On dephasing effects in complex projection data: implications for rapid B0 estimation

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Introduction: Gradient Echo Planar Imaging (GRE EPI) has proven to be a fast and reliable method for detecting signal fluctuations caused by changes in blood oxygenation. Unfortunately, this high sensitivity for T2* changes can lead to a high level of geometric distortions and signal dropouts, caused by field inhomogeneities. In order to keep the acquisition speed of EPI, several fast methods have been suggested for detecting and correcting those inhomogeneities, possibly in real time (e.g. [1-2]). Among those methods, the concept of inserting short acquisitions of only few projections and estimating at least linear terms from those measurements is very popular. What is usually not taken into account when dealing with projections though, are possible dephasing effects, leading to errors in the phase of the projections, not caused by field inhomogeneities along the direction of the projection. This can be understood by considering projections to have anisotropic voxels with a reasonable resolution along one direction and no resolution at all along the other one.

We here show how dephasing along the not resolved direction can become critical and present a possible solution to the problem. The arguments presented here are not limited to the described situation of fast field mapping but addresses a general problem when dealing with complex projection data.

Methods: The signal of in-plane projections along a given direction is often approximated as

$$s(x,t) \approx e^{i\omega(x)t} \int \rho'(x,y)dy , \qquad (1)$$

with $\rho'(x,y)$ being the complex product of spin density and coil sensitivity. $\omega(x)$ is a field inhomogeneity purely depending on x. Under this assumption, the phase of the complex conjugate product of the projection acquired at two different time points, t_0 and t_1, is proportional to $\omega(x)$:

$$\omega(x) \propto \angle(s(x,t_1)conj(s(x,t_0))) = \angle(e^{i\omega(x)(t_1-t_0)}(\int \rho'(x,y)dy)conj(\int \rho'(x,y)dy)) \qquad (2)$$

This implies that the integral over the spin density does not depend on time. This, however, is a rather strong assumption and prone to fail in a real-life situation: taking into account all in-plane inhomogeneities, Equation (1) reads:

$$s(x,t) \approx e^{i\omega(x)t} \int \rho'(x,y)e^{i\omega(x,y)t}dy \qquad (3)$$

Here $\omega(x,y)$ refers to all (in-plane) inhomogeneities that do not depend exclusively on x. From Equation (3) it can be seen, that the assumptions in Equation (2) hold true only if the inhomogeneities have a negligible effect on the integral, which is the case only in idealised situations and under mild conditions, or for very short time frames (order of milliseconds). The problem can be visualised by simulating the temporal phase evolution of a point on a projection, based on field map data. Figure 1 shows exemplarily such an evolution for a brain image in data obtained in a volunteer measurement at 3 Tesla (Tim Trio Magnetom, Siemens Medical Solutions, Erlangen, Germany). As can be seen, the phase can depart from linearity within a few milliseconds after excitation.

The solution we suggest is to precede projection acquisitions by a 180° refocusing pulse, if the phase evolution of the projections is relevant (e.g. for fast field mapping). We chose to test this statement by comparing first order fit results from image intensity masked field map measurements with results obtained from projection acquisitions. For this goal we modified an EPI product sequence such that the 90° excitation pulse was followed immediately by a 180° refocusing pulse, leading to a minimum time delay of 2.6ms between the centres of the pulses. After the 180° pulse the acquisition of projections along the two in-plane axes followed, in an interleaved pattern ([3]) centred around the spin echo. The echo spacing of the projections was set to the minimum possible value of 2.08ms, which was empirically found to be fast enough in order to avoid the dephasing problem at 3T. The projections were processed in real time according to Equation (2) and the values fed back into the sequence, such that the detected inhomogeneities could be corrected for during the rest of the slice acquisition. After a synchronisation delay of 3ms the normal EPI readout train was played out.

The brain of the same healthy volunteer as for Figure 1 was imaged using a 12 channel coil array, with the following parameters: EPI: 20slcs, 2x2x5mm³ (6/8 partial Fourier), 1mm gap, TE=40ms (defined as time between 90° pulse and centre of k space) and TR=2s; Projections: 3.9x∞x5mm³ and ΔTE=2.08ms; Field-map: 20slcs, 2x2x5mm³, 1mm gap, TE1=4.92ms and TE2=7.38ms. The slices were centred around the iso-centre, covering the whole brain.

Results & Conclusion: Figure 2 compares the detected first oder in-plane inhomogeneity values for the read-out (a) and the phase encoding (b) directions. As can be seen, the values show high agreement. Especially for lower slices (e.g. slices 1-4), not refocused projections would have failed. Figure 3 shows images from EPI acquisitions without real time correction (a) and with correction (b).

The results shown here indicate that reliable complex projection data, as needed for fast field mapping, for example, can only be obtained by refocusing the signal before acquiring the projections. Otherwise the dephasing over the large voxels can become crucial and lead to unpredictable phase contributions. We have implemented a solution to this problem and shown in-vivo results that suggest that it is indeed possible to calculate accurate estimates for in-plane first order field inhomogeneities in real-time. In contrast to approaches like Dynamic Shim Update ([4]), this allows for applying optimal shimming even in the presence of motion or other physiological effects, or hardware instabilities. Differences between field map and projection values are probably caused by the intrinsic magnitude weighting of the projections; furthermore the field map processing depends on the norm (here: L2), etc. The time delay introduced by the method is acceptable, it still allows to acquire high resolution EPI data. With the described setup, SAR limitations did not pose any problems, investigations will have to show whether this holds true for higher field strengths as well. Due to the refocusing, the contrast at TE=40ms should ideally correspond to approximately TE=35ms in a normal EPI sequence. A thorough comparison of T2* contrast and sensitivity to BOLD contrast is subject to further studies, though.

In summary, it has been shown how dephasing can lead to non-linear phase evolution of projection data. When used for fast field map estimations, this problem can become crucial. Usually, the time delay inherent to an excitation pulse does not allow for accurate projection acquisition, refocusing the signal using a 180° pulse prior to acquisition serves as an effective remedy.

References: [1] Ward et al. MRM 48, 771-780, 2002; [2] van der Kouwe et al., MRM 56, 1019-1032, 2006; [3] Zaitsev et al., ISMRM, Miami, 2005, #502; [4] Blamire et al., MRM 36, 159-165, 1996.
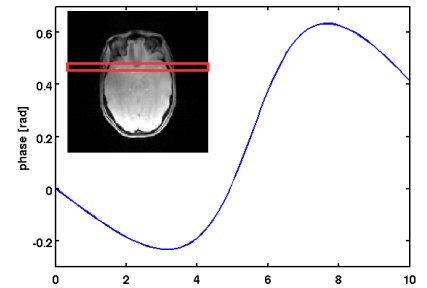
Figure 1: Phase of the projection of the red box for different time points. Accurate phase detection as in Equation (2) is only possible while the phase evolves linearly, i.e. here for 0-3ms. TE contribution of the excitation pulse and of the slice rewinder usually take about 2ms, not leaving enough time for acquisition of accurate projections.
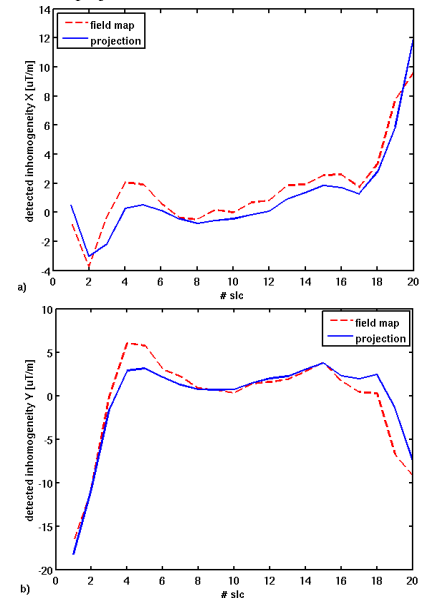

Figure 2: Comparison of detected linear field inhomogeneities (a: X, b: Y) based on field maps (red dashed curve) and on projections (blue solid curve).
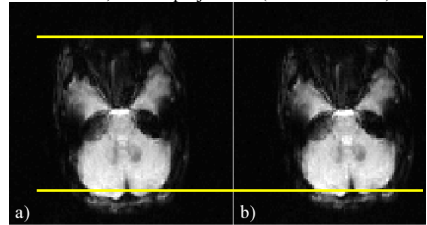

Figure 3: Comparison of not corrected EPI images (a) versus real time corrected images, in a slice with a high level of inhomogeneities. Effects of inhomogeneities in phase encoding (stretching) and read-out direction (shearing) can be observed upon careful examination.