Reconstruction for Muti-Coil Acquisition

David J Larkman PhD

Imaging Sciences Department, Imperial College London.

In the late 1980's array coils were developed for MRI, initially the motivation for this development was increase in signal to noise ratio (SNR). This is achieved by using array coils to construct shaped regions of sensitivity. In general the shape of the sensitivity of a single circular coil element is preserved as the coil size is increased, so if we make the coils bigger the depth of sensitivity increases (penetration depth). Both signal and noise can only be acquired from within a coils sensitive volume. In an imaging experiment where noise is present the sensitive volume of a coil extends over a limited region (before the sensitivity falls below the noise floor). We can localise signal in MRI, and indeed this is the strength of MRI, we can restrict the signal that is acquired in a coil to come from only a single slice in the volume (or slab or voxel depending on the localisation methods used). However, noise cannot be spatially localised and so we always collect noise from the whole volume of sensitivity of the coil. From this discussion it is clear that carefully tailoring the coils sensitivity to the volume of interest for imaging can increase SNR because noise is gathered from the smallest possible volume commensurate with the anatomy to be imaged.

The maximum SNR from an array coil is obtained only when the signals from each coil are combined in an appropriate way. If we simply add the images together then at all pixels all the coils contribute equal noise but the signal contribution depends on the object position relative to the coils. Clearly this is not the right way to combine signals. A thorough treatment of how to optimally combine array coil signals for maximum SNR was presented by Roemer[1] in 1990 and remained the standard reference for array coil signal combination since then. Roemers key observation was that the optimal signal to noise in an image reconstructed from multiple coils is achieved from by combining the signals on a pixel by pixel basis weighting each coil by the sensitivity of the coil at that location in space. In this way pixels close to a single coil element obtain the vast majority of their information (and noise) from only that coil and pixels between elements have shared information. This observation can be readily theoretically proved ,and also feels intuitively correct, it has underpinned a lot of the thought processes which have developed in partially parallel imaging since then. What Roemer also observed was that in general we do not know the spatial sensitivity of the coils but he observed that for most SNR's the sum of squares combination approximated the sensitivity combined result to withing 10% or so. This sidestepped the need to measure coil sensitivities and in doing so may well have inadvertently held up the development of PPI. If coil calibration had already been an integral part of array coil imaging before PPI then some of the early PPI proposals may well have been adopted more quickly.
For array coils to produce maximum SNR they must be "independent" i.e. there is no crosstalk between coils, this requirement imposes the need for multiple receiver chains on scanners, one for each receiver coil, this is expensive but, because of the requirement to combine coil data in the image domain, required.
An interesting footnote to this general discussion is that ultimately PPI reached back and re-invented the optimal image combination. A SENSE reconstruction can be run with a "speed up" of 1 (i.e. no acceleration") and what will result is an optimally combined image weighted by the sensitivity of the coil elements. This has been extended to now develop methods for doing such an optimal combination even in the absence of explicit separate coil sensitivity measurements [2]

Using array coils to replace gradient encoding has been in the literature for seventeen years with early papers suggesting completely replacing gradient encoding with coil encoding[3]. This is often referred to as massively parallel imaging and was never practically implemented at the time because of hardware limitations, more recently this has emerged again as a research activity with Single echo Acquisition (SEA) imaging[4]. A more practical proposal was to partially replace gradient encoding with coil encoding, this has become known as Partially Parallel Imaging (PPI). This was first proposed in 1993 [3, 5, 6] but it was not until commercial scanners began to incorporate multiple independent receivers that Sodickson made his ground breaking observations which lead to SMASH[7], this inspired what rapidly followed by Pruessmann with his SENSE paper[8]. With these two PPI proposals, real working imaging methodologies were developed and PPI in one form or another was taken up by all the major manufacturers leading to todays market where all newly purchases systems have some form of PPI.

PPI has now spawned many flavours, SENSE[8], SMASH[7], GRAPPA[9], SPACE-RIP[10], PARS[11], Generalised-SENSE[12], Generalized-SMASH[13] to name most, but by no means all, methods. These methods are all capable of reconstructing a full field of view image from a regularly sub sampled k-space (a reduced field of view image). Differences in approach become apparent when the domain in which the processing is considered. SENSE has both input data and coil reference data in the image domain. Generalized-SMASH and GRAPPA use both the input data and the coil reference data in k-space. SMASH, SPACE-RIP and generalized-SENSE have input data in k-space and coil sensitivity information in the image domain. It is generally accepted that SENSE is the simplest and quickest reconstruction method, however it is also the most limited because it can only operate on data that is regularly sub sampled. The other methods are in general more flexible, they fall crudely into two camps, there are exact methods (generalized SMASH, space RIP, generalized SENSE) which are capable of converging on an exact solution, and approximate methods which require fitting to establish reconstruction weights (SMASH, GRAPPA, PARS). It is important to note that if perfect reference data is available then the exact methods will yield the best results however, it is very often the case, particularly when pushing the limits of speed, that exact reference data will not be available, and in these cases the fitting elements of the other methods help control the reconstruction. A somewhat false distinction is often made between auto-calibrating methods and other methods. Any PPI method can be made auto-calibrating. The auto calibration refers to the integration of coil reference data (a low resolution full field of view image) with the actual exam. This has the distinct advantage that reference data is well matched to the target data and so is robust against motion but it also reduces the speed up as this extra data takes time to acquire.

We will look indetail at the basics of SENSE, SMASH and GRAPPA reconstruction and look at the differences between these methods, in particular careful attention will be paid to the calibration required for each of these methods and how this affects reconstruction fidelity.

No discussion of PPI is complete without consideration of the penalties associated with it Looking at the SNR reduction equation realized by Pruessmann:

$$SNR_{PPI} = \frac{SNR_{full}}{g.\sqrt{R}}$$

We can see it has two significant terms, the term which related to the reduced number of measurements $\sqrt{R}$ and the g-factor which is related to coil geometry but also the location of the imaging plane, the speed up factor and the k-space sampling pattern. Before looking at each term it should be noted that the g-factor is not useful on its own because there is no account taken of the baseline SNR. If SNR_{full} is small then the coil is not useful regardless of the value of g.

The $\sqrt{R}$ term is inescapable for traditional single echo PPI methods which reduce the number of k-space lines measured. However, there are ways that this term can be mitigated, the first it to parallel image not by reduction of phase encode lines but by exciting multiple slices simultaneously[14, 15] in these methods, because an increased number of spins are excited at the separate slice locations, the total signal sampled is increased and the $\sqrt{R}$ term no longer applies. The second “exception” is when we consider EPI imaging. If PPI is used to reduce the length of the echo train then the echo amplitude reduction due to T2* decay from the first to the last echo is reduced, this combined with shorter TE’s, accessible now the echotrain is shorter, make it possible to produce an SNR gain using PPI in EPI.

The g-factor arises as a result of the coil sensitivities being too similar.This ill-conditions the matrix inversion at the heart of PPI. An ill-conditioned system is defined as one where disproportionately large changes occur at the output when small perturbations disturb the input. In the case of PPI the small input perturbations are noise in the sub sampled input data. The ideal g-factor is 1, g-factors below 1.2 are typically tolerable depending on application. The g-factor varies from pixel to pixel across the image domain so a g-factor map is required to describe the performance of a particular configuration. Clearly the design of the coil is critical to achieving a g-factor close to 1 (the ideal). An important observation is that the g-factor is also dependent on the k-space sampling pattern although detailed discussion of this is beyond the scope of this talk. Conceptually the appeal of non-Cartesian PPI is clear but it is not a simple extension of Cartesian PPI. A consideration of image domain PPI processing quickly reveals the difficulties associated with PPI for non-Cartesian sampling patterns. Whereas with Cartesian SENSE for a speed up factor of 2, 2 pixels are aliased onto each other, if the same speed-up is considered for a spiral sampling pattern then in the image domain pixels are aliased to circular rings of pixels which in turn are aliased to other rings resulting in the potential for a single pixel to be aliased to all other pixels. This then requires a whole image approach to reconstruction rather than a pixel by pixel approach seen in simple SENSE. Pruesseman[12] has proposed a conjugate-gradient iteration based method for this type of reconstruction but these methods are numerically intensive and with current computing power this is preventing their widespread application. A more practical solution is presented by GRAPPA[9] and PARS[11] where local neighbourhoods in k-space are used to establish reconstruction weights suitable for similar regions in k-space. This approach, whilst approximate, does yield good reconstructions quickly.

Time permitting I would also like to point towards some research which illuminates the PPI reconstruction problem by modifying the reference data to produce artefact free images Those readers interested in the effects of reference data on the reconstruction may find the following references interesting. The use of PPI to correct artefacts in images opens up two avenues. The first, most obvious one, is to correct common problems with MR data such as those produced by motion or flow,[16, 17] these methods use a coil consistency check as the image quality metric. Multiple coils give different views of the object. Consider the scenario that part of the object is moving (eg flow in a large vessel) and that this is producing artefacts in the images. If we look at the individual coil images we will see that the intensity of these artefacts is different in each coil image because the proximity of the artefact source to each coil is different. The artefact has an intensity related to the sensitivity of the coil at the position of the source of the artefact, the artefact itself however is not at this position. If we remove the coil sensitivity information (by division in the image domain or deconvolution in k-space) now all coil images should be identical, where they are not identical this is artefact. This measure can then be used as a cost function in an iterative scheme to reduce artefacts.
The second avenue is that we are now free to impose coherent artefacts into images because we have a framework for correction given by SENSE. If the artefact is coherent, such as a Nyquist ghost, then correction can be done simply in the image domain[18-20]. These approaches allow us to change the way we acquire data, for example multi echo trains (TSE, GRASE) can be laid out sequentially in k-space rather than the more usual spreading across k-space.

1. Roemer, P.B., et al., *The NMR phased array*. Magn Reson Med, 1990. **16**(2): p. 192-225.
2. Bydder, M., et al, *Combination of signals from array coils using image-based estimation of coil sensitivity profiles.* 2002. **47**(3): p. 539-548.
3. Hutchinson, M. and U. Raff, *Fast MRI data acquisition using multiple detectors.* Magn Reson Med, 1988. **6**(1): p. 87-91.
4. McDougall, M.P. and S.M. Wright, *64-channel array coil for single echo acquisition magnetic resonance imaging.* Magn Reson Med, 2005. **54**(2): p. 386-92.
5. Ra, J.B. and C.Y. Rim, *Fast imaging using subencoding data sets from multiple detectors.* Magn Reson Med, 1993. **30**(1): p. 142-5.
6. Carlson, J.W. and T. Minemura, *Imaging time reduction through multiple receiver coil data acquisition and image reconstruction.* Magn Reson Med, 1993. **29**(5): p. 681-7.
7. Sodickson, D.K., M.A. Griswold, and P.M. Jakob, *SMASH imaging.* Magn Reson Imaging Clin N Am, 1999. **7**(2): p. 237-54, vii-viii.
8. Pruessmann, K.P., et al., *SENSE: sensitivity encoding for fast MRI.* Magn Reson Med, 1999. **42**(5): p. 952-62.
9. Griswold, M.A., et al., *Generalized autocalibrating partially parallel acquisitions (GRAPPA).* Magn Reson Med, 2002. **47**(6): p. 1202-10.
10. Kyriakos, W.E., et al., *Sensitivity profiles from an array of coils for encoding and reconstruction in parallel (SPACE RIP).* Magn Reson Med, 2000. **44**(2): p. 301-8.
11. Yeh, E.N., et al., *Parallel magnetic resonance imaging with adaptive radius in k-space (PARS): constrained image reconstruction using k-space locality in radiofrequency coil encoded data.* Magn Reson Med, 2005. **53**(6): p. 1383-92.
12. Pruessmann, K.P., et al., *Advances in sensitivity encoding with arbitrary k-space trajectories.* Magn Reson Med, 2001. **46**(4): p. 638-51.
13. Bydder, M., et al, *Generalized SMASH imaging.* Magn Reson Med, 2002. **47**(1): p. 160-70.
14. Larkman, D.J., et al., *Use of multicoil arrays for separation of signal from multiple slices simultaneously excited.* J Magn Reson Imaging, 2001. **13**(2): p. 313-7.

15. Breuer, F.A., et al., *Controlled aliasing in parallel imaging results in higher acceleration (CAIPIRINHA) for multi-slice imaging.* Magn Reson Med, 2005. **53**(3): p. 684-91.
16. Atkinson, D., et al., *Coil-based artifact reduction.* Magn Reson Med, 2004. **52**(4): p. 825-30.
17. Larkman, D.J., et al , *Artifact reduction using parallel imaging methods.* Top Magn Reson Imaging, 2004. **15**(4): p. 267-75.
18. Kellman, P., et al., *Phased array ghost elimination (PAGE) for segmented SSFP imaging with interrupted steady-state.* Magn Reson Med, 2002. **48**(6): p. 1076-80.
19. Kuhara, S. *A Novel EPI Reconstruction Technique Using Multiple RF Coil Sensitivity Maps* . in *ISMRM*. 2000.