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Highly Parallel Transmit/Receive Systems for Dynamic MRI

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Abstract- Dynamic MRI continues to grow in interest and capability with the introduction of 64 and 128 channel receivers, and, more recently, 8 and 16 channel parallel transmitters. This talk will describe progress in developing a 64 channel transmitter and applications in high-speed MR imaging, reaching 1000 frames per second.

I. Introduction

Magnetic Resonance Imaging has traditionally been a "serial" imaging method, with a single transmitter and receiver, and spatial localization performed with repeated acquisitions of the MR signal with only the imaging gradient amplitudes altered. The advent of parallel receivers enabled reduced image acquisition times by enabling simultaneous independent views of the object under study with different coils and receivers 1-3]. MR systems today have has many as 64 and 128 receivers, enabling MR images to be acquired in as little as a single echo [4-6], and at frame rates exceeding 1000 images per second for brief bursts [7]. These frame rates may be of use for imaging flow or non-periodic, rapid events, or even for imaging short lifetime species such as hyperpolarized agents [8, 9]. There are a number of limitations that limit the eventual applications of ultra-fast imaging methods. One is that the phase pattern of the receive coil can vary significantly over the voxel when the coil has dimensions on the order of the voxel [10]. This phase variation, obviously dependent on the location of the voxel with respect to the coil, must be corrected prior to signal acquisition or it will diminish or cancel the received signal. An additional issue is simply data acquisition requirements when acquiring data at extremely rapid rates from many channels. Finally, many of the applications of ultra-fast imaging will be limited by the duration of the excitation phase. To this end, the same techniques used for parallel receive are being applied to the transmit stage. Each of the major manufacturers are developing multiple channel transmitters to enable reduced excitation times in an exactly analogous manner to parallel receive MRI.

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This talk will discuss the implementation of a prototype parallel MRI transmit/receive system and efforts to overcome the limitations mentioned in the previous paragraph. The development of a 64 channel parallel transmitter for MRI, along with the modulators, control system, and interface to a previously developed 64 channel receiver will be described.

II. Methodology & Theory

Several manufacturers and research groups have demonstrated multiple channel transmit systems with eight or more channels [11-18]. For the purposes of investigating the limits of highly parallel transmit it was decided to develop a 64 channel parallel transmitter. One approach is to use commercial amplifiers, but this can be quite expensive. A commercial, broadband MRI amplifier capable of producing 100 watts can cost upwards of $10k, beyond our budget.

In order to investigate transmit arrays on a large scale, our group has developed a 64 channel transmit system capable of producing 100 watts per channel was successfully developed [19]. A block diagram of the system is shown in Fig. 1. Phase and amplitude control is provided on a per channel basis to allow for B1 shimming. A separate mode of operation allows for independent channel modulation using high speed D/A cards (currently National Instruments PCI-6733) controlled by LabVIEW running on the host computer. Our system presently supports eight channels for fast modulation, but can be expanded with the addition of more or higher channel count D/A cards. The transmit system is a modular design consisting of vector modulators (HP HPMX-2005), 0.8W 34dB gain stage (Freescale MHW-1345N), 100W 24dB gain MOSFET amplifier (Freescale MRF6V3150NR1), PIN diode (MicroSemi UM9104F) based T/R switches, and monolithic (Minicircuits GALI-74+) receiver preamplifiers. The transmitter has a dynamic range of 60dB, with gain controllable from +5dB to +65dB for individual channels. Further, noise blanking is performed at the input to first amplifier stage and output of second amplifier stage to limit noise during receive. The vector modulators allow for simple phase and amplitude levels to be set using the digital rheostat array which has a USB interface provided by an NI digital I/O unit (NI USB-6501). Control of this system is handled by a computer running a GUI based software we developed in house. The Varian scanner provides RF input to the system, and up to three digital control lines for modulator output trigger (in fast modulation mode), noise blanking and MOSFET gate bias control.

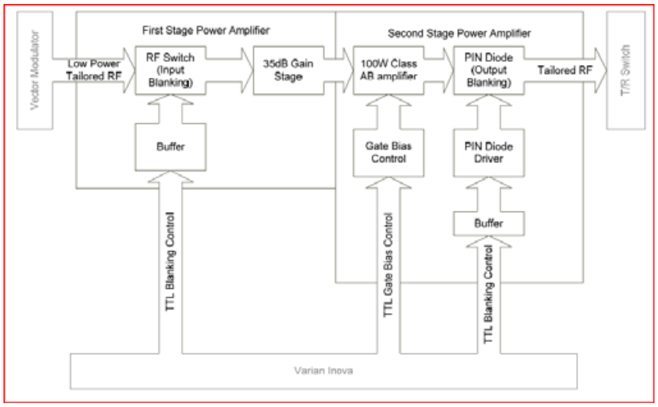

Figure 1. Overview of a single channel of the 64 channel transmit system.

A block diagram of the modulator is shown in Figure 2. Figure 3 is a photograph of a single, four channel modulator board [20]. The vector modulator approach was used as it enables direct modulation of the signal using audio frequency I and Q signals. This avoids the need to provide an intermediate frequency to each of the RF channels as would be used in a conventional single or dual conversion approach.

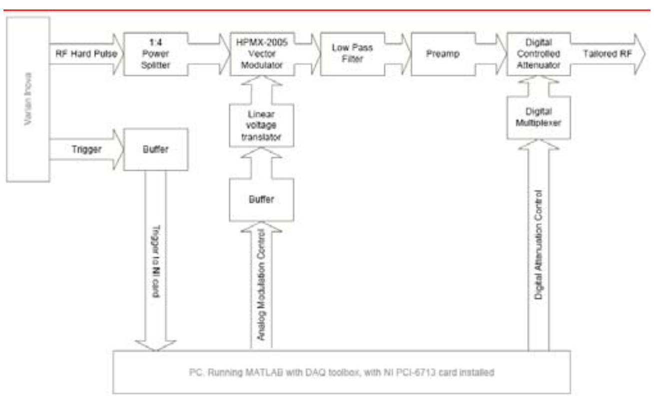

Figure 2. Overview of a single channel of vector modulator. Input is an unmodulated hard pulse from the system, and kHz range modulation signals.

The ability to remove the gate bias from the power amplifier MOSFET is provided to limit temperature rise. This is especially effective due to the low duty cycle at which the system is normally operated. The high power supplies used for the 100W amplifiers are commercial 48V, 3kW front end supplies (Cherokee elec. CAR3010L1NH). Solid core RG-58 (Belden 8240) cable was used to connect the power amplifier output to the T/R switches. Lower loss cable was prohibitively large, so the 2.4dB of cable loss must be compensated for with higher output power from the amplifiers.

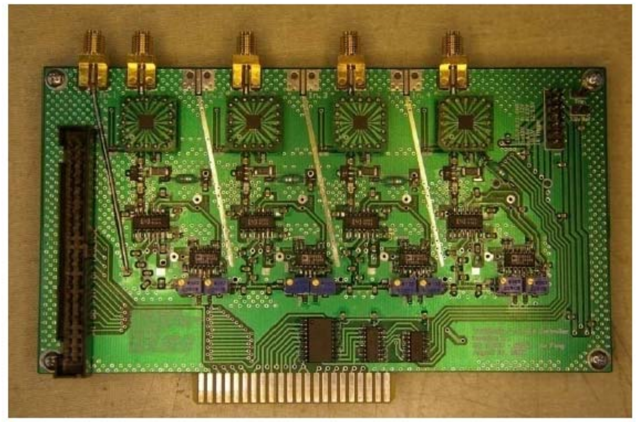

Figure 3. Photograph of a four-channel vector modulator board. The connector at the right is for the high-speed I/Q input. The edge card provides power and logic signals.

A modular architecture allows for the parallelization of construction and testing of individual subsystems and channels. Construction, troubleshooting, and maintenance is further eased by separating channels onto discrete boards which are easily removed. Figure 4 shows a single 100 watt RF amplifier, designed to be easily replaced in case of failure. The board is designed using stripline circuits for matching networks to enable greater repeatability and reliability [21]. Figure 5 shows the coil, tuning electronics, T/R switches and preamplifiers, all constructed in house, prior to being inserted in the magnet.

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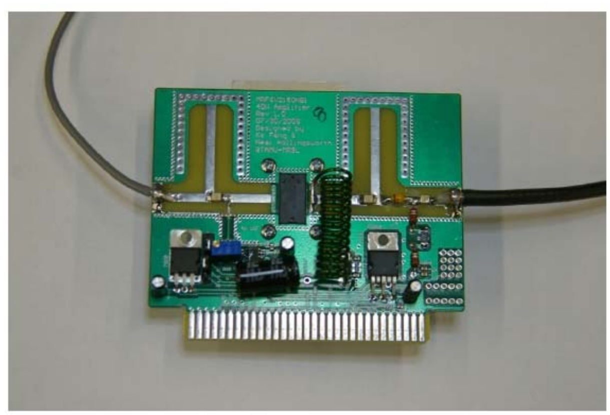

Figure 4. Single channel 100 watt RF amplifier. The matching networks are both done with stripline construction for repeatibility. Edge connector is for DC and control signals.

Fig. 5. 64 channel coil, tuning electronics, flange plate, T/R switch and preamplifier assembly prior to loading in the magnet.

III. RESULTS & DISCUSSION

Three experiments are shown below illustrating the use of the system. To test the vector modulator both conventional, 1D RF pulses and 2D RF pulses were generated and compared with the results from the system scanner. Figure 6 shows the results of a four-slice multi-slice sequence using the system modulator and the prototype vector modulator to generate the RF pulses, with good results.

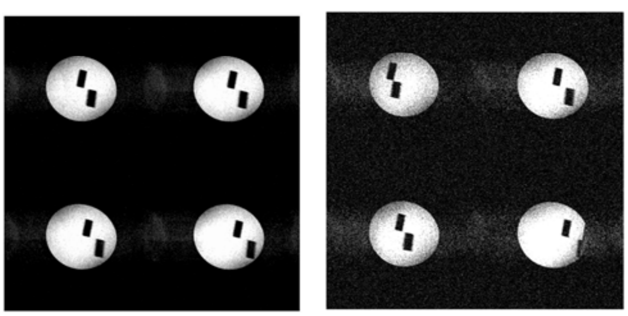

Figure 6. Example of a multi-slice excitation using the original system modulator (left) and the prototype vector modulator board (right).

Next, the system was used to generate a series of MR elastograms, MR images encoding the effects of a transverse acoustic wave [22]. The setup to generate the acoustic wave is shown in Figure 7. Using an external trigger from the MR scanner and a variable delay, the acoustic wave was started at advancing delays prior to obtaining a single echo image from the prototype system. The series of single echo images shown in Figure 8 clearly show the advancing wavefront [23].

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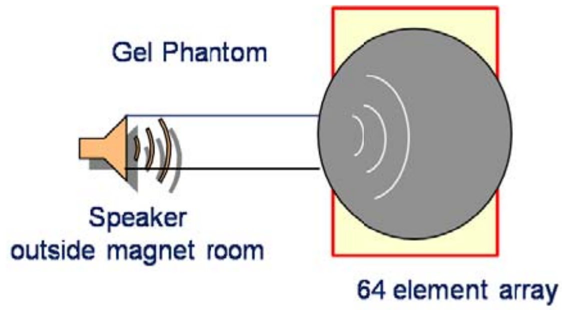

Fig. 7. Illustration of the setup for MR elastography demonstration using single echo acquisition imaging.

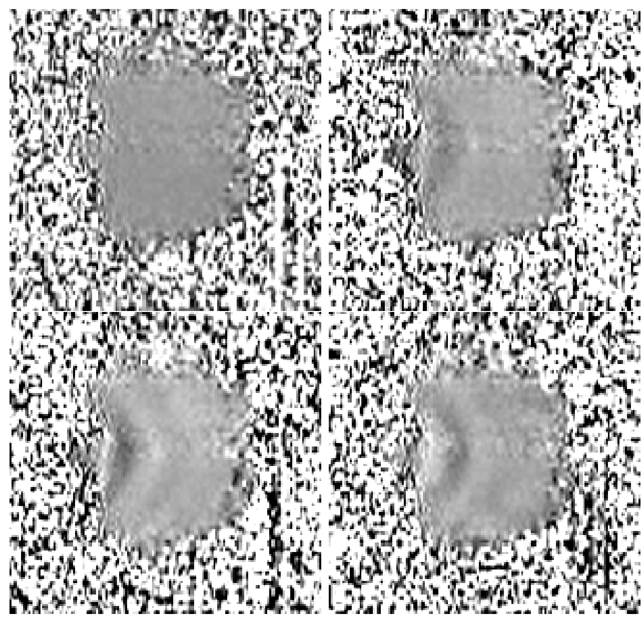

Figure 8. Four frames from a series of images visualizing the velocity encoding of a transverse wave propagating through a gelatin filled dish phantom.

As a final demonstration, MR imaging was performed during the excitation of a 2D RF pulse using a "flyback" excitation. During each flyback period an echo was obtained, and an image formed from the 64 channel receiver and RF coil array [7]. One difficulty is the necessity of moving the magnetization to be centered about the position in k-space required to compensate for the phase impressed upon the received signal by the RF coil phase pattern. Thus, instead of a progressive raster scan of k-space, the k-space trajectory returns to the same k-space line during each flyback period to form an image. Figure 9 shows the output from the gradient monitor showing the gradient patterns required for this k-space trajectory. Figure 10 shows three of a series of 32 images formed during excitation of the RF pulse, designed to form an "M" of magnetization (though the "M" is on its side in these images".)

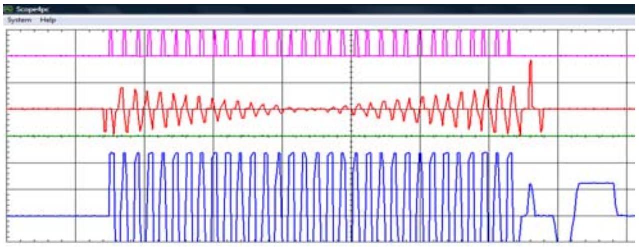

Figure 9. Gradient monitor output from a 2D RF pulse excitation using a "flyback" sequence. One gradient (red) returns the magnetization to be centered on the k-space value necessary to correct for the phase ramp of the RF coils.

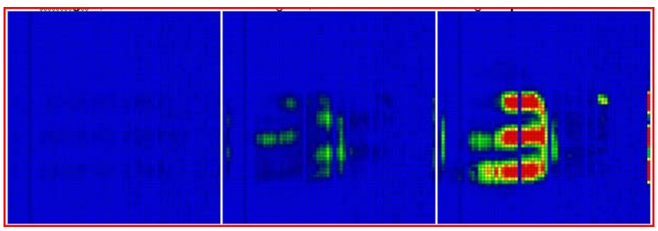

Figure 10. Three frames acquired during the excitation of a 2D RF pulse using single echo imaging. These are frames 1, 10 and 22 out of 32 made during the 60 msec excitation, illustrating the ability to monitor the development of transverse magnetization in real-time.

This preliminary work has illustrated the potential for extremely fast MR imaging, and at the same time potential limitations that must be overcome. The phase imparted by the coils remains a complication and our group is investigating multiple methods to overcome this, including the application of a conjugate phase with the 64 channel transmitter or to use a custom gradient coil to compensate for the phase distribution.

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