Optical MR Receive Coil Array Interconnect

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Motivation

Recently there has been a trend to dramatically increase the number of available receiver channels in MR systems. Groups from major manufacturers have demonstrated the benefits of coil arrays of up to 32 channels. For large arrays, common mode signals and ground loops present challenges in maintaining signal integrity and patient safety. The purpose of this work is the exploration of non-conductive signal paths in an MR coil array, such that no conductors are connected between the imaging coils and the rest of the system. Fiber optics has been proposed as a viable way to reliably carry RF signals in harsh environments, being virtually free of all electromagnetic interactions. By removing conductive ground connections, we expect to see improvement in array detuning effects of cables and transient noise immunity. The risk of RF burns in imaging subjects by the presence of cable loops is reduced. In addition, we expect to be able to implement high channel counts with small interconnect cross sectional area cables. Another added benefit is that cables do not need to be an integer number of half wavelengths for impedance match.

Architecture

The goal of the research is to eliminate all electrical conductors between receive coils and image processing equipment. Analog optical transmission with an external optical modulator, illustrated in Figure 1, requires the least amount of hardware and electrical power at the coil while still meeting the stringent noise figure requirements of MR imaging. The electrical to optical conversion takes place in a Mach Zehnder optical modulator. The modulator receives a constant power optical signal from a remote laser via an optical fiber. This constant amplitude signal is intensity-modulated by the RF signal from the receive coil, transmitted out of the imaging volume via fiber, and converted back to an electrical signal for processing in a photodiode receiver. An electrical RF amplifier is required after the coil for noise performance. The matching network passively matches the output of the amplifier to the input of the modulator. The critical performance characteristics of the components are shown in Table 1. The amplifier is the only component near the coil that requires electrical power (a few hundred milliwatts) assuming passive blocking and a bias-free modulator. This can be delivered using an additional fiber for coupling high power laser energy to a photovoltaic.

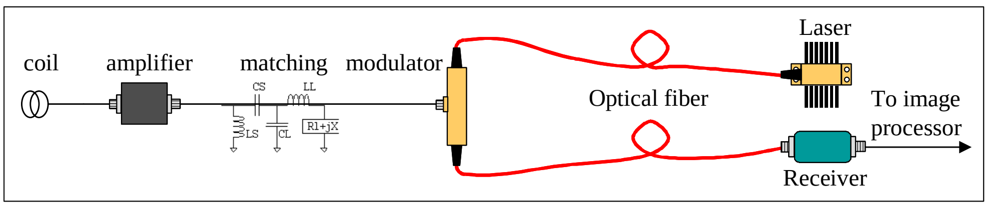

Figure 1. Single coil optical receive architecture for minimal hardware and required power at the coil

Electrical amplifier gain	30 dB
Electrical amplifier noise figure	< 0.5 dB
Matching network gain	16 dB
Modulator conversion efficiency	1 mW/Volt
Laser intensity noise	-160 dB/Hz
Photodiode responsivity	0.9 Amps/Watt

Table 1. Component Performance

Material Issues

Most commercially available optical components are not designed for use in high magnetic field environments. The components' packaging along with electrical and optical connectors often are constructed from Kovar or contain Ni plating making it difficult to preserve the uniform magnetic fields established in MRI systems. Therefore, our initial system tests placed the electro-optic components outside the bore of the magnet. Separate tests were conducted to examine the performance of commercially available $LiNbO_3$ optical modulators in high magnetic fields. This was accomplished by placing a commercial modulator in the bore of an electromagnet with field strengths up to 2.5 Tesla. The transfer function of the modulator was examined at various field strengths, voltage bias conditions and physical orientations. No change in performance was found, suggesting that commercial $LiNbO_3$ modulators are suitable for use in high magnetic fields such as those associated with MRI.

Results

A performance model was developed to predict noise figure and distortion of the amplifier, matching circuit and optical link. The model results for noise figure and distortion are plotted as a function of the combined gain of the amplifier and matching circuit in Figure 2. Intermodulation ratio (IMR) for the 3rd harmonic was used as a distortion metric with –32 dBm of total input signal power. To achieve the required noise figure of 1 dB, a front-end gain of 40 dB is required, but distortion performance is below our 60dB requirement. Several forms of linearization including linearized modulators and post-processing are under investigation. A feasibility demonstration of the architecture was performed using a single head coil on a 1.5 Tesla GE Signa Scanner. Because off-the-shelf modulators contain magnetic materials, the modulator was located outside of the imaging volume. The images shown in Figure 3 were obtained with no discernable image quality degradation.

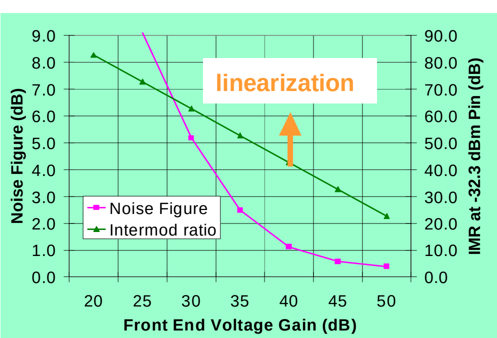

Figure 2. Model predictions of noise figure and distortion

Figure 3. MR images recorded through optical link

Conclusions

We have demonstrated the feasibility of generating MR images without the use of any electrical conductors between the subject and equipment outside of the imaging volume. Future work includes removing the magnetic materials from optical modulator package and testing the architecture with the hardware in the imaging volume.