

Magnetic Resonance Encephalography Reconstruction with Magnetic Field Monitoring

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Introduction: In the last years field probes [1,2] have been used for dynamic magnetic field monitoring (MFM) in order to reduce image artifacts due to imperfect gradient coils and amplifiers, eddy currents, B_0 drifts and patient breathing by taking the measured field dynamics into account in the image reconstruction [3]. In Magnetic Resonance Encephalography (MREG) [4,5] none Cartesian trajectories are used to monitor functional activity at a high temporal resolution. Long MREG experiments suffer from B_0 and gradient drifts. These can be corrected for by measuring concurrently the k-space trajectory during the experiment. In the present work 4 ^1H [3] field probes are used in combination with 28 channels of a 95 channel head coil [6] in order to record the trajectory in every second repetition.

Trajectory Reconstruction: The phase of the acquired signals of the field probes reflects the magnetic field evolution at the probe positions. The field evolution $k_0(t)$, $k_x(t)$, $k_y(t)$, $k_z(t)$ can then be estimated in a least square sense from the unwrapped field probe phases, taken into account their position and their off-resonance frequency.

Image Reconstruction: The signal equation is in matrix form $S = Ax$, with the measured signal S , the forward operator A , including the measured trajectory as a 4D vector ($k_0(t)$, $k_x(t)$, $k_y(t)$, $k_z(t)$) and coil sensitivities. The image x is found by using a non-linear conjugate gradient to minimize the cost function $f(x) = \|Ax - S\|_2^2 + \lambda \|TV \cdot x\|$, where λ is the regularization parameter and TV the total variation operator [7,8].

Methods: Measurements were performed on a 3T Tim TRIO MR scanner (Siemens Healthcare, Erlangen, Germany). 28 elements of a 95 head coil array [6] were used in conjunction with a field camera 4 ^1H field probes [3] (Fig. 1) to record field variations. The field probes are operated in transmit/receive mode and connected to the spectrometer of the scanner. The separate transmit chain (signal generator: N5181A, Agilent, Santa Clara, CA, USA; RF Amplifier: 75A400, Amplifier Research, Souderton, PA, USA; power splitter: MITEQ-ESTONIA, Estonia) is controlled via trigger signals from the scanner and allowed to excite the field probes with short RF-pulses (5 μs). A spherical stack of spirals trajectory [9] is used with $\text{FOV} = 192 \times 192 \times 64 \text{ mm}^3$, a resolution of $(2 \text{ mm})^3$, a varying undersampling factor of $R_r = 3$ to 6 for the radial and $R_z = 2$ to 5 for the z k-space component, slab thickness 50 mm, $\text{TR} = 0.2 \text{ s}$, $\text{TE} = 34 \text{ ms}$, ADC dwell time = 2 μs , a readout time of 75 ms and a flip angle = 25° . A 5 min MREG measurement was performed in a healthy volunteer. In order to avoid acquiring signal originating from the head with the field probes, the trajectory and the imaging data are acquired in every second repetition with a $\text{TR} = 0.1 \text{ s}$. Image reconstruction and data analysis are performed offline in MATLAB (The Mathworks, Natick, AM, USA).

Results: A typical measured stack of spirals is shown in Fig. 2. The 1st frame of the acquired time series, reconstructed with the corresponding trajectory can be seen in Fig. 3. While the difference between the nominal and the measured trajectory is negligible at the beginning of the measurement, the trajectory drifts over time (Fig. 4). After 25 s the field drifts are linear and show breathing related oscillations. After subtracting the linear component of the drift, the temporal spectrum can be derived (Fig. 5). The k_0 component shows the strongest respiration at 0.2, its 1st harmonic at 0.4 Hz and the ECG signal at 1 Hz. The temporal spectrum of k_x , k_y and k_z show the 0th harmonic of the respiration. The temporal spectrum of k_z shows a peak at 2 Hz which was reproduced with FIDs acquired in a phantom. Therefore it must be scanner related. The temporal spectrum of a region of interest (10 voxels)³ of the reconstructed time series, using the nominal and the measured trajectory, shows that the cardiac and the breathing related peaks can be reduced (Fig. 6). The noise of the trajectory determination propagates into the reconstruction leading to higher noise.

Conclusion: Four ^1H field probes were built into a 28 channel head coil and trajectories were measured in every second repetition during an in vivo experiment. It was shown that respiration and cardiac signals can be reduced in the reconstructed data.

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References: [1] Barmet C. et al. MRM 62:269 (2009); [2] Sipliä P. et al. Proc ISMRM 2010:217; [3] Wilm B. et al. MRM 65:1690 (2011); [4] Hennig J et al. NI 34:212(2007); [5] Zahneisen B. et al. MRM 65:1260 (2011); [6] Grotz T. et al. Proc ISMRM 2010, p1496; [7] Grotz T., Proc. ISMRM, p4381(2011); [8] Rudin L., Physica D 60:259–268 (1992); [9] Irarrazabal P., MRM 33:656–662 (1995)

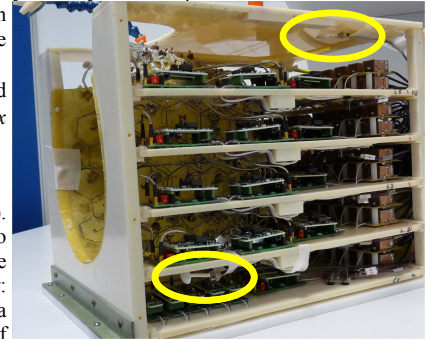


Fig. 1: 95ch head coil [6] with field probes highlighted by the yellow ellipses.

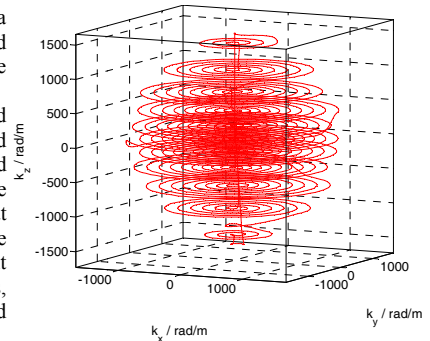


Fig. 2: Measured stack of spirals trajectory.

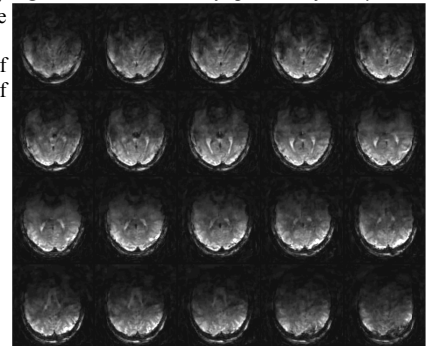


Fig. 4 Reconstructed images with the measured trajectory shown in Fig. 2.

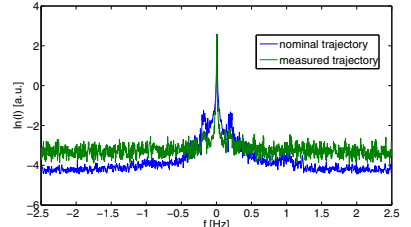


Fig. 6: Temporal spectrum from a region of interest of (10 voxels)³.

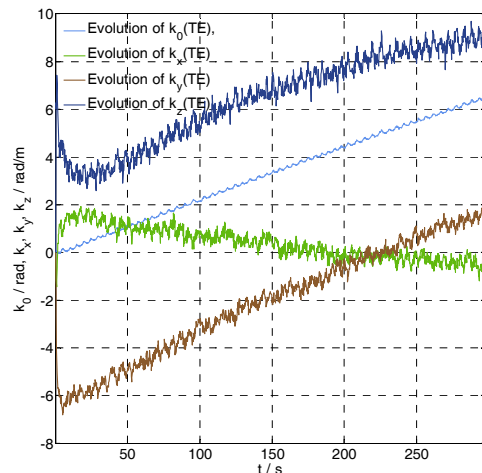


Fig. 4: Drift of the measured trajectory components over 5 min at the echo time.

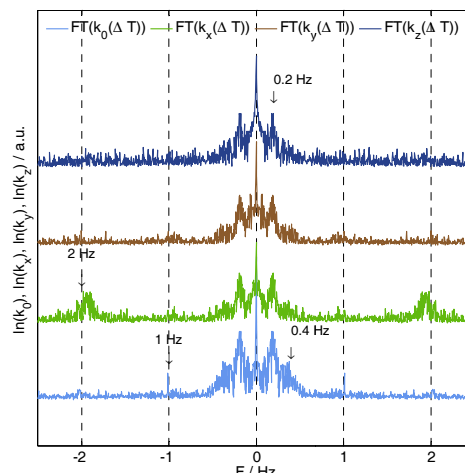


Fig. 5: Temporal spectrum of the drifts averaged over $\Delta T = \text{TE} \pm 200 \mu\text{s}$. Physiological effects as breathing (0.2 Hz) and ECG (1 Hz) are visible. The peak at 2 Hz originates from the scanner.