A simple method to reduce signal fluctuations in fMRI caused by the interaction between motion and coil sensitivities

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Introduction: In functional MRI (fMRI), the BOLD signal due to brain activation varies over the time-series by a few percent depending on the paradigm and field strength used (1). To pick up these small signal changes, multi-channel receiver coils are becoming increasingly popular since they offer much higher SNR (and allow for parallel imaging). For example, the SNR is about ~4-5 times higher in brain cortex regions located close the coil elements of a 32-channel RF coil compared to a standard quadrature head coil (in-house measurements). This has naturally a high impact on the fMRI experiment as the SNR squared is proportional to the scan time. The problem that arises is that the spatial signal variation due to the coil-sensitivity field, $C(\mathbf{r})$, will post image realignment introduce a temporal signal variation in the presence of head motion since the $C(\mathbf{r})$ field is imprinted in the data in the laboratory frame. Hence, when head motion is partially correlated with the fMRI stimulus, this coil-dependent signal variation may interfere with the desired BOLD signal. This issue was recently addressed for DTI using an iterative non-linear estimation of the diffusion tensor (2), but to best of our knowledge this coil (a.k.a. bias) field vs. motion interaction has not been clearly quantified or addressed for fMRI studies using e.g. 32-channel receiver coils. In this study, we have investigated how large this signal variation becomes for realistic head motion levels, and second, explored to what extent this coil-induced temporal variance can be suppressed by simply pre-normalizing the data by $C(\mathbf{r})$ in a pixel-wise manner prior to 3D image realignment.

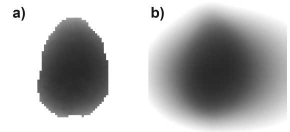

Figure 1. a) Original coil sensitivity map. b) Coil sensitivity map after extrapolation

Materials and Methods: A healthy volunteer was imaged using a GE 3T Discovery MR750 system with a 32-channel head coil (MRI instruments Inc., Minneapolis, MN). To simulate a real fMRI scan, similar scan parameters were used, however without any stimulus or paradigm. Three distinct motion types were used: a) "as still as possible", b) step-wise left-right head motion, and c) step-wise nodding motion. The magnitude of the deliberate head motion for b) and c) was near the upper limit possible in the 32-ch coil while wearing headphones without applying an unrealistic head force. 300 image volumes, each consisting of 50 slices, were acquired in one sequence with 100 volumes allotted for each motion type. With a TR of 3 s, the total scan time became 15 mins. Other imaging parameters were: FOV 24x24 cm, slice thickness = 4 mm, resolution 64x64, TE = 30 ms. The coil sensitivity map was first calculated as the ratio of the first image volume in the series with a corresponding volume acquired with the body coil (5 NEX) acquired prior to the fMRI scan (3). For a more well-behaved map, both volumes were smoothed with a 3x3x3 mm Gaussian kernel prior to taking the ratio. As large motion can result in brain data outside the measured coil sensitivity map, each slice of the map (Fig. 1a) was fitted (and extrapolated) with thin-plate splines (Fig. 1b) similar to ref (3). The fMRI dataset was 3D-realigned twice using SPM5, with and without pre-normalizing the fMRI data with the coil sensitivity map. This served two purposes: First, to check whether there were any differences in the realignment parameters. This would indicate that the motion estimates are erroneous as a consequence of that the non-normalized data is violating the sum-of-squares difference model used in the realignment process. Second, if and how the temporal relative standard deviation of the data, $\sigma_{rel}(\mathbf{r})$, is reduced for pre-normalized data.

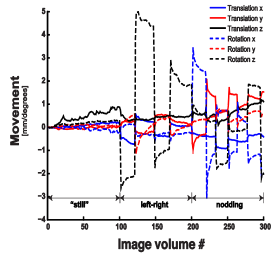

Figure 2. Motion estimates across the time series

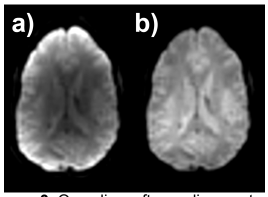

Figure 3. One slice, after realignment a) without and b) with pre-normalization by the coil sensitivity field

Results: In Fig. 2, the motion estimates across the time series are shown. The arrows bellow indicates the periods for the three intended motion types. For the estimated motion values, the difference with and without pre-normalization was too small to show in this graph, implying that the image realignment process itself is quite robust against $C(\mathbf{r})$, despite relying on image similarity. The maximum left-right and nodding movement was 5 and 3 degrees, respectively. Figure 3 shows one of the slices, after image realignment, a) without and b) with pre-normalization by the coil sensitivity field. This leads to Fig. 4, where the two maps in the top panel show the relative standard deviation maps, $\sigma_{rel}(\mathbf{r})$ and $\sigma(\mathbf{r})_{rel,prenorm}$. As these are *relative* standard deviation maps, the signal hyper-intensity seen in Fig. 3a has no direct impact on the result, rather the variation of this hyper-intensity over time. From the ROI selected in the relative standard deviation maps, a signal fluctuation plot is given below, showing the signal deviations in percent. Comparing Fig. 4 and Fig. 2, it is clear that these coincide in time. With no pre-normalization, the peak-peak signal fluctuation over the entire data set is over 5%. Conversely, with pre-normalization with the coil sensitivity map, the signal fluctuation, on the same dataset, is reduced to less than 2%. Both the fluctuations due to left-right and nodding movements are much reduced. Moreover, even during the "as still as possible" period, the signal drifts in a similar manner to the involuntary *z*-translation (Fig. 2, black), and even less than 1 mm *z*-translation over this period, resulted in ~1% signal change without pre-normalization (Fig. 4).

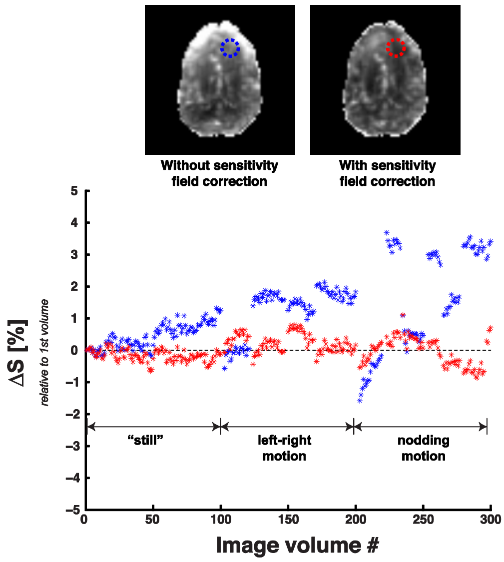

Figure 4. Relative standard deviation maps, $\sigma_{rel}(\mathbf{r})$ and $\sigma_{rel,prenorm}$, a signal fluctuation plot is given below, showing the signal deviations in percent in the ROI. Blue dots) without and red dots) with pre-normalization by the coil sensitivity field

Discussion & Conclusion
For the slice presented here, the largest $\sigma_{rel}(\mathbf{r})$ was seen in the frontal regions of the brain. Which is not too surprising as the gradient of the coil sensitivity field is large in this region and that the head is generally moving more in the front. Not taking the coil sensitivities into account could therefore lead to misinterpretations of the BOLD signal. We have here presented one simple idea to deal with this. Our method is in an early stage, and we seen other regions, primarily in the lower parts of the brain, where the correction performance was not as successful as shown in Fig. 4. Partially, this can be explained by the susceptibility gradients near the sinuses and the ear canals that we do not account for (i.e. as the head rotates, the distortion direction relative to the head changes, leading to another source of signal variation across time). Finally, the thin-plate spline fit that was used to extrapolate the coil sensitivities was performed in 2D. Next, we will use a 3D model fit of the measured coil sensitivity field and investigate if this improves the performance further.

References
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