Radiative principles for efficient RF transmission at UHF: radiative antennas and travelling waves

Authors: C.A.T. van den Berg, Anna Andreychenko, Alexander Raaijmakers:

University Medical Centre Utrecht, the Netherlands. Email*: c.a.t.vandenberg@umcutrecht.nl*

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 radiative principles for RF transmission at UHF: radiative antennas and travelling waves

Introduction: Reactive coils vs radiative antennas

RF coils can be classified in RF receive coils and RF transmit coils. RF receive coil are designed to create a maximal sensitivity to the NMR signals emitted by the spins while picking up a minimal amount of thermal noise. The function of RF transmit coils is to generate a homogeneous B1+ field in the imaging region while inducing minimal RF power deposition that results in undesired RF tissue heating. At lower field strengths ($\leq$ 3T) receive coils generally consists of loop type coils that are placed in close contact to the target region for maximal signal coupling. The RF transmit field is at low fields exclusively applied by the socalled birdcage resonator [1] The key feature of the birdcage coil is that it is tuned to support two degenerate resonant modes that are driven in quadrature to create a circularly polarized RF magnetic field (B1+) that rotates at the same frequency and direction as the precessing spins. The introduction of the birdcage coils was a tremendous step forward in creating homogeneous and efficient spin excitation enable over a large volume. The loop as well as the birdcage coil is based upon a resonant operation, meaning that the currents are maximized by tuning them with lumped elements inserted in the coil conductors (capacitors) to resonance. In this way the reactive energy stored in the electric fields and magnetic fields is maximized. Every period of oscillation the energy contained by the resonance converges from purely electric energy, mainly stored in the capacitors, into magnetic energy stored in the near field region of the coil. Ideally, this leads to an advantageous situation where the sample (usually a human subject) resides in the near field region with large magnetic field and the electric fields, causing undesired losses and tissue heating, are screened from the human subject. The quality factor of a coil, defined as the stored energy over the losses per oscillation cycle, is a good measure of the realization of this state. This type of coupling is known as reactive coupling and the magnetic fields in the near field are directly related to the currents in the coil through the law of Biot-Savart (eq. 1).

Eq. 1

$$\boldsymbol{B}(r) = \frac{\mu_0}{4\pi} \int \frac{I\mathrm{d}l \times r}{|r|^3}$$

where
$\boldsymbol{B}(r)$ = magnetic field strength at position r
μ_0 = permeability of free space
dl = differential coil segment
I = current in coil segment.

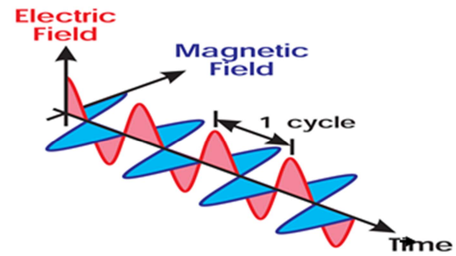
With the advent of ultra-high field imaging ($\geq$ 7 T) the frequency has increased to 300 MHz and the performance of an RF coil driven at resonance, starts to degrade. This is caused by the fact that the imaging target region can no longer located in the near-field region as this region shrinks with frequency. Outside this reactive near field the reactive fields will have decayed and the so-called radiative fields become more dominant. Between the reactive near field and the socalled far field, a transition zone of 1 to 2 wavelengths exists, sometimes referred to as the radiative near field, where both terms have amplitudes of equal order but the radiative component dominate. Although no strict border exists as it also depends on antenna dimensions, the most frequently used cross-over point between reactive and radiative zones is defined as a quarter of the RF wavelength. In air this would amount to 25 cm for 7T, but since biological tissue has a relatively high dielectric permittivity (~40-60), for on-body coils this amounts to only a few centimeters. Outside this region, the RF field has turned into an electromagnetic wave that propagates away from the coil.

This power flux of a wave is expressed by the Poynting vector **S** (eq. 2). It is the energy that is carried by the wave per unit time and area and it is therefore expressed in watts per meter square. It is a vector quantity, calculated by the vector product of the electromagnetic wave components **E** and **H** whose time-varying nature, as was demonstrated by Maxwell, leads to the propagation of the wave. The wave impedance η is a parameter that characterizes an electromagnetic wave. It is the ratio of the electromagnetic wave components E over H, equation 3. A wave traveling freely in a medium has a wave impedance that is equal to the equilibrium wave impedance of the material η medium [equation 4]. This equilibrium value is a property of the material. Close to sources (antennas) and interfaces, the wave impedance will deviate from this equilibrium value.

Eq. 2 $$\boldsymbol{S} = \boldsymbol{E} \times \boldsymbol{H}$$

Eq. 3 $$\eta = \frac{\boldsymbol{E}}{\boldsymbol{H}}$$

Eq. 4 $$\eta_{\text{medium}} = \sqrt{\frac{\mu}{\varepsilon}}$$



The radiative antenna: Maximizing signal penetration at depth by optimizing directivity

As known for many decades in antenna engineering, to maximize the arrival of radiative energy and thus to maximize the field amplitude in the radiative field zones, an electromagnetic wave should be established by the antenna that has a high directivity of its power flux towards the target location. The direction of the Poynting vector specifies the direction of energy flow. Thus, to create a directive power flux both the *electric* and *magnetic* field orientations of the coil, or more correctly antenna, need to be considered. Raaijmakers et al. [3] introduced a first design for a radiative antenna for UHF MRI that was particularly designed to create a directive energy flux towards the target location. This so-called single-side adapted dipole antenna (SSAD) consists of a dipole antenna mounted upon a dielectric substrate with a high permittivity (figure 1b). The substrate is placed adjacent to the skin of the patient. The main electric field and magnetic field orientation of the electric dipole create a dominant Poynting vector aimed towards the target location. The high permittivity of the substrate will augment the electric fields within the substrate resulting in a larger Poynting vector into the substrate and towards the patient minimizing radiation spill to air. A comparison of a loop coil and the SSAD antenna is shown in figure in terms of Poynting vector distribution (1c,d) and resulting B1+ field distributions (1e,f). Note the more homogenous and more efficient B1+ field penetration for the SSAD.

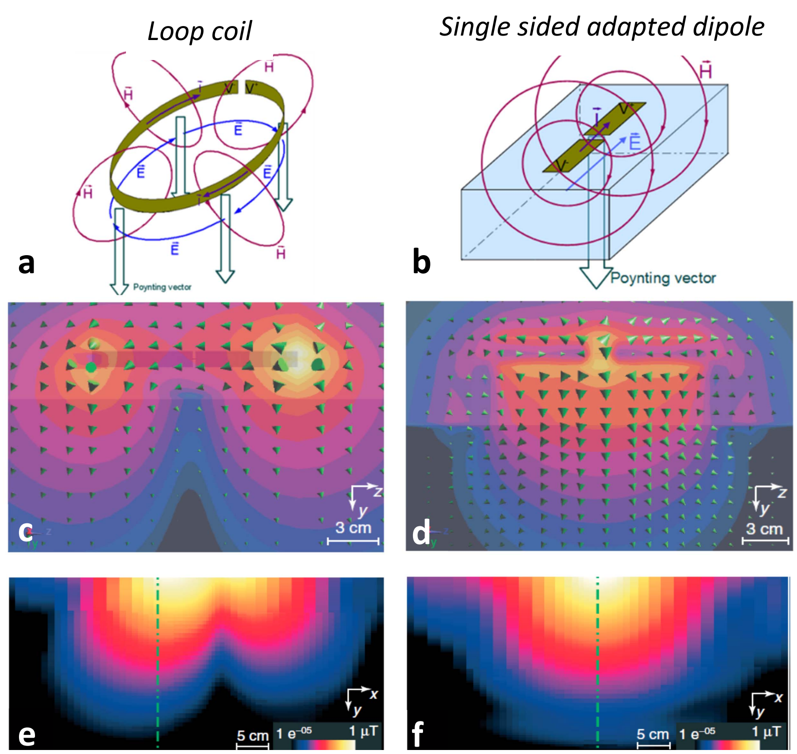


Fig. 1: Comparison between the field orientation of a loop (a) and SSAD (b) and their corresponding simulated (Semcad X, SPEAG AG) Poynting vector distributions (c,d) and B1+ fields (e,f) in a homogenous phantom normalized for 1 W power.

Travelling wave MRI/Waveguide MRI

The relatively small RF signal wavelength of UHF MRI does not only have downsides. New types of RF transmit schemes becomes possible because of the interesting phenomenon of the waveguide action of the scanner bore which was nicely demonstrated by a pioneering Nature paper of Brunner et al. naming it travelling wave NMR [4]. Interestingly, a possible use of waveguides as a low loss RF probe in high field NMR was already mentioned in 1977 [5]. But given the low field strengths feasible at those times, the required dimensions would become bulky. However, at 7T the RF frequency is sufficiently high that the RF signal wavelength becomes comparable to the circumference of the human MR scanner bore (1 m vs. 1.9 m), and the bore starts to act as a cylindrical waveguide for the RF signal. The waveguide effectively guides the power flux carried by the RF signal along the longitudinal axis of the bore. Waveguides are well-known in microwave engineering for their high power-handling capability and low losses. There are several potential benefits of such a waveguide concept in comparison to the surface RF coils/antennas. 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 [6] exciting relatively high MR signal in the areas not intended for excitation (e.g. head instead of body torso).

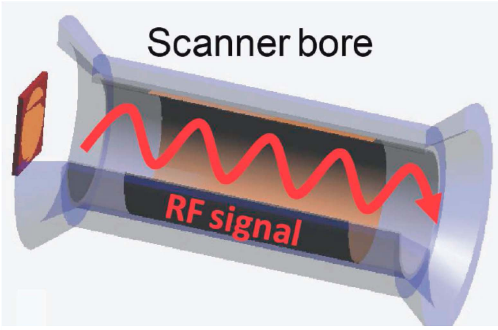


Fig. 2: A path antenna as the beginning of the bore excites a waveguide mode establishing a travelling wave in the RF bore.

In waveguide/travelling wave MR the waveguide creates a directional power flow from an antenna placed at the beginning of the bore to the target. Therefore, also waveguide MRI is based upon a radiative power transfer similar to the radiative antenna, and previously mentioned principles such the Poynting vector and the wave impedance are also relevant here. An important difference is, however, that as it follows from the waveguide theory, wave propagation occurs in form of discrete modes: a certain, distinct transverse electro-magnetic field pattern fulfilling the boundary conditions set by the waveguide walls. Each mode has its characteristic wave impedance and Poynting vector distribution. The modes can only propagate if their cut-off frequency (fc) condition is addressed. The cut-off frequency of a certain mode sets the limit of the lowest frequency where the mode can propagate. Below the cut-off frequency the mode is evanescent and rapidly decays in the waveguide. The waveguide type, geometry, size and electric and magnetic filling define the cut-off frequency of each mode. The RF shield of the MR scanner bore forms a hollow, metallic, cylindrical waveguide. In such a waveguide the TE11 mode has the lowest cut-off frequency. Thus, the TE11 mode is the dominant mode. Its cut-off frequency equals to: $fc=1.841/(\pi d\sqrt{\mu\varepsilon})$, where d-the waveguide diameter, μ-permeability of the waveguide substrate, ε-permittivity of the waveguide substrate. For a 60-cm hollow diameter bore, the cut-off frequency of TE11 mode is about 293MHz. Therefore, at 7T the frequency of RF signal (298 MHz) fulfills the cut-off condition and the human sized scanner bore starts to act as a waveguide for the RF signal (figure 2).

When travelling wave NMR was first introduced [4], an ideal B1+ homogeneity was looming as a travelling wave is characterized by homogeneous amplitude along its propagation direction. The curvature of the field, dictated by Maxwell, can be observed in the phase, which does not compromise MR image quality. However, there are certain fundamental limitations to this, which have to be acknowledged. First of all, a travelling wave will still have a standing wave in the transverse plane. Thus, there will be always a non-uniform transverse B1+ excitation pattern, which is specific for each waveguide mode. Second, in practice it is very difficult to achieve an ideal, uniform longitudinal homogeneity, as a non-uniformly loaded bore is not able to support the true travelling wave propagation. Each waveguide non-uniformity due to varying cross section or loading (e.g. the shoulder section of a human subject) causes partial reflection and/or mode conversion. These backward propagating modes will interfere with the forward travelling modes creating a standing wave pattern in the longitudinal direction.

Therefore, also the B1+ coverage in the longitudinal direction cannot be homogeneous. Therefore waveguide MRI would be a more adequate name than travelling wave MRI of this type of RF coupling into the subject.

Recent innovations in waveguide MRI

In whole-body waveguide MRI large volumes (the whole bore volume) are excited and, thus, the power efficiency of the waveguide MRI as any type of volume excitation will be intrinsically lower compared to surface transmit elements. Therefore, given the current limitations in RF power amplifiers at 7T, waveguide MRI can generate at this moment only moderate B1+ fields. For clinical sequences this will be a major obstacle. Consequently, much of the recent work on waveguide MRI has been focusing on boosting the efficiency. Coaxial waveguides have been introduced, where premature power losses in the feeding section due to dissipation and reflections are minimized [7,8]. Another solution includes dielectric lining of the scanner that increases both radial and longitudinal coupling towards the patient [9]. Dielectric lining/loading of the scanner bore also lowers the cut-off frequency of higher order modes converting them into propagating modes at 7T. This allows RF shimming based on waveguide modes [10]. An attractive property of the presence of multiple waveguide modes is that RF field patterns per channel demonstrate much more spatial diversity. This arises from the fact that each excitation source can couple to several modes. As the modes have linearly independent transverse electro-magnetic patterns (i.e. B1+ patterns) the "RF basis" to span up a desired B1+ patterns will have a higher rank. This promotes a more efficient B1+ steering in the transverse plane Moreover, the characteristic longitudinal wavelength discrepancy of each mode creates the freedom to shim B1+ pattern efficiently also in the longitudinal direction [11]. This optimal control of the fields in three dimensions is a distinct feature of parallel transmit MRI based on a multi-mode waveguide equipped with multiple excitation sources.

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