Real-time prospective shim correction using self-referencing navigators

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Introduction: In fMRI advantage is taken of the intrinsic BOLD contrast which can be obtained with GRE-EPI sequences. Unfortunately the contrast changes are of about the same magnitude as noise, so that block design paradigms are commonly used to increase the effect to noise ratio. For this procedure to be valid, acquisition settings must be identical during the complete measurement. This is usually not the case, as motion, breathing and hardware changes lead to varying field inhomogeneities. Several methods to measure and correct for inhomogeneities using so-called navigators have been presented, most importantly the autoshimming approach by Ward et al. ([1]) or the cloverleaf method by Kouwe et al. ([2]). The drawbacks of these methods, however, are that they either require prescans, an own RF-pulse along with possible signal saturation or that they are not truly real-time. Here we present an adaptation of the MESON technique introduced by Zaitsev et al. ([3]) using a single-shot 2D navigator composed of interleaved 1D projections for real-time prospective shim correction. Our navigator, which we embedded at the beginning of a normal EPI acquisition train at 3T, can correct for inhomogeneities in as little as 8.1 ms; the modified sequence is shown in figure 1.

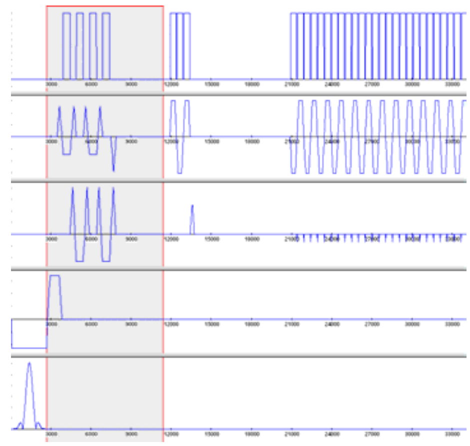

Figure 1: The modified EPI sequence; the gray area shows the navigator for two dimensions plus the delay for calculation and feedback.

Methods: Navigators have been described with different trajectories and for different purposes: orbital ([4]) or spherical ([5]) navigators for motion correction, dual-echo 1D projections for shim corrections ([1]) or cloverleafs ([2]), which can be seen as a combination of both. Here we consider only shim correction using multiple 1D projections; but unlike other methods, in our approach we restrict ourselves to keeping field inhomogeneities constant over time. Thus no comparison with reference scans needs to be performed, so that motion correction can be done in a separate step. This is a valid approach, since our navigators are self-referencing, i.e. we acquire two navigators per axis, separated by a very short ΔTE (2 to 3.2 ms, depending on resolution). A key feature of our approach is that during this ΔTE projections on other axes can be measured, that is the navigators are interleaved. Field variations can be measured by calculating the phase differences of each pair of navigators; the slope of a line fitted to the difference is proportional to the linear gradient, i.e. first order inhomogeneities, the offset of the fit to the frequency offset, i.e. zeroth order inhomogeneities. Current clinical MRI systems do not allow for real-time shim control, thus only up to first level shimming is relevant, since it can be simulated using the gradient coils. In this regime there are no inconsistencies due to missing prospective motion correction. Furthermore, since the readout bandwidth used is in the order of 0.1MHz, distortions of the navigators stay in the order of sub pixels.
Different methods for performing the fit have been tried, out of which the most efficient and reliable turned out to be to calculate the complex difference of the two navigators and then another complex difference over the elements of the result; the angle of the complex vector sum of the obtained values yields a magnitude weighted linear fit. The field drift can be estimated after subtracting a line with the estimated slope. Since data transfer and calculations could in parts be performed directly during acquisition it took less than an additional 4ms for receiving the results. We thus inserted a waiting cycle of this order after the navigator acquisitions; correction for the measured inhomogeneities could be performed immediately afterwards in the same acquisition. For the correction different approaches exist: our scanner allows for applying two gradients per axis at a time so we chose this approach; on other systems varying existing gradient moments may be an option ([1]).

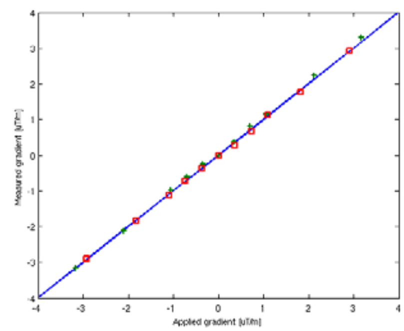

Figure 2: Plot of applied versus measured gradients: the straight line shows unity. Crosses determine gradients in x-direction, squares in y. Correlation for both was > 0.99.

Results & Conclusion: All measurements described in the following were acquired on a Trio Magnetom, 3T, Siemens Medical Systems, Erlangen, Germany.
To estimate the accuracy of our navigators we first performed phantom measurements with defined additional gradients in x and y direction ranging from -3 to 3 μT/m; figure 2 shows the results. As can be seen accuracy is very high. The standard deviation of noise in the measurements as well as in several measurements with no additional gradients was consistently estimated to be <0.04 μT/m. To obtain an estimate of the order of magnitude of physiological noise we then performed EPI brain measurements on a volunteer with free breathing of a duration of 5 min. Figure 3 shows the variation of linear gradients in y direction (anterior-posterior) estimated by our navigators and, as a ground truth, using a method proposed by Deichmann et al. ([6]); the breathing signal output by the scanners physiology monitoring unit is shown as well. As can be seen both methods have high correlation (0.85). Estimated standard deviations of the inhomogeneities were 0.084 (Deichmann) and 0.1 μT/m (our approach).
We have presented a new method for fast measurements and corrections of field inhomogeneities in true real-time: measurements and corrections take together between 8.1 and 10.6 ms, depending on resolution; further improvements are possible when using ramp sampling, which we did not use for the measurements here presented. This delay still allows for EPI data acquisition of 96x96 pixel using partial Fourier on a 3T scanner with a TE of 30ms. The preliminary results shown indicate a very good and stable correlation with existing methods.

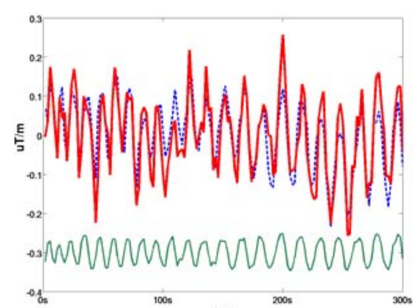

Figure 3: First order field variations measured in a volunteer over 5 minutes. The dashed line (blue) shows estimates using Deichmann's method, the solid line (red) from our approach. The solid lower curve (green) shows the breathing cycle (no units).

References: [1] Ward et al., MRM 48, 771-780, 2002; [2] Van der Kouwe et al., ISMRM, Kyoto, 2004; [3] Zaitsev et al., ISMRM, Miami, 2005; [4] Fu et al., MRM 34, 746-753, 1995; [5] Welch et al., MRM 47, 32-41, 2002; [5] Deichmann et al., NeuroImage 15, 120-135, 2002;