Dipoles and traveling waves

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Target audience: MR technicians, scientists, and engineers with an interest in understanding the physical principles of RF transmission at ultra high field strengths
Objectives: Understand physical principles that enable the use of dipole antennas and travelling wave imaging in ultrahigh field MRI and explore their applications.

With the advent of ultrahigh field imaging, the Larmor frequency has increased up to a value of 300 MHz and beyond. In this frequency range, two applications have become available that were not possible or suitable for the lower field strengths. These applications are 'travelling wave imaging' and the use of dipole antennas as surface array elements.

Travelling wave imaging

One problem with ultrahigh field systems is the occurrence of standing wave effects. Standing waves result in signal voids where signals destructively interfere. These effects make the construction of a body coil at ultrahigh field strengths unfeasible and body imaging has to rely on local transmit arrays with according B_1^+ inhomogeneity. Travelling wave imaging [1] is an approach that can potentially address these issues.

Travelling wave imaging relies on the presence of the cylindrical RF shield within the scanner. Just like clinical MRI scanners, ultrahigh field MRI systems are equipped with a RF shield directly behind the bore wall to shield the gradient coils from the RF systems and vice versa. This RF shield effectively is a hollow conducting cylinder. As such, it is a waveguide. Waveguides are used in the microwave frequency range as low loss transmission lines. Electromagnetic waves can propagate within waveguides following distinct patterns that are called 'modes'. Not all modes can propagate within every cylinder. For a given waveguide diameter, each mode has a minimum frequency: the cut-off frequency. Below this frequency, the mode cannot propagate; it is called 'evanescent'. Above it, the mode propagates along the longitudinal axis of the waveguide. With a bore diameter of 58 cm, the cut-off frequency for a cylindrical waveguide (the RF shield of the scanner) is exactly 293 MHz. This provides opportunities for a totally different way of transmitting and receiving RF signals to and from the imaging subject. With a propagating wave, no standing waves are to be expected, which could realize perfect B_1^+ homogeneity. However, the waves are only travelling in the longitudinal direction. In the transverse direction, standing wave effects still occur which partly downgrades the initial promise of the concept of travelling wave imaging.

Nevertheless, the waveguide concept has several benefits in comparison to surface RF coil arrays. First of all, it has a large field of view limited only by the scanner bore and encoding gradient coils. Second, the distant location of the excitation antenna leads to an improved patient comfort and safety as well as providing space for placement of additional equipment, for example, stimuli for functional MRI. Waveguide behavior of the MR scanner also explains the uncontrollable leakage of the RF signal shown to degrade the TEM body coil efficiency [2] exciting relatively high MR signal in the areas not intended for excitation (e.g. head instead of body torso).

Dipole antennas

Traditionally, B_1 fields in MRI are generated by resonant current-carrying coils. The idea behind this is that only currents can generate B_1 field and electric fields should be avoided at all times to prevent high SAR levels (tissue heating). However, this approach is regarding the RF coils as low-frequency devices. In the RF range, coils should be regarded as antennas. Antennas have a near-field and a far-field. In the near-field, the B_1 field distribution is directly determined by the currents on the coil. The magnetic field distribution can be approximated by the law of Biot-Savart. The extent of the near-field into the tissue scales with the wavelength. With shorter wavelengths (larger frequencies) the extent of the near-field decreases.

Beyond the near-field, a different approach is needed. Here, both electric and magnetic fields exist that propel each other and constitute a propagating wave through the tissue, away from the target. To generate a large field component in this region, a propagating wave needs to be established with preferably as little resonance as possible. Resonance only results in unnecessarily high field components close to the antenna with concomitant high SAR levels while the fields further away from the antenna do not benefit from this field enhancement. On the contrary. The extra conductive losses in the tissue close to the antenna would decrease the efficiency. Resonance should be avoided; the ideal quality factor is one.

Classical RF coils in MRI are typical near-field antennas. They aim to generate high field levels close to the coil by strong resonance. Close to the coil is always relative to the wavelength. At 1.5T and 3T, all imaging targets are still relatively close to the coil, in comparison to the wavelength in tissue of 80 or 40 cm. However, at 300 MHz, deeply located target regions are no longer located within the near-field. Efficient generation of B_1 may require a different approach.

Dipole antennas were not considered appropriate for MRI because they generate inherently electric fields, which should result in large SAR levels. Raaijmakers et al solved this issue by the use of ceramic high permittivity spacers between the antennas and the tissue [8]. The antenna generates both magnetic and electric fields that are oriented such that a propagating wave is emitted towards the imaging target. Therefore, the outer product of electric and magnetic field components: the Poynting vector S, needs to be oriented in the corresponding direction. A similar design was introduced by Winter et al. [9] using water as dielectric spacer and a bow-tie shaped dipole antenna. Since then, the use of dipole antennas in UHF MRI has taken off with many examples of applications e.g. [11-16]. An analysis of optimal current distribution patterns by Lattanzi et al. [10] confirmed the idea that linearly oriented current patterns (dipole antennas) are optimal for deeply located imaging targets at ultrahigh field strengths.

References:
1. D. O. Brunner et al. Nature, 2009, 457, 994
2. J.T. Vaughan et al. Whole-body imaging at 7T: preliminary results. Magn. Reson. Med. 2009;61(1):244-248
3. S. Alt et al. Coaxial waveguide MRI. Magn Reson Med. 2012 Apr;67(4):1173-82
4. A. Andreychenko et al. A, Coaxial waveguide for travelling wave MRI .... Magn Reson Med. 2012 70(3):875-84
5. A. Andreychenko et al. Improved RF performance of travelling wave MR with a high permittivity dielectric lining of the bore. Magn Reson Med. 2012 70(3):885-94
6. D.O. Brunner et al. Traveling-wave RF shimming and parallel MRI. Magn Reson Med. 2011 Jul;66(1):290-300.
7. A. Andreychenko et al. Improved steering of the RF field of traveling wave MR with a multimode, coaxial waveguide. Magn Reson Med. 2013 71(4):1641-49
8. A.J.E. Raaijmakers et al. Design of a radiative surface coil array element at 7 T: the single-side adapted dipole antenna. Magn Reson Med 2011;66(5):1488-97.
9. L. Winter et al. Design and evaluation of a hybrid radiofrequency applicator for magnetic resonance imaging and RF induced hyperthermia... PLoS One 2013;8(4).
10. R. Lattanzi et al. Ideal current patterns yielding optimal signal-to-noise ratio and specific absorption rate in magnetic resonance imaging: computational methods and physical insights. Magn Reson Med 2012;68(1):286-304
11. W. Lee et al. Parallel Transceiver Array Design Using the Modified Folded Dipole for 7T Body Applications. 21st Annual ISMRM Meeting Salt Lake City, Utah, USA, 2013, p292
12. Lakshmanan K, Cloos M, Lattanzi R, Sodickson D, Novivkov D, Wiggins GC. The Circular Dipole. 22nd Annual ISMRM Meeting Milan, Italy, 2014, p315
13. Wiggins GC, Zhang B, Lattanzi R, Chen G, Sodickson D. The electric dipole array: an attempt to match the ideal current pattern for central SNR at 7 tesla. 20th Annual ISMRM Meeting Melbourne, Australia, 2012, p2783
14. Ipek O, Raaijmakers AJ, Lagendijk JJ, Luijten PR, van den Berg CA. Intersubject local SAR variation for 7T prostate MR imaging with an eight-channel single-side adapted dipole antenna array. Magn Reson Med 2013
15. Raaijmakers AJE, Italiaander M, Voogt I, Luijten PR, Hoogduin JM, Klomp DJ, van den Berg CAT. The fractionated dipole antenna: a new antenna for body imaging at 7 Tesla. Magn Reson Med 2015, In press
16. Eryaman Y, Guerin B, Kosior R, Adalsteinsson E, Wald LL. Combined loop + dipole arrays for 7 T brain imaging. 21st Annual ISMRM Meeting, Salt Lake City, Utah, USA, 2013, p393