Projection-Based Estimation and Nonuniformity Correction of Sensitivity Profiles in Phased-Array Surface Coils

SungDae Yun, MS, 1 Walid E. Kyriakos, PhD, 2 Jun-Young Chung, MS, 1 Yeji Han, MS, 1 Seung-Schik Yoo, PhD, 2 and HyunWook Park, PhD 1*

Purpose: To develop a novel approach for calculating the accurate sensitivity profiles of phased-array coils, resulting in correction of nonuniform intensity in parallel MRI.

Materials and Methods: The proposed intensity-correction method estimates the accurate sensitivity profile of each channel of the phased-array coil. The sensitivity profile is estimated by fitting a nonlinear curve to every projection view through the imaged object. The nonlinear curve-fitting efficiently obtains the low-frequency sensitivity profile by eliminating the high-frequency image contents. Filtered back-projection (FBP) is then used to compute the estimates of the sensitivity profile of each channel. The method was applied to both phantom and brain images acquired from the phased-array coil.

Results: Intensity-corrected images from the proposed method had more uniform intensity than those obtained by the commonly used sum-of-squares (SOS) approach. With the use of the proposed correction method, the intensity variation was reduced to 6.1% from 13.1% of the SOS. When the proposed approach was applied to the computation of the sensitivity maps during sensitivity encoding (SENSE) reconstruction, it outperformed the SOS approach in terms of the reconstructed image uniformity.

Conclusion: The proposed method is more effective at correcting the intensity nonuniformity of phased-array surface-coil images than the conventional SOS method. In addition, the method was shown to be resilient to noise and was successfully applied for image reconstruction in parallel imaging.

Key Words: intensity correction; nonuniformity; parallel imaging; phased-array surface coils; sensitivity encoding (SENSE); sensitivity profile


A SURFACE COIL is a type of receiver coil that is placed directly on or over the region of interest (ROI) to increase magnetic sensitivity. It provides a very high signal-to-noise ratio (SNR) close to the region where it is placed, but suffers from decreased signal far from the coil. The phased-array surface coil is used to overcome the limited coverage of the surface coil by combining the images from two or more coils to produce a single image. While the phased-array technique improves the homogeneity of the images in the plane of the array compared to a single surface coil, the image intensity is still significantly bright near the coils compared to the far region (1). A birdcage head coil is conventionally used to achieve homogeneous spatial reception at loci distributed over the whole brain; however, surface coils and phased-array surface coils offer the potential for a fivefold increase in sensitivity at the expense of spatial homogeneity (2).

Since measured image intensities obtained from a single coil element of the phased-array coil can be described as the product of the magnetization of the object being imaged and the surface coil’s sensitivity profile (3), one can correct the intensity nonuniformity by dividing the surface-coil image by the estimated sensitivity profile of the coil.

Various intensity-correction algorithms have been developed to estimate sensitivity profiles (2–14). These algorithms can be classified into two categories (15): prospective methods (2–9) and retrospective methods (10–14). Prospective methods are based on prior knowledge. Several approaches of the prospective methods require prior information, such as coil size, shape, and position of the object, which is used to calculate the sensitivity profile prior to imaging (2,4–6). This type of approach is impractical for a flexible phased-array coil or a phased array in which the coil geometry may vary for each experiment. In other prospective approaches,
the sensitivity profile is estimated from a homogeneous phantom (7). Homogeneous phantoms have no complex anatomical structure, and yield images that represent very close estimates of the coils’ sensitivity profiles. These estimates, however, do not take into account differences in coil loading when the imaged object is changed, and thus render an inaccurate representation of the sensitivity profile of the coil for the object to be imaged. Similarly, images of the object from a body coil can be used to estimate the sensitivity profile of phased-array coil (3,8,9). However, all of these methods increase the imaging time for additional phantom/body coil imaging.

Retrospective methods are based on image-processing techniques. These methods can be classified as grayscale-based or transform-domain-based methods. Since the coil’s sensitivity profile is generally a slowly varying function of the space, while the anatomical structure contains higher spatial frequencies, a spatial-filtered image (10) or surface-fitted image (11) can be used as an estimate of the sensitivity profile. The performance of these approaches depends on a number of parameters, such as low-pass filter parameters and fitting basis functions. In addition, interesting methods have been introduced to estimate the sensitivity profile by using the maximum likelihood criterion (12) or by calculating the gradient across a pixel (14). Since gradient calculation tends to be affected by local variation, this type of approach suffers from noise effects.

Our goal in this study was to use the projection profiles of the phased-array coil images to estimate their sensitivity profiles (13). Because the projection is a signal integral along a line, the estimation using the projection becomes less sensitive to local variations, such as noise or anatomical structures. We believe this approach will work robustly for estimating the coil’s sensitivity profile even if the original data are noisy or have many complex anatomical structures. In addition, no prior knowledge of the coil’s parameter or additional scans, such as homogeneous phantom or body coil imaging, are required for this algorithm.

The intensity-corrected phased-array images obtained by the proposed method were applied to the computation of the sensitivity map, and compared with the previous sum-of-squares (SOS) approach to demonstrate the improvement in image quality. Robustness to noise and the number of projections was also investigated. The method was also applied to the partial k-space data and sensitivity encoding (SENSE) (16) reconstruction to demonstrate its feasibility for clinical applications.

**MATERIALS AND METHODS**

**Projection-Based Estimation: Theory**

The $i$-th channel image, $V_i(x,y)$, from a multichannel phased-array coil can be described as the product of the true anatomical image, $U(x,y)$, and the coil’s sensitivity profile, $S_i(x,y)$, which is the spatial modulation imposed on the image from the surface-coil. The resulting image intensity can be described as (ignoring the additive noise component):

$$V_i(x,y) = S_i(x,y)U(x,y), (1 \leq i \leq N).$$

where $N$ is the number of phased-array channels. Our goal is to obtain $\hat{S}_i(x,y)$, an estimate of $S_i(x,y)$. The intensity-corrected image of the $i$-th channel, $\hat{U}_i(x,y)$, which represents an approximation of the true anatomical image, $U(x,y)$, is then expressed in terms of the ratio of the measured signal and the estimated coil’s sensitivity profile as follows:

$$\hat{U}_i(x,y) = \frac{V_i(x,y)}{\hat{S}_i(x,y)}, (1 \leq i \leq N).$$

Repetition of this procedure for all channels produces the same number of intensity-corrected images as the number of phased-array channels. The results are averaged to obtain the overall intensity-corrected image, which is given by

$$\hat{U}(x,y) = \frac{1}{N} \sum_{i=1}^{N} \hat{U}_i(x,y).$$

The projection profile (Proj[$V$]) of the $i$-th channel image ($V_i$) can be described as a smoothly varying envelope in addition to small structural details in the image. We assume that the smoothly varying envelope is attributed to the coil’s sensitivity profile ($S$), and the high-frequency details are mainly attributed to the true anatomical structure ($U$) of the object. In order to find the coil’s sensitivity profile, the structural details must be removed from the projection profile. To achieve this, we used a nonlinear curve-fitting algorithm, the Levenberg-Marquardt algorithm (20), which was found to be effective for preserving the shape of the projection profiles. The filtered back projection (FBP) algorithm (17) was applied to the curve-fitted projection profiles to reconstruct the estimated coil’s sensitivity profiles ($\hat{S}_i$).

**Projection-Based Estimation: Illustration**

The detailed procedure of the proposed method for an eight-channel phased-array coil is described below (Fig. 1):

1. A full k-space image is acquired using the phased-array coil, and for each channel image (Fig. 1a), projection profiles (also called sinograms) ranging from $0^\circ$ to $179^\circ$ in increments of $1^\circ$ are calculated using the Radon transform (Fig. 1e).
2. The object region ($R_o$) is then separated from the background region ($R_b$) in the projection profile by setting a threshold, which is 20% of the mean value of the SOS image. Three equal-space regions ($R_1$, $R_2$, and $R_3$) are defined as shown in Fig. 1f. Each projection profile is divided into two parts based on its peak point (indicated by the black dots in Fig. 1f). The nonlinear fitting algorithm, which is based on a Gaussian function or a combination of Gaussian and fourth-order polynomial functions, is then applied to left and right parts of each projection profile. The functions of the fitting model are chosen according to the location of the
peak point. If the peak point is in region R2, the projection profile is relatively symmetric about the peak point. Therefore, the Gaussian fit is used for both parts. Otherwise, the Gaussian and fourth-order polynomial fits are used for the longer and shorter parts, respectively. The Gaussian and fourth-order polynomial functions were experimentally chosen as the optimum fitting functions that showed the minimum fitting error among various functions.

3. The sensitivity profile of each channel image is then obtained by applying the FBP to the curve-fitted projection profiles (Fig. 1b).

4. The channel image (Fig. 1a) is divided by the obtained sensitivity profile (Fig. 1b) to produce a uniform-intensity image.

5. Steps 1–4 are repeated for the other channel images of the phased-array coil (Fig. 2). This yields the same number of intensity-corrected images as the number of phased-array channels, which are averaged to obtain the overall intensity-corrected image (Fig. 1d).

**Image Acquisition**

The proposed technique was applied to both phantom and human brain imaging. Phantom images were acquired from a 1.5T system with an eight-channel phased-array coil using a spin-echo sequence (matrix = 129 × 129, FOV = 250 mm × 250 mm, TR = 600 msec, TE = 9 msec, slice thickness = 5 mm, 30 slices).

We compared the intensity uniformities of the SOS and intensity-corrected images. To compare the proposed method with the previous ones, both the proposed method and the previous methods were applied to the data set to obtain the intensity-corrected images. The previous methods used in this study were the transform-domain-based retrospective method proposed by Cohen et al (10), and the grayscale-based retrospective method described by Vokurka et al (14).

**Examination of the Optimum Number of Views in Projection**

The computation time of the proposed algorithm is spent mainly on the curve-fitting process, and is proportional to the number of projections. Decreasing the number of projection views generally results in a deterioration of the back-projected image details. However, since the sensitivity profiles are smoothly varying functions, it is possible to reduce computational time by decreasing the number of projection views without severe degradation of the sensitivity profiles.

Performance degradation was studied as the number of views decreased from 180 to two. The measure is the absolute difference between the intensity-corrected image from 180 projection views and that from the smaller number of views, as described by

$$y(p) = \frac{1}{N_x} \sum_{x=1}^{N_x} \frac{1}{N_y} \sum_{y=1}^{N_y} \sum_{s=1}^{N_s} \sum_{k=1}^{N_k} |(I_{180}(x,y))_s - (I_p(x,y))_s|.$$  (4)
**Figure 2.** Original eight-channel phased-array images and their corresponding estimated sensitivity profiles from the proposed method.

**Figure 3.**

a: SOS of original phased-array images (phantom with eight-channel coils).

b: The proposed intensity-corrected result of image a.

c: SOS of original phased-array images (human brain with four-channel coils).

d: The proposed intensity-corrected result of image c.

e: Vertical cut views of a3 and b3.

f: Vertical cut views of c3 and d3.

g: Horizontal cut views of a3 and b3.

h: Horizontal cut views of c3 and d3. The numbers 1, 2, and 3 denote different slices in the figure indices.
where $N_s$ is the number of slice, $N_x$ is the horizontal matrix size, $N_y$ is the vertical matrix size, and $(I_{ps})_s$ is the intensity-corrected magnitude image when the slice number is $s$ and the number of views is $p$.

**SENSE Reconstruction**

Sensitivity-based imaging, such as the SENSE technique, requires highly accurate sensitivity maps (16). Intensity-correction methods may be a way to achieve this goal. The first step in generating sensitivity maps is to acquire and reconstruct full-FOV images for each channel. Division of each of these images by the SOS yields sensitivity maps that are modulated by the intensity nonuniformities of the SOS. To reduce the nonuniform intensity effects, we compute sensitivity maps using intensity-corrected images from the proposed method. SENSE imaging with a reduction factor of 2 was performed with the use of the computed sensitivity maps. For comparison, we repeated the same process using SOS instead of intensity-corrected images to compute the sensitivity maps.

**Robustness to Noise**

We analyzed the performance of the proposed algorithm under a condition of noise to simulate practical situations. We applied the proposed intensity-correction method to the phantom images after adding various levels of white Gaussian noise. The noise standard deviation (SD) gradually increased in steps of 2%, from the initial noise-free image to 10% of the maximum intensity value in the phantom image. We used the absolute differences between the sensitivity estimate obtained from the noise-free image and that of the noisy images as the performance measure, which is formulated as:

$$d(\sigma) = \frac{1}{N_y} \sum_{y=1}^{N_y} \sum_{x=1}^{N_x} |S_{\text{noise free}}(x,y) - S(x,y)|,$$

where $N_x \times N_y$ is the image dimension, $S$ is the sensitivity estimate (i.e., the magnitude average of the sensitivity images) when the noise SD is $\sigma$, and $\sigma$ is 0%, 2%, 4%, 6%, 8%, and 10% of the maximum intensity value in the phantom image.

**Estimation From Partial k-Space Data**

The proposed method was also applied to partial k-space data at low frequency to show the applicability of this work to dynamic sensitivity estimation for cardiac imaging and abdominal imaging. From the full k-space lines of the phantom, we extracted only 16 lines in the center k-space and performed a Fourier transform to obtain low-resolution images. We then compared the intensity uniformity between the SOS and the proposed intensity-corrected images from the 16 k-space lines.

**RESULTS**

**Intensity Correction of the Phased-Array Images**

The SOS of the original phased-array images and the corresponding intensity-corrected images are displayed in Fig. 3. To assess the correction method, we plotted the intensity of a vertical line in the bottom-most images of Fig. 3a–d. Figure 3e and f show the cut views along the vertical lines, and Fig. 3g and h show the cut views along the horizontal lines where the solid line corresponds to the intensity-corrected image, and the dotted line corresponds to the SOS image. The result shows that the intensity profile from the proposed method appears more uniform than that from the SOS. For quantitative assessment (10), we calculated the signal deviation, before and after correction, from the mean signal intensity in the region of the image. This

![Figure 4](image-url)
calculation is applied only to the uniform phantom image (as shown in Fig. 3a2 and b2), so the measured value should not be affected by the anatomical structures. The signal deviation was 13.1% and 6.1% for the SOS and the proposed intensity-corrected images, respectively.

**Intensity-Corrected Images With a Reduced Number of Views in Projection**

As shown in Fig. 4, the absolute difference increases as the number of views decreases. However, image distortion does not increase significantly until the number of views decreases.
views decreases below 30. This suggests that it is possible to reduce the processing time significantly. We implemented the intensity-correction algorithm with Matlab on a 2.67-GHz PC. The computation time for one channel image, when the number of views is 180, was eight seconds. The processing time decreased to 1.4 seconds when 30 views were used.

The proposed algorithm was also tested using 30 projection views and compared with the results obtained using 180 projection views, as shown in Fig. 5. The difference between cut views from 180 projections and 30 projections is plotted in Fig. 5e for the phantom and Fig. 5f for the human brain image. The maximum value of the difference between the cut views does not exceed a maximum of 1.4% in the phantom images and 1.0% in the human brain images.

**Results of SENSE Reconstruction**

Figure 6 shows the SENSE reconstruction results of the phantom and brain images. Figure 6a and c show the SENSE reconstructions using the sensitivity maps computed with the SOS approach, and Fig. 6b and d show the SENSE reconstructions using the sensitivity maps computed with our intensity-correction approach. We plot the cut views along a vertical line of Fig. 6a–d. Figure 6e and f show the cut views (the solid line corresponds to the SENSE reconstruction from our intensity-correction approach, and the dotted line corresponds to the SENSE reconstruction from the SOS approach). The experimental results show that the intensity profile from the proposed method appears more uniform compared to that from the SOS.

**Robustness to Noise**

Figure 7a shows the noise-contaminated images with various SDs. The estimated sensitivity profiles obtained...
by our method are shown in Fig. 7b, below each corresponding noise-contaminated image. Figure 7c shows the absolute difference between the sensitivity estimates from the noise-free image and the noisy images. In Fig. 7c, the difference value also increases as noise increases, but the maximum value of the difference does not exceed 3% even when the noise SD is 10% of the maximum intensity in the image. From this result, we can see that the distortion of the sensitivity estimate does not increase significantly until the noise level increases to 10% of the maximum intensity value in the phantom image.

Result of Estimation From Partial k-Space Data
The SOS results of the low-resolution images from the 16-center k-space lines and the corresponding intensity-corrected images are displayed in Fig. 8. We plot the cut views of the vertical line in the rightmost images of Fig. 8a and b. Figure 8c shows the cut views (the solid line corresponds to the intensity-corrected image, and the dotted line corresponds to the SOS image). It shows that the sensitivity profile from the proposed method appears more uniform than that from the SOS.

Performance Comparison of the Proposed Method With Previous Methods
The SOS of the original phased-array images and the corresponding intensity-corrected images are displayed in Figs. 9 (phantom data) and 10 (human brain data). Figure 9a shows the SOS images and Fig. 9b–d show the intensity-corrected images from the proposed method, Cohen et al’s (10) method, and Vokurka et al’s (14) method, respectively. Figure 9e shows vertical profiles of Fig. 9a–d. The same analysis for the human brain data is shown in Fig. 10. The results show that the intensity profile from the proposed method has the best uniformity of all of the profiles from the other methods.

DISCUSSION
Since the phased-array coil was first developed by Roemer et al (2) in 1990, it has gained enormous popular-
ity. The phased-array coil offers high SNR and is well suited for fast parallel imaging techniques (e.g., simultaneous acquisition of spatial harmonics (SMASH) (19) and SENSE (16)). However, although the phased-array coil is an attractive option, its sensitivity profile is a major problem. The phased-array technique has been further developed to improve the homogeneity of the images; however, the image intensity near the coils is generally higher than that in the deeper areas of the imaged object.

In this work we propose a novel approach for correcting intensity nonuniformity in phased-array coil images. We verified the performance of the proposed method by applying it to several phased-array coil images using a 1.5T scanner with eight-channels and a 3.0T scanner with four channels. In both cases the intensity uniformity was substantially improved over that achieved with the widely used SOS method. The proposed intensity-correction scheme can also be used to calculate sensitivity maps for parallel imaging techniques such as SMASH and SENSE. Currently, these techniques divide the image obtained from each phased-array channel by the SOS or body coil image to obtain the sensitivity maps. This type of approach holds true when the SOS and the body coil have a homogeneous intensity distribution. However, this assumption is not generally true for the SOS image, and could introduce errors in sensitivity maps. Although a body coil image is more homogeneous than an SOS image, current system configurations do not allow the simultaneous acquisition of body and phased-array images. This requires a longer imaging time due to the additional scan, and makes the method less robust because of possible interscan object motion. In addition, the homogeneity of a body coil image may decrease as the field strength increases.

We estimated the sensitivity profile directly from the phased-array coil images. Consequently, neither prior knowledge of coil’s parameter (such as size, shape, and position) nor additional scans (such as homogeneous phantom or body coil images) are required for this method. In contrast to spatial-filter-based methods, the
The proposed method does not have to determine the cutoff frequency. It only has to find the smoothing curve that has the minimum difference with each projection profile of the phased-array coil images. This calculation is fully automated and can be easily performed using the nonlinear curve fitting method.

As shown in the experimental results, the homogeneity of the intensity-corrected images using the proposed approach was robust to noise owing to the use of projection profiles. Although the method requires intensive computation for projection and nonlinear fitting for each projection profile, we believe the processing time of less than 10 seconds is acceptable for clinical applications. However, since sensitivity profiles are smoothly varying functions, it is possible to reduce the number of projection views without severely degrading the estimates, which effectively saves computation time. In addition, since our method is completely parallel, one can further reduce the processing time by using a larger number of processors that allow for real-time applications.

When the sensitivity estimates were obtained from only low-frequency k-space data, our results were in line with previous results obtained using full k-space data. This suggests that the proposed method can be extended to dynamic sensitivity estimation for cardiac imaging (18) and abdominal imaging.

In conclusion, the images obtained by the proposed method were shown to have more uniform intensity than those obtained by the SOS approach. The sensitivity maps from our approach were applied to the SENSE reconstruction, and it shows better uniformity in the reconstructed images than the conventional SENSE reconstruction using the SOS approach. In this work we used only SENSE reconstruction to investigate the proposed intensity-corrected image, but we believe it can be effectively employed for other parallel imaging techniques, such as SMASH, in calculating sensitivity maps to improve the intensity uniformity of the reconstructed images.

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