A General Method for Correction of Intensity Inhomogeniety in Two Point Dixon Imaging

O. Friman[1], O. D. Leinhard[2,3], P. Lundberg[2,3], and M. Borga[3,4]

[1]MeVis Research, Bremen, Germany, [2]Department of Medicine and Health, Linköping University, Linköping, Sweden, [3]Center for Medical Image Science and Visualization (CMIV), Linköping University, Linköping, Sweden, [4]Department of Biomedical Engineering, Linköping University, Linköping, Sweden

INTRODUCTION: The two point Dixon method gives pure fat and water volumes that can be used for quantitative fat imaging [1]. However, inhomogeneities in the sensitivity field of the RF coils, which cause a slowly varying intensity over the image, pose a problem that needs to be corrected for before quantitative measurements can be obtained. A scheme for estimating the inhomogeneity field has previously been proposed [2]. This method is based on an interpolation between points with known high fat content by means of a normalized convolution with a large Gaussian kernel. In this contribution, we present a more general framework where a correction field is found by fitting a set of smooth 3D spatial basis functions to voxels with high fat content in a least square error sense.

DATA: Out-of-Phase (OP) and In-Phase (IP) images were acquired at TE 2.3 ms and 4.6 ms on an Achieva 1.5 T MR-scanner (Philips Medical Systems, Best, The Netherlands) a sixteen channel sense body coil. A standard TFE sequence was acquired during a 22 s breath hold in a healthy volunteer. 40 slices with a slice thickness of 4 mm was acquired using an in-plane resolution of 2×2 mm^2. Specific sequence parameters were: sense factor 1.8, water fat shift (WFS) 0.415 pixels, TR 6.5 ms and flip angle 13 degrees. Phase sensitive reconstruction was performed using the inverse-gradient method [3,4] giving a water image and a fat image (F).

METHODS: We make the following two assumptions: The image inhomogeneity varies smoothly over the image and it is the same in the fat image (F) in the in-phase image (IP). Consequently, the quotient image F/IP is free of inhomogeneity artefacts. Furthermore, by thresholding this quotient, voxels containing pure adipose fat tissue can be found. Let $\mathbf{x}_i$, $i = 1 \ldots n$, where $\mathbf{x}$ is a point in 3D space, denote the spatial locations of fat voxels. In the absence of artefacts, these voxels should all have the same intensity in the F and IP images. Without loss of generality, we can assume that the true intensity for the fat voxels is 1, and we denote the observed intensity values in F or IP by $w_i \equiv w(\mathbf{x}_i)$. We now search for a bias correction field $g(\mathbf{x};\mathbf{h})$, where $\mathbf{h}$ is a set of parameters. Let $g_i(\mathbf{h}) \equiv g(\mathbf{x}_i;\mathbf{h})$ be short for the correction field at the fat voxels. Hence, we should ideally have $g_i(\mathbf{h})\, w_i = 1$ for all i. To find a field that comes close to this ideal, one may minimize the cost function in Eq. 1 with respect to the parameters $\mathbf{h}$. Next, to ensure a smooth correction field, we express the field as a truncated sum of Discrete Cosine Transform (DCT) basis functions. The 3D DCT basis functions are sampled at the locations $\mathbf{x}_i$ and stored as rows in the $(n \times m^3)$ matrix $\mathbf{B}$, where m is the number of DCT frequencies used. We can then write the correction field at the fat voxels as $g_i(\mathbf{h}) = \mathbf{Bh}$, where the $(n \times 1)$ parameter vector $\mathbf{h}$ contains the DCT basis weights. Insertion in the minimization problem in Eq. 1 yields the linear least squares problem in Eq. 2, where $\mathbf{W} = diag(w_1,\ldots,w_n)$ and $\mathbf{1}_n$ is a vector of ones. The optimal DCT weights $\mathbf{h}_{opt}$ are found via the normal equations in Eq. 3. Finally, once $\mathbf{h}_{opt}$ is known, the correction field $g(\mathbf{x};\mathbf{h}_{opt})$ can be extrapolated to all voxels.

$$\min_{\mathbf{h}} \sum_{i=1}^{n} (g_i(\mathbf{h}) w_i - 1)^2 \qquad (1)$$

$$\min_{\mathbf{h}} \left\| \mathbf{WBh} - \mathbf{1}_n \right\|^2 \qquad (2)$$

$$\mathbf{h}_{opt} = (\mathbf{B}^T \mathbf{W}^2 \mathbf{B})^{-1}\, \mathbf{B}^T \mathbf{W} \mathbf{1}_n \qquad (3)$$

RESULTS: The correction procedure described above was applied with m=2-7 DCT frequencies. A 2D slice of the original 3D fat volume and corrected images are shown in the first row in Fig. 1. The right-most image is the result of the previous Gaussian correction method. Fat tissue is in these images displayed in yellow-red color. The corresponding correction fields are shown in the bottom row in Fig. 1. Note the relatively high frequency edges in the Gaussian correction field. To quantify the results, the histograms of the original and corrected images were analyzed; see the middle row in Fig. 1. If there were no artefacts, all fat voxels should have the same intensity. In the presence of artefacts, we will see more dispersed intensity values in the fat voxels. To measure the effect of the correction, the histograms were first scaled so that the peak of the fat tissue bump in the histogram is located at value 1. Next, the full-width-half-maximum (FWHM) of the fat tissue bump was measured to quantify the dispersion. The result is presented in the table to the right.

Method	**FWHM**
Original	0.45
DCT (m=2)	0.40
DCT (m=3)	0.34
DCT (m=4)	0.27
DCT (m=5)	0.26
DCT (m=6)	0.25
DCT (m=7)	0.23
Gaussian	0.24

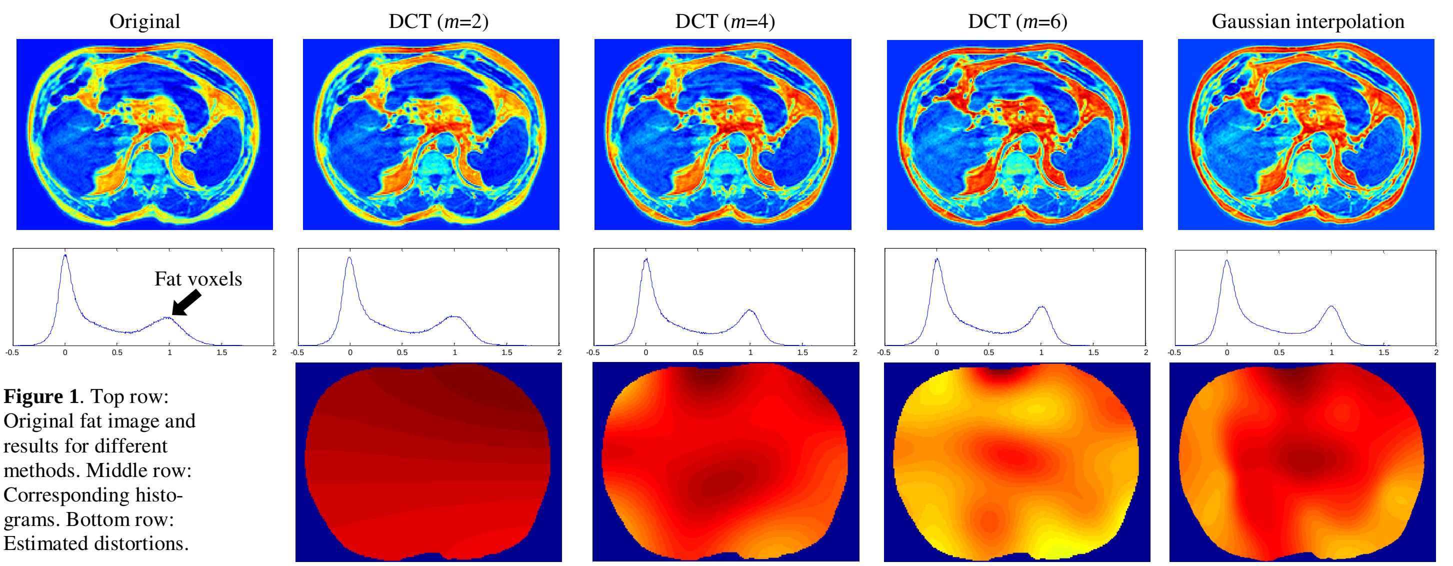

Figure 1. Top row: Original fat image and results for different methods. Middle row: Corresponding histograms. Bottom row: Estimated distortions.

CONCLUSIONS: We have presented a general approach for addressing field inhomogeneities in quantitative fat imaging. The method finds a correction field by fitting a set of smooth 3D spatial basis functions to voxels with high fat content in a least square error sense. By choosing the number of basis functions, the smoothness of the field is controlled in an intuitive way. In terms of fat peak width, we obtain at least as good results as with the reference method, but with a smoother correction field.

REFERENCES: [1] *Dixon WT. Simple proton spectroscopic imaging. Radiology 1984;153(1):189-194.*
[2] *Dahlqvist Leinhard O. Intensity inhomogeneity correction in two point Dixon imaging. ISMRM 2008*
[3] *Rydell J. Phase sensitive reconstruction for water/fat separation in MR imaging using inverse gradient. MICCAI 2007*
[4] *Rydell J. Three dimensional phase sensitive reconstruction for water/fat separation in MR imaging using inverse gradient. ISMRM 2008*