Three concepts for tuning and matching intravascular catheter coils

C. Pitsaer[1], R. Umathum[1], A-K. Homagk[1], C. Ozturk[2], and M. Bock[1]
[1]Medical Physics in Radiology, German Cancer Research Center, Heidelberg, Germany, [2]Institute of Biomedical Engineering, Bogazici University, Istanbul, Turkey

Introduction

Over the past years, intravascular MR imaging has been demonstrated using different coil configurations. Intravascular coils provide a local SNR gain which is e.g. used to image the different layers of the vessel wall with spatial resolutions of the order of several 100 µm [1,2]. In almost all of these studies, impedance matching of the coil and signal amplification was performed at the proximal end of the imaging catheters due to the limited space at the catheter tip, even though a local tuning and matching as well as a preamplification would be favourable. This study presents a comparison of three coil concepts for intravascular imaging: (1) the conventional method with matching and preamplifying at the proximal catheter end, (2) matching at the distal end of the catheter with a proximal amplifier, and (3) an amplified signal at the tip of the catheter (Fig. 1). The gain of SNR and high resolution imaging are demonstrated.

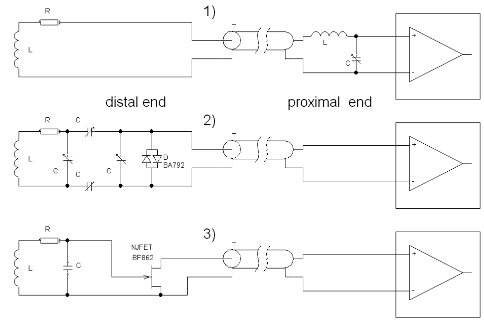

Fig. 1: The three coil concepts.

Materials and Methods

For the experiments a catheter coil was built that consisting of a 17 µm thick copper layer loop on a 50 µm thick polyimide support foil (Fig. 2). The coil is 8mm wide and 15mm long. It is connected to a 50 Ω micro-coaxial cable (Ø: 300 µm, length: 120 cm) via two conductors of 3 cm length. The large separation between coil and coaxial cable was chosen to minimize image artefacts from magnetic components of the preamplifier. The whole design fits in a 9 French tube with the coil being rolled up when retracted into the tube [3].

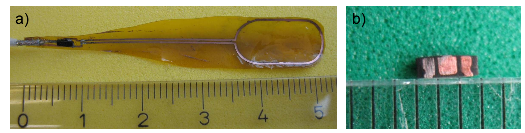

Fig. 2: a) 15 mm-long foil coil with the pre-amplifying transistor and a 100 pF capacitor at 25mm distance from the coil. b) Enlarged view of the commercial 3mm-long, 1mm-wide and 0.8mm flat filed transistor.

For the conventional design (coil 1), matching was realised at the proximal end of the catheter with a trimmer capacitor (Murata, Japan) and an inductor (Ø 2mm, 14 turns). Remote matching at the distal end (coil 2) was achieved with 4 trimmer capacitances to ensure a balanced network. Two crossed diodes (BA792 Philips) were added to the matching circuit to attenuate the possible B_1-field distortion during RF excitation. For both coil 1 and coil 2 signal amplification was realised at the proximal catheter end using a preamplifier (BGA 16, Infineon, gain: 19 dB). For amplification at the catheter tip (coil 3) an N-channel Junction FET transistor (NXP, BF862,) was soldered to the tip of the coil with a 100pF capacitor (size 0201) to create resonance at 63.7 MHz (Fig 1). At the proximal end of the coaxial cable a second transistor with tuning components was used to complete the preamplifier (gain: 20 dB). The plastic transistor package was partially ground off (filed) and pins removed to decrease its size and to reduce image distortions from magnetic components.

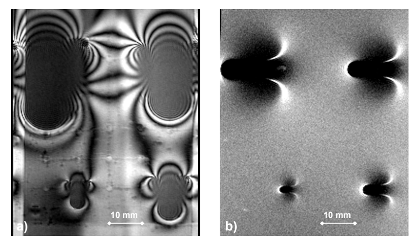

Fig. 3: Above artefacts for unfiled/filed transistors and below two chip capacitors in a) gradient echo and b) SE image

To compare the three coil concepts, phantom experiments were conducted on a clinical 1.5T whole body MR system (Symphony, Siemens, Erlangen, Germany). The image artefact sizes of the magnetic components (transistor and capacitor) were measured with a spin echo (SE: TR = 500 ms, TE = 12 ms, matrix=256×256, BW = 130 Hz/pixel) and a gradient echo (FLASH: TR = 26.1 ms, TE = 8.14 ms, α=20°, matrix=192×256, BW = 40 Hz/pixel) pulse sequence. For signal reception a spine array coil was used. To assess possible B_1-field distortions near the coil due to resonant coupling during RF excitation, a series of FLASH images with varying flip angle (α = 1°..25°) was acquired in aqueous solution (H_2O:GdDTPA 1:1000, T_1=340 ms) (TR=5 ms, TE=1.9 ms, matrix = 64×128, BW = 400 Hz/pixel). From the image series the apparent Ernst angle was calculated for a position close to the coil, where B_1-distortion is expected, and for a distal region in the solution, where B_1-distortion by the coil is not present. For all coil concepts sensitivity profiles were measured, and high resolution images were acquired in a vessel phantom. All coils were coated with varnish for isolation.

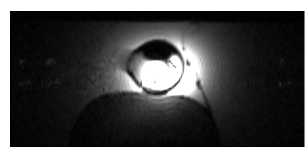

Fig. 4: Transversal cut of a lung aorta phantom with a kiwi fruit underneath.

Results and Discussion

Figure 2 shows the image artefacts due to the presence of the magnetic transistor and capacitor. The largest artefact size (25 mm) is seen in the FLASH image at the transistor before filing. The filed transistor was then placed about 30 mm away from the coil centre to avoid interference with the signal. In all coil designs B_1 was reduced close to the coil, and the largest reduction of 16% was seen with the third coil concept (Tab. 1). However, distal matching increased the SNR by a factor of 2.4, and distal preamplification even led to an increase in SNR by a factor of 3. Despite possible limitations due to space restrictions and artefacts from magnetic components, these preliminary results show that a significant SNR gain can be achieved with local matching and preamplification at the tip of a catheter.

	Solution	Coil 1	Coil 2	Coil 3
F	1.004	0.94	0.88	0.84
α_{Ernst}(°)	9.8	10.4	11	11.6
SNR		97	237	287

Table 1: B_1-field reduction factors F, resulting Ernst angles and SNR.

References
[1] Martin, A.J., et al. J Magn Reson Imaging, 1998. **8**(1): p. 226-34.
[2] Atalar, E. Magn Reson Med, 1996. **36**(4): p. 596-605.
[3] Homagk AK et al. 17th ISMRM 2009, 2566.