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Massively parallel MRI detector arrays

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ABSTRACT

Originally proposed as a method to increase sensitivity by extending the locally high-sensitivity of small surface coil elements to larger areas via reception, the term parallel imaging now includes the use of array coils to perform image encoding. This methodology has impacted clinical imaging to the point where many examinations are performed with an array comprising multiple smaller surface coil elements as the detector of the MR signal. This article reviews the theoretical and experimental basis for the trend towards higher channel counts relying on insights gained from modeling and experimental studies as well as the theoretical analysis of the so-called "ultimate" SNR and g-factor. We also review the methods for optimally combining array data and changes in RF methodology needed to construct massively parallel MRI detector arrays and show some examples of state-of-the-art for highly accelerated imaging with the resulting highly parallel arrays.

1. Introduction

The past decade of Magnetic Resonance Imaging (MRI) has seen tremendous advances toward improved sensitivity and reduced scan-time in clinical and research imaging examinations. While gradient performance improvements in strength and slew-rate always played an important role in terms of acquisition speed, magnet field strength and well-crafted detector geometries have always been critical for maximizing the sensitivity of the MR experiment. Because of this important role of a coil detector in determining sensitivity, we focus on receive arrays; although we note that transmit arrays have recently become an object of study in their own right [1–7].

Roemer and colleagues [8] showed that simultaneous detection of the MR signal with an array of surface coils could out-perform a single, larger surface coil covering the same area and gave a detailed description of how to optimally combine the data from the multiple elements given their sensitivity profile and measured noise correlation information. They further introduced the concept of "preamplifier decoupling" to suppress inductive coupling within the array using low-input impedance preamplifiers [8]. Only a year later, this concept was extended to volume-like acquisitions to show that the sensitivity of the array could compete favorably with volume coils [9]. While the spine was Roemer's initial focus and a logical place to distribute the reception coils along the length of the body, the concept was quickly extended to imaging other parts of the body [9–13] and to spectroscopy applications [14,15]. Interestingly, the brain, with its nearly spherical geometry was a less obvious target for arrays, and was among the last applications [16–18].

These demonstrations alone were enough to make array coils the favored method for many types of MRI examinations. But a second advantage of detection with arrays was soon introduced; namely the ability to reconstruct under-sampled *k*-space data and thus significantly speed up the MR image encoding process. These parallel imaging reconstruction techniques effectively used the additional spatial information contained in the signal intensity and phase profiles of the array elements to allow reconstruction of the image from under-sampled *k*-space data sets. In these methods, the image encoding is shared by the array coil and the gradient encoding steps. It is interesting to note that most imaging modalities rely exclusively on detector arrays for image encoding. For example CCD cameras, EEG and MEG, rely on a dense array of detectors, as does the human eye (where the retina is viewed as an array of photo-detectors), and perhaps most intriguing, the insect eye, which consists entirely of a directionally sensitive array of light-pipes with a few photo-detector cells inside (ommatidia). Medical imaging modalities such as MRI are the outliers where the role of detection and encoding can be completely separated. However, early MR researchers began to realize that the multiple elements could play a role in image encoding [19–23]. While this early work provided glimpses of what could be done, it was only after Roemer's work propelled the wide-spread implementation of parallel reception that the time was right for incorporating the

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spatial information present in the array to augment gradient encoding. The fact that the two types of encoding could be merged elegantly, as shown by the introduction of SMASH [24] and SENSE [25] in the late 1990s, opened the flood-gate to the variety of robust methods that exist today.

As a result of the sensitivity increase and encoding acceleration, virtually every clinical coil sold for MRI scanners today is an array. The trend toward higher and higher element counts has also increased over the years; from the initial systems with 4–8 elements, to the more commonly used 16–32 elements found on most new scanners, to exploration of what can be done with 64–128 channels on clinical systems.

In this review, we examine the theoretical basis for the trend toward higher channels from so-called "ultimate SNR" and "ultimate g-factor" analysis, as well as modeling regular geometries. We compare these identified gains to what has been achieved with 32, 64, 96, and 128-channel arrays. We also look at how these trends have forced us to re-think how we tune, match and decouple coils for MR detection. Finally, we look at the trends in MR acquisition sequences which now take advantage of the distribution of coil sensitivity patterns in all three directions.

2. Optimum combination of individual coil element data

In his original paper, Roemer described a way to elegantly combine the data from the multiple receivers of the array. It is almost always advantageous to combine data obtained from multi-channel array reception in the spatial domain, since the sensitivity profiles of each individual array element can be a steep function of the pixel's coordinates. Then the optimization can be done on a pixel-by-pixel basis taking into account the spatially changing amplitudes and relative phases of the sensitivity profiles. In addition to measuring the coil signal map $S_i(x,y,z)$ for each element, i, which tells us how the signal vectors will add, we need to know the noise covariance matrix, $\boldsymbol{\Psi}$, which describes the thermal noise variance in each channel and the covariance between pairs of channels, which informs us of how the noise from the channels adds. For a given pixel, we can form a vector of the coil sensitivity, **C**, and measured signal level, **S**, for each channel. Here **C** and **S** exist for each pixel and are vectors of length N_{ch}. We then generate the image intensity, I, of the combined channels for that pixel from a normalized weighted sum of the measured signal levels. The weights will be chosen to maximize the SNR of the combined pixel. We also express the complex weights as a vector, **w**, of length N_{ch}. Then the general expression is:

$$I = \lambda \mathbf{w}^{\mathbf{H}}\mathbf{S}, \tag{1}$$

where λ is a normalization constant which might vary as a function of location but does not effect the pixels SNR. Roemer showed that if the noise variances are equal and uncorrelated ($\boldsymbol{\psi}$ proportional to the identity matrix), then **w** = **C** and

$$I = \lambda \mathbf{C}^{\mathbf{H}}\mathbf{S}. \tag{2}$$

To create an image with spatially uniform noise levels, λ is chosen as:

$$\lambda = (\mathbf{C}^{\mathbf{H}}\mathbf{C})^{-1/2}. \tag{3}$$

In coil arrays, noise correlations always occurs between channels (either the channel's variances are unequal, or shared noise or coupling exists in the array). Taking those coupling effects into account, the image combination is fully optimized when $\mathbf{w} = \boldsymbol{\Psi}^{-1}\mathbf{C}$. This results in

$$I^{\mathrm{optSNR}} = \lambda \mathbf{C}^{\mathbf{H}}\boldsymbol{\Psi}^{-1}\mathbf{S}, \tag{4}$$

and λ becomes as

$$\lambda = (\mathbf{C}^{\mathbf{H}}\boldsymbol{\Psi}^{-1}\mathbf{C})^{-1/2}. \tag{5}$$

This can be thought of as "pre-whitening" the signal vector **S** prior to the combination (replacing **S** with $\boldsymbol{\Psi}^{-1}\mathbf{S}$). Roemer also showed that if the SNR is high in each channel, then the coil sensitivity vector **C** is well approximated by the signal vector **S** and does not need to be measured, i.e. the coil sensitivity is essentially just a map of the signal with perfect SNR. If the noise covariance is also proportional to the identity matrix and we use the λ for uniform noise, then we get what Roemer called the "root sum-of-squares" method;

$$I^{\mathrm{rSoS}} = \sqrt{\mathbf{S}^{\mathbf{H}}\mathbf{S}}. \tag{6}$$

This is a particularly useful form because no pre-scan measurements are required of **C** or $\boldsymbol{\Psi}$. One simply takes the sum of the square of the signal levels of each channel's measurement of that pixel. Since the noise covariance matrix requires only a second to acquire (by digitizing noise in the absence of excitation), it is useful to add this information in, but preserve the estimation of **C** by **S**. Then we have the "covariance weighted root-sum-of-squares" combination: $I^{\mathrm{cov\text{-}rSoS}} = \lambda\, \mathbf{S}^{\mathrm{H}}\, \boldsymbol{\Psi}^{-1}\, \mathbf{S}$ with $\lambda = \sqrt{(\mathbf{S}^{\mathbf{H}}\boldsymbol{\Psi}^{-1}\mathbf{S})}$, which simplifies to

$$I^{\mathrm{cov\text{-}rSoS}} = \sqrt{\mathbf{S}^{\mathbf{H}}\boldsymbol{\Psi}^{-1}\mathbf{S}}. \tag{7}$$

Given the choice of weights, w_{i}, the image SNR is given by [8]:

$$\mathrm{SNR} = \frac{\mathbf{w}^{\mathbf{H}}\mathbf{S}}{\sqrt{\mathbf{w}^{\mathbf{H}}\boldsymbol{\Psi}\mathbf{w}}}. \tag{8}$$

For the other combination methods, the resulting SNR can be shown to be:

$$\mathrm{SNR}^{\mathrm{rSoS}} = \frac{\mathbf{S}^{\mathbf{H}}\mathbf{S}}{\sqrt{\mathbf{S}^{\mathbf{H}}\boldsymbol{\Psi}\mathbf{S}}}, \tag{9}$$

while the image SNR for the noise cov-SoS is

$$\mathrm{SNR}^{\mathrm{cov\text{-}rSoS}} = \sqrt{\mathbf{S}^{\mathbf{H}}\boldsymbol{\Psi}^{-1}\mathbf{S}}. \tag{10}$$

Note that the SNR for the covariance weighted sum-of-squares image is (remarkably) the same as the image itself.

In the combination methods above, the complex-valued weights are chosen to maximize the image SNR using the coil sensitivity profiles and noise covariance matrix (or estimates thereof). The problem with the simple noise ROI in the "black" background area of an image is that usually there is no estimate of the coil sensitivities in this region. This results in sub-optimum combination of the array elements in this region and an amplification of the noise by an unknown factor. For example, if the rSoS method is used, the channels are essentially weighted by noise and combined. This is of little concern for discrimination of anatomy, since the weights used inside the body are accurate, but it eliminates this easy method of ROI outside the head to measure noise. Note the ROI in the noise-only part of the image is perfectly valid for single channel coils (after correction for the Rician distribution in magnitude data [26] or if the coil sensitivity is known in that region (for example through a theoretical calculation).

Assessing the SNR of an image acquired with an array coil in a manner that can be readily compared to either another group's measurements or to the SNR obtained by analyzing the signal mean and variance of a time-series of images requires further considerations. Kellman and McVeigh [27] elegantly described the series of correction factors needed to produce SNR in "absolute units", which are exactly what is needed to make the calculated SNR maps agree with those obtained from a time-series measurement and subsequently to make SNR maps directly comparable. When these correction factors are used, the SNR measurement some important