Effective T1 based intensity compensation of spin saturation effect due to out-of-slice head motion in fMRI time series obtained via the MSV motion correction algorithm

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Introduction: Subject head motion occurring between the acquisition of each slice of a multi-slice fMRI time series, causes some nuclear spins to be re-excited before they have had sufficient time to recover to their equilibrium states. This results in a loss of voxel intensity due to *spin saturation*. The spin saturation effect at a particular voxel in an EPI slice depends on the subject's past and present head positions with respect to the scanner and an intensity compensation factor based on the effective flip angle and effective T1 of the corresponding brain region. While head trajectory can be estimated and corrected by retrospectively mapping each EPI slice into a 3D reference anatomical volume (map-slice-to-volume, MSV) [1], the effective T1 based compensation factor of each voxel in the time series data is unknown. Since EPI volumes have low resolution; each voxel may be an unknown, dynamic combination of white matter (WM), gray matter (GM) and CSF depending on subject head position when it was acquired. We present a method of using MSV results to estimate the unknown effective T1 based compensation factor for each voxel in the EPI time series. Results show that correction of the spin saturation effect for a simulated time series with motion [2] is greatly improved by using MSV-based compensation factors accounting for effective voxel T1 as opposed to compensation factors based on a single brain tissue type.

Methods: The anatomical volume used in MSV is of a much higher resolution than the EPI volumes, ensuring that after motion estimation each large EPI voxel is mapped onto a small neighborhood of anatomical voxels. Each voxel in the anatomical volume may be approximated to as isochromat belonging to GM, WM or CSF. The effective compensation factor of a large EPI voxel can be written as a convex combination of the T1 based compensation factors of each of the finer GM, WM or CSF voxels in the corresponding anatomical volume neighborhood. The intensity $I(v_j)$ of an EPI voxel at location v_j is given by $I(v_j) \propto m_z^-(v_j)(\sin\alpha(v_j))e^{-TE/T2^*_{eff}}$.
Where, v_j is the vector of coordinates of the voxel in scanner space, $\alpha(v_j)$ is the effective flip angle, $T2^*_{eff}$ is the effective T2 of the large EPI voxel and TE is the echo time. The longitudinal magnetization just before flip angle excitation pulse, m_z^- assuming that it has been pre-maturely excited n consecutive times is given by;

$$m_z^-(v_j) = m_{z0}(v_j)(1+(\cos\alpha(v_j))^{n-1}(\cos\alpha(v_j)-1)e^{-(\Delta t_1+\Delta t_2+\ldots+\Delta t_n)/T1_{eff}} + \ldots + (\cos\alpha(v_j)-1)e^{-\Delta t_n/T1_{eff}}) = m_{z0}(v_j)\, f(\alpha(v_j),T1_{eff})$$

Where Δt_m, $m = 1, 2, \ldots, n$ is the time delay between the (m-1)th and mth excitation pulse, $m_{z0}(v_j)$ is the initial longitudinal magnetization at equilibrium, $T1_{eff}$ is the effective T1 of the large EPI voxel and $f(\alpha(v_j),T1_{eff})$ is its unknown compensation factor. The magnetization of the EPI voxel can be approximated as the average magnetization over its neighborhood of small anatomical voxels $N(j)$, that belong entirely to GM, WM or CSF. After some algebra and further approximations, the unknown compensation of the EPI voxel is given by $f(\alpha(v_j),T1_{eff}) \approx \sum_{k\in N(j)} \frac{\rho_0(k)\times f(\alpha(v_k),T1_k)}{\sum_{l\in N(j)}\rho_0(l)}$, where $\rho_0(k)$ is the relative proton density of the anatomical data voxel at v_k and $f(\alpha(v_k),T1_k)$ is its compensation factor.

A mathematically simulated fMRI time series was obtained using T2 – weighted MRI volumes from the International Consortium of Brain Mapping (ICBM) [3]. Each time series slice was simulated by rotating the T2-weighted MRI volume about the x, y and z axes and extracting a slice from it. The in-plane and out-of-plane rotation angles for each slice were randomly chosen in the range of -9 to 9 degrees. The 130 simulated fMRI volumes consisted of slices stacked in an interleaved acquisition fashion, with motion and acquisition order dependent spin–saturation artifacts calculated using Bloch equations [2]. Each final volume has a resolution of 1.56x1.56x6 mm^3 in a 128x128x14 matrix. The anatomical data used was a T1 weighted MRI volume from ICBM with a resolution of 0.78x0.78x1.5 mm^3 in a 256x256x72 matrix. Thus each fMRI time series voxel was mapped onto a neighborhood of 2x2x4 anatomical data voxels, after motion estimation using MSV. Since the relative proton density ρ_0 in GM, WM and CSF is proportional to their water content, we used normalized values of 0.8, 0.72 and 1.00 respectively to estimate the effective compensation factor. The T1 values used were 900ms (GM), 600ms (WM) and 4000ms (CSF).

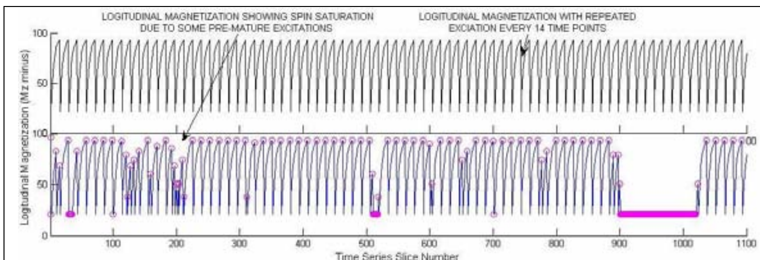


Fig. 1A Simulated longitudinal magnetization (m_z^-) of an EPI voxel with (bottom) and without (top) spin saturation.

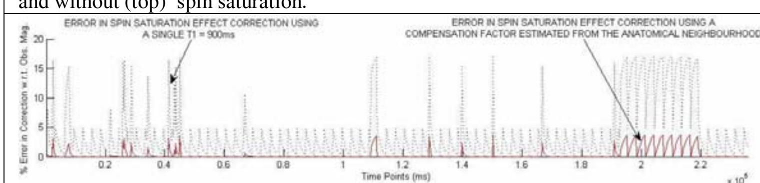
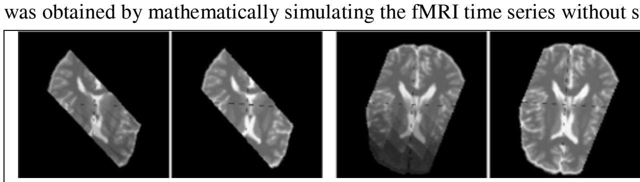


Fig. 1B Comparison of the percentage error in spin saturation correction using a compensation factor estimate based on the anatomical neighborhood and a single (most common) T1 value of GM (900ms). This EPI voxel simulation used 15 voxels of GM matter and 1 voxel of WM. (1 EPI voxel = 16 anatomical vol. Voxels)

Results: The performance of the estimated compensation factor was evaluated by simulating the longitudinal magnetization of an EPI voxel assumed to correspond to a group of 16 smaller voxels using the Bloch equations in MATLAB. Fig 1A shows the simulated longitudinal magnetization with and without spin saturation where 15 voxels were GM and one was WM. A 90 degree flip angle was assumed, with a rectangular slice profile. Pink circles denote the time points when the voxel was excited to mimic spin saturation effects. As seen in Fig 1B, the estimated compensation factor based spin saturation correction outperforms that implemented using only the T1 (900ms) of the 15 gray voxels. Figs. 2 and 3 show the great reduction in spin saturation artifacts in the simulated time series data obtained by using the MSV-based compensation factor estimates. Ground truth for Fig. 3 was obtained by mathematically simulating the fMRI time series without spin saturation artifacts.



A1 (before) A2 (after) B1 (before) B2 (after)

Fig. 2 Parts of two simulated motion corrected EPI slices before and after correction, showing reduction in spin saturation artifacts.

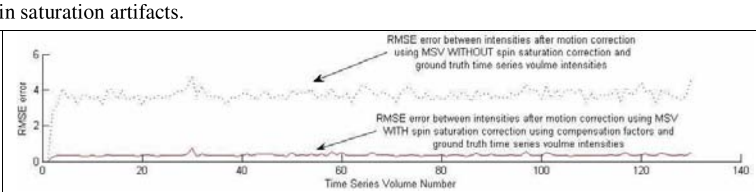


Fig. 3 RMS Error between the intensities of the ground truth time series data without spin saturation and data obtained after spin saturation correction.

Conclusion: We have developed and validated an MSV-based spin saturation correction technique. This method's accuracy is dictated by MSV's capability to estimate the motion of the subject's head between slice acquisitions. Also, the correction will improve with better anatomical data resolution with respect to the EPI volumes.

References

1. Kim, et al, MRM, 41, p964 (1999); **2**. Kim, et al, Proc. ISMRM 2006; **3**. ICBM, (www.bic.mni.mcgill.ca/brainweb/); **4.** Muresan, et al IEEE TBE 52 (8), 1450, 2005