Compensation of Abdominal Breathing Artifacts in Multi-Shot Acquisitions Using Parallel Imaging and Automatic Coil Weight Selection from Adjacent Slices

M. Honal[1], U. Ludwig[1], and M. Zaitsev[1]

[1]Dept. of Diagnostic Radiology, Medical Physics, University Hospital Freiburg, Freiburg, Germany

Introduction

Breathing motion during multi-shot acquisitions causes artifacts which appear as ghosts in the reconstructed images. Various techniques for motion correction using parallel imaging have been proposed [1-5]. They reconstruct images from subdivisions of motion corrupted k-space data and average them subsequently for artifact reduction [2], replace motion corrupted k-space lines [3] or correct for motion using parametric motion models [4, 5]. For multi-shot acquisitions of the abdomen these methods are of limited use: averaging may cause substantial blurring, often too much data is affected by motion to reject and replace it, and it is difficult to find a parametric model for breathing motion. Here parallel imaging and automatic coil weight selection are used to reconstruct images with reduced artifacts from undersampled k-spaces acquired during one shot of a multi-shot acquisition. Together with techniques for flexible image combination (e.g. image matching), that method can be used for breathing motion correction in abdominal imaging.

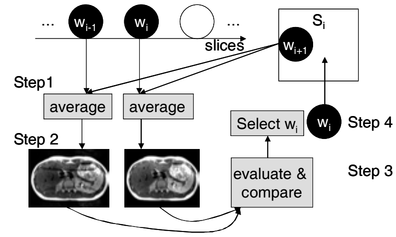

Fig. 1: Weight selection algorithm.

Methods

An abdomen of a normal volunteer was imaged during free breathing using a TSE sequence with continuously moving table on a 1.5T scanner (Magnetom Espree, Siemens Medical Solutions, Erlangen, Germany). Data were received from 12 channels of standard spine and body coil arrays. For each slice position k-space was acquired with two shots, one for the even, one for the odd k-space lines. The GRAPPA algorithm [6] with an undersampling factor of two was used to interpolate missing lines for the undersampled k-spaces acquired with each shot. Coil weights were estimated in three different ways. (A): For each slice position data from both shots were combined to a fully sampled k-space matrix and weights were computed on 32 reference lines extracted from the center of that matrix. The obtained weights were applied separately to odd and even k-space lines at the current slice position. (B): Weights were computed for each slice position as described in (A). Weights to be applied for reconstruction of missing lines for the shots at the current slice position were obtained by averaging the weights estimated for the current slice, its left and its right neighbor. (C): Weights were computed for each slice position as described in (A). To reconstruct missing lines for a shot at slice position i, a subset S_i of the weights from slice positions i-1, i and i+1 was determined iteratively, and the average of all weights in S_i was applied. During each iteration for the computation of S_i, every new weight candidate and the weights already contained in S_i were averaged (Fig.1, Step 1). Each averaged weight set was applied to reconstruct an image for the considered shot (Fig.1, Step 2). Thus, multiple images were obtained for each shot and evaluated using an image quality criterion (Fig.1, Step 3). The newly considered weights leading to the largest improvement of the criterion were finally included in S_i (Fig.1, Step 4). That procedure was continued until no more weights were left to be included or the image quality criterion could not be improved anymore.

Image quality was evaluated in terms of artifacts occurring due to inconsistent weights. For the used undersampling factor of two these are mostly N/2 ghosts. Their intensity was determined by evaluating the similarity between the lower and upper half of each reconstructed image using a correlation based measure. The simple correlation coefficient between lower and upper image halves is not a suitable similarity measure. Due to phase differences, the overlap between the ghost and the image might have a reduced intensity, such that ghosting would decrease the correlation coefficient instead of increasing it. Therefore lower and upper image halves were segmented into cells of 16x16 pixels (total image size was 320x320 pixels), such that either significant ghosting or no ghosting was present in each cell. Then the sum of the absolute values of the correlation coefficients for the pixels of each cell pair was used to measure ghosting. The segmentation is illustrated schematically in Fig. 2a for a non-ghosted image and in Fig. 2b for a ghosted image. Two representative cell pairs containing ghosts and causing structure in Fig. 2b are highlighted. The correlation coefficients for these cell pairs have higher absolute values compared to the values obtained for corresponding cell pairs in Fig. 2a, indicating a higher ghost intensity.

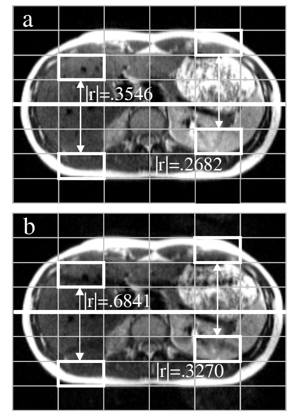

Fig 2: Segmentation of a non-ghosted image (a) and a ghosted image (b). Absolute values of the correlation coefficients are given for the highlighted cell pairs.

Results

Figure 3 shows images for one slice where the above described variants for obtaining coil weights were applied. The application of methods (A) (Fig. 3a) and (B) (Fig. 3b) leads to ghosting artifacts, only with automatic coil weight selection (method C) a ghosting free image is obtained (Fig. 3c).

Discussion

Parallel imaging and automatic coil weight selection have been applied to reduce breathing artifacts in multi-shot acquisitions of the abdomen. The proposed method does not require detailed a priori knowledge of the motion characteristics such as a parametric motion model. It can be applied, if motion is sufficiently slow such that all k-space lines acquired during one shot are consistent, which is the case for breathing motion in the abdomen. The rationale behind weight selection from adjacent slice positions is, that weight estimates are potentially less corrupted for some slice positions than for others, since the motion is not synchronized with the acquisition. Using weights from adjacent slices does not result in additional artifacts, because coil sensitivities vary relatively slowly in slice direction [7]. Therefore, automatic coil weight selection from a given slice range becomes applicable for the proposed breathing motion correction technique. A future challenge is to combine multiple artifact free images obtained for one slice position using matching techniques. That would result in a final image with improved signal to noise ratio. In contrast to simple averaging for artifact reduction, the application of matching would avoid blurring. The major limitation of the proposed technique is the requirement of at least one consistent set of weights in the given slice range (or of multiple weight sets which form a consistent set after averaging). The chance of finding appropriate weights can be increased however, when multiple averages are acquired per slice position or a larger range of slice positions is considered for weight selection.

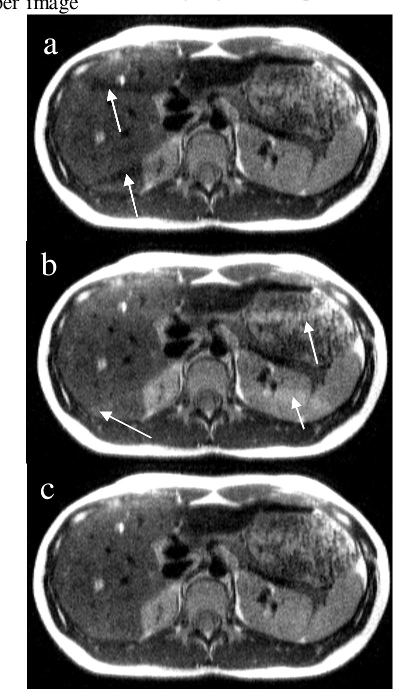

Fig. 3: Images reconstructed with GRAPPA from the data of one shot, using methods (A) (3a), (B) (3b) and (C) (3c) for coil weight computation. Arrows indicate the locations of ghosts.

References

[1] D. Atkinson: NMR Biomed, 2006. 19(3): 362-367. **[2]** H. Fautz, et al.: MRM., in press. **[3]** M. Bydder, et al.: MRM, 2002. 47(4): 677-686. **[4]** M. Bydder, et al.: MRM, 2003. 49(3): 493 - 500. **[5]** D. Atkinson, et al.: MRM, 2004. 52(4): 825-830. **[6]** M. A. Griswold, et al.: MRM, 2002. 47(6): 1202-1210. **[7]** M. Honal, et al.: Proc. German Chapter of the ISMRM, 2006. 7-8.