Historical Perspectives on the Development of a Parallel MRI System for Head Imaging

J.W. Carlson
Berkeley *Drosophila* Genome Project, Lawrence Berkeley National Laboratory
Berkeley, CA 94720

The second half of the 1980's was a time in which MRI had progressed from the research laboratories to routine clinical practice. The fundamental physics of the system was well understood, as was the engineering needed to provide instruments suitable for clinical efficacy. The perception of the time was that next step in the expansion of the clinical utility of MRI was in the direction of ultrafast scanning modalities. At the Radiologic Imaging Laboratory at UCSF during this time, Larry Crooks was leading an effort in implementing an echo–planar MRI system for pediatric applications[1]. This work motivated explorations of methods for decreasing the minimum scan time for image formation by decreasing the number of phase–encoded echoes.

One idea was a hybrid method of using dual receiving coils to reduce the number of phase–encoded echoes by 1/2. The first publication on dual receiver reconstruction [2] laid out the principles of the approach. Two coils formed on the surface of a cylinder – one a uniform saddle coil and the other a coil configured to have a response proportional to location in the phase–encoded direction – are intrinsically decoupled and can be used to simultaneously receive the NMR signal. The rf gradient coil could be built using a standard technique of saddle coils or it could by using a birdcage receiver tuned to its second resonant mode. Since the coils have a well defined spatial response, we can design a reconstruction algorithm without having to measure the spatial sensitivity. In the first publication a very simple scheme was proposed. Since the location of the signal in the readout (x) direction can be localized easily, only the phase–encoded (y) direction was relevant. In the simplest approximation, the first phase–encoded signal depends on the spatially dependent magnetization, $m(x, y)$,

$$\int e^{iy} m(x,y)dy \approx \int m(x,y)dy + i\int ym(x,y)dy, \tag{1}$$

or, the first phase–encoded echo is a sum of the uniform and gradient signals with no phase encoding and a 90° relative phase shift. Similarly, higher order odd phase–encoded signals can be approximated by summing the uniform and gradient signals from the adjacent even phase–encoded signal.

In retrospect this simple method would not have been acceptable since the residual artifacts after applying this approximation are too high. Within a few months an improvement [3] seen based on simulations showed that the artifact could be reduced by approximating the first phase–encoded echo from a linear combination of the uniform and gradient signal from the lower phase–encoded echo and the higher phase–encoded echo. Or,

$$\int e^{iy} m(x,y)dy \approx \frac{1}{2}\int m(x,y)dy + \frac{i}{4}\int ym(x,y)dy + \frac{1}{2}\int e^{2iy}m(x,y)dy - \frac{i}{4}\int ye^{2iy}m(x,y)dy. \tag{2}$$

Only after this later publication was it seen that this improvement could be extended further. The structure of the approximation in Eqn.(2) arises from an expansion of e^{iy} in polynomials of e^{2niy} and ye^{2iny}. The next order is:

$$e^{iy} = \frac{1}{512}\Big[243(1+e^{2iy}) + 162iy(1-e^{2iy}) + 13(e^{-2iy}+e^{4iy}) + 6iy(e^{-2iy}-e^{4iy})\Big] + \mathcal{O}(y^5). \tag{3}$$

The expansion can be carried out to arbitrarily high order by including additional even phase–encoding terms. Based on simulations, the artifact was reduced to acceptable levels at tenth order. As an alternate to doing a power series expansion, one can produce a set of coefficients so that the error in approximating e^{iy} with terms containing e^{2iny} and ye^{2iny} is uniform over an interval of $-\pi < y < \pi$. It is not possible to make the error vanish everywhere in the region and still have a well–behaved solution, so one must choose a subinterval over which the error is minimized. This finalized the algorithm used to generate the skipped odd phase–encoded echoes.

In the actual implementation there are a set of details that must be considered in the reconstruction procedure:

1. The signals received on both channels need to be adjusted for the different receiver coil sensitivities and phase shifts.
2. The center defined by the rf gradient coil will not coincide with the center defined by the magnet's gradient coil.
3. Maxwell's equations prohibit an rf field which consists only of one component which varies linearly in y. Any reconstruction algorithm must account for realistic reception patterns.
4. Uniform coils are not perfectly uniform, nor are rf gradient coils perfectly linear.
5. While the uniform and gradient rf coils are intrinsically decoupled, imperfections in construction cause a lack of symmetry and some coupling between the coils.

Fortunately, the first four details can be taken care of with one technique. One can shift the center of the rf gradient coil during the reconstruction by adding a contribution from the uniform coil. The term which is used to in Eqn.(3) for the gradient signal is then a linear combination of the coil signals on the two receivers,

$$G_r = \alpha G_c + \beta U_c. \tag{4}$$

The subscript c designates the measured signal intensity to differentiate it from the quantity used in the reconstruction which is designated by the subscript r. The quantities α and β are complex quantities that will be used to adjust the relative gain and phase differences of the two channels. For transaxial images or sagittal images off–center, Maxwell's equations require that there be a x component of the rf field from the rf gradient coil which also varies linearly in x. This is effectively another term proportional to the uniform coil's signal, except that this proportionality constant depends on x. In order to do the reconstruction, we acquire a few odd phase–encoded echoes close to center of Fourier space and use this data to do a weighted least–squares fit for α and β, where both of these quantities are complex, linear functions of x.

The remaining issue of of coil construction is determined by details of the implementation. For the first phantom images acquired, in 1989 [4], two coils were used to simulate simultaneous acquisition. The MRI system used did not have dual receivers at the time, and, since this was a test of the reconstruction algorithm and not the coils, both coils were not simultaneously active. In later work in collaboration with Takahashi Minemura [5], we were able to construct dual quadrature coils in which all four channels could be simultaneously active, but still did not have dual receivers. Our attempts to simulate simultaneous acquisition on human subjects by acquiring two separate images consistently had a frequency displacement between the two acquisitions that we attributed to center frequency drift or environmental disturbances of the field. Only after we had dual receiver capability in September, 1994, were we able to test the reconstruction on human images. Examples are shown in Figure 1.

Signal to noise of the reconstructed images never achieved the levels that we had expected. In retrospect, our assumptions that we would be in the sample–dominated noise limit were not correct. We had not been able to achieve significant coil loading in the rf gradient coil at 15MHz. Nonetheless, we were able to demonstrate reliable reconstruction with an acceptable artifact level.

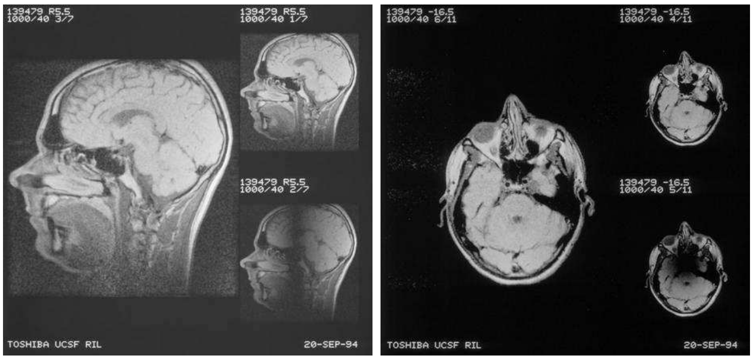

Figure 1: Representative human images from September, 1994, using simultaneous reception on quadrature receivers in sagittal (left) and transaxial (right) orientation. In both cases, the inset images show the full 256 × 256 acquired images, while the larger image is the reconstructed based on 138 phase–encoded echoes. Images reconstructed using only the even phase–encoded echoes would show severe overlap at this resolution (data not shown).

References

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