A Buyer's and User's Guide to RF Coils

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The RF coil is one of the primary factors determining the speed and image quality in MRI. Over the past 10 years there have been many important technical developments in RF coil design, particularly the rapid increase in the number of available RF channels. This has enabled the development of a host of accelerated acquisition methods which largely rely on the multiplicity of RF channels to significantly reduce scan times. Advances in RF coil design have also proven essential in unlocking the potential of ultra high field MRI. Currently a hot research area is parallel transmit, where RF is broadcast through a multitude of antennas promising to provide ever greater control over the RF excitation profile in the body. For someone considering the purchase of a new scanner or updating an existing scanner with state of the art RF coils, there are many factors to consider regarding the performance benefits of the multitude of RF coils presently available. In this presentation the basic principles determining RF coil performance and the practical trade-offs of various coil designs in the clinical setting will be explored.

Array Coils

Until the relatively recent surge in the number of RF channels available on the scanner, the majority of RF coils in day to day use in human imaging were volume coils such as birdcage or TEM coils. These can be used both for excitation and reception. In receive mode they provide highly uniform volumetric coverage at fields of 1.5T and lower, and are still the main workhorse for RF transmit up to 3T. With the arrival of phased array coils [1], great improvements in sensitivity were possible. The phased array consists of a number of surface coil loops which are combined together. Each surface coil is very sensitive to nearby tissue, improving enormously over the sensitivity of a volume coil, but only over a limited volume. By combining surface coil loops into an array this high sensitivity can be extended to cover a large field of view. Today the vast majority of coils used in human imaging are array coils, with element counts ranging from 4 to 32 channels. Experimental systems have been developed with as many as 96 elements in a head coil or 128 elements in a torso array [2] [3] [4]. So is it always better to have more elements in an RF coil? To understand this we must consider several principles of array coil performance.

Sensitivity:

We define sensitivity more formally as SNR, or Signal to Noise Ratio. While the individual elements of an array can provide substantially higher sensitivity to tissue which is close to the element, if we want to image the same volume of tissue that we were able to with a birdcage or TEM volume coil we need an array which completely surrounds the object. If we build an array of identical dimensions to a volume coil, the general principle is that both coils will have nearly the same central SNR, but the array will have substantially higher SNR in the periphery [5]. However, in practice the array can do better than this because it is possible to place the array coils close to the body, following the contours of the body. The volume coil, on the other hand, is disturbed by close proximity to tissue and loses symmetry and efficiency, and must generally be built on a rigid circular or elliptical cylindrical former some distance from the body. Generally speaking the central SNR will be determined by how well the anatomy fills the volume of the coil, though there are exceptions to this rule.

What then is the advantage of increasing the number of elements in the array? As we increase the number of surrounding elements from 4 to 8 to 16, we generally find that the central SNR increases, and that the peripheral SNR increases even more [6]. As we push to higher numbers of elements, there is generally little or no central SNR gain, except at higher fields where the behavior of electromagnetic waves favors even higher element counts. Eventually increasing the number of array elements is counter-productive as the individual surface coils become very small and cumulative losses and noise sources in the array erode the intended SNR gains. These principles can be seen in Figure 1, where three head coils are compared. The first is a 12 channel commercial coil (Siemens Healthcare, Erlangen Germany) of fairly generous proportions (27 x 25cm). The other two coils were built on a very tight fitting helmet (22 x 18cm) and had 32 and 96 elements respectively. The higher central SNR of the helmet coils compared to the commercial head coil can largely be attributed to their smaller dimensions, which conformed so tightly to the head that a small proportion of adults could not even fit in them. Comparing the 96 and 32 channel coils, we see that the central SNR is almost the same for the two coils, while the 96 channel coil provides SNR gains in the periphery. One could conclude that, given the expense and complexity of a 96 channel coil system, the performance gains are not significant enough to merit the use of such a coil in routine clinical scanning, and that 32 channels (or less) may be sufficient. Similar trade-offs will be encountered when considering other high element count arrays.

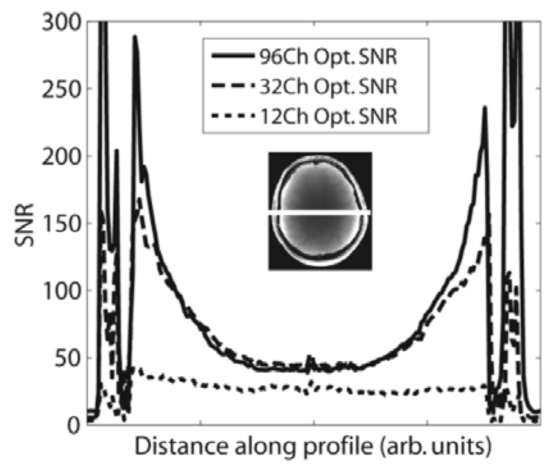

Figure 1. SNR Profiles for 3 head coil designs at 3T

Acceleration:

As well as providing higher SNR than volume coil designs, array coils also enable the use of accelerated imaging techniques such as SENSE and GRAPPA [7] [8]. MRI is a time consuming process because it is necessary to scan through a very large parameter space, obtaining lines of data at a large number of different gradient settings. If we simply skip some of those lines, we can acquire the data faster, but the reconstructed image will be aliased, which means that the outer edges of the image are folded in on top of the middle (Figure 2). The SENSE and GRAPPA algorithms make use of the diverse sensitivity profiles of the various coil elements in the array to unfold the image and produce an un-aliased image which has been obtained in substantially less time

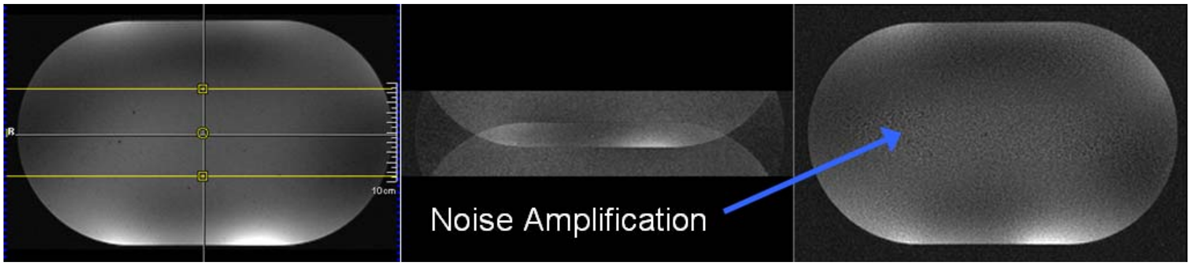

Figure 2. Left: Full acquisition. Center: Accelerated acquisition directly reconstructed. Right: Un-aliased image showing regions of enhanced noise due to g-factor effects

than the normal acquisition. There are two penalties for this accelerated data acquisition. Since SNR increases as the square root of the data sampling time, shortening the scan intrinsically reduces the SNR. There is however an additional SNR loss associated with the unfolding algorithm which is typically characterized by the so-called g-factor. This is related to the geometry of the array, and causes noise amplification in selected regions of the image. As a general rule, having more elements in the array results in lower g-factor. It is also important how the elements are arranged in relation to the phase encode direction, since accelerated imaging involves skipping phase encode steps. If there are not several different coil elements arranged along the phase encode direction, the g-factor will be very high and the image will be degraded by noise. A practical demonstration of this can be seen in the design of knee coils. An 8 channel knee coil (Invivo Corp. Gainesville FL) has 8 elements arranged in a single row around the knee. A 15 channel coil (QED, Cleveland OH) has 3 rows of 5 elements. G-factor maps for acceleration in the AP direction (across the diameter of the coil) and the HF direction (along the length of the coil) are shown in Figure 3. The 15 channel coil shows substantially lower g-factor for acceleration in the HF direction, and this is of practical importance because the phase encode direction is most commonly oriented in this direction to avoid in-flow artifacts. In terms of SNR the 8 channel coil has slightly higher SNR but covers a smaller field of view. The more flexible acceleration capability of the 15 channel coil could be seen as an advantage of this design.

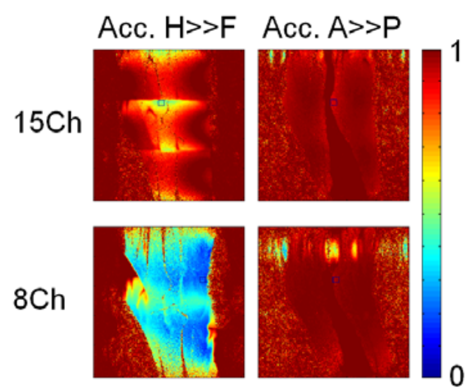

Figure 3. Maps of 1/g-factor for 3X acceleration for 15Ch and 8Ch knee coils. Both coils perform similarly for A>>P acceleration (right) but for H>>F acceleration the single row of coils in the 8 channel coil causes very high g-factors (blue colors)

When acquiring 3D MR images it is often possible to accelerate the acquisition in two directions, the in-plane phase encode direction and the through-plane partition direction. These accelerations are by and large independent of each other in the sense that the resulting g-factor is not necessarily significantly higher for a 3X acceleration in one direction compared to a 3X3 = 9 times acceleration in two directions. You still have to deal with the SNR loss due simply to the shorter scan time, but with high element counts acceleration of 3D acquisitions by a factor of 4 to 12 times is often perfectly feasible, and motivates the choice of a coil with more elements.

Summary:

Significant gains in SNR can be obtained through the use of anatomically optimized coils with a reasonably high channel count. This SNR can be traded off for higher resolution, or shorter scan times with increased patient throughput. There are some provisos, however, which will be discussed in more detail in the presentation. These include possible problems with reconstruction times with high element count arrays, edge artifacts due to extreme image brightness in the periphery, and workflow issues if the use of a particular coil requires removal of other standard coils from the scanner.

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