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Device localization and dynamic scan plane selection using a wireless MRI detector array

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Abstract

Purpose—A prototype wireless guidance device using single sideband amplitude modulation (SSB) is presented for a 1.5T MRI system.

Methods—The device contained three fiducial markers each mounted to an independent receiver coil equipped with wireless SSB technology. Acquiring orthogonal projections of these markers determined the position and orientation of the device, which was used to define the scan plane for a subsequent image acquisition. Device localization and scan plane update required approximately 30 ms, so it could be interleaved with high temporal resolution imaging. Since the wireless device is used for localization and doesn't require full imaging capability, the design of the SSB wireless system was simplified by allowing an asynchronous clock between the transmitter and receiver.

Results—When coupled to a high readout bandwidth, the error caused by the lack of a shared frequency reference was quantified to be less than one pixel (0.78 mm) in the projection acquisitions. Image-guidance with the prototype was demonstrated with a phantom where a needle was successfully guided to a target and contrast was delivered.

Conclusion—The feasibility of active tracking with a wireless detector array is demonstrated. Wireless arrays could be incorporated into devices to assist in image-guided procedures.

Keywords

Magnetic Resonance Imaging (MRI); Wireless Link; Single Sideband Amplitude Modulation (SSB); Microcoil Tracking

Introduction

For the past 25 years, the role of magnetic resonance imaging (MRI) in image-guided needle biopsies has rapidly expanded (1). When first developed, MR-guidance was achieved by the simple manual alignment of the imaging plane with the susceptibility artifact produced by the biopsy needle (2, 3). Now, sophisticated systems exist that give a physician the ability to dynamically control the imaging plane and monitor the needle trajectory in real-time. This is accomplished by tracking the position and orientation of the needle insertion apparatus and using that information to update the imaging plane between each acquisition.

There are three primary methods of tracking the position and orientation of a needle apparatus inside the MR bore. They all share the common goal of localizing multiple markers with fixed positions on the apparatus. Optical tracking (4–6) uses multiple cameras with different positions and orientations to measure the locations of optical markers. By using reflective optical markers the entire apparatus can become self-contained and wireless (6). The drawback to this technique is that the optical markers must remain visible to the cameras throughout the entire procedure, which can severely limit the ability of the physician to freely wield the device in the crowded magnet bore. Passive fiducial marker tracking (7, 8) uses the signal enhancement produced by individual tuned coils that couple directly to local fiducial marker. The coils that are already being used to collect the image (body coil, spine array, etc.) are also used to collect the enhanced signal. The fiducial markers are localized by utilizing specific sequence parameters to enhance the signal of the passive fiducial marker while only slightly perturbing other surrounding signal sources. This allows for an apparatus to be entirely self-contained and wielded within the bore without regard for camera visibility. The drawback of this technique is that the enhanced marker visibility and automated localization is highly dependent on a number of parameters including the marker position and orientation, which still limits the ability to freely wield the needle apparatus inside the bore. Active fiducial marker tracking (9–13) also uses tuned coils containing fiducial markers, but connects them directly to the MR scanner treating them as standard coils. This allows for easy integration with the MR system, and the apparatus can be robustly localized regardless of marker orientation and position as long as the marker remains inside the bore's imaging region. However, this requires coaxial cables to connect the markers to the MR system. Coaxial cables may hinder the wielding of the device, and more importantly, cables can present patient safety issues (14).

In this work, we describe a device that can perform active fiducial marker tracking using a wireless coil array. This provides a self-contained device that can be freely wielded inside the magnet bore without worry of visibility or cabling constraints. Device localization with

active fiducial markers also allows for straightforward system integration. Unique encoding frequencies keep the marker signals independent during wireless transmission, which allows for simple localization. However, the use of wireless technology does introduce concerns in preserving the signal quality of the markers. Our group along with others have previously demonstrated that wireless technology can successfully transmit the MR signals in a coil array, but at the expense of some loss in the signal-to-noise ratio (SNR) (15–17). Conveniently, active tracking has a large inherent SNR (>100) and requires only small amounts of SNR to perform precise and accurate localization (18). This makes active tracking a suitable application for wireless MRI technology (19). This work outlines the design and construction of a wireless active fiducial marker tracking prototype and shows it to be well suited for MR-guided biopsies.

Methods

Wireless Design for Active Fiducial Marker Tracking

Typically, a wireless system used to transmit multiple MRI coil signals would have numerous rigorous design requirements. However, the localization of active fiducial markers only requires detecting the maximum signal magnitude in a simple projection acquisition (18), without requiring any coherent phase information. This greatly reduces the requirements for wireless transmission and enables the use of technologies that would otherwise be unsuitable for conventional wireless MRI. The wireless modulation scheme used in this work was single sideband amplitude modulation (SSB) (20–22). SSB was chosen because of its simplicity in implementation and the previous demonstration of high quality results with the platform (17).

Figure 1 depicts the SSB transmission system implemented in the three tracking coils used in this work. It is similar to a previously presented design used to transmit MRI signals for traditional imaging (17). For each tuned fiducial marker, the signal was first amplified with a low-noise preamplifier (Quality Electrodynamics, Cleveland, Ohio, USA). The signal was then mixed (MAX2671, Maxim Integrated) with a unique high frequency carrier. The carrier was generated locally using an integer-N PLL (ADF4111, Analog Devices) and VCO (MAX2623, Maxim Integrated) combination with a local 10 MHz reference clock (FOX924B, Fox Electronics). The 10 MHz reference clock was chosen because of its small surface mount package and good frequency drift performance. The PLL loop filter had a bandwidth of 100 kHz and a phase margin of 45 degrees. The carrier frequency was controlled locally with a microprocessor (PIC12F635, Microchip), which was programmed to enter sleep mode immediately after programming the PLL (23). The modulated marker signal was amplified (RF2314, RFMD) and filtered to select the lower mixed sideband using a lumped-element 7th order Chebyshev lowpass filter. The Chebyshev filters were hand-built using a filter design program (Vlad's Lumped Element Filter Designer, available online at www.microwaves101.com). When powered by a 10 V source (which was necessary to power the coil preamplifier), each wireless active fiducial marker board sourced about 60 mA and had an approximate overall gain of 45. The SSB-encoded signal was then wirelessly transmitted with a surface-mount 1/4-wave monopole planar antenna (SP series, Linx Technologies). The three wireless signals were then received outside the MRI bore with an

in-house constructed patch antenna (24). It was based on $1/2$ wavelength rectangular design with a center coaxial feed point and air dielectric thickness of 1.27 cm. The received signal was amplified (RF2314, RFMD) and split evenly using a cascaded series of lumped-element two-port Wilkinson power splitters. The split signals were then mixed (MAX2682, Maxim Integrated) with the same carrier frequencies used for upconversion, each corresponding to an original marker signal. The mixers used for upconversion and downconversion were different because they were designed for these specific frequency translation applications. The down-converting carrier frequencies used a 5 MHz clock as the PLL reference frequency. The 5 MHz clock was generated away from the MRI scanner with a 10 MHz crystal oscillator (CXOH20, Crystek Corporation) and a divide-by-two counter (SN74LS92D, Texas Instruments), and it was delivered to the receiver unit with a coaxial cable. This was the same reference clock strategy used for the SSB receiver in reference 17. The demodulated signals were filtered with a 5th order bandpass Chebyshev filter centered at the Larmor frequency. The recovered tracking signals were then fed directly into the MRI receiver ports for signal processing and marker localization.

When SSB is used for conventional imaging, the MR image is encoded in both phase and frequency information, so it is important that both the up-converting and down-converting carriers are synchronous in both phase and frequency. This requires a master clock signal to be transmitted between the wireless transmitters and receivers. However, in the application of active tracking, the fiducial markers' spatial coordinates are contained in only the projection's frequency information, so instead of transmitting a shared clock signal, separate asynchronous reference clocks can generate carriers that have the same frequency but arbitrary phase. However, small frequency differences between these two frequencies can affect the accuracy of the localization. These differences can be separated into two components: fixed frequency offset and frequency drift. Fixed frequency offset between the two carrier frequencies adds a fixed error in the encoded frequency information. This fixed error can be minimized through manual adjustment of the clocks during device construction. Frequency drift between the carrier frequencies causes an additional random error in the projection's frequency information. The extent of this random drift error depends on the quality and stability of the clocks in both the wireless transmitter and receiver. Therefore, the desired overall accuracy of the scan plane updates must be taken into consideration when selecting the clock signal sources. On the bench, the extent of frequency drift was observed with a spectrum analyzer (CXA series, Agilent). A tone of 63.6 MHz was fed into each channel of the tracking device, and the maximum drift of the recovered tone was recorded over a two minute period using the marker tools on the spectrum analyzer. Following this measurement, a rare earth magnet was placed close to the prototype to determine if a strong magnetic field introduced any additional drifting to the recovered signal.

Active Fiducial Marker Tracking

The paradigm for active fiducial marker tracking used in this work was introduced originally by Dumoulin (18). The algorithm was based on Siemens interactive real-time sequence with active tracking (25). A series of projections (Flip Angle = 5° , N = 512, FOV = 400 mm, BW = 400 Hz/Px) were acquired in three spatially orthogonal directions, and the encoded signals from three fiducial markers were collected wirelessly. The spatial locations of the markers

were determined by the location of maximum signal in each projection and channel. This assumes that each tracking coil collects a clear maximum signal from only one of the fiducial markers. The positions of the fiducial markers were compared to the known geometry of the device, and used to calculate both position and orientation of the device. This information was fed back to the MR control console, and a new imaging plane was updated in relationship to the device location for the subsequent image acquisition. Selection of the new imaging plane is flexible, and for this work a plane was selected through all 3 markers (details below). Other possibilities for future work include varying the slice center offset relative to the device and varying the slice orientation. The time required to acquire all three projections, calculate marker positions, send feedback to the MR control console, and calculate the new imaging plane was approximately 30 ms. Other projection implementations, such as the Hadamard sequence for off-resonance correction (18), were not used in this work, but could be included in future realizations.

Prototype Construction

Figure 2a is a picture of the wireless active tracking prototype constructed for a 1.5 T MR scanner (MAGNETOM Espree, Siemens Healthcare, Erlangen, Germany). The device was designed for hand-held use with dimensions 12.7×3.5×6.0 cm, where the individual transmitter boards (12.7×3.5 cm) were placed 2 cm apart inside the device. Figure 2b is a photo of one of the active fiducial markers. The fiducial markers were small vitamin E spheres with a 0.75 cm diameter. Each fiducial marker was positioned on top of an active tracking coil. The coils were made into a gradiometer with two adjacent counter-wound four-turn 0.65 cm diameter solenoids in series. This design was chosen because it ensured that the coils always received signal regardless of their orientation in the magnet. In addition, the coil's sensitivity range would be more restricted to only local signals when compared to conventional solenoids, which reduced the possibility of signal interference from the patient or operator's hand. The three coils were tuned and matched to 63.6 MHz. The carrier frequencies were chosen to be 925, 940, and 955 MHz, which corresponded to SSB-modulated tracking signal frequencies of 861.4, 876.4, and 891.4 MHz respectively. These frequencies were spaced 15 MHz apart to avoid interference from preamplifier noise (26) and centered at the transmission bandwidth between the two antennas (which was determined on the bench using a network analyzer S_{21} measurement). Non-integer multiples of the 10 MHz reference clock were possible due to an integrated reference divider in the integer-N PLL. This reference divider had phase instabilities present that made it unsuitable for standard imaging applications (17), but no deleterious effects from these instabilities were observed when used with projection magnitudes. A connection for DC power was available so that either rechargeable batteries or an external power supply could power the prototype.

The wireless receiver base station is pictured in figure 2c. It is a general-use SSB wireless receiver, capable of receiving up to eight wireless channels (17). Three of the receiver channels were programmed to the prototype's carrier frequencies to receive the modulated signals. The receiver unit was placed at the end of the patient bed, resulting in a wireless transmission distance of approximately 80 cm. Bench measurements suggested that about 30 dB path loss was experienced by the wireless signals with this transmission distance.

SSB System Bench Characterization

Previous investigations using wireless MRI detector arrays have required a thorough characterization of the wireless system in order to show that the MR signal was transmitted with minimum degradation to the original SNR (16, 17, 27). However, in the application of device localization, the projections of the active fiducial markers have a very high SNR, and the encoded spatial coordinates in those projections can be extracted at very low SNR levels. Therefore, the SNR preservation and dynamic range requirements for the SSB system can be greatly reduced when compared to a normal imaging application.

The impact that the wireless link had on the projection SNR was studied with the system noise figure, which is defined as the ratio between signal SNR before and after the transmission system, expressed in decibels (28). The overall system noise figure (NF_{sys}) was calculated using the Friis system noise figure equation (29).

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

'g' was the gain in normal units, and 'F' was the noise figure in normal units. 'PA' represented the coil preamplifier. 'TX' represented the prototype's transmitter system, and 'RX' represented the receiver unit modules. 'Air' represented the wireless link that includes both path loss and the impedance-matched antennas. This equation assumed that there were no noticeable noise contributions from the MR suite environment. 'Air' was regarded as a simple attenuation stage, where F_{air} was assumed to be equivalent to the inverse of the path loss g_{air} . Using a combination of bench measurements and device specifications, the overall system noise figure for the wireless prototype was calculated using Equation 1 for different degrees of path loss.

The dynamic range of the SSB system was also measured. This was accomplished with two dynamic range measurements. First, the dynamic range of each carrier was determined through measurements of phase noise and reference spurs using a spectrum analyzer (CXA Series, Agilent). The normalized phase noise was measured at three different frequency offsets away from the carrier (1, 10, and 100 kHz), and the maximum reference spur power was also recorded. Then, the linear dynamic range of the SSB system (DR_{P1dB}) was calculated.

$$DR_{P1dB} = P1dB - (-174dBm + 10 \log_{10} B) - NF_{PA} \quad [2]$$

-174 dBm was the thermal noise in a 1 Hz bandwidth at 290 K, B was the bandwidth of the acquired signal, and NF_{PA} was the noise figure of the coil preamplifier. P1dB was the SSB system input-referred 1 dB compression point, which is defined as the input power level where the system gain varies by 1 dB. The SSB system gain was recorded for varying amounts of 63.6 MHz input power where the prototype was connected to the receiver unit with a -30 dB attenuator to simulate path loss.

Localization Precision

The performance of the tracking prototype was characterized in the scanner by measuring the precision of localization. By measuring precision, the errors introduced by the wireless system which would cascade into the inherent accuracy of projection-based localization were observed. The tracking prototype was placed near isocenter, and 100 sets of sequential projections were acquired with the device stationary in order to capture the full range of drifting error between any of the clocks. The maximum variation between measured coordinates was recorded.

Needle Guidance Phantom Test

The performance of the wireless tracking prototype was demonstrated in a needle-guidance phantom test. The needle delivery target was a kumquat placed in a 473 mL gelatin phantom. The phantom container was opaque, so sight of the target was completely obscured. Within the MRI bore, a researcher manipulated the wireless tracking prototype with one hand while operating a biopsy needle (MR2015, E-Z-EM Inc.) with the other hand. The prototype was used to guide the needle to the target and deliver 0.5 mL of gadolinium-based contrast. A GRE imaging sequence was used (TE=5.5 ms, TR=10 ms, FA=15°, BW=250 Hz/Px, 10 mm slice thickness) with a 350×262.5 mm field of view and 128×96 matrix size. Images were collected with two elements from the patient bed spine array and reconstructed with sum-of-squares. An in-room monitor was positioned at one end of the magnet bore so that the operator could view the acquired images during the procedure. The wireless tracking prototype was battery powered during the entire acquisition.

Using the prototype's position and orientation to define a scan plane is flexible, but to present this research clearly, the information was used in only one way. The slice center was positioned approximately at signal source #3 (figure 2a) with the phase encode direction defined along the line created by sources #1 and #3, and the scan plane included all three fiducial markers. This allowed for a simple visual analysis of the acquired images to confirm accurate device localization and scan plane updates.

Results

The SNR preservation of the prototype's SSB system was characterized by evaluating the increase in the system noise figure. The coil preamplifiers had 27 dB gain and 0.5 dB noise figure. The SSB transmitter electronics had an average 18 dB gain and 9.3 dB noise figure, and the receiver modules had an average 7 dB gain and 9.1 dB noise figure. Using these values with Equation 1, it was possible to calculate SNR preservation for different levels of wireless path loss. Under the normal operating conditions of 30 dB path loss, the overall system noise figure was only 1.45 dB. This corresponded to only a 20% drop in SNR when compared to a wired connection directly from the preamplifier. For a substantial amount of path loss (45 dB), the overall system noise figure became 9.72 dB, corresponding to 88% loss in SNR. These values would have varied slightly as the orientation of the device changed in the magnetic field (30). However, the active fiducial marker projections had such high SNR initially that localization still occurred reliably in the presence of these losses.

Multiple dynamic range measurements were also performed. The measured carrier phase noise for all carriers and frequency offsets never exceeded -90 dBc/Hz. The maximum spurious signal in the carriers was -55 dBc. The average P1dB was -54 dBm input-referred to the coil preamplifier. This value was limited primarily by the combination of both the up-converting mixer and the amplifier immediately following it. When using Equation 2 and a 20 kHz signal bandwidth, the SSB system had 76.5 dB linear dynamic range. These dynamic range measurements demonstrate that the SSB system was able to encode and transmit projection signals with very high SNR.

On the bench, the maximum frequency drift observed with the spectrum analyzer did not exceed 400 Hz. No additional drifting was observed when a strong rare earth magnet was placed in close proximity to the prototype. Since the tracking projections had a bandwidth of 400 Hz/Px, the errors caused by drift in marker coordinate calculation was less than 1 pixel or 0.78 mm (400mm / 512 pixels). This error was confirmed when marker locations were measured with the stationary prototype. In the sets of 100 projections, the acquired one-dimensional spatial coordinates varied no more than 1 pixel. If the projections were acquired with a lower bandwidth, the SNR for the tracking projections would have increased even further, but at the expense of possibly increasing drifting errors to above 1 pixel.

Figure 3 is an example tracking projection collected with the tracking device. In this example, the coil face was oriented perpendicular to B_0 . Even at different coil orientations, no failure in peak detection was observed in this study. The combination of a fiducial marker with a high signal density and a coil with a highly localized sensitivity creates high SNR projections with easily identifiable maximum amplitudes.

In total, 150 frames of images were collected during the phantom needle guidance procedure, and a video of these images can be found online as supplemental material. The spatial resolution of the images has been increased four-fold through bicubic interpolation to more closely mimic the images displayed by the in-room monitor. Figure 4 presents selected frames from the procedure. First, the target was visualized with coronal slice positions (figures 4a and 4b), where the operator's right arm is also visible. Then, the wireless tracking prototype was repositioned to view the target at a new angle and allow for better needle guidance and visibility (figures 4c and 4d). The needle was then delivered to the target (figures 4e and 4f), contrast was injected (figure 4g), and the needle was retracted (figure 4h). Finally, the target was visualized at different viewing angles to confirm contrast delivery. Figure 4i, which has the same image orientation as figure 4b, shows the change in target contrast (along with views of the operator's arms and neck). Figure 4j has another slice orientation that captures the complete needle trajectory along with the contrast-filled target. The presence of the tracking markers in each image validates the operation and scan plane tracking accuracy of the wireless prototype.

Discussion

The results of the work demonstrate that active tracking with a wireless coil array is feasible. A wireless coil array with a simplified SSB design introduced errors up to 0.78 mm into the measured coordinates of the fiducial markers. Successful operation of the wireless tracking

prototype was demonstrated in a gelatin phantom, where a needle was guided to a target and an injection was delivered. It is clear that a plethora of other more sophisticated wireless designs could be used to achieve similar results, but the goal of this work was to demonstrate that even a basic wireless design could satisfy the relatively simple requirements for active fiducial marker tracking. In this work, transmission of a frequency reference signal was not necessary, so it was not included in the design. Wirelessly transmitting this reference would have added additional complexity to the SSB design, and it would have provided another opportunity for device failure.

In this realization, the wireless tracking prototype was separate to the needle insertion device. This required both the user's hands to operate the prototype and wield the needle. The advantage to this setup is that it gives an interventional radiologist a flexible way to control the imaging plane independent of the interventional device. Alternatively, the wireless active tracking fiducial markers could be integrated directly into the surgical device, which would then require only one hand to perform the intervention.

The wireless tracking prototype could benefit from further design improvements beyond solely improving the wireless electronics. An interface, placed locally on or near the device, could provide intuitive controls to toggle between scan plane orientations and positions. Incorporating a wireless gyroscope and accelerometer would allow for the device's orientation and position to be tracked outside the linear region of the gradient fields (31). Hybrid systems that include both optical and signal-based tracking have been shown to be effective in performing image-guided interventions immediately outside the MRI bore (32).

While not experienced in this preliminary study, wireless interference effects like multipath distortion could cause one of the wireless fiducial marker signals to experience very large amounts of path loss, and this would cause the received signal to fall below the receiver's electronic noise floor. This would disrupt localization and cause the tracking sequence to fail. A simple consistency check could be built into the scan plane adjustment to allow for an immediate reacquisition of the projection set. Since destructive interference is typically narrowband and very sensitive to particular transmission distances, consecutive failures would be unlikely during actual operation, and this would cause minimal disruption to the real-time acquisition. If failure had occurred too frequently, the robustness of localization could have been improved by using additional receiving antennas, additional wireless active fiducial markers, or an altogether different wireless platform that is less prone to destructive interference (for example, digital wireless transmission (16)).

Localization with a wireless coil array does not only have application in needle guidance. It could also prove beneficial in other active tracking applications, such as motion correction (33) and intravascular device tracking (34). However, we believe that simple wireless guidance devices, such as the one presented in this work, are where wireless coil arrays will have immediate clinical relevance.

Conclusions

This work has presented real-time device tracking using a wireless SSB receive array to transmit projection signals from small active fiducial markers. The design of the wireless SSB system took advantage of the reduced system requirements in active tracking as opposed to standard imaging applications. Simple wireless designs, such as the one described here, could be substantially miniaturized and incorporated into devices for image-guided procedures.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1. Block Diagram for SSB-Based Wireless Active Tracking Array

(a) Each coil's signal was amplified, mixed with a unique carrier, filtered, and amplified for transmission. The combined wireless signal was received and distributed to receiver boards that recover the signal from one of the original coils. (b) The carriers were produced using a PLL/VCO combination programmed by a local microprocessor. There was local clock controlling each individual PLL.

Figure 2. Photo of the Wireless Active Tracking Prototype

(a) The picture of the wireless active tracking prototype shows the positions of three active fiducial markers (labeled by the numbers.) (b) A close-up photo of the active fiducial marker shows the fiducial marker mounted on the coil. (c) The wireless signals were received by a general-purpose eight channel receiver constructed for SSB-based MRI (17).

Figure 3. Tracking Projections

An example tracking projection demonstrates that wireless transmission preserved the acquisition's high SNR, allowing for easy identification of the maximum amplitudes. This projection was from active fiducial marker #2 in the scanner's x-direction. In this particular projection, SNR was estimated to be more than 2000 by dividing the peak amplitude with the standard deviation of the leftmost 75 data points.

Figure 4. Phantom Images

A selection of images collected during the needle placement procedure demonstrates accurate operation of the wireless prototype. The entire series of images can be viewed as supplemental material online. (a–b) First, the target was visualized with coronal slice positions. (c–d) Then, the prototype was repositioned to view the target at a new angle for better needle guidance. (e–h) The needle was delivered to the target, contrast was injected, and the needle was removed. (i–j) Additional viewing angles of the target confirm contrast delivery.