**Transmit Arrays**

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**Introduction:**

The clinical and research potential of magnetic resonance imaging/spectroscopy (MRI/S) for whole-body applications at high (≥ 3 tesla) fields and of head applications at ultrahigh (≥7 tesla) fields appears to be limitless. It is currently; however, hampered by significant physical and technical challenges. The most notable of these difficulties include 1) safety concerns regarding exceeding radiofrequency (RF) power deposition in tissue [1, 2] and 2) large image inhomogeneity/voids due to “undesired” RF field inhomogeneity [3-5] across the anatomy. Recent results of multi-transmission approaches [6-23] such as transmit SENSE and B 1 (refers to the RF magnetic field in tissue during MRI/S) shimming have emerged as answers to alleviating the RF inhomogeneities as well as the, potentially unsafe, local and global RF power deposition associated with high/ultrahigh field human MRI.

To this date however these multi-transmission methods have not evolved to realize a full scale scientific/clinical research. In particulars, several major obstacles have dampened the enthusiasm for wide implementation of parallel transmission methods into high and ultrahigh field imaging/spectroscopy. These include:

1. Considerable scanning and preparation time for every subject (while the subject is in the scanner) for mapping the B 1 field which in many circumstances, especially at ultrahigh field, is inaccurately measured,
2. Significant RF excitation (B 1+ field) intensity losses associated with current decoupled multi-transmit arrays due to increased local/global power deposition in tissue at lower flip angles, and
3. Concerns regarding the unclear RF safety assurance of the multi-transmit experiment (to this date, the power deposition/electric field in the human body are not measurable using MRI techniques.)

In this presentation we will evaluate many of these issues through the context of Transmit Arrays.

**Background and Theory**

**B 1 Inhomogeneity and SAR**

In addition to relaxation parameters (T 1 and T 2) and proton density (ρ) and other parameters, the MR signal is a function of the transmit (commonly is referred to as the B 1+) and receive
(commonly is referred to as the $B_1^+$) fields. For example, MRI signal for a gradient echo sequence is [24]:

$$\text{Signal} \propto F(T_1, T_2, \rho) \times \sin(B_1^+)	imes B_1^-$$

(1)

The general criterion is to have the image contrast exclusively dependant only on the tissues' inherent properties, i.e. $T_1$, $T_2$, and $\rho$. In order to achieve this goal, the transmit RF coil must produce $B_1^+$ field that possesses high intensity (to achieve high signal) and uniform distribution across the region of interest, i.e. the contrast in the region of interest should not depend on the $B_1^+$ field. This concept also applies for the $B_1^-$ field.

Higher field strengths correspond to increased Larmor frequencies and therefore operational RF frequencies. A 7 tesla for instance, coil/transmit array excites $B_1^+$ field at 298 MHz for proton imaging. At this frequency (wavelength is approximately 12-cm in tissue) and higher, the wavelengths of the electromagnetic waves produced by RF coils/arrays become smaller than the human head; or in other words the human head becomes electrically “large” [3, 25-32]. Unlike the case at lower field strengths, the electromagnetic waves now have to “travel” significant electrical distances in the human head. As a result, the electromagnetic fields become non-uniform which will result in inhomogeneous $B_1^+$ and $B_1^-$ fields in biological tissues as well as inhomogeneous electric fields and, therefore, localized SARs. Both which can have a devastating effect on the integrity of the images and on the safety of the patient. Nonuniformity of the MRI RF fields can be the largest cause of error in MRI [24] in general and at high fields in particular.

**Multi-Transmission Methods**

Variable phase/amplitude multi-port excitation or $B_1$ shimming (in electromagnetic terms: phased array antenna excitation) is based on the fact that for high frequency MRI operation and asymmetrical/inhomogeneous/irregularly-shaped loading (human head/body), integer multiples of phase-shifts and uniform amplitudes are not necessarily the ideal characteristics to impose on the voltages driving the transmit array in order to obtain a homogeneous transmit field. Furthermore, overall as well as localized RF field excitation in high field human MRI may be achieved with rather distinctive and non-obvious amplitudes/phases associated with the excitation voltages.

Early numerical work for potential high field MRI experiment has shown that specified superposition of electromagnetic fields radiated from long thin wires can alter the field in the sample [33] and that coil/head interactions could be manipulated by changing the coil’s excitation mechanism [17, 18]. This concept has been clearly verified in recent experimental head studies at 3, 7, and 9.4 tesla by varying the phases of voltages driving transmit array(s) [22, 34, 35]. As such, $B_1$ shimming and other methods that manipulate the MRI RF field, such as transmit SENSE and tailored RF pulses are capable of producing specified RF excitation [13, 14, 22, 36-42] for high field MRI.

Multi-transmission approaches [6-18] (transmit SENSE and $B_1$ shimming) have generally aimed at reducing RF power deposition in tissue and homogenizing the RF field across the anatomy of interest [6-18, 21, 43, 44].

**Transmit Arrays**

Since the advent of ultrahigh field MRI, distributed circuit resonators have shown good potential for use with human applications above 7 Tesla, as exemplified by the transmission line resonator of Roschmann [45], the transverse electromagnetic (TEM) resonator of Vaughan [43], and the free element resonator of Wen [46]. As magnet strength and the Larmor frequency increase, the length of the conductors used in head sized volume coils become a significant fraction of the operating wavelength. In this case, the coil struts develop self resonance which
may degrade coil homogeneity and fill factor since the current is no longer uniform along each strut [47]. As a result, the transmission line properties of the coil’s conductors become significant, invalidating lower field strength assumptions which neglected them. In contrast to lumped element designs, distributed circuit resonators utilize and enhance the transmission line properties of conductors by using the intrinsic reactance of transmission line elements. Furthermore, a significant practical advantage of the distributed circuit coils for ultrahigh field MRI operation is the relative ease in the tuning process of the coil when loaded, because each of these coils allows simultaneous adjustment of all of its resonant elements. Aside from these operational and practical issues however, the electromagnetic interpretation of the behavior of loaded ultrahigh field distributed circuit coils is still rather uncertain.

Many designs of transmit arrays [8, 11, 13, 34, 38, 48-59] have been used in exciting the RF field at various MRI field strengths. Transmit arrays can be coupled or uncoupled. For example, it is expected that each excited element of a 7 tesla loaded TEM head/body coupled-coil [41, 43, 60] to experience unique coil/load impedances that may differ significantly from those incurred by other coil elements. Moreover, the frequencies at which each coil element experiences real input impedances (resonances) may vary as well. In such cases, the resonance frequencies of the different modes will differ from one element to another. This particular issue will not be resolved with the use of matching circuits, since it is difficult to be “on mode” (the imaginary component of the input impedance to be zero) using all the coil elements simultaneously at the same frequency. In this regard, coupled-arrays are at a disadvantage compared to decoupled arrays. A coupled-array element however tends to provide significant B1 field intensity within deep head/body tissue as well as a generally wide B1 field distribution. This is an advantage over decoupled-arrays in which strong coupling exists between the array elements and the subject (thus their performance and operation are sensitive to the imaged subject.) It would be more difficult to obtain relatively deep penetration for volumes as electrical large as the human abdomen at ≥ 3 tesla and as human head at ≥ 7 tesla with decoupled-arrays.

CITED REFERENCES


Exploring the Limits of RF Shimming: Single Slice and Whole Brain Field Optimizations at up to 600 MHz with Transmit Arrays of up to 80 elements, in *Proceedings 14th Scientific Meeting, International Society for Magnetic Resonance in Medicine*, Seattle, 2006, p. 1383.


Ultrahigh-field MRI whole-slice and localized RF field excitations using the same RF transmit array, *IEEE transactions on medical imaging*, vol. 25, pp. 1341-7, Oct 2006.


