Intensity inhomogeneity correction in two point Dixon imaging

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INTRODUCTION: Intensity inhomogeneity prevalence in MR imaging is due to factors such as static field inhomogeneity, RF excitation field nonuniformity, inhomogeneity in reception coil sensitivity and patient movement. The effect of the nonuniformity is usually a slow varying non anatomic intensity variation over the image. Although it sometimes can be difficult to see the intensity nonuniformity by visual inspection there are implications that significantly can decrease segmentation and registration results as many medical imaging techniques is based on the assumptions that the same tissue has the same intensity throughout a volume. More importantly it affects the linear quantification of the MR signal. A voxel containing a certain amount of fat should have the same signal strength, independent of where it is located in space. This is not true in case of intensity inhomogeneity occurrence. In difference with other acquisition methods the two point Dixon method gives pure fat and water volumes after a correct phase sensitive reconstruction. Therefore there is no problem in distinguishing different tissues in the same volume. A simple but efficient way to correct for intensity inhomogeneity in fat volumes acquired with a two point Dixon technique [1] would be to locate voxels corresponding to pure adipose tissue and estimate a correction field from these points. A method has been developed addressed to this idea.

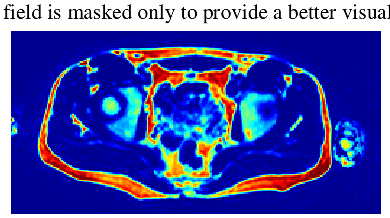
METHODS: The most common model in describing the intensity inhomogeneity effect in MR images is [2]:
$x = ax' + e$ where x' is the true signal, a denotes the inhomogeneity effects and e noise.

To find the slow varying multiplicative field the following steps are proposed:

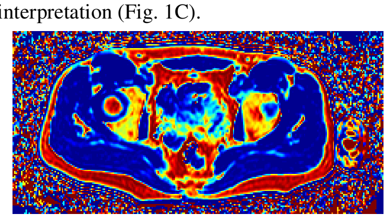
Identify pure adipose tissue voxels:
The intensity nonuniformity in the in-phase volume, *IP*, and extracted fat volume, *F*, are close to equal. By calculating the *F/IP* ratio an estimate of fat content relative to water content is given without the impact of intensity nonuniformity. As pure adipose tissue consists of about 90 percent fat and 10 percent water [4] a simple thresholding can be made to collect pure adipose tissue voxels. An additional masking is needed to only account for values in human tissue. This is made by an initial thresholding operation with an empirically determined threshold for background values followed by bimodal segmentation.
Interpolate a field from the identified voxels:
When the adipose tissue voxels have been identified the correction field is created with normalized convolution [3] where pure fat voxels is weighted as 1 and remaining voxels as 0. Convolving with a Gaussian smoothing kernel suppresses noise at the same time as interpolating a field from the collected points.
Reconstruate the volume:
The extracted fat volume is divided by the calculated intensity nonuniformity field to acquire an intensity normalized volume.

Data acquisition and water/fat separation: 50 volumes from 20 different patients were acquired. Single breath hold (28 s) multi-slice MR-images were collected out of and in phase (TE = 2.3 and 4.6 ms) using a 1.5 T MR-scanner (Philips Medical systems, Best, the Netherlands). Slice thickness was 5 mm and the field of view was 290x200x410 mm (ap, fh, rl). TR was 286 ms and the flip angle was 80 degrees. The signal intensities were rescaled to the sensitivity of the quadrature body coil using constant level appearance (CLEAR) reconstruction and the inverse gradient method [5] was used for water and fat separation.

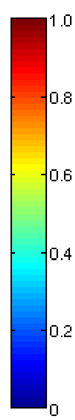
RESULTS: Since the intensity inhomogeneity can be difficult to observe with a normal grayscale mapping the resulting images are presented with color mapping. The original and corrected fat images are normalized with the fat peak value corresponding to a hundred percent adipose tissue. Peak values have been obtained by histogram analysis of voxels in the subcutaneous region and the corresponding subcutaneous voxel histograms before and after intensity normalization are presented below (Fig. 2A-B). The chosen slice (Fig. 1A) is close to the end of the acquired volume where the nonuniformity effect is larger. An evident improvement is shown in the corrected fat image (Fig. 1D). Similar improvement was seen inter slices and was most apparent close to the edges of the reception coil. The intensity nonuniformity field is masked only to provide a better visual interpretation (Fig. 1C).

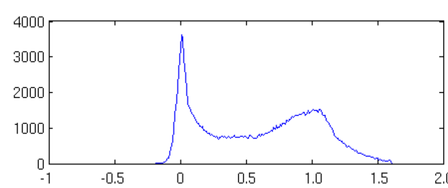


1A: Original fat image.

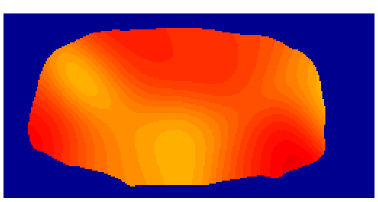


1B: F/IP ratio

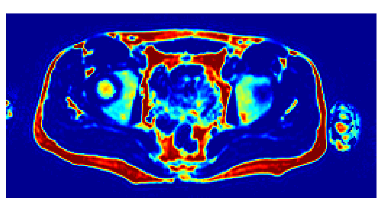




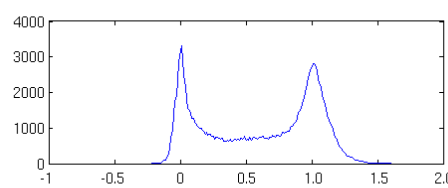
2A: Subcutaneous fat histogram before correction



1C: Intensity nonuniformity field



1D: Corrected fat image



2B: Subcutaneous fat histogram after correction

DISCUSSION: A simple and efficient solution to account for intensity inhomogeneity in two point Dixon imaging is to identify pure adipose tissue values and estimate a correction field by normalized convolution. The method has shown good stability in the evaluation of 50 datasets. After application of the correction field there are noticeable signs that imply that the intensity distribution over the volume has improved (Fig. 1D). Intensity variations seen inter and intra slices was efficiently eliminated. Looking at the subcutaneous fat histogram (Fig. 2B) for all voxels after intensity nonuniformity correction a distinctive peak for pure adipose tissue can be seen without dispersing effects caused by intensity variations.

REFERENCES: 1. Dixon, W.T. Radiology, 1984. p.189-194. **2.**Hou, Zujun. International Journal of Biomedical Imaging, 2006. p.1-11. **3.**Knutsson, H. SCIA'93. **4.**Querleux, B. Skin Research and Technology, 2001. p.118-124. **5.** Rydell, J. MICCAI 2007.