# Volume Microstrip RF Coil for MRI microscopy

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**Introduction:**
Standard microstrip coils are made of PCB (Fig. 1). Electrical current flowing thought the strip creates the magnetic field in the coil vicinity that is used for MRI [1] and MRS [2,3]. However RF field produced by the coil is very inhomogeneous and restricted to small FOV. This construction does not allow any quantification measurements, such as T1 or T2. Therefore we designed and manufactured a **volume micro strip coil**, that is based on a double microstrip concept.

**Electromagnetic field simulation and SNR calculation.**
Numerical simulations of the microstrip structure were performed using AC/DC and RF modules of the Comsol Multiphysics Finite Element Method (FEM) package (Comsol AB, Sweden). To confirm the performance of the calculations first we considered a single microstrip design (Fig.1). Full 3D quasi static, time harmonic simulations at 500MHz showed, as expected, a decrease of the RF field amplitude with the distance from the strip.
To overcome the problem of inhomogeneous RF field we introduced an additional, parallel microstrip (Fig.2) carrying RF current in opposite direction. Numerical simulations confirmed, that such a double microstrip structure generates homogenous RF field in the volume between the strips.
The simulation software provided impedance of the coil, spatial distribution of $B_1$ field and current density which allowed calculation of intrinsic SNR. According to [4,5] signal received in MRI experiment after excitation with $\pi/2$ RF pulse, neglecting relaxation effects can be derived from following equation: $S\sim\sum_n\left|\sin\left(V\left|\hat{B}_1^+\right|\gamma\tau\right)\left|\hat{B}_1^{-*}\right|\right|$. Where: $\hat{B}_1^+=\frac{1}{2}\left(\hat{B}_x+i\hat{B}_y\right), \hat{B}_1^-=\frac{1}{2}\left(\hat{B}_x-i\hat{B}_y\right)^*$ are rotating frame $B_1$ field components and $\hat{B}_1^+$ is the flip angle inducing component. $\hat{B}_x, \hat{B}_y$ are laboratory frame $B_1$ field components (complex values - phasor notation), $i=\sqrt{-1}$ and the asterisk denotes complex conjugate. $\gamma$ is the gyromagnetic ratio of ${}^1H$, $\tau$ is the duration of the rectangular RF pulse. The summation is conducted over all voxels in ROI. The noise detected by the receiver in case of the microcoil is dominated by the coil resistance [6]: $N\sim\sqrt{R_c}$, yielding: $SNR\sim\frac{S}{\sqrt{R_c}}$.
To compare the single and the double microstrip configuration, signal intensity was calculated along z axis (parallel to $B_o$ in Fig 1, 2) passing through the centre of the coils. The result (Fig. 3) showed that simulated MR signal intensity is stronger and more homogenous for a double microstrip (blue line) than for a single strip (green line).

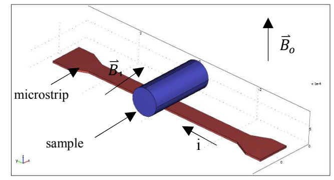

Fig.1. Single microstrip geometry. $B_1$ RF field is created by current i passing through the strip.

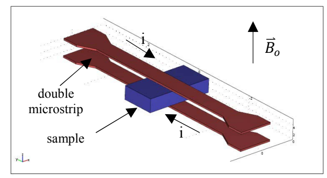

Fig.2. Double microstrip geometry. Current flows in opposite directions creating uniform $B_1$ RF field in the volume between strips.

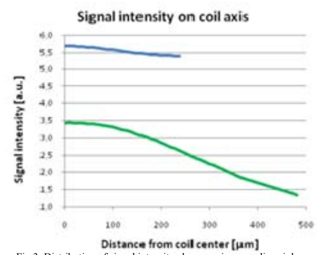

Fig.3. Distribution of signal intensity along z axis, green line sigle microstrip, blue line double microstrip

**Coil design and construction**
To optimize the microcoil performance SNR was calculated for several double microstrip configurations. For example the simulations showed that a coil with both microstrips of 500 µm width and 250 µm gap produced 30% higher SNR in the centre than 500 µm gap coil. The impedance was similar for both configurations. Fig. 4 shows the plots of simulated MR signal intensity along *z* axis, revealing better performance of the first microcoil.
To test the coil performance a double microstrip microcoil was cut out of two pieces of PCB (Injectorall, USA), 1.6 mm thick FR-4 substrate, 70µm copper layer with a digital milling machine (LPKF, Germany). The microstrips were 500 µm wide, 5 mm long with 1 mm gap, mounted with washers to the probe enclosure. The coil was tuned to 500 MHz and matched to 50 Ohm. A prototype of the double microstrip is shown in Fig. 5. The 1 mm gap between PCB boards allowed easy insertion of different phantoms and samples into the coil. The Q parameter measured on-the-bench was 120 with agreement to numerical simulations.

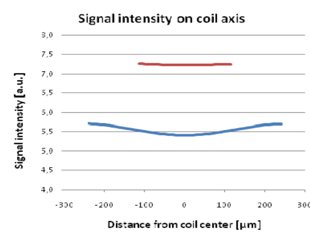

Fig.4. Distribution of signal intensity along z axis for double microstrip coils having different gap between strips: blue 500 µm, red 250 µm.

**Results and Discussion**
For MR imaging experiments, the double microstrip coil was positioned inside the 11.7T vertical bore magnet (Oxford Instr. UK) equipped with a commercial 72mm ID gradient set (maximum 600 mT/m) and Avance console (Bruker, Germany). Several MR images were acquired proving the high performance of the double microstrip microcoil for MRI microscopy.
An image of a phantom made of parallel micro capillaries (200 µm outer diameter, 160 µm inner diameter) filled with doped water is shown in Fig. 6. The image was acquired with MSME sequence: TE/TR 6.4/1000ms, NEX 16, FOV 3mm, MTX 128x128. The image resolution of 24x24x300 µm was limited by the performance of the gradient coils.

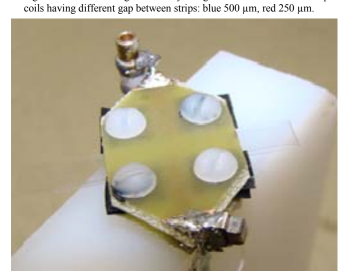
Fig.5. Double microstrip setup mouned on probehead with sample inserted.

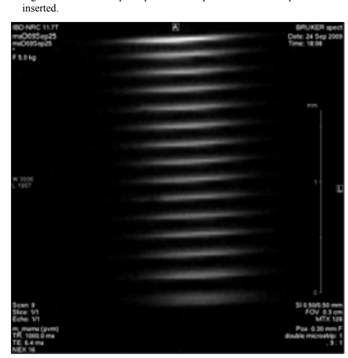
Fig.6. Image of micro capillary phantom acquired with double microstrip microcoil.

**Conclusion**
We demonstrated that the double microstip volume coil can be used for MRI microscopy. The proposed construction is cost effective, fast and efficient. The results obtained with MRI agree with theoretical calculations and FEM simulations.

**References**
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