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Identification and mitigation of interference sources present in SSB-based wireless MRI receiver arrays

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Abstract

Purpose—Single sideband amplitude modulation (SSB) is an appealing platform for highly parallel wireless MRI detector arrays because the spacing between channels is ideally limited only by the MRI signal bandwidth. However this assumes that no other sources of interference are present outside that bandwidth. This work investigates the practical interference between multiple SSB-encoded MRI signals.

Methods—Noise from coil preamplifiers and carrier bleed-through are identified as sources of interference. Two different SSB systems were designed for 1.5T with different noise filtering properties. We show how the differences between the filtered noise profiles impact the received MR signal's dynamic range (DR_{sig}) and image signal-to-noise ratio (SNR) through simulation, bench measurements, and phantom imaging experiments.

Results—When operating individually in the MR scanner, both SSB systems were shown to minimally impact the original DR_{sig} and SNR. On the other hand, when all eight channels were operating simultaneously, an average SNR loss was observed to be 12% in the one system, while a second system with more complex filtering was able to achieve a 3% loss in SNR.

Conclusion—Successful wireless transmission of multiple SSB-encoded MRI signals is possible as long as channel interference is properly managed through design and simulation.

Keywords

Magnetic Resonance Imaging (MRI); Wireless Link; Frequency Multiplexing; Single Sideband Amplitude Modulation (SSB)

Introduction

In recent years, there has been an increasing research effort into the transmission of multiple MRI signals. This demand is fueled by the increasing size of receiver arrays, which now have as many as 128 coil elements (1,2). These arrays require a substantial amount of

coaxial cabling. Besides the significant cost of these cables, large cable bundles can also experience coupling that can reduce signal quality and can increase the risk of RF burns when not designed properly. Therefore, any technology that could reduce the need for cables could improve the complexity, safety, and cost of an MR system. However, a practical implementation of such a cable-less system has proven elusive.

The primary constraint with reducing the number of coaxial cables is that the modified signal transmission system should have a minimal negative impact on the MRI signal and the reconstructed image. Time-division multiplexing signals from multiple coils into a single receiver port has been shown to reduce the number of coaxial cables required for an array (3), but the maximum sampling rate of the system analog to digital converter (ADC) limits the extent of cable reduction. Direct optical modulation of the MRI signal replaces coaxial cables with optical cabling (4,5), but an analog optical link's dynamic range is typically insufficient for high dynamic range MRI signals (6). Hardware-based compression can reduce the transmission dynamic range of the MRI signal while preserving the reconstructed image quality (7). By combining these technologies, high dynamic range multi-channel optical transmission systems have been created (8). For higher degrees of cable reduction, frequency division multiplexing (FDM) has been proposed to transmit all the signals in a large coil array using only a single cable (9) or wirelessly (10–12). A digital FDM implementation has been presented based on 802.11b (12), but placing an individual digital-to-analog converter on each coil has the potential to become too expensive in cost and power consumption. This also adds additional design concerns over introducing broadband noise into the MR environment from the required digital switching signals. Analog FDM techniques require precise synchronization between the transmitter and receiver systems to avoid cascading phase errors in the recovered signals (11).

When only considering the bandwidth of the MR signal, it would seem that analog FDM methods, such as single sideband amplitude modulation (SSB) (13–15), would be superior to digital transmission. Digital transmission with the 802.11b protocol requires 25 MHz spacing between neighboring channels to avoid overlap between the modulated signals (16). Ideally, the spacing in SSB is only limited by the bandwidth of the MRI signal, allowing for a much denser transmission spectrum. However, this advantage assumes that there are no potential sources of interference outside the MRI signal bandwidth.

In this work, we identify potential sources of interference experienced between multiple SSB-encoded wireless MRI signals, including the increased bandwidth of preamplifier noise and noise resulting from the carrier signal itself. We introduce a simulation method that identifies the appropriate channel spacing to be used when transmitting multiple SSB signals by minimizing the interference from these noise sources. We use this simulation with two separate SSB systems that have different levels of filtering. We predict the optimal performance for both SSB systems in hypothetical transmission settings, and the simulation is used to determine the channel spacing in an eight channel SSB-based receive array transmitting wireless signals in the 900MHz/GSM band. The systems are used to collect phantom images and are compared against images collected directly from wired connections. The results presented in this work have importance in not only SSB-based wireless receive arrays, but can be extended to any analog FDM application in MRI.

Theory

SSB-Encoded MRI Signals

If one only considers the bandwidth of the MR signal, the spacing between individual SSB channels would only need to be a few hundred kHz. Unfortunately, the mixed MR signal contains off-band noise that could interfere in adjacent SSB channels when they are

combined during air transmission, which impacts the modulated signals' noise floors. Figure 1a is an illustration of a typical MRI signal with SSB modulation using the filtering method (13). The noise from the coil preamplifier has a bandwidth that is much larger than that of the MRI signal, and there is bleed-through of both the carrier and its phase noise. If two of these spectra were combined with a small spacing, figure 1b illustrates how the preamplifier noise would overlap into a neighboring MRI signal bandwidth, increasing the noise floor. Alternatively, if one used a wider spacing that was on the order of the Larmor frequency, phase noise from the carrier would interfere with the signal bandwidth, as shown in figure 1c. Thus, one must take great care in both understanding and compensating for these effects when designing an SSB system and deciding the spacing between multiple channels.

Simulation of Noise Contributions in SSB

A simulation method was developed in MATLAB to determine the channel spacing that yields the least amount of noise interference when transmitting multiple SSB-encoded MRI signals (17). The simulation generates the spectra from all the possible channel spacing combinations and calculates the expected noise increase in each of the encoded signal bandwidths. Channel spacing combinations are not limited to just uniform spacing, but could also contain mixed amounts of spacing between the transmitted channels. To determine all the possible spacing combinations, the simulation requires multiple input parameters: the number of SSB channels, the frequency step-size between possible carriers, the antenna transmission bandwidth, and the SSB noise power spectrum. The antenna bandwidth is the frequency range between the transmitting and receiving antennas where air transmission meets a certain path loss requirement determined by the SSB design. The SSB noise power spectrum should be representative of the typical SSB-encoded MRI signal when connected to an element of the coil array. The spectrum should have a frequency span sufficiently wide enough to capture all the possible sources of noise interference and a frequency resolution on the order of the MRI signal bandwidth. With these input parameters, the individual SSB spectra for each possible carrier frequency can be generated and combined additively to create the spectra for possible channel spacing permutations. Using the combined spectra, the amount of interference in each channel is calculated by totaling the noise power increase in the encoded MRI signal bandwidth along with any additional bleed-through contributions that occur during the MR receiver's down-sampling. The simulation records the maximum channel noise increase for each combination, and the overall channel spacing is chosen as the combination that creates the smallest increase.

Materials & Methods

SSB Design

To utilize our simulation method and investigate the effects of overlapping noise profiles in SSB-based MRI signals, two different SSB systems were constructed for 1.5 T. The two systems were designed to have similar operation, but with different noise profiles. The operation of both systems is summarized schematically in figure 2a. The signal from each coil's preamplifier was fed into a transmitter module where SSB modulation was performed. The modulated signals from all coils were then combined, amplified, and transmitted wirelessly with a single antenna. In this implementation, a single antenna pair was used to transmit and receive all the signals to avoid negative effects from antenna coupling. The transmitting antenna was a surface mount 1/4-wave monopole planar antenna (SP series, Linx Technologies). The receiving patch antenna (18) was constructed based on a 1/2 wavelength rectangular design with a center coaxial feed point and air dielectric thickness of 1.27 cm. Both antennas were tuned to a center frequency of 868 MHz and matched to 50 ohms. The amplifier immediately before the transmitting antenna was ADA-4743 (Avago Technologies), and the amplifier immediately after the receiving antenna was RF2314

(RFMD). Signal combination and distribution was performed with cascaded series of lumped-element two-port Wilkinson power splitters. The cumulative wireless signal was received outside the MRI scanner bore and split evenly into multiple receiver modules. Each receiver module corresponded to a particular transmitter module, recovering the coil's original signal content and filtering away the remaining signals from the other transmitters. The recovered signals were then fed directly into the MRI receiver ports for image reconstruction as if they were normal MRI coil signals.

The first SSB system was designed with the intention of implementing a minimally complex SSB system. The transmitter module, depicted in figure 2b, contained a mixing stage (MAX2661, Maxim Integrated) and an amplification stage (ERA-5XSM, Mini-Circuits), and the lower sideband was selected with a lumped-element, 7th order Chebyshev low pass filter. The receiver module, depicted in figure 2d, contained a downconverting mixing stage (MAX2682, Maxim Integrated), and the recovered MR signal was selected with a lumped-element 5th order Chebyshev band pass filter. The Chebyshev filters were hand-built using a filter design program which is available online (19). When a module was powered by a 10 V source (which was necessary to power the coil preamplifier), each transmitter module consumed about 800 mW power, and each receiver module consumed about 300 mW.

The second SSB system was designed to have an improved noise profile over the first SSB system as depicted in figure 2c. To filter the off-band preamplifier noise, an intermediate frequency (IF) stage at 43.6 MHz was included with a commercially available surface acoustic wave (SAW) filter (X6964M, EPCOS). At the time of design, we were not able to source an affordable commercial option for a 63.6 MHz SAW filter. The 43.6 MHz SAW filter IF stage was chosen because when mixed with a high frequency carrier, there was still a large spectral gap present for SSB filtering, and the 20 MHz source was simply created through multiplication (PI6C4511, Pericom) of the 5 MHz reference clock (described in detail later this section). The drawback to the 43.6 MHz filter was that it has a relatively large -30 dB attenuation bandwidth of 7.6 MHz. The IF mixer was a low-power broadband mixer LT5560 (Linear Technology), and an extra amplification stage (ADA-4543, Avago Technologies) was necessary to compensate for SAW filter losses. The RF mixer was replaced with MAX2671 (Maxim Integrated), which had similar performance to MAX2661, but with additional conversion gain and carrier suppression. The SSB filter was replaced with a SAW filter from TriQuint Semiconductor. The particular SAW filter used depended on the module's carrier frequency (part numbers used in this work: 855821, 856932, 855728, 856824.) This new transmitter module consumed about 1150 mW using a 10 V supply.

Carrier generation for both SSB systems is illustrated in figure 2e. The carrier frequencies utilized were located in the 900MHz/GSM band. Each module's carrier was generated locally using an integer-N PLL (ADF4111, Analog Devices) and VCO (MAX2623, Maxim Integrated) combination. The loop filter had a bandwidth of 100 kHz and a phase margin of 45 degrees. The carrier frequencies produced were limited to integer multiples of the clock's 5 MHz due to phase instabilities present in the integer-N PLL. In order to eliminate cabling to control the transmitter modules, the carrier frequency was controlled locally with a microprocessor (PIC12F635, Microchip). The microprocessor had 1.75 KB internal FLASH memory that was used to store a look-up table of predefined settings to define the multiple carrier frequencies for the module to generate. External resistors and capacitors connected to the microprocessor defined a specific RC time constant, which selected a specific entry in the look-up table. Upon module power up, the microprocessor immediately probed the time constant and programmed the module. Following that, the microprocessor entered "sleep" mode, which allowed image acquisition to occur without clocking noise interference (10).

To ensure phase stability and synchronization between the carriers produced by the transmitter and receiver modules, a highly stable 5 MHz clock signal was fed to all the modules. Using a spectrum analyzer (CXA series, Agilent), the phase noise of the clock was measured to be less than -90 dBc/Hz for frequency offsets above 100 Hz with -75 dBc spurious signal at 120 Hz offset. In addition, the frequency drift was measured to be less than 0.5 ppm over a 60 second interval. The clock was generated using a divide-by-two counter (SN74LS92D, Texas Instruments) and a 10 MHz crystal oscillator (CXOH20, Crystek Corporation). In this implementation, the clock signal is fed via coaxial cable to the array to eliminate any possible contamination from unstable reference sources.

SSB Array Construction

To test our channel spacing simulation method, both SSB systems were used with an eight element receive array. Eight similar rectangular coils (27.3×13.3 cm) were evenly placed on 25.4 cm outer diameter FR4 cylinder. Each element was tuned and matched to 63.6 MHz with an approximate $Q_{\text{unloaded}}/Q_{\text{loaded}}$ of 4.2 (256/61). Neighboring coils were overlapped for nearest neighbor geometric decoupling, and low impedance preamplifiers (Quality Electrodynamics, Cleveland, OH, USA) provided decoupling for the remaining interactions. Transmission decoupling from the transmit pulse was performed only passively with crossed diodes. In this implementation, active decoupling was excluded due to its cabling requirement for switching PIN diodes. However, a non-electrical signaling solution, such as optical detuning (20) or a wireless tone decoder, could be utilized as a replacement. One cable was used to transmit the clock and DC power to each transmitter module. The eight receiver modules were connected to a standard receiver cable, which could be directly plugged into the patient bed. Figure 3 shows a picture of the completed array.

Wireless Channel Spacing Simulation

The channel spacing simulation was performed for both SSB systems when transmitting the eight array signals. The step-size between possible carriers was 5 MHz for both SSB systems, due to the use of an integer-N PLL. We chose a path loss requirement of -30 dB as this was above the expected performance between the antennas when operating in the MRI scanner within the VCO output frequency range (21). Using an S_{21} measurement and an 80 cm transmission distance, the usable bandwidth between the system's antennas was found to be 75 MHz (841–916 MHz). The path loss variation measured in this bandwidth was approximately between -25 to -30 dB. This allowed for 16 possible carrier values for the 8 channels to utilize. The SSB noise power spectra for both systems were acquired through two measurements. First, the noise power of the modulated MRI signal before SSB filtering was measured with a spectrum analyzer. The noise power was measured with a 0.5 MHz bandwidth at intervals of 0.5 MHz for a 200 MHz frequency span centered on the carrier. This fully characterized the carrier and preamplifier noise. Next, the profiles of the SSB filters were measured with S_{21} using the same frequency span and resolution. The two measurements were necessary to account for the variability of filter performance across a wide range of frequencies (particularly for the SAW filters in the second SSB system). Noise power for each channel was calculated as the noise power at the encoded center frequency with an additional 1% contribution from the upper sideband in the final downconversion stage (from 63.6 MHz to 1.4 MHz.) This final fold-over occurs due to bleed-through in the MR receiver down-sampling stage (corresponding to 65.0–67.5 MHz.) Using these parameters, the simulation was performed for both SSB systems to find the optimal channel spacing, and the results were used in the array.

Antenna Bandwidth Simulation

It is clear that the upper limit of the number of channels in such an SSB system is proportional to the total system bandwidth, which is currently limited by the antenna

bandwidth. Thus, to predict the maximal theoretical performance for other systems, additional simulations were performed for both SSB systems using a variety of antenna bandwidths (75, 150, 250, 500 MHz) and a narrower carrier resolution of 0.5 MHz. With such a large number of channel spacing combinations to examine, it was impractical to go through every possible combination, so the wireless array simulation was modified to use simulated annealing (22) to estimate an approximately optimal channel spacing combination. The cost function for the simulated annealing was the average noise power increase experienced in all channels. Noise power for each channel was calculated as the average noise power over a 2.5 MHz signal bandwidth (corresponding to a 62.5–65.0 MHz center frequency) with the additional 1% signal contribution from the down-sampling stage in the MR receivers (65.0–67.5 MHz). For both SSB systems, a single representative SSB profile was used, as shown in figure 4. Since simulated annealing only examines a fraction of the total number of combinations, it is possible that the global optimum cost function may not be found with only a single optimization. Therefore, the optimization was repeated at least 20 times for each arrangement of SSB system, antenna bandwidth, and coil number, and the minimum cost function was recorded.

Single Channel Bench Measurements

Before testing the array's performance with the MRI scanner, the individual channels from both SSB systems were tested with a series of bench measurements. The standard for evaluating a non-coaxial transmission system is the loss in the image signal-to-noise ratio (SNR). For a single MRI detector, an equivalent way to represent SNR is through the acquired signal's dynamic range (DR_{sig}). Gabr et al described the mathematical relationship between the two values (6):

$$DR_{sig} = 20 \log_{10} \frac{2}{\sqrt{N}} \sum_{n=1}^N SNR_n \quad [1]$$

SNR_n is the individual pixel complex-valued SNR in an image with N pixels. The value of DR_{sig} is dependent on the same parameters as SNR, such as image contrast, coil size, proximity to patient, etc. As described in the following subsections, both SSB systems were characterized to evaluate performance in preserving DR_{sig} and SNR during single channel transmission.

Noise Figure—The incorporation of additional electronics between the preamplifier and the MRI system's ADC will introduce noise to the signal, increasing the signal's noise floor. This effect is typically described with the system noise figure (NF_{sys}), which is defined as the difference in signal SNR before and after the transmission system, expressed in decibels (23). Assuming the system imposes no nonlinearity effects to the signal, NF_{sys} can also be defined as the difference between the acquired signal's dynamic ranges:

$$NF_{sys} = DR_{sig,in} - DR_{sig,out} \quad [2]$$

For a system with multiple stages with separate gains and noise figures, the noise figure cascade is calculated with the Friis system noise figure equation (24). Using that, it is possible to predict the NF_{sys} for the preamplifier and SSB system together:

$$NF_{sys} = 10 \log_{10} \left(F_{PA} + \frac{F_{TX} - 1}{g_{PA}} + \frac{F_{air} - 1}{g_{PA} g_{TX}} + \frac{F_{RX} - 1}{g_{PA} g_{TX} g_{air}} \right) \quad [3]$$

'g' is the gain in normal units and 'F' is the noise figure in normal units, also known as noise factor. This equation reveals that NF_{sys} is dominated by the first stage if the gain is sufficiently large to minimize the noise figure impact from the remaining stages. 'PA' represents the coil preamplifier, and 'TX' and 'RX' represent the SSB transmitter and receiver modules. 'Air' represents the wireless link that includes both path loss and the impedance-matched antennas. It is assumed that there are no noticeable noise contributions from the MR suite environment during air transmission. 'Air' is regarded as a simple attenuation stage, where F_{air} is assumed to be equivalent to the inverse of the path loss g_{air} . If significant, path loss can have substantial impact on NF_{sys} even though it occurs after the first few stages of the receiver chain. To avoid this, it is important that the combined gain of the preamplifier and the transmitter module is sufficiently large to overcome the path loss. Therefore, the increase in NF_{sys} for both SSB systems was predicted using equation 3.

Carrier Dynamic Range—One assumption in all analog mixing schemes is that the carrier spectrum is a simple delta function. However, realistic carriers contain reference spurs and phase noise at frequency offsets surrounding the tone. Since these errors in the carrier are included in the mixed MR signals, the dynamic range of the carrier ($DR_{carrier}$) determines the maximum DR_{sig} that can be successfully encoded and transmitted in SSB. If the MRI signal was mixed with a low $DR_{carrier}$, the carrier's phase noise could interfere with the lower power spectral components of the MRI signal. To characterize the impact of these errors, $DR_{carrier}$ for both systems' transmitter and receiver modules was measured with a spectrum analyzer. Phase noise was measured at frequency offsets of 1 KHz, 10 KHz, and 100 KHz. Reference spur power was also measured at integer multiple offsets of the 5 MHz reference.

−1dB Compression Point—The upper linearity limit of the SSB system was evaluated by measuring the −1 dB compression point input referred to the coil preamplifier (P1dB). A single wireless channel was measured from each SSB system, and the system gain was recorded for varying amounts of 63.6 MHz input power. The transmitter and receiver modules were connected with a −30 dB attenuator to simulate path loss. Using the measured P1dB values in decibels, the linear dynamic range of the system (DR_{P1dB}) was then calculated by subtracting P1dB with the coil's noise floor and the added noise from the coil preamplifier:

$$DR_{P1dB} = P1dB - (-174dBm + 10\log_{10}B) - NF_{PA} \quad [4]$$

where −174 dBm was the thermal noise in a 1 Hz bandwidth at 290 K, B was the bandwidth of the acquired MRI signal, and NF_{PA} was the noise figure of the coil preamplifier.

Imaging Experiments

All imaging experiments with the array were performed on a Siemens 1.5 T Espre. The receiving patch antenna and the eight receiver boards were placed at the end of the scanner patient table 80 cm from the transmitting antenna and SSB array.

First, an imaging experiment was performed to determine if operation of the SSB systems increased the noise floor of the MRI signal. A water phantom was placed inside the array, and the system's body coil was used to acquire noise scans when the SSB system was powered and unpowered. The standard deviation for each acquisition was calculated and compared.

In order to validate our noise model of multiple wireless coils, a set of imaging experiments was performed to measure DR_{sig} before and after wireless transmission. To accomplish this, four separate images were acquired: (1) a "wireless image" collected using the wireless

array, (2) a “wireless noise image”, which was the same as (1) but with the body coil transmitter voltage set to zero, (3) a “wired image”, which had the same imaging parameters as (1), but cabling was connected directly to the coil preamplifier output, and (4) a matching “wired noise image”. DR_{sig} was calculated and compared for both the wireless and wired image acquisitions by doubling the peak k-space amplitude and dividing by the standard deviation of the noise acquisition as described by Gabr et al (6).

SNR maps for each element were calculated for both the wired and wireless acquisitions and examined for artifacts. SNR maps were calculated by taking the magnitude of the reconstructed coil images and dividing by the standard deviation of the corresponding complex-valued noise image.

These experiments were performed using a large cylindrical water phantom (200 cm² cross-sectional area, 36 cm length). A series of two-dimensional gradient echo (GRE) experiments were performed: TE = 10 ms, TR = 150 ms, FA = 47° (Ernst angle), 225 mm × 225 mm, 128 × 128, BW = 100 Hz/Pixel. Slice thickness was adjusted to vary DR_{sig} . In addition, the same GRE experiments were performed with a higher bandwidth (BW = 1350 Hz/Pixel). First, these acquisitions were performed with only a single channel of each SSB system. Then, the same acquisitions were repeated with all eight channels operating simultaneously.

In order to estimate the error in the measurements described above, fifteen additional phantom GRE acquisitions were performed with a slice thickness = 2.5 mm and BW = 100 Hz/pixel, where the signal from an individual coil was collected directly from the preamplifier. The standard deviation of DR_{sig} and the SNR map mean was calculated across the acquisitions.

Results

Single Channel Bench Measurements

The low noise preamplifiers used in this research had a gain of 27 dB and a noise figure of 0.5 dB. In the first SSB system, the transmitter modules had a predicted 10.0 dB noise figure and 24.7 dB gain. The second SSB system had a predicted 11.5 dB noise figure and 28.5 dB gain. The difference between the gains of the two systems was due to the additional amplification in the IF stage and the difference in attenuation between the Chebyshev and SAW filters in SSB filtering. The receiver modules had a predicted 9.5 dB noise figure and 7.2 dB gain. Using equation 3 and the −30 dB path loss estimation, the expected performance of NF_{sys} was predicted to be 0.8 dB for the first system and 0.7 for the second system. Despite the gain difference between the two systems, the expected difference in SNR performance should still be small when transmitting a single channel. These results have been summarized in table 1.

The carrier noise of each transmitter and receiver module was measured. In the first system, the maximum phase noise measured was −80 dBc/Hz at 1 KHz offset, −82 dBc/Hz at 10 KHz offset, and −80 dBc/Hz at 100 KHz offset. The maximum spurious power measured for the first system was −47 dBc at 5 MHz spacing and −61 dBc at 10 MHz spacing. In the second system, the maximum phase noise measured was −78 dBc/Hz at 1 KHz offset, −78 dBc/Hz at 10 KHz offset, and −80 dBc/Hz at 100 KHz offset. The maximum spurious power measured for the second system was −54 dBc at 5 MHz spacing and −61 dBc at 10 MHz spacing.

The P1dB of a single channel with −30 dB attenuation was measured to be −43 dBm for the first system and −49 dBm for the second system, both input-referred to the coil preamplifier input. Both these measurements were limited primarily by the linearity of the RF mixer in

the transmitter module. The difference between the measurements was due in part by the additional gain provided by the second system's IF stage. When using equation 4 and a 100 kHz signal bandwidth, the calculated DR_{P1dB} was 80.5 dB for the first SSB system and 74.5 dB for the second system. These values for DR_{P1dB} are on similar order, but slightly lower than the values obtained with $DR_{carrier}$. DR_{P1dB} will vary based on the acquisition used. When transmitting multiple SSB channels, the linearity limit of the system shifts to the downconverting mixer in the receiver module, because no filtering was used at the input of the receiver modules and all the wireless signals are fed into the mixer. The measured compression point of a single receiver module was -12.5 dBm.

Channel Spacing Simulation

Example single channel noise profiles from both systems are shown in figure 4. In the first system, the preamplifier noise is visualized well, along with the bleed-through of both the carrier and the unwanted upper sideband. In the second system, the improved filtering provided by the SAW filters can be seen in the reduced bandwidth of the preamplifier noise and the further SSB suppression.

Figure 5 summarizes the results from the channel spacing simulation. The graphs are histograms of the maximum noise power increase experienced in the channel spacing combinations. Figures 5a and 5b summarize the simulation results using the SSB array's operating parameters. For the first SSB system, the best channel spacing combination was found to be when the channels were spaced apart by an even 10 MHz, while the remaining combinations have larger amounts of expected interference. For the second SSB system, the histogram reveals that all the combinations have improved interference performance when compared to the first system due to the improved filtering present in the second SSB system. Again, the best channel spacing combination was found to be an even 10 MHz spacing across channels, but there were additional combinations that could have been chosen with similar interference performance. In order to further demonstrate the utility of these simulations, figures 5c and 5d summarize the results when the parameters were modified so that nine channels were transmitted instead of eight. In this case, channel spacing combinations with low amounts of interference no longer exist, because at least two channels must be spaced apart by only 5 MHz, resulting in lower performance. This demonstrates how simulation can help determine the limits of an SSB system and assist in the design stage.

In order to show the most direct comparison between these systems in our scanner testing, a constant 10 MHz spacing was used for both systems in the array, where the receivers had carrier frequencies 905, 915, ... 975 MHz.

Figure 6 illustrates the agreement of spectra between the simulation and the actual SSB array. Figures 6a and 6c show the cumulative spectra predicted by the simulation. Figures 6b and 6d show the actual measured spectra from the SSB arrays. For both SSB systems, there is good agreement between the simulated and acquired spectra. In particular, the interference behaviors between neighboring preamplifiers agree well. It should be noted that the individual channel gains of the first system were manually adjusted to lessen the impact of interference from carrier bleed-through. These adjustments (an additional gain of 2.5 dB to the 955 MHz channel and 5 dB to the 965 MHz and 975 MHz channels) made the predicted noise interference more uniform across the individual channels.

Antenna Bandwidth Simulation

Figure 7 illustrates the simulation results for multiple antenna bandwidths and a 0.5 MHz carrier resolution. For each antenna bandwidth, there is an abrupt increase in interference,

which can be used as an indicator for the upper limit of performance for that particular configuration. For example, the predicted amount of interference for the first SSB system and a 75 MHz antenna bandwidth begins to increase at eight coils, corresponding to our observed results. As expected, a larger antenna bandwidth and better filtering allows for more channels to be transmitted without a substantial amount of noise power increase. Given an appropriate antenna bandwidth configuration, more than 32 channels could be transmitted using either system. This figure shows how simulation can also be used to guide the selection of an appropriate antenna bandwidth and frequency band when designing an SSB system for MRI applications.

Imaging Experiments

The noise acquisitions collected with the body coil revealed no increase in noise standard deviation when either SSB system was powered, demonstrating no impact on the signal due to operation of the electronics.

Figure 8 summarizes the imaging experiments performed with the first SSB system. When transmitting only a single channel (figure 8a), there is nearly identical performance between the wired and wireless measurements. When all eight channels were transmitting simultaneously (figure 8b), the DR_{sig} acquired wirelessly was noticeably lower than the wired, with an average of 12% SNR reduction present for the smallest slice thicknesses. Visual analysis of the SNR background maps (figure 8d) reveal no structured artifacts in the $TH = 2.5$ mm acquisition, eliminating the possibility that this loss was due to system non-linearity and implying it was entirely due to interference from the other channels. Compression of DR_{sig} begins to be observed at $TH = 10$ mm for single channel transmission and $TH = 7.5$ mm for simultaneous transmission. This is likely due to slight non-linear amplification occurring at these high signal levels, which is confirmed by the structural artifacts present in the SNR maps.

Figure 9 summarizes the imaging experiments performed with the second SSB system. When transmitting a single channel (figure 9a), there is good agreement between the wired and wireless measurements. When all eight channels were transmitting simultaneously (figure 9b), only a 3% SNR reduction was present at the smallest slice thicknesses. Signal compression at larger slice thicknesses is more pronounced for both single channel and simultaneous transmission. Visual analysis of the SNR background maps (figure 9d) suggest that this is again likely due to non-linear amplification. This agrees with the bench measurements where the second system had both increased gain and reduced linearity. To further demonstrate that the compression during simultaneous transmission was primarily due to receiver non-linearity, an additional set of simultaneous transmission experiments were performed with an additional 3 dB attenuation at the input of the transmitting antenna (figure 9e). At the expense of slightly reduced SNR preservation, the non-linearity losses at the larger slice thicknesses were reduced.

Figure 10 summarizes the SNR performance of the individual array channels during simultaneous transmission for the low BW and $TH = 2.5$ mm acquisitions. Mean SNR was calculated as the average across the entire magnitude-based SNR map, including background pixels. For each SSB system, the eight wireless channels show a similar scale of decrease. In the first SSB system, mean SNR was preserved on average by 88% (minimum 84%), and in the second system, mean SNR was preserved on average by 97% (minimum 94%).

The standard deviation of DR_{sig} for the repeated phantom measurements was 0.1 dB, and the standard deviation of the average SNR was 1%, thus indicating that these results are

reproducible within a narrow range. These values were used as an error estimate for the figures described in this section.

Discussion

The results presented here demonstrate that multiple SSB-encoded MRI signals can be simultaneously transmitted wirelessly with proper design and interference considerations. Preamplifier noise and carrier bleed-through were identified as potential sources of interference. Through both the simulation and filtering of these sources, good SNR preservation was obtained when transmitting eight SSB channels. When imaging a phantom with the smallest slice thicknesses, the first system preserved mean SNR by 88%, but the second system, with improved noise filtering, preserved mean SNR by 97%.

Simulation is also necessary to evaluate the limitations of SSB systems. For example, if either of the SSB designs presented here were used in a receive array with more than eight channels (as illustrated by the simulation results in figure 6c and 6d), both systems would be incapable of maintaining such high levels of SNR preservation. The simulations summarized in figure 7 demonstrate how either SSB system would need to be modified to allow for more channels to be transmitted. Interference can be reduced by either implementing more aggressive filtering of the preamplifier noise and carrier noise or by implementing a larger antenna bandwidth by using a higher transmission frequency and/or additional sets of antennas. This will also require a narrower carrier frequency resolution by using a smaller reference frequency or a fractional-N PLL.

The designs presented here should provide researchers with a good starting point on their own development of a fully functional wireless SSB array. For example, the effects of non-linear amplification were clearly observed during the simultaneous transmission of higher power signals. The inclusion of a channel selection filter and a variable gain stage into the receiver design would address this problem (25). These components would also provide a means to mitigate the negative effects of air transmission not experienced in this work including transmission antenna coupling and multipath distortion.

Another important topic not addressed here is the clock synchronization between the elements and the MR scanner. Here, we used a wired connection for the reference signal to focus on the interference from the preamplifier and carrier noise. Any additional phase noise from lack of synchronization would cascade into the carrier, and frequency drifting would introduce additional drifting phase errors into the acquired signal. A stable crystal oscillator signal was distributed to all modules via coaxial cable to ensure that the phase error accumulation did not drift outside the pixel bandwidth. However, some other solution would have to be developed to move to a fully wireless coil array setup. The synchronization clock could be transmitted wirelessly to the array, but this high signal quality would need to be preserved even after path loss. Synchronous detection with the carrier could be used (11), but once again, signal quality must be maintained during wireless transmission.

The DC power consumption for these SSB systems should be considered. Given the P_{1dB} measurements, the maximum output power delivered by each transmitter module to the antenna is 7.4 mW for the first system and 4.5 mW for the second system. Yet, each transmitter module dissipates approximately 1 W to achieve nearly 80 dB DR_{P_{1dB}}. If a larger DR_{P_{1dB}} system were needed (eg. for 3D acquisitions), the power requirements would likely further increase. It should be noted that the SSB designs presented here were not optimized for efficiency. For example, the transmitter power in either system could have been reduced by nearly a factor 2 if the coil preamplifier did not require 10V (the majority of the components in each design required 5V or less). Even with an improved design, it is

likely that the large power requirements in highly linear SSB systems will make long-term battery operation difficult.

Previous work has estimated that the majority of clinical MRI acquisitions have DR_{sig} between 50 dB and 85 dB (6). The SSB systems presented here were measured to have $DR_{carrier}$ of approximately 80 dB, not covering the entire clinical range. The effects of this limitation were not clearly visualized during imaging experiments because non-linearity effects were dominant at similarly high signal strengths. $DR_{carrier}$ is an important measurement in SSB, as it will likely be the limit of SNR preservation in more sophisticated designs.

The usefulness of wireless receive array technology extends beyond its application in massively parallel receiver arrays. We have already begun exploring other MRI applications where wireless transmission could prove beneficial. For example, we have demonstrated that small numbers of coil elements can be combined into wireless tracking devices (26), which we believe will play a key role in advancing interventional MRI applications. And, we have also demonstrated that a small array of wireless NMR probes could be used for the real-time monitoring of gradient fields (27) which could provide this information without significant system changes.

Conclusion

Our results indicate that successful wireless transmission of multiple SSB-encoded MRI signals is possible as long as channel interference is properly managed through design and simulation. Simulation of multi-channel transmission is an effective means to minimize channel interference and to determine a wireless system's limitations. We have demonstrated this by constructing a system with eight high-quality channels in the 900MHz/GSM band with a final SNR preservation of 97% on average. It is hoped that the techniques described here can lead to more widespread use of wireless SSB techniques in targeted MRI applications.

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References

1. Schmitt M, Potthast A, Sosnovik DE, Polimeni JR, Wiggins GC, Triantafyllou C, Wald LL. A 128-channel receive-only cardiac coil for highly accelerated cardiac MRI at 3 Tesla. *Magnetic resonance in Medicine*. 2008; 59(6):1431–1439. [PubMed: 18506789]
2. Hardy CJ, Giaquinto RO, Piel JE, Rohling A, Kenneth W, Marinelli L, Blezek DJ, Fiveland EW, Darrow RD, Foo TKF. 128-channel body MRI with a flexible high-density receiver-coil array. *Journal of Magnetic Resonance Imaging*. 2008; 28(5):1219–1225. [PubMed: 18972330]
3. Porter JR, Wright SM, Famili N. A four-channel time domain multiplexer: A cost-effective alternative to multiple receivers. *Magnetic resonance in medicine*. 1994; 32(4):499–504. [PubMed: 7997116]
4. Yuan J, Wei J, Shen GX. A 4-Channel Coil Array Interconnection by Analog Direct Modulation Optical Link for 1.5-T. MRI *IEEE transactions on medical imaging*. 2008; 27(10):1432–1438.
5. Memis OG, Eryaman Y, Aytur O, Atalar E. Miniaturized fiber-optic transmission system for MRI signals. *Magnetic Resonance in Medicine*. 2008; 59(1):165–173. [PubMed: 18098294]

6. Gabr RE, Schär M, Edelstein AD, Kraitchman DL, Bottomley PA, Edelstein WA. MRI dynamic range and its compatibility with signal transmission media. *Journal of Magnetic Resonance*. 2009; 198(2):137–145. [PubMed: 19251444]
7. Kose K, Endoh K, Inouye T. Nonlinear amplitude compression in magnetic resonance imaging: Quantization noise reduction and data memory saving. *Aerospace and Electronic Systems Magazine, IEEE*. 1990; 5(6):27–30.
8. Biber, S.; Baureis, P.; Bollenbeck, J. Analog optical transmission of 4 MRI receive channels with high dynamic range over one single optical fiber. *Proceedings of the 16th Scientific Meeting of ISMRM; Toronto, Ontario, Canada*. 2008. p. 1120
9. He W, Qin X, Jiejing R. Four-channel magnetic resonance imaging receiver using frequency domain multiplexing. *Review of scientific instruments*. 2007; 78(1):015102. [PubMed: 17503941]
10. Riffe, MJ.; Heilman, JA.; Gudino, N.; Griswold, MA. Using on-board microprocessors to control a wireless MR receiver array. *Proceedings of the 17th Scientific Meeting of ISMRM; Honolulu, Hawaii, USA. Honolulu, Hawaii, USA*. 2009. p. 2936
11. Scott, G.; Yu, K. Wireless transponders for RF coils: systems issues. *Proceedings of the 13th Scientific Meeting of ISMRM; Miami, Florida, USA*. 2005. p. 330
12. Wei J, Liu Z, Chai Z, Yuan J, Lian J, Shen GX. A realization of digital wireless transmission for MRI signals based on 802. 11b. *Journal of Magnetic Resonance*. 2007; 186(2):358–63. [PubMed: 17433744]
13. Sabin, W.; Schoenike, E. *HF Radio System & Circuits*. 2. Noble Publishing Corporation; 1998.
14. Lathi, BP. *Modern Digital and Analog Communication Systems*. 3. Oxford University Press; 1998.
15. Haykin, S. *Communication Systems*. 4. John Wiley & Sons, Inc; 2001.
16. Shen, GX.; Wei, J.; Pang, Y. Design of Digital Wireless Transmission for 64 Channel Array using IEEE 802.1. *Proceedings of the 16th Scientific Meeting, International Society of Magnetic Resonance in Medicine*; 2008. p. 1121
17. Riffe, MJ.; Gudino, N.; Twieg, DB.; Heilman, JA.; Griswold, MA. An investigation of on the channel spacing limits in SSB-based wireless MRI receiver arrays. *Proceedings of the 20th Scientific Meeting of ISMRM; Melbourne, Victoria, Australia*. 2012. p. 435
18. Godara, L. *Handbook of antennas in wireless communications*. Vol. Chapter 6. Boca Raton, Florida: CRC Press LLC; 2002.
19. *Microwaves101*. 2007. http://www.microwaves101.com/encyclopedia/vlads_filter_designer.cfm
20. Wong EY, Zhang Q, Duerk JL, Lewin JS, Wendt M. An optical system for wireless detuning of parallel resonant circuits. *Journal of Magnetic Resonance Imaging*. 2000; 12(4):632–638. [PubMed: 11042647]
21. Steer, M. *Microwave & RF Design: A Systems Approach*. SciTech Publishing; 2012. Chapter 2: Antennas and the RF Link; p. 61-100.
22. Kirkpatrick S, CG, Vecchi M. Optimization by simulated annealing. *Science*. 1983; 220(4598): 671–680. [PubMed: 17813860]
23. Egan, WF. *Practical RF System Design*. John Wiley & Sons, Inc; 2003. Chapter 3: Noise Figure; p. 47-90.
24. Friis H. Noise figures of radio receivers. *Proceedings of the IRE*. 1944:419–422.
25. Drentea, C. *Modern Communications Receiver Design and Technology*. Artech House, Inc; 2012.
26. Riffe, MJ.; Yutzy, SR.; Blumenthal, C.; Hsu, DP.; Nakamoto, DA.; Sunshine, JL.; Flask, CA.; Gulani, V.; Duerk, JL.; Griswold, MA. Real-time scan plane selection with a novel hand-held device for needle guidance. *Proceedings of the 19th Scientific Meeting of ISMRM; Montreal, Quebec, Canada*. 2011. p. 3753
27. Riffe, MJ.; Neumann, D.; Blumenthal, C.; Lee, G.; Seiberlich, N.; Griswold, MA. Wireless Magnetic Field Monitoring. *Proceedings of the 20th Scientific Meeting of ISMRM; Melbourne, Victoria, Australia*. 2012. p. 2749

Figure 1. SSB-Encoded MRI Noise Spectra

(a) A typical noise power spectrum of a MRI signal encoded with SSB. The center peak is the carrier and its phase noise. The two sidebands are the modulated preamplifier noise. The upper sideband is suppressed with $>20\text{dB}$, and the encoded MRI signal bandwidth is indicated in the lower sideband. For presentation purposes, the diagram is not to scale. **(b)** If two SSB channels have small frequency spacing, preamplifier noise will interfere with the encoded MRI signal, and **(c)** if the spacing is too large, the phase noise from the carrier will interfere.

Figure 2. Block Diagram of the SSB Systems

(a) Both SSB systems are designed to have similar operation. Each coil's signal is fed into a transmitter module where SSB modulation is performed. The modulated signals are combined, amplified, and transmitted wirelessly. The combined wireless signal is received and distributed to receiver boards that recover the signal from one of the original coils. (b) The transmitter module in the first SSB system contains a single mixing stage. SSB modulation is realized through a low pass filter. (c) The transmitter module in the second SSB system contains an additional IF stage at 43.6 MHz. Two bandpass SAW filters are used to reduce noise sources outside the modulated signal bandwidth. (d) The receiver module was used in both SSB systems. It contains a single downconverting mixer and a band pass filter to recover the original MRI signal. (e) The carriers are produced using a PLL/VCO combination, and the frequency is programmed by a local microprocessor, which is defined by the time constant of an RC network.

Figure 3. Photo of Completed SSB Array

The array was equipped with the first SSB system transmitter modules. When operating, the receiving antenna was placed 80 cm away from the transmitting antenna on the patient table.

Figure 4. Measured SSB Noise Spectra

Example noise spectra from (a) the first SSB system and (b) the second SSB system reveal the difference in noise profiles. In addition, these spectra also illustrate the difference in the spectral gaps between the modulated signal and the carrier, highlighting the importance of simulation to manage the interference tradeoffs that exist between multiple SSB channels.

Figure 5. Channel Spacing Simulation Results

(a & b) Histograms of the channel spacing simulation for both the SSB systems demonstrate that only a few channel combinations exhibit small amounts of interference when transmitting eight channels. Because of the improved filtering in the second SSB system, the overall interference performance has improved, and there is more than one option for the optimal channel configuration. (c & d) The simulation's parameters were modified to predict interference behavior if nine channels were transmitted instead of eight. Channel spacing combinations with low amounts of interference no longer exist, since at least two channels must be spaced apart by only 5 MHz.

Figure 6. Multi-Channel SSB Spectra

Both SSB systems were programmed with an even 10 MHz channel spacing. The simulated (a & c) and measured (b & d) spectra of all eight channels reveal similar behaviors. The smoother profiles in the simulated spectra are due to each data point representing the average power in a 0.5 MHz interval.

Figure 7. Antenna Bandwidth Simulation

The simulation results illustrate the relationship between antenna bandwidth, number of channels, and noise interference for both the first SSB system (a) and the second SSB system (b). Darker data points represent the average increase in the channel noise power for the optimal channel combination found through at least 20 repetitions of simulated annealing. The lighter data points represent the corresponding maximum noise power increase experienced by the worst channel in the optimal combination. It is clear that the channel number where the average noise power begins to increase sharply indicates the upper limit of performance for each of the particular SSB design configurations.

Figure 8. Imaging Experiments with First SSB System

Summary of the DR_{sig} measurements are presented (a) when transmitting a single channel and (b) when transmitting all eight channels simultaneously. (c) Using equation #1, the reduction in SNR is predicted when comparing differences between wired and wireless DR_{sig} . (d) SNR maps are also presented comparing the wired and wireless acquisitions for TH = 2.5 mm and TH = 10 mm at low BW (100 Hz/Pixel). The smaller images are the same SNR maps, but scaled for the bottom 2% of the original in order to reveal any artifacts introduced into the background noise.

Figure 9. Imaging Experiments with the Second SSB System

Summary of the DR_{sig} measurements are presented (a) when transmitting a single channel and (b) when transmitting all eight channels simultaneously. (c) Using equation #1, the reduction in SNR is predicted when comparing differences between wired and wireless DR_{sig} . (d) SNR maps are also presented comparing the wired and wireless acquisitions for $TH = 2.5$ mm and $TH = 10$ mm at low BW (100 Hz/Pixel). The smaller images are the same SNR maps, but scaled for the bottom 2% of the original in order to reveal any artifacts introduced into the background noise. (e) An additional set of multi-channel imaging experiments were performed with an extra 3dB attenuation at the input to the transmitting antenna.

Figure 10. Mean SNR Measurements

A channel-by-channel comparison of the mean SNR values for (a) the first SSB system and (b) the second SSB system when collecting the low BW (100 Hz/px) and TH = 2.5 mm acquisition. The difference between the wired SNR measurements is due the coils being slightly modified (new tuning and matching components) when the array was switched from the first SSB system to the second.

Table 1
Summary of the predicted noise figures and gains calculated for the elements of each SSB system.

	Coil Preamp	TX Module	Estimated Path Loss	RX Module	Overall
SSB System #1					
Gain (dB)	27	24.7	-30	7.2	28.9
NF (dB)	0.5	10.0	30	9.5	0.8
SSB System #2					
Gain (dB)	27	28.5	-30	7.2	32.7
NF (dB)	0.5	11.5	30	9.5	0.7