# A Shielded MRI Breast-Coaxial Coil

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## Introduction

To acquire good breast images, a dedicated r.f. probe is necessary, and the probe design is a major factor in determining image quality and field of view. The optimal design should have excellent signal-to-noise ratio (S/N) and good $B_1$ field homogeneity, yet allow imaging of not only the breast but also of the chest wall and the axillary region as tumours also occur in these areas. Unfortunately, these requirements tend to be conflicting. A fundamental design issue is the direction of the probe's $B_1$ field. When parallel to the chest wall, as the coil conductor cannot be in the chest, the field will not penetrate far into the former and will exhibit gross inhomogeneity therein. On the other hand, as the wall tissue is an electrical conductor, we might expect superior S/N. If the $B_1$ field is perpendicular to the chest wall, a Helmholtz pair probe will give good field homogeneity over the breast and good chest wall penetration, but significant chest power dissipation with concomitant loss of S/N. In considering the origin of the losses, we realised that weak $B_1$ fields covering a large volume could contribute to substantially larger losses than bigger fields confined to a smaller volume. Thus shielding of the chest wall away from the breast was investigated and a substantial improvement in performance obtained.

## Design and Construction

High spatial resolution is essential for detecting small tumours (*1*), and thus the highest S/N is needed. Our probe is therefore for imaging a single breast at 128 MHz (3T). The second breast is held away from the coils and shielded. The probe comprises three separate, unconnected rings of conductor: a main Helmholtz pair, of radius and nominal separation 6.8 cm, and a smaller, tuned coupling ring of radius 5.5 cm, arranged as shown in Fig. 1. Their axis is vertical and the patient therefore prone. The separation of the Helmholtz pair is slightly and remotely adjustable to allow tuning, while the coupling ring can be moved axially to permit matching. The three coils are coupled by mutual inductance, and the coupling ring is attached to 50 Ω cable with a sleeve-type balun. To minimize dielectric loss and sample-related frequency change, each Helmholtz ring is tuned with four capacitors spaced 90° apart, the pair resonating in phase at the Larmor frequency. The tuned inductive coupling renders tuning and matching independent (*2*). The probe holder was high-density polyethylene, and the three rings, each of conductor diameter 9.6 mm, were machined from plexiglas®, carefully cleaned and then copper-plated to avoid the ferromagnetic impurities found in copper tubing. Finally, annular aluminium foil shields of thickness 20 µm were placed symmetrically above and below the probe, as shown in Fig. 1, on the surface and within the wooden patient bed.

## Probe Loading

When a breast was inserted into the probe without the two foil shields, the Q factor (unloaded Q = 360) dropped almost independently of breast size (cup size B, Q = 46; cup size D, Q = 44). This suggested that the dominant source of noise was the chest. With two men, similar results (Q = 60 and 54) were obtained, confirming the hypothesis. Thus shields were added with the goal of reducing the fringe $B_1$ field penetrating the chest beyond the breast.

It is difficult to calculate the field distribution produced by the ensemble of coils and shields. Thus we built several shields of various internal diameters. (The external diameter was set by width of the patient table, which just fit the bore.) With each pair of shields, the separation of the Helmholtz coils was adjusted to maintain $B_1$ field homogeneity and the coils were also retuned. We then compared performances using field mapping with a small search coil, the following specifications being desired: 1. The shield should not reduce the unloaded Q factor by more than 10%; 2. The field on the probe axis at the chest wall should be about 50% of the field at the centre of the probe; 3. The field at the edge of the breast at the chest wall should be no less than 33% of the field at the centre of the probe. These specifications were met with a top sheet internal diameter 2 cm larger than that of the probe. Thus the whole edge of the breast is imagable. Surprisingly, the unloaded Q-factor of the probe rose with the shielding in place to 500, presumably due to inhibition of radiation. However, placement of a contiguous, vertical, aluminium foil cylindrical shield round the probe decreased the Q factor to 160 and it was therefore not added. The Q-factors with the two male volunteers now rose to 134 and 205, indicating the efficacy of the shield, while for a female volunteer Q rose from 44 to 80. Frequency changes due to breast insertion were minimal.

## Results and Conclusion

Images were made with both the multi-slice, multi-echo and FLASH techniques and displayed excellent S/N in accord with that expected from the probe's superior Q-factor. The measured image uniformity was consistent with the bench measurements of the $B_1$ field. In particular, a region up to 1.5 cm above the top shield was visible for the width of the probe. No image artefacts due to gradient-induced eddy currents in the foil shields were seen. These results demonstrate that it is possible, with the aid of r.f. shields, to view the chest wall adjacent to the breast while minimising losses due to adjacent conductive tissue.

## References

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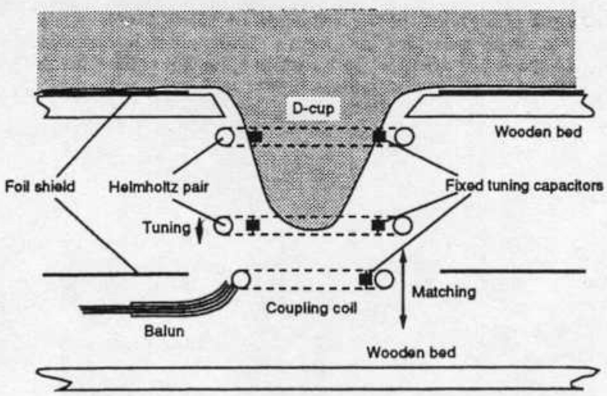


**The IBD / U. Manitoba Breast Coil Components and Shields**

Chen, X., R. Gordon, D. Hoult & B. Tomasek (1997). A shielded MRI breast-coaxial coil. In: R.I. Grossman & et al. (eds.), Proceedings of the International Society for Magnetic Resonance in Medicine, 5th Scientific Meeting and Exhibition, Vancouver, B.C., Canada, April 12-18, Berkeley: International Society for Magnetic Resonance in Medicine, 3, p. 1531.