Dynamic field monitoring by 20 channel field probes integrated with 12 channel head coil

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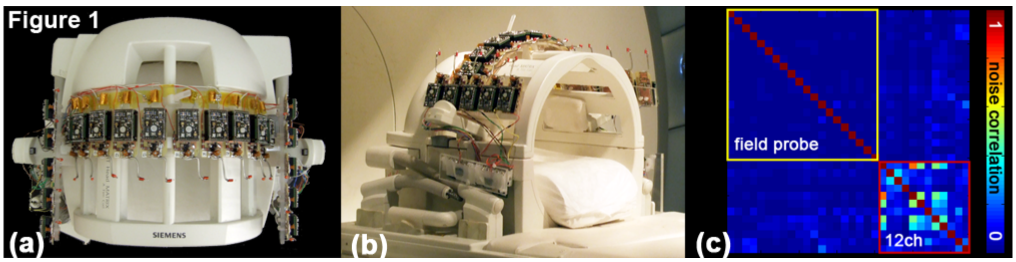
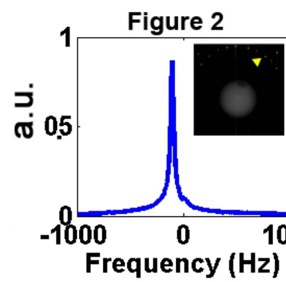
TARGET AUDIENCE: Engineers and scientists interested in hardware to monitor the spatiotemporal distribution of magnetic field during MR acquisition.

PURPOSE: Magnetic field probe has been designed to dynamically detect the frequency shift in MRI[1]. Here we aim at developing a 20-channel field probe system integrated with a 12-channel head coil array to dynamically detect the frequency fluctuations caused by the subject, eddy current, or other systemic changes during the scan. The frequency shift was continuously estimated by the phase difference monitored by the field probes[2]. This information was found useful in dynamic tracking of the magnetic field distribution and the *k*-space trajectory. Our results show that, under the limit of a fixed number of RF channels, an integrated RF field probe and coil array system holds promise of high quality MRI by measuring and calibrating MRI signal simultaneously.

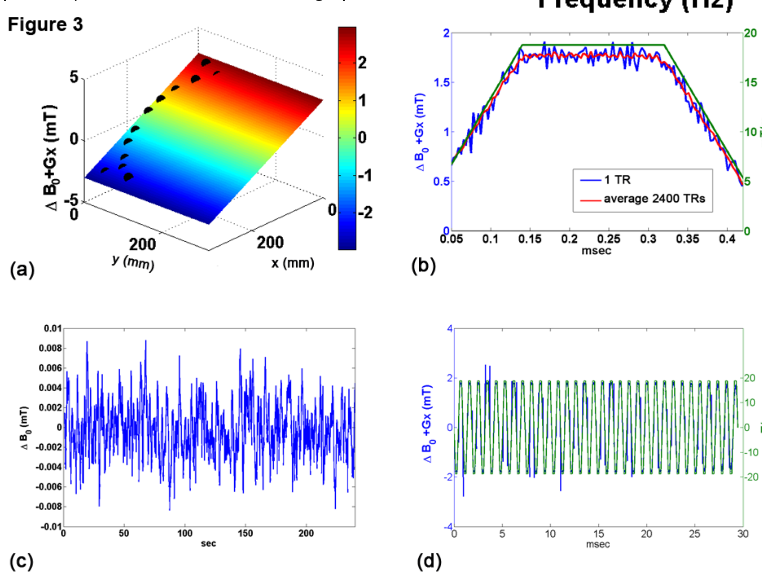


METHOD: Twenty magnetic field probes were constructed by a seven-turn solenoid using the 22 AWG copper wire. A cylindrical glass capillary tube (outer diameter: 1.5-1.8mm; wall thickness: 0.2mm; length: 3 mm) filled with water was surrounded by the field probe. Each field probe tuned to 123.2 MHz was connected to a low noise preampfiler (Siemens Healthcare, Erlangen, Germany) through a 5-cm coaxial cable and a matching network, which transformed the impedance to 50 Ω in order to obtain the lowest noise figure. The insertion loss between 20 field probes was less than -35dB, which indicated a good decoupling. These field probes were actively detuned by a parallel LC circuit in series with the PIN diode. During RF transmission, a forward biased PIN diode was turned on to detune the circuit by forming a high impedance on the field probe. This 20-channel field probe array was integrated with a 12-channel head coil array (Siemens, Erlangen, Germany; Figure 1 (a) and (b)). A mechanical housing was carefully designed to place field probes in order to avoid coupling to the head coils and to minimize the signal from the imaging object.



Data were acquired on a 3T MRI (Tim Trio, Siemens, Erlangen, Germany). Free induction decay (FID) was measured with TR = 1500 ms, acquisition duration = 512 ms, TE = 0.35 ms, bandwidth = 1000Hz, flip angle = 90° after the whole volume excitation.

The positions of field probes were located by a gradient echo pulse sequence (FoV = 320x320 mm^2, image pixel size = 1×1×2 mm^3, TR = 100 ms, TE = 10 ms, flip angle = 25°). MR inverse imaging (InI)[3] pulse sequence (InI; FoV = 256*256 mm with pixel size = 4mm × 4mm × 256mm, TR = 100 ms, TE = 30.00 ms, and flip angle = 30°) was also used to record a 4-minute dynamic scan of a subject. The frequency shift was estimated dynamically by the phase different between two contiguous frequency encoding in the first phase encoding line on the field probes: $\Phi(t)-\Phi(0)=\omega t=\gamma(\Delta B_o + G_x x)t$. With the located field probes, the difference between the linearly fitted phase and thus the magnetic field over the space and the read-out gradient gave the estimated frequency shift ΔB_o.



RESULTS: Fig 1 (c) shows the noise correlation matrix between field probes and 12 RF coils in the head array. Good decoupling was found among field probes (noise correlation max. =0.088) and between field probes and RF coils (noise correlation max=0.207). The spectrum of a representative field probe is shown in Fig 2. The magnetic field gradient in frequency encoding direction was well fitted to a linear slope (Fig 3 (a)). The incremental phases during the read-out matched the pulse sequence read-out gradient wave form (Fig 3 (b)). It is even more clear after averaging over 2400 TRs. The calculated magnetic field $\Delta B_o + G_x x$ was found slightly different from the theoretical gradient waveform, potential due to the eddy current. Fig 3 (c) shows the magnetic field fluctuation ΔB_o over time at one field probe (indicated by the yellow arrow head in Fig. 2) because of dynamic B_0 shift, susceptibility effect, or low SNR. To monitor the phase evolution after RF excitation, we compared between the gradient waveform and the field fluctuation monitored by the field probe averaged across TRs (Fig 3 (d)). Peaks in Fig 3 (d) may be caused by imperfect transient timing between frequency and phase encoding gradients.

DISCUSSION: Limited by the available RF channels in a commercial MRI system (32 in our case), we consider our 20-channel field probe system as a good combination with a 12-channel head coil array to maximize the efficiency in both generating MRI signals and collecting useful spatiotemporal calibration information for high quality MRI. Our preliminary results demonstrate the feasibility of tracking the *k*-space trajectory in dynamic parallel MRI. While the results were only shown in pulse sequence for fast fMRI, we expect that this integrated system including field probes and RF coils can also be used in, for example, diffusion weighted imaging, where eddy currents and other field disturbances are critical.

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