CHALLENGES FOR HIGH FIELD CLINICAL MRI

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ABSTRACT

Clinical MRI can potentially benefit greatly from the relatively recent advent of commercial high-field (7 and 9.4 Tesla) systems. However, there are several major engineering challenges associated with high fields, the main one of which is to produce a strong, homogeneous transmit B\textsubscript{1} field, while remaining within regulatory guidelines for tissue power deposition.

Index Terms— RF distribution, transmit arrays, clinical MRI, high frequency, quadrature excitation.

1. INTRODUCTION

The general advantages of high field (7T and above) clinical MRI include higher signal-to-noise and spatial resolution. In addition, specific applications can benefit from particular properties of high fields, including: (i) increased sensitivity to iron deposition for studies of neurodegenerative diseases, (ii) increased background tissue suppression in magnetic resonance angiography, and (iii) increased spectral resolution in localized spectroscopy. An example of high resolution imaging is shown in Figure 1, in which right coronary angiogram was acquired using a simple quadrature surface coil transmit/receive geometry.

Figure 1. (left) Schematic of quadrature surface coil for coronary artery imaging, (right) right coronary artery acquired at 7 tesla.

For larger regions of the body, conventional MRI volume coils, such as a quadrature birdcage coil [1] or transverse electromagnetic mode (TEM) resonator [2], are designed to produce homogeneous RF transmit field distributions at low frequencies (long wavelengths), and typically have electrical currents with equal magnitude flowing in opposite directions on opposite sides of the sample. In a symmetric sample, this can create a standing wave pattern with constructive interference near the center of the sample and regions of destructive interference approximately one quarter wavelength away. At lower frequencies, e.g., <130 MHz for 1.5 and 3 tesla, \( \lambda \) is relatively long compared to the dimensions of the head and torso, and so these regions of destructive interference lie outside the imaging volume and thus are of little concern. At 300 MHz (7T), however, \( \lambda \) in brain is only \( \sim \)13 cm, and regions of constructive and destructive interference can occur in structures with similar or greater dimensions. This problem requires the design and optimization of much more sophisticated RF technology, as described in the following sections.

2. B\textsubscript{1} INHOMOGENEITIES AT HIGH FIELD

The image signal intensity (I) is a function of the transmit and receive sensitivities, and for a simple gradient echo sequence is proportional to the product of the sine of the transmitted tip angle multiplied by the receive sensitivity:

\[ I \propto \sin \left( \gamma B_1^+ \tau \right) \left( B_1^- \right)^* \]  

where \( \tau \) is the duration of the RF pulse. In the low tip-angle limit, the signal is proportional to the product of transmit and receive sensitivities:

\[ I \propto \left( B_1^+ \right) \left( B_1^- \right)^* \]  

The effect of the B\textsubscript{1} inhomogeneity as a function of field strength has been studied extensively [3]. Figure 4 shows that, even using a low tip-angle gradient echo imaging sequence, there are significant image non-uniformities at fields higher than \( \sim 4 \)T.
Figure 2. Simulated gradient-echo images as a function of field strength using a birdcage coil with ideal current distributions in the rungs. The signal is calculated as the product of the transmit and receive fields, assuming a low tip-angle excitation.

Figure 3 shows a measured $B_1^+$ map from the head of a volunteer at 7 tesla using a quadrature birdcage transmit coil. The strongest field is present at the centre of the brain, and areas of low signal intensity are evident close to the surface of the brain. The asymmetric excitation pattern, with respect to both horizontal and vertical axes, is also very clear. On the right of Figure 3 is an example of a spin-echo sequence, which clearly shows the reduced signal-to-noise and contrast in the temporal lobe regions arising from the $B_1$ inhomogeneity.

The fact that the interaction of high frequency magnetic fields with the human body gives rise to sample-induced spatial variations in the image intensity was recognized very early in the development of MRI [4]. The challenges of delivering electromagnetic energy into the human body at frequencies where the body dimensions are larger than the wavelength have been studied for a very long time, specifically in the area of electromagnetic or radiofrequency hyperthermia. Apart from the fundamental difference that hyperthermia aims to control the electric fields in order to produce localized therapeutic heating, the essential approach to “guiding” the distribution of RF energy is the same as that which has been adopted for high field MRI [5], namely the use of an array of single-element, electrically-decoupled applicators which are fed from separate sources [6-9]. An example of a modern, commercial multi-element array used for patient hyperthermia is shown in Figure 4. Also shown in Figure 4 are illustrations of the way in which the focal point of the electric fields can be steered by changing the phase of each of the dipole elements.

Figure 3. (left) Calculated $B_1^+$ map in the brain of a volunteer acquired at 7 tesla using a quadrature birdcage coil. Turbo spin-echo image of a volunteer acquired at 7 tesla using a quadrature birdcage excitation and 16-channel phased array receive. The lower signal intensity and, in particular, the lower gray/white matter contrast close to the temporal lobes indicate the intrinsically lower $B_1^+$ field in these areas.

Figure 4. (upper) A commercial six-element electric dipole hyperthermia applicator, in which the magnitude and phase of the input signal can be changed. (bottom) Different heating patterns that can be produced using “shimming” of the electric fields. An eight element dipole array is used, with different phases (but equal magnitudes) applied to each dipole. With all phases equal, the electric fields are “focused” in the center of the phantom (bottom left). By varying the phase, the location of the focus can be moved around, as shown by the figures to the right. The sample is a uniform sphere with dielectric constant of 80. The greater the number of elements in the array, the tighter the focus, and the greater the ability to steer the focus [10].

3. TRANSMIT ARRAYS

In principle, an RF system for high field MRI in which the magnitude and phase of the current in conductive elements about the head are controlled individually (rather than having equal and opposite values on opposite sides of the subject and thus creating a strong standing wave pattern) should be able to produce much more homogeneous fields than a conventional volume coil at high frequencies.

A variety of different coil geometries can be used for a transmit array: the two most common are the shielded loop and stripline designs shown in Figure 5 [11-13]. The stripline has found most use due to ease of construction, higher $B_1$ field close to the individual elements, and ease of element decoupling.
In vivo results, shown in Figure 6, obtained at 9.4 tesla using a similar stripline array illustrate the increase in signal homogeneity that is achievable simply by changing the phase of two elements of the array, in this case the two elements closest to the area in which the $B_1$ field is most inhomogeneous [14].

4. SAR AND HEATING EFFECTS

Associated with any RF magnetic field is a corresponding RF electric field which produces electrical currents in conductive tissues. A key consideration in assessing the practical viability of any transmit array and/or associated imaging sequence is the power deposition in tissue, quantified via the local and average specific absorption rate (SAR) values, measured in Watts per kilogram. There are strict regulatory guidelines on these values [15] in terms of peak instantaneous and time-averaged values for both local and global regions-of-interest. The SAR can be calculated from the electric field ($E$) distributions which are estimated from, for example, full-Maxwell calculations of the electromagnetic fields using finite difference time domain (FDTD) techniques, and is given by:

$$SAR = \frac{\sigma}{2\rho} |E|^2$$  \[3\]

where $\rho$ is the material density. An equally important measure is the temperature ($T$) in the head, which can be modeled with a finite difference implementation of the Pennes bioheat equation [16]:

$$\rho c \frac{dT}{dt} = \nabla \cdot (k \nabla T) + \left[ -\rho_{bio} \omega_c (T - T_{bio}) \right] + Q_m + SAR$$  \[4\]

where $c$ is the tissue heat capacity, $k$ the thermal conductivity, $\omega_c$ the blood perfusion, and $Q_m$ the heat generated by metabolism. Figure 7 illustrates the SAR distribution in the brain at 3 and 7 tesla, showing very different heating patterns at the two different frequencies [17].

Figure 5. Examples of loop (left) and stripline (right) transmit arrays constructed for neurological applications at 7 tesla [12].

Figure 6. Effect of the transmit phase on image homogeneity at 9.4 tesla [14]. (a) Scout FLASH image of a head inside a circularly polarized elliptical coil. The loss of signal near the left ear is the result of destructive interference reducing the net $B_1^+$. The relative transmit phase for each coil labeled near the two lines representing the conductor and ground planes of each coil is shown. (b) By adjusting only the relative transmit phase of the two coils closest to the area of low signal intensity, local destructive interference can be reduced.

Figure 7. Calculated distributions of (left) SAR (W/kg) and (right) corresponding temperature increase (°C) at 3 T (128 MHz) and 7 T (300 MHz) modeled during exposure to a head-average SAR of 3.0 W/kg. (Top) An axial plane passing through the center of the coil and brain, (center) corresponding sagittal and (bottom) coronal planes. The simulated RF coil was a sixteen-element TEM resonator with rung currents equal in magnitude and with a phase given by their azimuthal position.

5. REFERENCES


