A Multi-scale Method for Automatic Correction of Intensity Nonuniformity in MRI Data

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Introduction
Surface coils, especially phased-arrays, are used extensively for acquiring magnetic resonance (MR) images with high spatial resolution. The signal intensities on images acquired from using these coils have a non-uniform map, due to the coil sensitivity profile. Although these smooth intensity variations have little impact on visual diagnosis, they become a critical issue when quantitative information is needed from the images.

A new method is presented here to estimate the near optimal coil's sensitivity profile from analysis of the multi-scale approximate templates and residues produced by the wavelet transform. It does not rely on any data other than the image generated by the MR scanner.

Method
We assume that the non-uniformity signal intensity of a noise-free case is given by

$$U(x, y) = M(x, y)B(x, y)$$  \hspace{1cm} (1)

where $M(x, y)$ is the true signal emitted by tissue and $B(x, y)$ is a sensitivity profile of a surface coil. In this new method, the $B(x, y)$ corresponds to approximate templates of the wavelet transform. Therefore, there are some sensitivity profiles within multi-scale analysis. The scale of optimal $B(x, y)$ is estimated according to the following minimal power spectrum \[1\] of coefficients of the wavelet transform:

$$k_0 = \min \{ p_k * q_k, k \in [M, N] \}$$  \hspace{1cm} (2)

where

$$p_k = \text{abs}(\text{mean}(w_k * w_k) - \text{mean}(w_k * w_{k-1}))$$

$$q_k = \text{abs}(\text{mean}(s_k - [s_{k-1}]_2) - \text{mean}(s_k - [s_{k-1}]_2)^2)$$

where $w_k$ is residues from high-pass filters, $s_k$ is approximate templates from low-pass filters, $[h]$ is an operator that resizes the data two times with linear interpolation and $[M, N]$ is limited decomposition in practice.

Based on the quadrature mirror filter theory \[2\], a complementary high-pass filter is obtained by shift and modulation of a low-pass filter:

$$G(z) = zH(-z^{-1})$$

A regularization filter \[3\] is used as the prototype filter for the wavelet transform, because its smoothing window can be tuned to any scale through a single parameter with no effect on execution and it has excellent filter properties.

$$H(z) = \frac{1}{1+\lambda(z^{-2} - 4z^{-4} + 6 - 4z + 2)}$$

The multi-dimension wavelet transform can be realized by cascading the low-pass and high-pass filter \[2\].

Experiment and Results
A series of phantoms and human images were used to validate the multi-scale method, which was written in IDL on a SGI workstation. MR images were acquired in a 1.5T SIGNA scanner (GE Medical System) with custom designed phased array coils. The regularization parameter $\lambda$ is fixed at 1000.

The processing time of this method per image is approximately 10s for resolution of 512 x 512. Table 1 lists the intensity variance and mean of the coiled 3D phantom, the corrected 3D phantom, and uniform phantom. Fig.1 shows the effect of correction on a neck image. The left is an original image and the right is the corrected images using multi-scale method. To better visualize the effect of the new method, the intensity of one row at same image location is extracted and superimposed on the images.

Conclusion
One parameter for filters in this method needs to be selected before implementation. It can be fixed because its size only affects the scale location of the optimal sensitivity profile. The method is automatically implemented.

Table 1 show mean and standard deviation of coiled phantom, corrected phantom and uniform phantom.

<table>
<thead>
<tr>
<th>Image No.</th>
<th>Coiled phantom</th>
<th>Multi-scale method</th>
<th>Uniform phantom</th>
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<td>Var.</td>
<td>Mean</td>
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<td>764.5</td>
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</tbody>
</table>

Fig.1 show effect of correction on neck image.

Reference:

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