SNR Measurement on Single Images of Phantom Using Wavelet Transforms

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Target Audience: This work is relevant to investigators interested in examining alternate methods of MR image quality assurance.

Purpose: SNR measurement on phantoms is very important in quality assurance for clinical MRI scanners [1]. The recommended approach is obtaining a noise image by take the difference of two data sets [2]. However, small drifts in the position of the phantom or coil during measurement and signal variation caused by flow inside a large solution phantom can result in noise overestimation. Efforts have been made to obtain the noise level from one image only. Wavelet transforms have been widely used for image compression and more recently in compressed sensing [3]. The hope in applying this transform is that undesirable background signal variations will be compressed into relatively few elements in the data array while Gaussian noise remains evenly distributed with the same standard deviation. Here, we describe a wavelet-based SNR quantification method which extracts the local noise level from one image and validate it using the difference noise image.

Methods: We acquired MR images on a 1.5T Philips Achieva (Best, The Netherlands) whole body clinical scanner on 3 RF coils: (i) C1 coil which is circular loop surface coil with diameter of 20 cm; (ii) 8 channel SENSE head coil and (iii) 8 channel SENSE wrist coil. The MR pulse sequence parameters and the phantom used for each coil are listed in Table 1. For each coil, the images were acquired with acquisition and reconstruction matrix size of 256×256, and 17 repeated data sets were obtained. Data analyses were carried out with internally developed software written in IDL (Exelis, Boulder, CO). SNR was quantified from multiple ROI's as shown in Figure 1. S_0 denotes the mean signal level of an ROI. The noise level of the ROI was extracted from the image intensity data. First, the signal intensity within the ROI was fitted using a polynomial of x and y coordinates containing a constant and terms up to the 5th order. The result of the polynomial fit was subtracted from the image intensity removing any slow signal variation from the data. What remains in the data are Gaussian noise [4] and possibly fast varying signal intensity such as Gibbs ringing, SENSE unfolding artifacts, etc. Next, we used a 2D Haar wavelet transform to compress the artifactual features. Finally, the histogram of wavelet transformed data was analyzed. We hypothesize that most elements in the data array contain noise, and artifact features are outliers of the Gaussian distribution. The FWHM of the central peak in the histogram was determined, and from this we obtained an initial estimation of the s.d. of noise as σ_0= FWHM/2.35. The portion of the histogram within $2\sigma_0$ from zero is fit to a Gaussian centered at zero to yield the final value of σ. The SNR of the ROI is given by S_0/σ. To validate the wavelet noise extraction procedure described above, the difference image method was used as the reference, where the noise image was obtained by subtract the average of 16 images from one single image. Figure 1 and Table 2 represent 3 data sets for each coil.

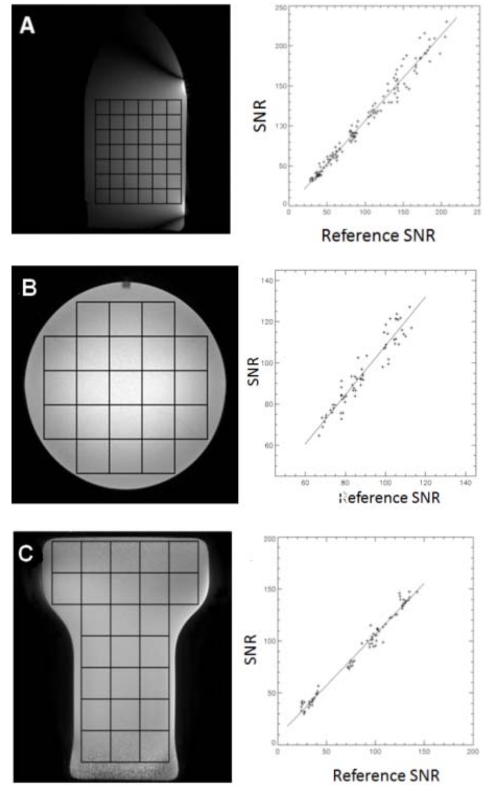

Figure1: Imaging and regression results: (A) C1 coil, (B) SENSE head coil, (C) SENSE

RF coil	Phantom	Acquisition parameters	SNR quantification
C1	5 liter $CuSO_4$ solution	Sagittal SE, fov = 350 mm, Slice thickness = 2 mm, TR/TE/flip = 500/20/90°	ROI pixels = 16×16 Number of ROI = 42 5th order polynomial fit
SENSE head 8 channel	BIRN gel	Axial SE, fov = 220 mm, Slice thickness = 5 mm, TR/TE/flip=500/20/90°	ROI pixels = 32×32 Number of ROI = 21 2nd order polynomial fit
SENSE wrist 8 channel	Wrist $CuSO_4$ solution	Corronal SE, TSE factor = 2, fov = 154 mm, Slice thickness = 1 mm TR/TE/flip = 494/22/90°	ROI pixels = 32×32 Number of ROI = 25 2nd order polynomial fit

Table1: Summary of parameters and phantoms

Results: The ROIs of SNR evaluation for each coil are shown in Figures 1. Table 2 lists the relationship of results between the wavelet method and the reference method. A positive difference in the Bland-Altman analysis means $SNR>SNR_{ref}$.

Discussion: There is a substantial variability of the noise level across the uniform solution phantom when SENSE coils are used. Therefore, we use wavelet transforms in local areas supported by small array sizes. However, as the array size decreases, the statistical fluctuation of the measured noise level increases. For this reason, a larger array size is preferred if possible. This method requires that the noise on each voxel is independent. If the noise is correlated (e.g., in the case where the reconstruction matrix is larger than the acquisition matrix), there will be sub-bands with different noise levels after the wavelet transform. This causes disagreement between the noise measured by our method and noise directly calculated from the standard deviation of the difference image. The images can be reformatted to decrease the reconstruction matrix size and remove the spatial correlation of noise. Parallel imaging reconstruction commonly introduces correlation in noise, and details about the noise correlation are unknown to the scanner user in most cases. This could be the reason that the SNR levels from the single image method are slightly higher than the reference method in Table 2. Validation of the wavelet based SNR measurement by the conventional difference image method is needed in these situations.

RF coil	Reference SNR range	Linear correlation	Bland-Altman analysis
C1	28 – 207	$SNR = -0.6+1.07\times SNR_{ref}$ r = 0.987	mean difference = 6.0% s.d. of difference = 7.8%
SENSE head 8 channel	67 - 113	$SNR = -10.8+1.19\times SNR_{ref}$ r = 0.951	mean difference = 6.4% s.d. of difference = 5.5%
SENSE wrist 8 channel	24 - 143	$SNR = 8.3 + 0.98\times SNR_{ref}$ r = 0.991	mean difference = 11.4% s.d. of difference = 12.0%

Table2: Relationship summary of wavelet and reference methods

Conclusions: An SNR quantification approach based on wavelet transform is described for local areas on MRI of phantoms without using a difference image for noise.

References: [1] Och JG. *Med Phys* 19, 217-229 (1992).
[2] Reeder SB, *Magn Reson Med* 54, 748-754 (2005). [3] Lustig M, *Magn Reson Med* 58, 1182-1195 (2007).