Coil Arrays for Parallel MRI: Introduction and Overview.

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Introduction. The introduction of coil arrays in MRI for improving signal-to-noise ratio (SNR) [1, 2] led manufacturers to include multiple channels in MR receivers during the 1990's. More recently, the spectacular success of partially parallel imaging methods [3-6] is driving the industry to develop systems with far more receiver channels to enable higher acceleration factors and broader coverage. The increasing number of high-field MRI systems, providing the higher SNR necessary for high acceleration factor parallel imaging, will continue to push the development of hardware and techniques for high-speed parallel imaging. The increasing number of channels and higher frequencies lead to a variety of challenges that researchers are investigating in order to realize the full potential of parallel MRI. This talk review the basics of coil arrays, from improving signal-to-noise to using the sensitivity patterns for image localization. Topics will include decoupling of coils using geometrical and electronic methods, and the effect of coil patterns, the g-factor, on parallel imaging. Methods for modeling the RF coils and the difficulty with modeling (and using) RF coils at high-field will be introduced. Finally, an overview of current technical challenges in the development of parallel MRI will be discussed. These include the widespread availability of high channel coil receivers, the difficulty in developing coil arrays with large numbers of elements that are well-decoupled and maintain high SNR from the individual elements, and processing the large amounts of data. Other, less obvious factors that affect the development of parallel MRI include the ability to decouple arrays of coils for transmit SENSE [7] and parallel imaging at high fields [8, 9], and the optimization of systems for multi-sample MRI [10-12], another form of parallel MRI.

Parallel Imaging at High Acceleration Factors. The potential acceleration rates that can be obtained in parallel imaging are limited by a number of factors. First, the SNR is proportional to the square root of the total observation time. Therefore, in general, an acceleration factor of L results in a decrease in SNR of $\sqrt{L}$ [4]. This becomes quite costly at high acceleration factors. Independence of the coils, both at their ports and in terms of their sensitivity patterns becomes more of a challenge with larger numbers of elements. Several groups are investigating techniques to offset the effects of coil coupling [13] or design coil arrays that exhibit reduced coupling, such as microstrip arrays [9,14]. Optimizing coils for low g-factors can lead to different designs than optimizing for high SNR, a topic that remains under discussion. An excellent example of this can be found in articles by Bottomley [15] and Weiger [16]. Both authors consider the optimization of coils for cardiac imaging.

For whole volume imaging, several authors have suggested that 1D acceleration factors beyond four or five may be difficult to achieve in practice. However, Weiger and others have applied SENSE in two orthogonal directions, 2D SENSE, during 3D image acquisitions [17, 18]. 2D SENSE promises much higher acceleration factors. In certain applications, particularly the imaging of surfaces to which an array can be conformed, the acquisition of an image in a single echo is possible [19,20]. Figure 1 is an illustration of this, showing a SEA image of a resolution phantom obtained at 1.5 Tesla using a 64 channel receiver and 64 channel array installed at the University of Würzburg. Because an image can be obtained from each echo, SEA imaging could prove useful for observing the evolution of magnetization during conventional imaging sequences. This technique has enabled MR "movies" at 125 frames per second [21].

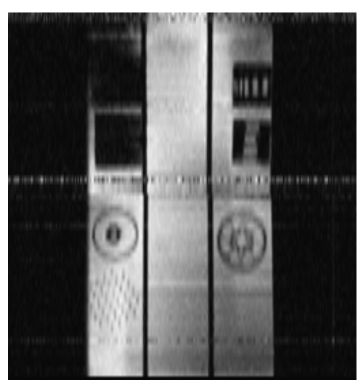

Figure1. Example image obtained in a single echo using a 64 channel array coil at 1.5 Tesla.

Parallel Imaging at High Fields. Parallel imaging methods, such as SMASH and SENSE are of obvious interest at high-fields, where the higher SNR promises to increase the number of potential applications. However, the high dielectric constant of most tissues, combined with the increasing electrical dimensions of the patient at high fields, leads to inhomogeneous fields and potentially to dielectric resonances [22, 23] which may have significant effects on parallel imaging at high fields [8]. These results indicate that it may be possible to use RF shimming methods [24] to locally enhance the geometry factor. Alternatively, strong dielectric resonance effects could increase similarity of field patterns, adversely affecting g-factors.

Transmit Phased Array Coils. An exciting application for array coils is the extension of parallel MRI to the excitation phase, first suggested by Katscher as “Transmit SENSE” [7]. This technique has the potential to greatly shorten the duration of sophisticated spectral-spatial RF pulses [25], allowing them to be combined with short TE/short TR pulse sequences. Other benefits of transmit sense include the potential for reduced SAR [26] and improved B1 homogeneity at high field strengths. A number of groups have investigated the development of transmit coil arrays [27, 28, 14]. Unlike receive arrays where the use of isolation preamplifiers can significantly reduce current on the array elements reducing the effects of inductive coupling [2, 29], transmit array elements obviously must produce a defined current on each element. Mutual impedances between the elements change the active impedance of each element in proportion to the rung currents. If this impedance modulation is not taken into account (a very difficult process in the face of unknown sample loading), the desired current waveforms are not achieved. A number of approaches have been suggested to overcome this problem. Microstrip line elements, which reduce coupling between elements, have been tested with good success [9, 14]. In another approach, Kurpad has developed a current source amplifier for MRI applications [28]. In this amplifier, the input voltage defines the output current rather than the output voltage as in conventional RF amplifiers. Figure 2a shows an eight channel transmit array with integrated current source amplifiers for each rung. Not shown is a controller which allows the relative amplitudes and phases of each channel to be independently adjusted. Because of the current source amplifiers, this control is a direct, rather than an iterative process. Figure 2b shows the calculated (top) and measured (bottom) field patterns received from a homogeneous receive coil after setting the rung currents to excite the first three birdcage modes. Although the array does contain current probes to monitor the rung currents, no adjustment was made to compensate for any residual induced currents.

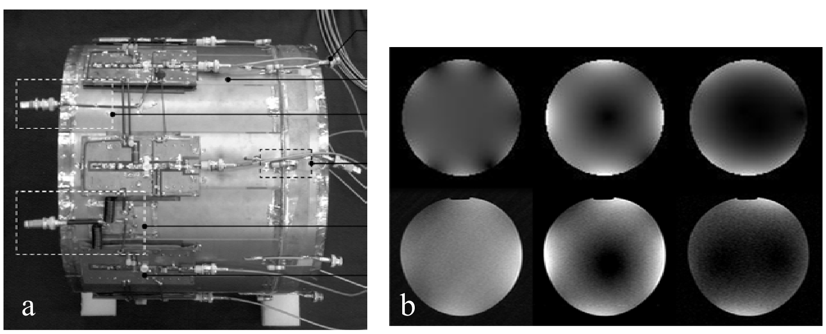

Figure 2. (a). Eight-rung transmit array coil with integrated current amplifiers. (b) Calculated (top) and measured (bottom) transmit patterns obtained with currents set to created the first three birdcage modes. [28].

Acknowledgements: Support by the Dept. of the Army (DAMD17-97-2-7016), the NSF (BES 0101059) and the NIH (EB003296) is gratefully acknowledged. This paper does not necessarily represent the position or policy of the sponsors. The tremendous generosity, expertise, and patience of Mark Griswold, Mathias Mueller and the physics group at the University of Würzburg while testing the 64 channel array and receiver on their system is greatly appreciated, as is the assistance of Ed Boskamp in designing and testing the transmit head coil array.

References:

1. Hyde, J.S. et al., *J. Magn. Reson*. 70, p. 512, 1986.
2. Roemer, P.B. et al., *Magn. Reson. Med*. 16, p. 192, 1991.
3. Sodickson, D.K et al., *Magn. Reson. Med.* 38, p. 591, 1997.
4. Pruessmann, K.P. et al., *Magn Reson.Med.*, 42, p. 952, 1999.
5. Griswold, M.A. et al., *Magn Reson Med.* 47, p. 1202, 2002.
6. Kyriakos, W.E. et al., *Magn Reson Med.,* 44, p. 301, 2000.
7. Katscher, U. et al., *Magn Reson Med*., 49, p. 144, 2003.
8. Wiesinger, F. et al., *Proc. 12th ISMRM,* p. 323, 2004.
9. Zhang, X., et al., *Proc. 12th ISMRM,* p. 1602, 2004.
10. Li, Y., et al., *Anal Chem,* 71, p. 4815, 1999.
11. Bankson, J.A. et al., *Magn Reson Imaging*, 19, p. 1001, 2001.
12. Bock, N.A et al., *Magn Reson Med*, 49, p. 158, 2003.
13. Ohliger, M.A., et al., *Magn Reson Med*, 52, p. 628, 2004.
14. Wichmann, T. et al., *Proc. 12th ISMRM,* p. 1578, 2004.
15. Bottomley, P.A. et al., Magn Reson Med, 37, p. 591, 1997.
16. Weiger, M., et al., Magn Reson Med, 45, p. 495, 2001.
17. Weiger, M., et al. *MAGMA*, 14, p. 10, 2002.
18. Zhu, Y., et al., *Magn Reson Med,* 2004. 52(4): p. 869.
19. Wright, S.M. et al., *Proc. 2nd. IEEE EMBS*, p. 1181, 2002.
20. McDougall, M.P. et al., *Proc. 11th ISMRM*, p. 4982 2003.
21. Wright S.M. et al., *Proc. 12th ISMRM*, p. 511, 2004.
22. Collins, C.M. et al., *Magn Reson Med,* 45, p. 684, 2001.
23. Yang, Q.X., et al., *Magn Reson Med,* 47, p. 982, 2002.
24. Caserta, J. et al., *J Magn Reson,* 169, p. 187, 2004.
25. Stenger, V.A. et al., *Magn Reson Med,* 50, p. 1100, 2003.
26. Zhu, Y., *Proc. 12th ISMRM,* p. 331, 2004.
27. Ibrahim, T.S., et al., *Magn Reson Imaging*, 18, p. 733, 2000.
28. Kurpad, K., Ph.D. Dissertation, Texas A&M, 2004.
27. Reykowski, A. et al., *Magn Reson.Med.*, 33, p. 848, 1995.