Tuning and Amplification Strategies for Intravascular Imaging Coils

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Introduction and Motivation:

There is a clear benefit in using intravascular imaging coils over larger surface coils for acquiring high-resolution, small-field-of-view images of the vasculature. Several intravascular coil designs have been proposed for applications that include both vessel wall imaging and guiding the revascularization of chronic total occlusions in arteries [1,2]. The manufacturing of such coils poses several challenges. Due to the size of such devices (typically < 2mm in diameter), it can be very difficult to incorporate local matching networks and signal amplifiers. The challenges in doing so increases drastically as the complexity of the imaging device increases (such as with the inclusion of a catheter lumen). Typical approaches to intravascular coil manufacturing have been to connect the imaging coils with a matching network located at the proximal end of the catheter to a length of thin micro-coaxial cable that runs through the walls or lumen of the catheter. This approach, although convenient, severely degrades the signal as a result of resistive noise associated with the thin transmission line.

The purpose of this study is to investigate tuning and amplification strategies for intravascular coils and to assess the signal-to-noise benefits of incorporating a matching network and/or miniature amplifier into catheter-based intravascular imaging devices at various locations in the signal chain.

Materials and Methods:

In order to investigate the effect on SNR of component placement along the signal chain, good comparability between test cases needed to be maintained. Therefore individual blocks were designed to be modular so that they could easily be re-positioned, as seen in Figure 1. All measurements were taken within a short time interval to avoid variations of the MR system. Since all experiments were to take place in an MR environment, all blocks were designed with MR-compatible electronics.

Our test setup was designed for a 3T MRI. A ~2-mm diameter 5-turn microcoil (μ-Coil) was constructed out of 24AWG magnet wire and wound around a 6Fr catheter section. The loaded inherent resistance of the coil was measured to be approximately 9Ω. A low-noise amplifier (LNA) was designed with a small-scale surface mount NPN BJT with a noise figure of approximately 2dB and a gain of approximately 20dB. Three different types of coaxial cables were used: a RG58 cable with negligible resistance was used to acquire our baseline SNR measurements; a 400-um diameter 1.5m length of cable with an attenuation of 1dB/m and a resistance of 3Ω/m; and a 150-um diameter 1.5m length of coaxial cable with an attenuation of 3dB/m and a resistance of 18Ω/m.

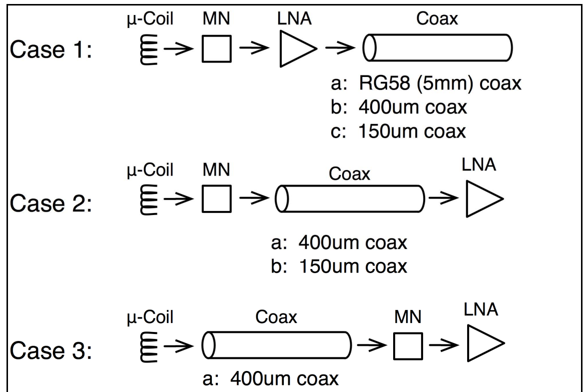

Fig 1. Schematic diagrams of test configurations used for experimental setup. Each configuration consisted of a microcoil (μ-Coil), a matching network (MN), a low-noise amplifier (LNA), and a length of coaxial cable.

MR images were collected on a gel phantom using an SPGR pulse sequence (FOV=8cm, FA=30, TR=50ms, TE=7.9ms, 192x192 matrix, slice thickness=1mm). The transmit and receive gains were kept constant throughout. Slice prescription was done by selecting a 1-mm thick slice beyond the tip of the μ-Coil that did not exhibit any signal nulls. By doing so, slice prescription between test cases would be positioned within +/-0.5mm of each other. To average out this error, three independent SNR measurements were taken from three independently acquired prescriptions. Finally, SNR measurements were normalized to the mean of our baseline SNR measurement (Case 1a) and compared against calculated theoretical estimates.

Results and Discussion:

The results shown in Figure 2 suggest good agreement between empirical data and theoretical expected values. The SNR shows good robustness to noise when the LNA is placed in front of a lossy coax cable. By designing the amplifier for minimum noise figure and high gain, we are able to decrease the overall noise figure of the system. The noise contribution of the coax cable is effectively reduced by the gain of the amplifier leading to a negligible hit in SNR, as seen in test configuration 1b and 1c. In test configuration 2a and 2b, the coaxial cable is placed in front of the LNA and as a result the noise in the system is dominated by the noise contribution of the cable. The most likely source of discrepancy to the expected values can be attributed to assumptions made for the noise characteristics of the transmission line.

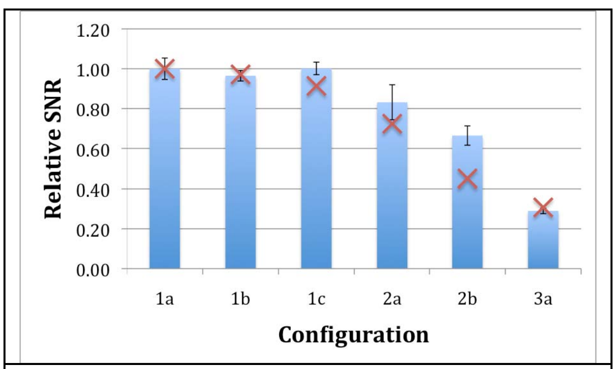

Figure 2. Relative SNR measurements with mean and standard deviation over three independent measurements normalized to the mean of the 1a test configuration. Theoretical expected values are shown overlaid in red onto experimental results.

Test case 3a represents the most practical configuration for intravascular microcoils, where both the matching network and amplifier would be packaged outside the catheter. However, the SNR suffers greatly as a result of having the greatest sources of noise at the input of the signal chain. The results presented here suggest that a LNA designed with a low-noise figure placed close to the μ-Coil would allow miniature coax cables to be used despite being high loss. Otherwise, if a LNA isn't placed close to the μ-Coil, the choice of coax is limited to low loss ones with diameters that would not be practical for an intravascular application.

References: [1] Hillenbrand et al. MRM (2004) vol. 51 (4) pp. 668-75 [2] Anderson et al. MRM (2008) vol. 60 (2) pp. 489-95