Transmit Power Optimization for Tracking, Wireless Marker and Imaging applications of a Multi-mode Endovascular coil.

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Introduction:
In previous work, we introduced the concept of the multi-mode intravascular MRI coil [1,2] which consists of a single active device that is connected to the external system via a single coaxial cable and performs three functions: 1) active tip tracking, 2) imaging, and 3) inductively-coupled wireless marker. The purpose of such a multi-functional coil is to enable single insertion; wholly MRI-guided minimally invasive procedures. As shown in figure 1, the multi-mode coil consists of a tightly-wound tracking solenoid with a highly localized sensitivity pattern that is connected in series with an elongated imaging loop that has a broader sensitivity pattern. Furthermore, the multi-mode coil is inductively coupled to the external transmit coil in order to enable the wireless marker functionality as described in [3]. This is achieved by transforming the open circuit at the decoupling circuitry to a low impedance at the coil terminals via a (2n-1)λ/4 transmission line, thereby allowing current induced by the transmit field to flow in the multi-mode coil. This results in the magnification of the transmit B_1 field within the region of sensitivity of the multi-mode coil, leading to larger flip angles compared to regions outside the coil sensitivity. In addition, the magnification of the B_1 field due to the tracking component of the multi-mode coil is greater than that due to the imaging component due to the greater conductor (and hence, current) density of the tracking solenoid. The dominant tracking peak may be used to localize the position of the catheter tip using 3 orthogonal projections. The objective of this work is to determine the optimal transmit power for each mode of operation of the multi-mode endovascular coil.

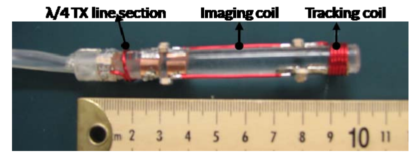

FIG.1 Implementation of an enlarged model of the multi-mode coil.

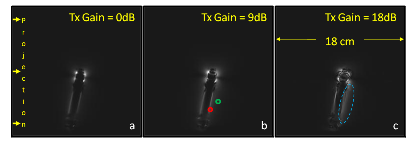

FIG.2 Multi-mode coil images for transmit gains corresponding to 0 dB (a), 9 dB (b) and 18 dB (c).

Methods:
An enlarged model of the multi-mode coil (figure 1) was constructed on a rigid acrylic rod (3.175mm outer diameter) using 22 gauge magnet wire. The tracking component of the multi-mode coil consisted of a 5-turn, tightly wound solenoid placed at the distal end of the catheter. The single-turn imaging loop was 40mm long and spanned the diameter of the acrylic rod.

Transmit power calibration was performed with the above multi-mode coil placed in a basinet filled with tap water and doped with 0.3% copper sulphate. The T1 of the above solution was measured to be about 150ms using a look-locker sequence. The phantom was devoid of features in order to enable accurate measurements of projections and signal intensities. In the tip tracking and imaging modes, the multi-mode coil was connected to the single channel receive only port of a 1.5T GE Signa MRI scanner via the decoupling circuit. The body coil was used exclusively for B_1 field transmission. For the wireless marker experiment, the body coil was used in the transmit-receive mode.

A standard fast gradient recalled echo (FGRE) pulse sequence (TR/TE = 200/1.2 ms) was used for the tracking, imaging, and wireless marker experiments. With the flip angle set to 90°, the transmit gain was varied between 0 and 20 dB in steps of 1 dB. Images were acquired with both the multi-mode and body coils for each transmit gain step. The FOV's for the multi-mode coil and body coil images were 18 cm and 48 cm respectively.

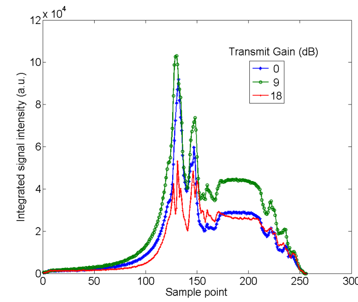

FIG.3 Signal intensity projections of the multi-mode coil images in figure 2 along the direction indicated in figure 2a.

Results and Discussion:
Phantom images obtained with the multi-mode coil corresponding to transmitter gains of 0 dB, 9 dB and 18 dB are shown in figure 2. These transmit gains correspond to transmit powers of 160 W, 1.3 kW and 10 kW respectively. Signal intensity projections of the above three images are calculated along the direction indicated in figure 2a and plotted as shown in figure 3. The plots indicate a dominant tracking peak which can be spatially localized with a minimum of three orthogonal projections. Comparing the signal intensity plots corresponding to transmit gains of 0 dB and 18 dB, one can note the marked reduction in tracking peak amplitude in the latter case although the signal intensity projections of the imaging coil are approximately equal, indicating that lower transmit power is sufficient to robustly track the tip of the catheter. Additionally, the results seem to indicate that the multi-mode coil could be designed to simultaneously extract good imaging and robust tip tracking performance.

Figure 4 is a plot of the mean signal intensity in the regions of interest indicated in figure 2b. The ROI's are placed adjacent to the multi-mode coil conductor (red) and 1 cm away from it (green). These two particular ROI's are chosen to demonstrate how the transmit power may be adjusted to equalize transverse magnetization to compensate for the sharp drop in coil sensitivity and hence, increase the area of coverage of the imaging component of the multi-mode coil. The plot shows that for transmit gains between 15 dB and 19 dB, the signal intensities are equalized. This is confirmed by a visual examination of the region indicated in figure 2c.

Finally, phantom images obtained with the body coil in transmit-receive mode, corresponding to transmitter gains of 0 dB, 9 dB and 18 dB are shown in figure 5. As indicated in figures 5a and 5b, the bright signal adjacent to the conductors of the multi-mode coil disappears with increased transmit power. For large transmit power, banding due to overflipping is evident in figure 5c.

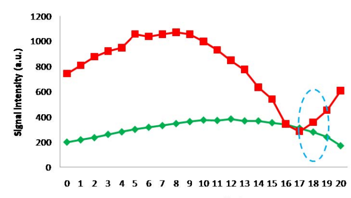

FIG.4 Plot of the mean signal intensities in the ROI's indicated in figure 2b. The blue oval indicates the transmit gain for which imaging coverage is extended and corresponds to the results indicated in figure 2c.

Conclusions:
In conclusion, our study shows that the transmit power in the external transmit coil may be adjusted to obtain optimal performance of the multi-mode coil in all three of its functional modes. In addition, the transmit power may be optimized to extend the useful region of coverage of the imaging component of the multi-mode coil. The fact that the multi-mode coil is coupled to the transmit coil may raise legitimate concerns about RF safety. A preliminary numerical study of the RF heating mechanisms of the multi-mode coil is presented in an accompanying abstract [4] and seems to indicate that the multi-mode coil may be successfully designed under the additional constraint of RF safety.

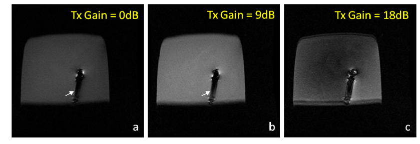

FIG.5 Body coil images for transmit gains corresponding to 0 dB, 9 dB and 18 dB for wireless marker application.

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References:
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