INVESTIGATION OF DETECTOR COLLIMATOR EFFECTS FOR DUAL MODALITY MR -NUCLEAR IMAGING

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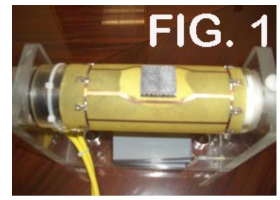
Purpose:

Most of the nuclear imaging systems, especially those that are based on single photon emission detection, require the use of a collimator- usually made out of lead (Pb) or some other high Z material such as tungsten (W). Even some PET systems employ some sort of slat-septa collimation, although the important part of PET image formation is done electronically via coincidence detection. Multi-modality imaging combines two or more complementary imaging systems to provide spatially and temporally co-registered multidimensional images by the chosen modalities. The purpose of this study was to investigate the affect of Pb collimators for a hybrid system that combines single photon scintigraphic imaging with MRI. We have tested a parallel-hole collimator in combination with a shielding plate both made of lead inside a 4T MR system to evaluate their effect on the MR image quality. Although the current common belief for the choice of such MR-compatible shielding and collimator materials is to use a high-Z, non-electrically conductive material, we tested a Pb parallel-hole collimator since it is a cost-effective and proven material in nuclear imaging.

Methods:

The effect of Pb collimator on MR images was studied by designing and constructing a true-multimodality RF coil specifically developed for this purpose. The coil design is based on a birdcage configuration where the copper rungs of the birdcage are displaced to allow for the integration of a nuclear collimator right into the RF coil [1]. A picture of such a coil designed for a parallel-hole collimator is shown in Fig. 1. This is an 8-section birdcage RF coil where the separation between the two rungs are opened up to allow a 2.54 mm x 2.54 mm parallel-hole collimator to be inserted through. In other types of applications where pinhole collimators may be used, a hole of appropriate size may be cut out for the pinhole. This specific RF coil was designed to work with a single CZT detector (16x16 array – 1.6mm pixel size) and had an inside diameter of 30 mm and a wall thickness of 7 mm. It should be emphasized that this fully integrated design allows for bringing the collimator as close to the object as desired to improve spatial resolution without worrying about RF coil attenuation of gamma rays. A cylindrical phantom with a 24mm diameter was filled with $CuSO_4$ and placed inside the RF coil with and without the Pb collimator. The pulse sequence parameters were: T1 Weighted Spin Echo, TR/TE=500/20ms, matrix = 128*128, FOV = 40*40mm, thickness=4.0mm, NEX=2, BW = 33.3 kHz. The center frequency of the RF coil and its Q-factor were measured under different loading conditions. Additionally MR images were acquired under different conditions to evaluate the MR performance in terms of SNR as well as signal uniformity changes resulting from the presence of Pb collimator. MR images without the Pb collimator were acquired as a reference set then the Pb collimator was inserted into the RF coil as shown in Fig. 2. In order to improve the signal uniformity in the images we also placed small Pb sheets on the other 3 sides to balance the Pb collimator as shown in Fig. 3 and repeated the measurements.



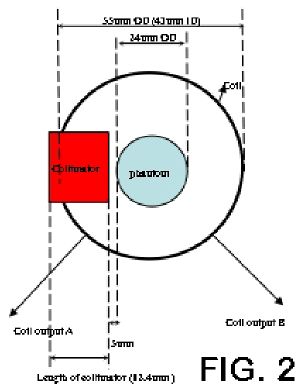


FIG. 2

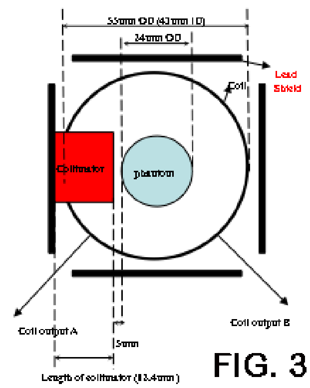


FIG. 3

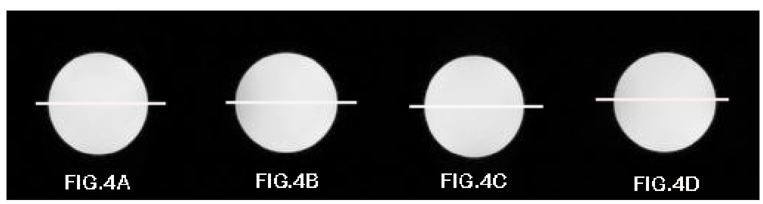
Results:

The center frequency of the RF coil and its Q-value were measured without and with the collimator/detector assembly inserted into the RF coil. The results are given in Table 1. The first column in Table 1 corresponds to the 4 cases shown in Fig. 4.

	f 0 (MHz)	Q-value
NO SHIELD (Fig. 2)		
4.A-Reference (no coll.)	170.288	196.9
4.B-With Pb collimator	171.120	140.2
WITH SHIELD (Fig. 3)		
4.C-Reference (no coll.)	170.295	153.6
4.D-With Pb collimator	170.579	149.8

Table 1

We also analyzed the MR images obtained under these conditions to assess the signal non-uniformity arising from the presence of the Pb collimator and additional shielding near the phantom. MR images corresponding to the 4 cases shown in Table 1 are shown in Fig. 4 with a horizontal profile drawn to compute the signal drop off due to the Pb collimator that was on the left-hand side of Figs. 4B and 4D. The signal non-uniformity was calculated as the % difference in the signal between the sides close to the collimator vs. farthest from it. Table 2 summarizes the signal non-uniformity and the image SNR for the four cases presented here.



	SNR	Non-uniformity (%)
NO SHIELD (Fig. 2)		
4.A-Reference (no coll.)	442.7	2
4.B-With Pb collimator	389.3	13
WITH SHIELD (Fig. 3)		
4.C-Reference (no coll.)	378.1	2
4.D-With Pb collimator	351.4	6

Table 2

Discussion:

As expected, the introduction of a Pb collimator near or inside the RF coil causes a shift in the center frequency and a drop in the Q-factor. Consequently this results in various degrees of SNR changes as a Pb collimator and additional Pb shield are introduced. The SNR reduction is due to the RF absorption by the collimator and other surrounding metal. It should be noted that the addition of Pb shields as shown in Fig. 3 improves the image signal non-uniformity considerably even in the presence of a Pb collimator 5mm away from the object. Although the introduction of the Pb shields to improve the uniformity results in the degradation of the image SNR (see Table 2) since the time it takes to acquire nuclear images is much longer than for MRI data collection one can easily increase the number of averages in the MRI to overcome this problem. In conclusion our results demonstrate that lead (Pb) collimators could be used in MR -nuclear dual modality imaging systems with appropriate precautions such as balancing the collimator material with other shielding made of the same material as the collimator.

References: [1] O. Nalcioglu, W. W. Roeck, and S.-H. Ha, "Multi-modality RF Coil," U. S. Patent Office (pending): The Regents of the University of California, 2008.

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