field in the longitudinal cross section of the phantom in Fig. 1 at $f_0$ = 300 MHz ($B_0$ = 7 T). Analysis is done by HFSS (FEM) code.

To further assess the magnetic field distribution, we rigorously compare $|H_{rcp}|$ and $|H_{lcp}|$ along the $z$-axis in the phantom. This comparison is shown in Fig. 3. We can conclude from the figure that high and uniform $|H_{rcp}|$ and very low $|H_{lcp}|$ is achieved throughout the phantom. Very strong fields are present at the beginning and the end of the saline phantom due to strong reflections appearing at abrupt material discontinuities. This behavior can be easily mitigated by introducing caps on both phantom ends in the form of cylindrical or conical buffers made of the same material as the phantom or optimized to provide impedance matching between the phantom medium and surrounding air.

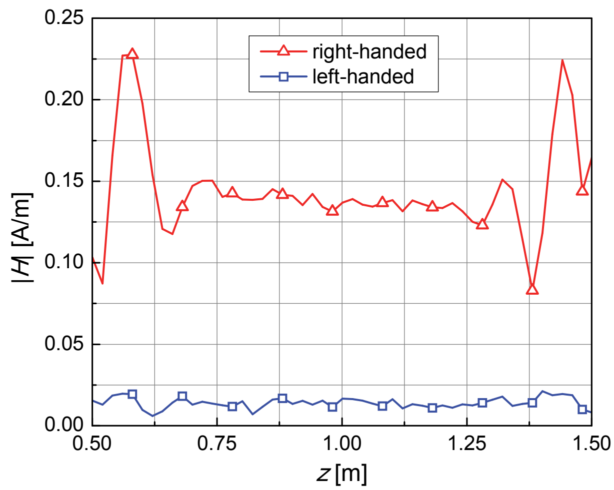


Fig. 3. Distribution of right-handed CP and left-handed CP components of the transverse RF rms magnetic field along the axis of the phantom ($z$-axis) in Fig. 1 at $f_0$ = 300 MHz ($B_0$ = 7 T) – analysis by HFSS (FEM) code.

## Acknowledgement

This work was supported by the National Science Foundation under grant ECCS-1307863 and by the Serbian Ministry of Education, Science, and Technological Development under grant TR-32005.

## References

[1] Z-P. Liang and P. Lauterbur, *Principles of MRI*, IEEE press, NY 2000.

[2] D. O. Brunner, N. De Zanche1, J. Frohlich, J. Paska, K.P. Pruessmann, "Travelling-wave nuclear magnetic resonance," *Nature*, 457 (7232), 2009, 994-U2.

[3] A. Tonyushkin, J. Muniz, S. Grant, and A. Kiruluta, "Traveling Wave MRI in a Vertical Bore 21.1-T System," *Proceedings of the 20th Scientific Meeting of the International Society for Magnetic Resonance in Medicine*, Melbourne, Australia, 2012.

[4] A. Tonyushkin, G. Adriany, D. Deelchand, M. Garwood, and A. Kiruluta, "Below Cut-off Traveling Wave NMR at 16.4T: Interference of Propagating Modes in a High Dielectric-filled Waveguide," *Proceedings of the 53rd Experimental Nuclear Magnetic Resonance Conference*, Miami, FL, 2012.

[5] B. Zhang, G. Wiggins, Q. Duan, D. Sodickson, "Design of a patch antenna for creating traveling waves at 7 tesla," *Proceedings of the 17th Scientific Meeting, International Society for Magnetic Resonance in Medicine*, Honolulu, HI, USA, 2009, p. 4746.