Spatial field monitoring using navigator echoes

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Introduction: Body movement and respiration can lead to significant field fluctuations and induce image artifacts in the brain (1,2), an effect that is more apparent at higher magnetic field strengths. Strongly T_2*-weighted sequences are often used at high field strengths, because of their high contrast between e.g. gray and white matter and low SAR. However, these sequences are very sensitive to even slight fluctuations in the magnetic field and resulting *f*0 changes (3). In earlier work we have shown that correcting for global *f*0 fluctuations (assuming constant *f*0 shifts in a slice) considerably improved image quality in Alzheimer's patients: however some artifacts remained (3), presumably due to differences in local fluctuations (within a slice). In this study we have investigated the spatial distribution of the field variations caused by body movements and respiration and have implemented a technique to estimate these local field variations using the combination of a navigator echo and the sensitivities of the different coil elements in the receive array.

Methods: Single slice single shot EPI imaging was performed on a Philips 7T scanner using a 16 channel Nova Medical receive coil with the following parameters, TR/TE/FA = 40ms / 25 ms/ 10°, EPI factor = 27, SENSE factor 2.3. Before the image acquisition a navigator echo was acquired using all 16 coil elements at TE_{nav} = 13.5 ms. Three subjects performed the following tasks during scanning: touching nose, moving arm, touching neck and taking deep breaths. A total of 4000 - 6000 images were acquired, resulting in 2'30''- 4'00'' scanning time. The first image was taken as a reference and the phase difference with respect to subsequent images was calculated: these measurements were considered the ground truth ($f0_{epi}$). The navigator echo was used to estimate the *f*0 variations in the following three ways: I) A global *f*0 variation was measured by averaging the signals from the different receive channels of the navigator echo ($f0_{glo}$). II) The spatial distribution was estimated by combining the coil sensitivities of the different elements (measured with a separate scan) with the navigator signal from the respective elements ($f0_{nav}$). III) As a refinement the spatial encoding of the navigator signal along the readout gradient (anterior-posterior direction) was taken into account ($f0_{nav2}$). The following equation was used for the spatial estimations (4):

$$f0_{na2v}(x,y,t)=\sum_{ch} NAV_{ch}(x,t)\cdot CSM_{ch}(x,y)$$ where, CSM is the coil sensitivity map.

Sum of squared difference (SSD) images were generated between the navigator *f*0 estimations and the ground truth ($f0_{epi}$) to judge the error of frequency shift estimation of the different approaches.

Results and Discussion: Figure 1 shows the time course of the average navigator signal for the different channels. A clear difference is visible in the measured *f*0 from different channels. The large changes in *f*0 correspond to touching nose (at 17s and 125s), moving arm (50s) and touching neck (90s). Figure 2 shows the spatial field distribution for a single subject while yawning. Considerable variation is visible over the transverse slice. The frequency difference ranged from -20 to 20 Hz. Figure 3a-c show the sum of squared difference images for the three navigator based *f*0 estimations (global navigator ($f0_{glo}$), spatial navigator ($f0_{nav}$) and the improved spatial navigator ($f0_{nav2}$), from left to right). Incorporating the spatial dependency of the navigator echo signals obtained from all coil elements yields a better estimation of the *f*0 distributions. Including the spatial information along the readout direction resulted in a further improvement of the *f*0 estimation. The SSD was reduced by 13-17% and 50-75% compared to the global navigator estimation for the $f0_{nav}$ and $f0_{nav2}$, respectively.

Conclusion: We have shown that the complex spatial variation in the magnetic field can be efficiently estimated using the combination of navigator echoes and the receive coil sensitivity map. In future, these results will be used to improve the correction of navigated sequences.

References: 1. Birn et al. MRM 40:55-60 (1998); 2. Van de Moortele et al. MRM 47:888-895 (2002); 3. Versluis et al. NeuroImage 51:1082-1088 (2010); 4. Hennig et al. Neuroimage 34:212-219 (2007).

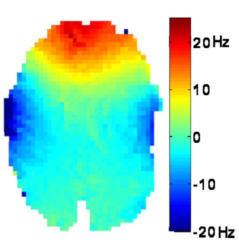

Figure 2: Single frame frequency shift map. The ground truth $f0_{epi}$ is shown during yawing (scale in Hz) for a single time point of the EPI scan.

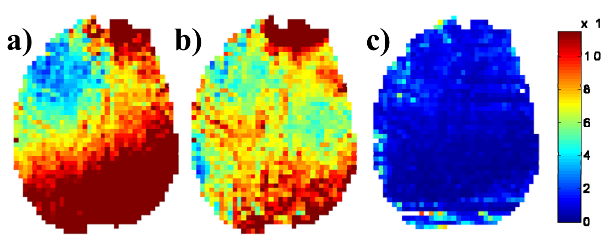

Figure 3: SSD maps calculated over the entire time course. The sum of squared difference images for one subject, comparing the difference between the ground truth $f0_{epi}$ estimation and the three navigator echo based approaches: $f0_{glo}$ (a), $f0_{nav}$ (b) and $f0_{nav2}$ (c). Lower SSD values represent better estimation of the spatial frequency distribution. A large improvement in the estimation is seen in (c) for the $f0_{nav2}$ navigator approach.

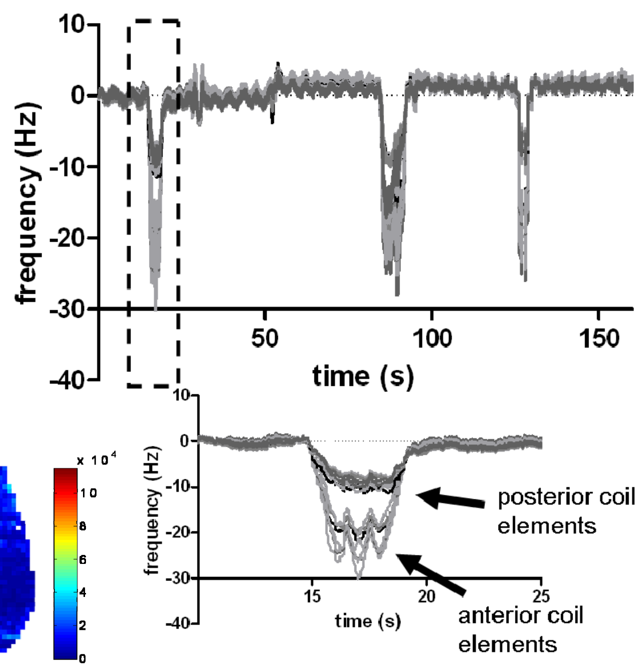

Figure 1: Time course of the navigator signal for the different coil elements. The total time course (top figure) shows the effect that body movements (see text for details) have on the resonance frequency. The bottom figure shows a zoomed part of the time course during nose touching. The arrows point to the elements located posterior and anterior, respectively showing the difference in measured frequency change.