# A Complementary Images Shimming Method to Mitigate $B_1$ Inhomogeneity for High Field MRI

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**INTRODUCTION** The $B_1$ inhomogeneity is well known to be a source of artifact in high field MRI, and requires the use of multiple-channel parallel transmit coil to reduce the artifact by making either the $B_1$ distribution (static shimming) or the flip angle distribution (dynamic shimming) as uniform as possible [1]. The former method complies with most imaging sequences, albeit limited to a small ROI for a reasonable number of channels, the latter usually implemented with a k-space spoke trajectory, offers more degrees of freedom and might cover an entire head slice or volume, but appears as a very specific technique in which the pulses design could not be easily and fully automated. We present here an intermediate and versatile approach to mitigate $B_1$ inhomogeneity based on the principle of complementary images superposition (CIS) while maximizing the signal to noise ratio (S/N). Both simulation and experimental results are reported in this abstract for a 7 T MRI scanner.

**METHOD** For a region of interest $\Omega$, we want to produce a minimal artefact and maximal S/N image by combining several images acquired with optimized static $B_1$ shimmings. The magnetization distribution in image *i* for a unit constant magnetization is given by:

$M_i(\vec{r}) = \sin(\gamma\tau\left|\sum_{j=1}^{n}\alpha_{ij}B_{1,j}^{+}(\vec{r})\right|)$ , where $B_{1,j}^{+}$ corresponds to complex-valued maps measured with the AFI sequence [3] for each channel, and $\alpha_{ij}$'s the static shimming complex weighting coefficients to be optimized in the CIS approach. According to Roemer's work [2], the optimum final combined image in term of S/N could be obtained if we choose to minimize the criterion:

$J_{SNR} = \iiint_{\Omega}\left|R_c(\vec{r})\sqrt{\sum_{i=1}^{m}M_i(\vec{r})^2} - I_t\right|d\vec{r}$ , where $R_c(\vec{r})$ is the reception sensitivity pattern related to $B_{1,j}^{-}(\vec{r})$ , $I_t$ is a constant image intensity assuming a uniform distribution of proton density, $T_1$ and $T_2$. As an example, we used an 8-channel RF coil and 2 image acquisitions (m = 2, n = 8). If m = 1, then the CIS approach would be nothing else but the classical static shimming. For the experimental demonstration, we used a round bottom flask of 16 cm diameter, filled with a CuSO4-doped saline solution providing a spherical homogeneous phantom with a $T_1$ of 330 ms. The experimental reception sensitivity pattern was not measured, but considered uniform and then assigned a unit value in the optimization criterion equation. Under this condition, we superposed the flip angle image at the end of the process instead of the intensity image itself to check the quality of the CIS technique. A simulation test has also been carried out with an inhomogeneous phantom composed of 8 human tissues in Finite Elements Method based simulations (Aarkid head model with Ansoft HFSS code). In this case, the simulated $R_c(\vec{r})$ was considered: $R_c(\vec{r}) = \sqrt{\sum_{j=1}^{n}\left|B_{1,j}^{-}\right|^2(\vec{r})}$ .

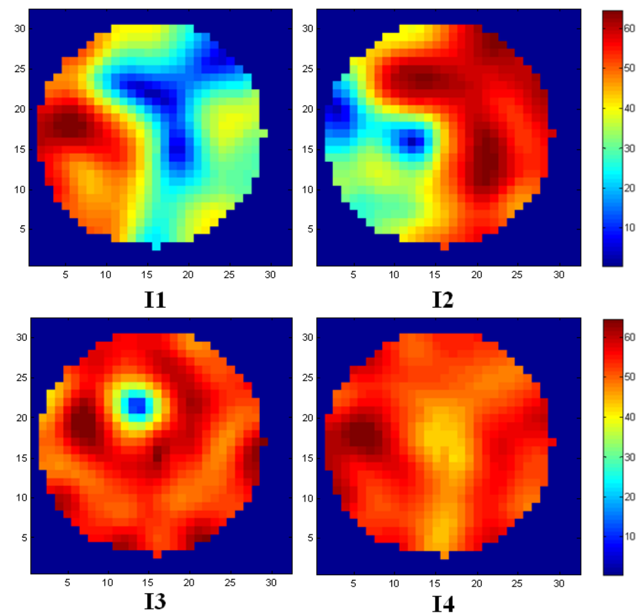


**Fig. 1**: Experimental flip angle maps (arbitrary unit, small FA). I1, I2 are first and second image obtained with the CIS technique over one slice. I3 shows an image if static shimming was used. I4 shows the combined images I1 and I2.

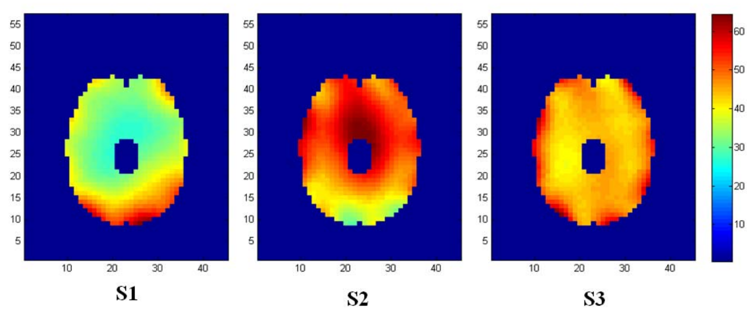


**Fig. 2**: Simulation results with an inhomogeneous phantom over a brain axial slice without diencephalon. S1 shows the reception sensitivity pattern. S2 displays the combined flip angle map while S3 the final intensity image accounting for the reception pattern (arbitrary unit, low FA).

**RESULTS** The magnetisation (or flip angle) maps in Fig. 1 were obtained over an axial slice with optimal weighting coefficients for a classical static shimming (I3), two complementary shimmings (I1 and I2) and finally a superposition of I1 and I2. The simple static shimming was not able to cover the entire ROI (islet of missing magnetization in I3). In the CIS approach, we reached excitation homogeneity of 5.4% (I4). In the simulation study (Fig 2), very satisfactory results were also obtained. The intensity enhancement at the rim of the brain axial slice comes from the structure of the RF coil in which the radiating elements were placed very close to the head.

**DISCUSSION** The CIS technique is very easy to use for a large type of MRI sequence. It allows minimizing the $B_1$ artefact on a slice or in a volume, without increasing the SAR, contrary to a dynamic shimming. For a large sphere or for the whole head, it is however difficult to get a good enough $B_1$ inhomogeneity compensation with only 2 acquisitions. Since the CIS technique does not require a time consuming optimization routine, 3 or 4 acquisitions could be considered with an increase of the acquisition time, but also with an increase of the final image quality. This method is totally compatible for most of gradient echo sequences.

**REFERENCES** [1] U. Katscher, Parallel RF transmission in MRI, NMR in Biomedicine, 2006; 19: 393-400. [2] P. B. Roemer, The NMR Phased Array, Magn Res Med, 1989; 16: 192-225. [3] V. L. Yarnykh, Actual Flip-Angle Imaging in the Pulsed Steady State, Magn Reson Med, 2007; 57: 192-200.