Comparison Between CCMI and CAHM for design Shielded gradient coils for MRI.

S. S. Hidalgo[1,2], C. Gadzinski[2], and B. Rutt[2]

[1]CI3M, UAM Iztapalapa, Mexico, City, Mexico, [2]Imaging Department, Robarts Research Institute, London, Ontario, Canada

Introduction One of the most important concepts for obtaining an image using MRI is the use of magnetic field gradients. There have been developed different techniques to improve the quality of the magnetic field gradients that allow high strength gradient fields that can be rapidly switched on and off for fast imaging modalities, large homogeneous-gradient-volume or minimum inductance. This work is focussed on a comparison between two methods based in the target field method proposed by Turner[1,2] to build shielded gradient coils. The main problem of the target field method is that a current of infinite extent is designed with a set of constraints but the current is modified, and in consequence the final coil might not have the desired characteristics. Carlson[3] proposed a current distribution using a Fourier Series for a coil of finite length. Another solution was proposed by Chronik[4] that adds a set of current constraints forcing the current to lie over a certain length. Another important issue to consider is the interaction of the rapidly switched gradient fields with other conducting structures in the MRI system that generates the eddy currents. To avoid problems in imaging due to these eddy currents shielded gradient coils were proposed by Morich[5] *et al.* and Van Vaals Bergman *et al.*[6]

Method Most gradient coils that are used in MRI consist of wire arrangements on the surface of a cylindrical former. Carlson approach with harmonics minimisation (CAHM) allows a restricted length to be included at the beginning of the design process [3]. Considering the current distribution on the inner cylinder to be limited to the region $|z| < L$, the inner coil current distribution is defined as a weighted harmonic series a_n of finite axial extent ($2l$), Eq (1). A functional is minimized in terms of the coefficients a_n over a mesh of N points defining the region of a desired uniform gradient in Eq (2), where L is the inductance, $\sum_{n=1}^{N}\left[gx_q - B_z(\vec{r_q})\right]^2$ is the local deviation of the field from the desired value. □ enables the adjustment of the inductance, β enables adjusting the reduction of the magnetic field outside of coil, although these actions implies a decrease of the gradient uniformity. Chronik's method, the constraint current minimum inductance target field method (CCMI) is a modification of Turner's minimum inductance method with addition of a set of current constraints [4]. These constraints allow a longer region of uniformity for a shorter gradient coil. The first constraint is related to the *z*-component of the magnetic field. The second current constraint is the closure constraint that prevents any current density from crossing the boundaries of a specified area on the surface of the cylindrical former. The third current constraint forces the current density to remain constrained within some region of the coil surface using the values of the azimuthal current density over a set of coordinates. Once the constraints have been identified, the functional be minimized takes the form

$$U\left(j_\phi^m(k)\right) = L\left(j_\phi^m(k)\right) + \sum_{n=1}^{N}\lambda_n\left[B_z - B_{zn}\right] + \sum_{n=1}^{P}\lambda_p\left[J_\phi - J_{\phi p}\right] + \lambda_q\left[\Lambda_\phi - \Lambda_q\right] \quad (3)$$

where $j_\phi^m(k) = \sum_{n=1}^{N}\lambda_n a_n^m(k) + \sum_{p=1}^{P}\lambda_p a_p^m(k) + \lambda_q c_q^m(k)$. We have analyzed shielded gradient coils with aspect ratio(AR) equal to one (x:z:1). The coils designed by CAHM were designed using 12 coefficients for the current density. For both methods it was calculated the current density for the shield coil, both coils gradient and shield are designed with the same length, property that helps to save physical space in the bore, avoiding claustrophobia.

$$J_x(x,z) = \begin{cases} \sum_{n=1}^{N} a_n\left(\frac{n\pi z}{l}\right) & |z| < l \\ 0 & |z| > l \end{cases} \quad (1)$$

$$U = \sum_{p=1}^{N}\left(gz_p - B_z(\mathbf{r}_p)\right)^2 + \beta\sum_{q=1}(B_z(\mathbf{r}_q))^2 + \lambda L \quad (2)$$

Results and Discussion A transverse head gradient coil for imaging application has been implemented in the MATLAB software language and used to create a shielded gradient coil using the CCMI and CAHM methods. It is important to mention that the gradient coil and the shield coil have the same length. The CAHM approach with minimization allows a flexible method of coil design in which the trade off between of the gradient uniformity and the inductance, can be made in a more intuitive and direct manner than is possible in the target field or minimum inductance methods. It was necessary to establish some criteria to try to make a fair comparison between these two methods, such as the same *AR*=1, similar inductance similar length coil, a similar achieved *z-ROU(z-Region of Uniformity)*. In CAHM method, the variation of the number of harmonics allows having a better control of the current density and avoids oscillations that affect the feasibility of the coil. For head gradient, coils are simulated with inductance≈300uH, length(l)≈radius≈32cm. Figure 1.a shows the comparison of the figure of merit vs achieved *z-ROU* for both designs, CAHM method allows more designs due to the variation of the weighting α and β. In this plot it is seen that coils designed by CAHM not only achieved a better z-ROU but also the figure of merit is increased, CCMI coils can not achieve *z-ROU*>45% of the total length of the coil. Figure 1.b shows the magnetic field ratio for the shielded and unshielded gradient coil along x axis for z = 0.01. It can be seen that CCMI method offers a slightly better screening. It was taken from Figure 1a the optimal design achieved by CCMI and CAHM method, the best case for CCMI coils just achieve a z-ROU=45% however the figure of Merit is very poor compared to the CCMI coils. Table 1 shows a CAHM gradient coil with *AR*=1 achieving a *z-ROU*=45%. Figure 2 shows the wire patterns for a coil that has *z-ROU*=40% for CAHM and CCMI. The length of the coil is more exact in CAHM than in CCMI method (Figure 2), this is due to the length of the coil in CAHM is included in the equation of the current density Eq(1), in comparison to CCMI, the current density is forced to lie over an approximate region, an generally the wires cover a longer or shorter length. Shielded gradient coils by CCMI are less wire dense than CAHM coils; this property could compromise the feasibility of the coil. It has been shown that the compromise between the achieved *z-ROU*, inductance, efficiency, length of the primary and screen coil, and figure of merit can be made according to the desired clinical application of the design of gradient coils.

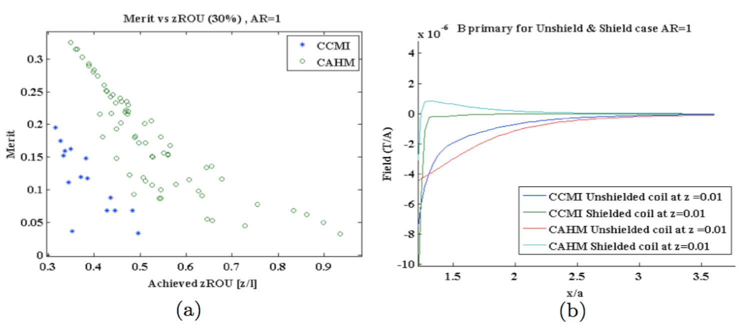

Figure 1. (a)Comparison plot, Merit vs achieved *z-ROU* between CCMI and CAHM for shielded transverse gradient coils for *AR*=1. (b) Magnetic field ratio for the shielded and unshielded gradient coil along x axis for z = 0.01

		CAHM
	Radius [m]	0.16
	Shld Radius [m]	0.176
	eta [T/m/A]	0.2634
	L [uH]	275.1015
	M	0.20204
	Wire density	365.6895
	Wire length	64.5129
	Coil length [m]	0.32011
	Wire number	20
	Gradient error ROU	
Longitudinal extents [cm	10% [cm]	9.9591
	20%[cm]	12.5902
	30%[cm]	14.471
	50%[cm]	17.3814
Transverse extents [cm	10%[cm]	18.4801
	20%[cm]	20.887
	30%[cm]	22.4684
	50%[cm]	24.6287
	Field error ROU	
Longitudinal extents [cm	10%[cm]	9.9591
	20%[cm]	12.5902
	30%[cm]	14.471
	50%[cm]	17.3814
Transverse extents [cm	10%[cm]	12.951
	20%[cm]	14.6861
	30%[cm]	15.8343
	50%[cm]	17.2037

Table 1 Comparison table for shielded coil for *AR*=1 achieving a *z-ROU*=45%,

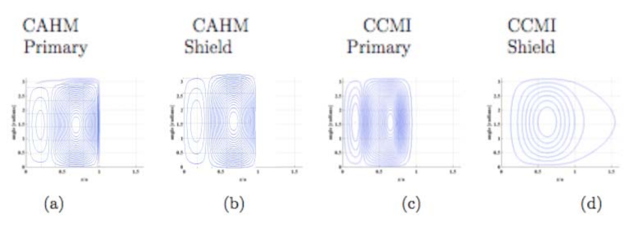

Figure 2. Wire patterns for shielded coils for *AR*=1 achieving a *z-ROU*=45% by CAHM and CCMI.

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