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An eight-channel T/R head coil for parallel transmit MRI at 3T using ultra-low output impedance amplifiers

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Parallel transmit is an emerging technology to address the technical challenges associated with MR imaging at high field strengths. When developing arrays for parallel transmit systems, one of the primary factors to be considered is the mechanism to manage coupling and create independently operating channels. Recent work has demonstrated the use of amplifiers to provide some or all of the channel-to-channel isolation, reducing the need for on-coil decoupling networks in a manner analogous to the use of isolation preamplifiers with receive coils. This paper discusses an eight-channel transmit/receive head array for use with an ultra-low output impedance (ULOI) parallel transmit system. The ULOI amplifiers eliminated the need for a complex lumped element network to decouple the eight-rung array. The design and construction details of the array are discussed in addition to the measurement considerations required for appropriately characterizing an array when using ULOI amplifiers. B_1 maps and coupling matrices are used to verify the performance of the system.

1. Introduction

The increase in signal-to-noise ratio (SNR) and spectral resolution that comes with high field magnetic resonance imaging (MRI) can be traded for improvements in spatial and temporal resolution [2]. In many cases, technical challenges prevent these benefits from being straightforwardly realized in practice. One significant and well-known challenge is potential inhomogeneity in the transmit B_1 field due to the higher frequencies (and shorter radiofrequency (RF) wavelengths) associated with higher magnetic field strengths [3–7]. In neuroimaging applications, this often manifests as a central brightening artifact due to constructive interference in the center of the head preventing uniform tip angles [8–9].

A number of research groups have used multiple transmit channels to address this challenge. Approaches range from relatively straightforward B_1 shimming [10–13] to more complex transmit SENSE techniques in which separate RF excitation pulses are sent to each channel [14–16]. With either approach, however, the level of independence between transmit channels is a concern. Any current applied to one element will induce a voltage (more accurately, an electromotive force or EMF) in any other elements if they have any mutual impedance. In turn, this induced voltage can drive unwanted currents in the other coils, contaminating the desired coil pattern. This is commonly referred to as "coupling" between coils. One can preserve the patterns either by reducing the induced voltage by eliminating or cancelling the mutual impedance, or, by ensuring that no additional currents are generated as a result of the induced voltage. There are a number of possible ways to eliminate the mutual impedance, such as geometrically overlapping the coils or constructing a lumped element network on the coil to cancel the mutual inductance [17]. To eliminate the currents driven by the mutual impedance, one can introduce a high impedance across the terminals of the coils as in the case of using isolating preamplifiers [18]. A benefit of the latter approach is that it results in less concern for the mutual impedance between coils, just as with receive arrays. To generalize, array coil design, specifically with respect to the degree of on-coil decoupling required, depends on and operates in concert with the preamplifier regime in the receive case and the amplifier regime in the transmit case [15].

Most transmit array coils are designed for use with standard RF power amplifiers, which have been designed to produce maximum

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output power assuming a 50 Ω load. To maintain expected performance, these 'conventional' amplifiers generally rely on decoupling of the coil elements themselves, just as with receive arrays operating with conventional preamplifiers, to ensure that the amplifier sees the expected load. Some combination of geometric overlap and lumped element decoupling networks must be employed on these arrays to enable independent operation of the channels. Geometric overlap is limited in application to adjacent elements and imposes constraints on the array geometry. Lumped element networks require no overlap, but as the channel counts increase, the network required to fully decouple all elements increases in complexity due to the increasing number of decoupling capacitors required [17]. As an example, to decouple seven elements with a lumped element ladder network, three decoupling capacitors are required for each ladder stage [19], giving a total of 21 capacitors needed for full decoupling. In extending the approach to eight elements, four decoupling capacitors are required at each ladder stage for a total of 32 decoupling capacitors. The values for the decoupling capacitors are guided by closed-form equations that require iterative non-linear and numerical field solvers to compute [20], and further fine adjustments in element tuning are done iteratively and experimentally. The amount of available decoupling is also sensitive to loading. As indicated in one study [19], the decoupling between opposing elements of a four-element array was found to increase from −30 dB in the unloaded case to −10 dB in the loaded case, indicating 10% of the power from one element being coupled into its opposing element. Due to these complications involved with the various on-coil decoupling strategies, stripline elements and shielded loop elements have emerged as the two most common designs for parallel transmit arrays in neuroimaging due to their inherent favorable coupling properties [11,21–25]. Even still, an additional decoupling network is typically needed, adding complexity to the array design [11]. Recently, alternative approaches to amplifier design have been investigated that provide some degree of isolation between channels, decreasing or even eliminating the need for decoupling between the coil elements themselves.

One approach under investigation is current source amplification in which each series resonant element is driven with a prescribed current, insensitive to the effects of loading, including element-to-element coupling [21–25]. This allows for large amounts of flexibility in the design of the array coils used with current sources, as in principle no on-coil decoupling techniques are needed. Current source amplifiers provide high isolation by presenting a high impedance to the series tuned coil, but are not matched for optimal power output, so they are limited in the amount of current they can produce as compared to standard RF power amplifiers [26]. Recent work on current mode class-D (CMCD) amplifiers has shown to improve efficiency and may be able to produce higher peak output levels than linear current source amplifiers [24–25,27]; however, the CMCD design has other complications that must be addressed in the process [25].

Chu et al. introduced the "ultra-low output impedance (ULOI) amplifier", which, in contrast to the current source amplifier, provides decoupling in a manner analogous to preamplifier decoupling in parallel receive applications [28]. The array elements are matched to 50 Ω input impedance using a network that forms a trap when connected to the ULOI amplifier or preamplifier. ULOI amplifiers are power matched and present a low impedance to the coil port. The power match enables peak output levels comparable to standard power amplifiers while the low impedance provides isolation when combined with an appropriately designed matching network on the coil. The isolation obtained with ULOI amplifiers is substantially lower than that of current source amplifiers, but the peak current delivered to the coil is higher. In addition, as discussed in [28], the ULOI amplifier drain is biased into saturation so that output current and parametric variation is minimized in cases of non-passive loading such as the case with coupled transmit array elements. In practice, the ULOI amplifiers represent a "middle ground" between current source and standard power amplifiers with respect to isolation and output power. We chose to implement a parallel transmit system with ultra-low output impedance amplifiers to achieve greater power output over previously built current sources [23] and to achieve decoupling benefits to simplify the on-coil decoupling network. An added benefit of this approach is that it allows for a straightforward transmit/receive configuration in that the matching network required for the ULOI amplifier is identical to the one needed for low-input impedance preamplifiers. This paper discusses the construction and characterization of an eight-channel transmit/receive head array for use with an ultra-low output impedance parallel transmit system. The eight-element array was decoupled using a simple decoupling network in combination with the isolation provided by the ULOI amplifiers. The characterization of the isolation provided by the stages of the system, from amplifier to coil, is discussed in detail.

2. Materials and methods

2.1. System overview

The eight-channel parallel transmit system was designed as a retrofit for a 3 Tesla GE clinical research scanner, requiring a rapid and transparent switchover from a standard single channel transmitter. Two inputs were required from the host GE scanner: the input to the RF amplifier and the master exciter unblank (RF gate) signal. A single hard pulse played out from the scanner is divided eight ways and then modulated by an in-house built vector modulator [29]. The control system employs a PXI-7853R FPGA-based board with programs written in LabVIEW (National Instruments, Austin, TX) to drive the hardware and provide the baseband in-phase and quadrature signals to the vector modulator from the user-defined amplitude and phase information of each RF pulse [30]. The modulated waveforms pass through a first gain stage prior to the ultra-low output impedance amplifiers. The head array coil is connected to the amplifiers through transmit/receive switches. Low input impedance preamplifiers on the array coil provide the first gain stage on the receive side prior to passing through to the scanner receiver chain.

2.2. Array fabrication

The eight-channel head array was designed with shielded rungs fabricated using copper sheet metal mounted to a ½ inch wide, 25 cm long acrylic piece with six breaks with 79 pF of capacitance at each break (Passive Plus, 1111C Series, Huntington, NY). The mounting piece for all the element hardware was a 12-inch outer diameter cylinder fabricated from white polycarbonate using a fusion deposition modeling (FDM) rapid prototyping machine.

The shield consisted of two layers of single-sided ½ ounce copper Pyralux (AC182500E, DuPont, Research Triangle Park, NC) mounted to the 12 inch cylinder. Each layer was slotted longitudinally to mitigate eddy currents and the two layers were oriented to alternate the position of the longitudinal slits. Rectangular slots were removed from the Pyralux shield and replaced with copper mesh to provide a view port through the coil for patient comfort and visual stimulus in functional imaging studies.

The elements were mounted on the inside of the cylinder with connections to the shield at one end and the matching network at the other. With the ½ inch thickness of the acrylic rung support and the ¼ inch thick cylinder, each rung was ¾ inch from the