Parallel Imaging

Mark A. Griswold, Ph.D.
Department of Radiology, Case Center for Imaging Research, Case Western Reserve University School of Medicine, 11100 Euclid Ave, Cleveland, OH, 44106, USA
email: mark.griswold@uhhospitals.org, mag46@case.edu, Tel. +1 (216) 844-8085

Introduction

MRI has traditionally been a speed-limited imaging modality. Starting with the very slow early clinical scanners in the 1980s, the field has seen a steady increase in imaging speed. Most of this increase in speed has been due to the advent of faster and stronger gradients. However, by the late 1990s it became clear that gradient systems had essentially been fully developed with no further increases in speed possible. On the other hand, the last few years has seen the rapid development of methods based on imaging with an array of RF coils. These parallel imaging methods have now been developed for nearly every MR application, as demonstrated by the high quality clinical images which will be shown throughout the week. The goal of this lecture will be to provide the clinician with a basic understanding of these methods, including how they were developed, and how they can be used clinically to provide better imaging quality in a reduced time.

A Simplified Example

While most modern parallel imaging methods require somewhat complex mathematical skills, most of the basic concepts and requirements can be understood from a very simple example, which is a simplification of the basic Parallel Imaging with Localized Sensitivities (PILS) method [1].

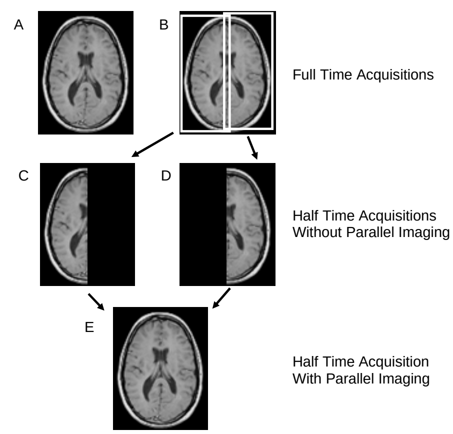

Figure 1: A simplified version of parallel imaging with two completely localized coils.

Let us assume we are interested in imaging an axial cross-section of a head with the phase-encoding direction in the left-right direction. If we use a standard head coil, we would have to image with a large enough FOV to cover the entire width of the head. Also assume we need N_y points in the phase-encoding direction to achieve the resolution we need (see Fig. 1a). Assuming we acquire one echo per TR interval, this would require a time of $N_y \times$ TR to acquire the image.

Now assume we have a coil which only has sensitivity on the left half of the head. In this case, we could image just the left side of the head with half the total FOV. This would require only ½ N_y phase encoding steps to achieve the same resolution, resulting in a two-times increase in our imaging speed. However, we would only have an image of one-half of the head (see Fig. 1c). In addition, this image would have a signal-to-noise ratio (SNR) which is reduced by $\sqrt{2}$ compared to the full acquisition, due to the reduced imaging time.

However, assuming that instead of a single coil, we have an array that includes another coil that can simultaneously image the right side of the head (Fig. 1d). In this case, each coil would only see half of the head, but used together, we could stitch together an image of the whole head

(Fig. 1e). This would have been accomplished in one half of the normal imaging time. This is the most simple version of parallel imaging, but it highlights five basic concepts of parallel imaging: **1)** Multiple receiver coils must be positioned so that each coil has a different sensitivity over the FOV. In this case, each coil has sensitivity on only one side of the head. **2)** Each receiver coil must be supplied with its own independent receiver chain, so that all coils in the array can operate simultaneously **3)** A reduced number of phase-encoding steps are acquired, generally resulting in an aliased, or folded-over image in a reduced imaging time **4)** The following parallel imaging reconstruction has to have knowledge of the sensitivity patterns of the individual coils in the array. In this example, we must know which coil is on the left and which one is one the right. In the most extreme cases, we must know the complex sensitivity of each coil at each pixel in the image **5)** Image SNR is reduced by at least the square-root of the reduction factor used in the image acquisition. In general, SNR is typically reduced even more than this, and is referred to as geometry factor, or g-factor losses

While this example illustrates these basic points, a method such as this is nearly useless in practice, since we typically do not have access to a coil with sharp boundaries between different regions of the object. Other more general methods have been developed to deal with this and other problems encountered in practice. Historically, these methods have been categorized by the path taken from the data acquisition to the reconstructed image. The image domain methods, such as SENSE [2], operate, as expected, entirely in the image domain using folded images and coil sensitivity maps. On the other hand, k-space methods, such as SMASH [3] or Generalized SMASH [4], or GRAPPA [5], work entirely in k-space. There are also hybrid methods which perform the reconstruction directly from the k-space data to an unaliased image. SPACERIP [6] and generalized SENSE [2,7] are examples of this type of reconstruction.
Many different parallel imaging reconstructions are possible. The various methods will all have differences in their theoretical and practical performances, especially in which kinds of coil sensitivity information is needed and how this information is acquired. A method should be selected based on the relative advantages or disadvantages for the specific application for which it is used. These various tradeoffs will be discussed during the lecture. Also, clinically relevant issues, such as artifacts, etc. will be discussed and methods to deal with these in a way that preserves patient throughput will be discussed.

SNR Losses and the Geometry Factor

As mentioned above, one property which is common to all parallel imaging methods is an intrinsic SNR loss. In general, there are two sources for this loss in SNR. First, since we are decreasing the total scan time, the SNR falls with the square-root of the imaging time. So for a speed-up of 4x, we observe a minimum of a 2x decrease in SNR. Unfortunately, there are additional losses due to the parallel imaging process. These are due to the configuration of the array, and are thus referred to as "geometry factor" or "g-factor" related losses. In general, these losses depend on how different the sensitivity is between the various elements. If the elements all look the same, the g-factor will go up. On the other hand, if the coils are all completely isolated with no overlapping signals, then we can approach the situation where there are no geometry-factor related losses. The exact amount of loss depends strongly on the exact imaging situation, but does tend to increase with more aggressive accelerations.

References

1. Griswold MA, Jakob PM, Nittka M, Goldfarb JW, Haase A. Partiallay parallel imaging with localized sensitivities (PILS). Magn Reson Med 2000; 44: 602-609
2. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. Magn Reson Med 1999; 42: 952-962
3. Sodickson DK, Manning WJ. Simultaneous acquisition of spatial harmonics (SMASH): Fast imaging with radiofrequency coil arrays. Magn Reson Med 1997; 38: 591-603
4. Bydder M, Larkman DJ, Hajnal JV. Generalized SMASH imaging. Magn Reson Med 2002; 47: 160-170
5. Griswold MA, Jakob PM, Heidemann RM, Nittka M, Jellus V, Wang J, Kiefer B, Haase A. Generalized autocalibrating partially parallel acqusitions (GRAPPA). Magn Reson Med 2002; 47: 1202-1210
6. Kyriakos WE, Panych LP, Kacher DF, Westin CF, Bao SM, Mulkern RV, Jolesz FA. Sensitivity profiles from an array of coils for encoding and reconstruction in parallel (SPACE RIP). Magn Reson Med 2000; 44: 301-308
7. Pruessmann KP, Weiger M, Bornert P, Boesiger P.,Advances in sensitivity encoding with arbitrary k-space trajectories. Magn Reson Med 2001; 46: 638-651