Comparison of Microstrips and Surface Coils at 11T as Building Blocks for Phased Array Surface Coils

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Introduction

The incorporation of phased array technology into small animal MRI systems has lagged behind that in the clinical MRI world. This is partially because the multichannel hardware was not incorporated into the standard research system until the late 1990s, whereas, this capability appeared in most clinical imagers in the early 1990s. In addition, the availability of phased array coils for these same small animal systems is limited. For researchers with nonstandard systems, i.e. a very high field with large bore, there are no phased array coils available. But the promise of gains in Signal-to-Noise offered by phased array technology has pushed researchers to continue the development of phased array coils for high field systems, despite the greater complexity, coupling, and general lack of excitation coils. Microstrip construction techniques have been offered as a way to reduce coupling between coil elements in arrays [1-5]. In this abstract, we compare a shunt terminated loop microstrip [4], a tunable loop microstrip [5], and an unshielded surface coil, as well as, microstrips on different circuit board substrates. The implementations were compared as single elements and arrays, and evaluated by their performance in the MRI system and on the bench.

Methods

We built three coil designs on a microwave substrate (Rogers Ultralam 2000), as shown in Fig. 1; shunt terminated microstrip (hereafter called the Rogers microstrip loop), tunable loop microstrip (hereafter called the shielded loop), and an unshielded loop. In addition, we duplicated the Rogers microstrip loop on FR4, a common epoxy laminate. The substrate thickness was 0.060", top conductor width 0.125", and bottom conductor width 0.375". The loops measured 1.2" (3cm) on each side. All loops were tuned and matched to 50 Ω at 470 MHz. Single loops were place on a phantom filled with a tissue equivalent solution ($\varepsilon = 48.6$, $\sigma = 0.6$ S/m @470 MHz). A magnetic field probe embedded in the phantom measured the field 2 cm from the coil. Images were acquired (field probe not present), from which depth penetration and SNR were evaluated. We then evaluated each design by placing two loops side-by-side. Isolation between the loops was measured. Images were acquired (driving one loop, then the other) on an 11.1T Magnex magnet with a 40 cm clear bore and Bruker Biospec console, and the images summed. Note, the coils were plugged into standard 50 Ω input impedance preamps. Finally, the Rogers microstrip was modified by trimming back the substrate on the edge of the loop that was placed adjacent to the other loop, and adding a vertical shield (Fig. 2). This effectively reduced the spacing between the two microstrip loops, providing better signal coverage. In addition, a decoupling capacitor was added between the two microstrip loops. Isolation measurements and imaging were repeated.

Results

Table 1 shows bench and magnet measurements for the single coils. It is seen that the Rogers microstrip loop and shielded loop have similar Q, SNR, and field measurements. The FR4 microstrip loop and the unshielded loop have lower Q, SNR, and field measurements. The field plot in Fig. 3 shows the signal intensity down the axis of the coils. The plot indicates that the Rogers microstrip and shielded loop have the highest signal intensity along the entire plot, whereas the FR4 microstrip and the unshielded loop have the lowest signal intensity. The last column of Table 1 shows the isolation measurements of the side-by-side coil setup. The isolation measurements are similar for all designs. Images of the two coil setup for the Rogers microstrip indicated little shared signal between the coils. Fig. 4 is a summed image of the Rogers microstrips, indicating a signal void between the coils caused by the wide width of the shield trace on the bottom of the microstrip. Fig. 5 is a summed image of the modified mictrostrip, showing improved signal intensity between the coils because the width of the shield trace on the bottom has been reduced and the coils can be placed closer together. The isolation between the modified microstrips was -25 dB, a significant improvement.

Conclusion

The Rogers microstrip loop and shielded loop show superior performance (higher field intensity and SNR) to unshielded loop and FR4 microstrip loop at 470 MHz. Their geometric isolation allows use of standard 50 Ω input impedance preamplifiers in array designs. However, the traditional implementation lends itself to a signal void between the two loops of a small array. This problem can be overcome by a modification of the microstrip (trimming, shielding, feedback component) that improves the isolation and signal intensity between the coils. The improved isolation and ability to use 50 Ω input preamps greatly simplifies the construction of the loops in the array.

Loop	Q	SNR	Field (dB)	Isolation (dB)
Rogers Microstrip	33	19.2	-18.0	-10.5
Shielded	37	18.5	-17.1	-12.5
Unshielded	14	13.4	-22.5	-10.5
FR4 Microstrip	20	14.9	-22.0	-14.3

Table 1 Bench and magnet measurements

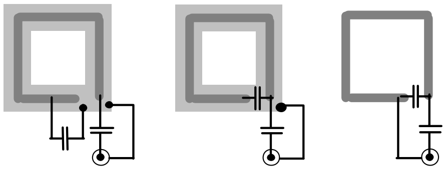

Fig. 1 Microstrip, shielded and unshielded loops

Fig. 2 Modified strip

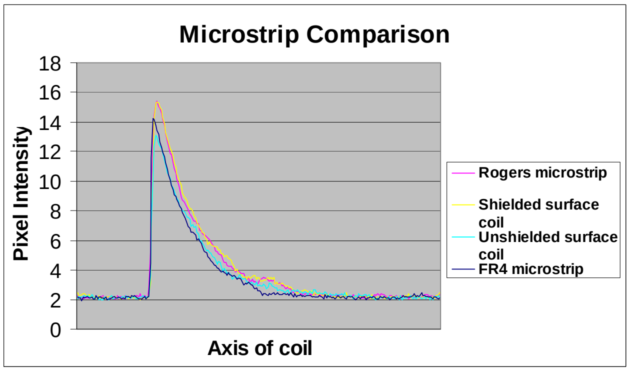

Fig. 3 Signal intensity down axis of coils

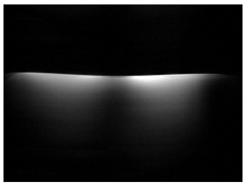

Fig. 4 Microstrip

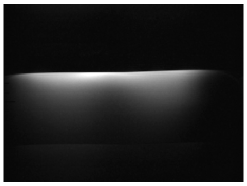

Fig. 5 Modified microstrip

References

[1] Lee, RF et al., Magn Reson Med 45, 673 (2001)
[2] Zhang X et al., Magn Reson Med 46, 443 (2001)
[3] Adriany G et al., Proc ISMRM 11, 474 (2003)
[4] Wichmann T et al., Proc ISMRM 12, 1578 (2004)
[5] Wu B et al., Proc ISMRM 12, 1576 (2004)

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