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A Proposed Design of Collimator and Crystal for Single Photon Detection in Breast Cancer Imaging

Kwang Hyun Kim[1),4)], Gyuseong Cho[1)], Wan No Lee[1)], Yong Hyun Chung[1)], and Young Soo Kim[1)]
Yong Choi [2)]
Soongpyung Kim[3)]
[1)]Korea Advanced Institute of Science and Technology 373-1 Kusong-dong, Yusong-gu, Taejon, Korea
[2)]Samsung Medical Center, 50 Ilwon-dong, Kangnam-gu, Seoul, Korea
[3)]Chosun University, Kwnagju 501-759 Korea
[4)] Hyun Dae Nuclear Co., Ltd 893-1 Bongchun-dong, Kwanak-gu, Seoul, 893-18 Korea

Abstract

Monte Carlo simulation of collimator and crystal has performed for the design of discrete gamma camera for breast tumor imaging. The purpose of simulation is to induce specific design parameters for collimator, crystal and optical bonding material between collimator and crystal. From the induced parameters we can estimate the performance of the detector of the gamma camera and suggest its accuracy for practical use in the field. The geometry of the collimator should be matched to the crystal geometry in case of discrete gamma camera. The performance of the collimator is a main factor in determining the overall imaging performance For design and simulation, collimator hole of 3mm×3mm, 0.25mm septal thickness of square type Tungsten collimator corresponding to the pixilated photosensor, and a fixed location of breast tumor 25mm from the collimator surface in the phantom have been fixed in this study. For more detail drive of design parameters we varied collimator height, crystal height, crystal surface treatment, and bonding material between crystal and photosensor. We also analyzed geometric efficiency and spatial resolution compared to general theoretical formulation, and indicated their accuracy for realistic several tumor sizes.

I. Introduction

Functional scintimammography of breast cancer using ^{99m}Tc MIBI and gamma camera of the discrete scintillator/photodiode structure has been reported recently (Figure.1). [1] This technique is able to detect cancer with more than 90% specificity [2] while the specificity of X-ray mammography is much low. The gamma camera of the discrete scintillator/photodiode consists of a collimator, crystals, photodiode, and electronic circuits.

Firstly, the segmented collimator passes only the parallel gamma-ray from the sources in the object to the detector surface. And it is made of lead or tungsten depending on the gamma-ray energy. Its height and hole shape, for example round, hexagonal, and square type is chosen according to the application purpose. Secondly, the crystal which converts single gamma-ray into light is typically CsI(Tl) for 140keV gamma-ray from ^{99m}Tc. Thirdly, the photodiode that convert light generated in the crystal into charge carries are generally solid-state detector such as Si pin photodiode, HgI_2, and etc. Finally, the electronic circuits handle electronic signal from photodiode and produce image data using Winner Take All (WTA) algorithm [1] to search the position in the detector.

In this paper we presented Monte Carlo simulation results of collimators and crystals to suggest acceptable parameter value which can optimize the camera head design even for the worst condition of tumor position in breast. Additionally, we analyzed the relation between the geometric efficiency and the spatial resolution of collimator. We calculated true counts for various tumor size and collimator heights, and compared with the estimated background counts.

II. Rationale

The specificity of a scintimammography system is governed by the spatial resolution and geometric efficiency of the collimator. Generally, the total response of a collimator has the geometric, the penetration, and the scatter components [3]

The geometric component means the passing probability of photons through the collimator hole properly without experiencing any interaction within the patient or the septal material. It is the major contribution to the total counts of the detector and has the most influence on the geometric efficiency of the collimator, spatial resolution, and field of view. The penetration component through the septa is negligible for low energy photons such as 140keV from ^{99m}Tc. The scatter component which is the number of photons that may experience Compton scatter interaction within the patient body and the septal material of the collimator before reaching the scintillation crystal causes both the poor energy resolution and the high background count.

III. Modeling Discussion

Three design parameters for simulation have been used to suggest optimal parameter values for application purpose in the breast tumor imaging as the follows; the collimator, the crystal, and the optical bonding material between the crystal and the photodiode.

A. Collimator design

The parallel-hole collimator is by far the most common type of collimator used clinically. In recent research,[1] the simulated 1 to 1 matched square hole collimator with the pixellated photodiode had demonstrated better spatial resolution than round or hexagonal hole collimator. The walls of the collimator hole, called septa are made of a material with high atomic number such as lead. However, we used Tungsten (W:Cu= 6:4) material instead of lead for future manufacture with ease to make square hole, and the septal thickness of the collimator was fixed at 0.25mm depending on the gap between the pixellated crystals and the photodiode. Also the hole size was 3mm×3mm with the same reason as mentioned above because of the difficulty of wiring at the end of the photodiode connecting the electric circuits. [1] The collimator heights varied range of 10mm to 60mm for the geometric efficiency in this study.

B. Crystal design

Among the various scintillation materials, CsI (Tl) was chosen because it has larger gamma ray absorption coefficient per unit size and high light yield (Photons/MeV). [4] This CsI (Tl) was taped with Teflon of 0.25mm thickness to prevent light loss in the crystal and cross talk from each pixellated crystal. Its pixel size also follows as collimator hole size. The crystal height varied range of 0.1mm to 10mm for the absorption energy in the crystal and the light transmission efficiency considering the spatial resolution (FWHM).

Because the condition of treatment for top and side surface of crystal effect on the light transmission efficiency, we treated top and side surface of crystal with Ground treatment (harsh surface), Polished (polish surface), Metal-0.95RC (metal coating with 95% reflection coefficient), Polished-0.98RC (polish surface with 98% reflection coefficient), and Painted-0.98RC (coloration surface with 98% reflection coefficient) respectively.

C. Optical bonding design

The optical bonding material between the crystal and the photodiode also influences on the light transmission into the photosensor. So as to find out high light transmission efficiency, the refractive index and thickness of the optical bonding material varied range of 1.6 to 2.0 and 0.1mm to 0.5mm respectively.

D. Source and background conditions

In view of sensitivity, the worst source or the tumor condition in the breast is at the middle of the breast volume with the lack of activity because that condition, far from the collimator surface, dose not give desirable quantity of gamma ray to detect and also can be interrupted by the unnecessary gamma ray called background from other organ such as heart, liver, and etc in the patient body. For that reason, in order to simulate and analyze the detector performance, we located the source at 25mm in the water phantom from the collimator surface. The background activity per volume in the overall body was assumed 22.5 counts per second in this paper. [5]

IV. Simulations and Results

For realistic simulation we divided top area of the collimator and the crystal into 11×11 pixels which has 1,278mm^2. And MCNP4B simulation code was used for the energy deposition in the crystal depending on the thickness of the crystal, the geometric efficiency and the spatial resolution of the collimator. DETECT97 also based on Monte Carlo simulation code was used for the light transmission efficiency for the crystal thickness and the surface treatment of the crystal.

A. Deposition energy

140keV gamma ray from ^{99m}Tc can penetrate the crystal depositing partial energy if the crystal thickness is not enough. For this reason, decision of the crystal thickness is important so that it is possible to maximize of light generation. So as to the decide deposition energy or the sensitivity depending on the crystal thickness we fixed the collimator height of 35mm and varied the crystal thickness raged of 0.1mm to 10mm. Figure (3) shows its results and displays that the crystal thickness saturated at 5 ~ 6mm and 8 ~ 9mm. Figure (4) shows that the light transmission was degrading as the crystal thickness increase. Therefore, we may chose thickness of 5 ~ 6mm considering trade-off between the energy deposition and its light transmission.

B. Surface treatments of crystal top and side

For good light transmission from the crystal into the photodiode, it is necessary to decide also crystal surface. With a fixed crystal height of 6mm, the results in figure (5) shows the side and the top surface of the crystal are desirable with Polished-RC0.98 and Painted-RC0.98, respectively resulting in over 94% efficiency.

C. Optical bonding

The optical bonding material between the bottom side of the crystal and the photodiode also was considered including the surface treatment of crystal bottom side. As shown in figure (6), the surface treatments of the crystal bottom with both the Polish and the Ground have no differences. For the refractive index of the bonding material, 1.8, the same as crystal’ s was the best resulting over 94% efficiency.

D. Geometry efficiency

The calculation using the theoretical formulation and the simulation for the geometric efficiency were done under the condition of the crystal thickness of 6mm, a planar sheet source, and various collimator heights. Generally, the theoretical formulation of geometric efficiency for parallel hole collimator with square hole is given as follows;

$$G = G_0\left(\frac{A_{open}}{A_{unit}}\right) = \frac{p}{2\sqrt{3}}\left[\frac{a^2}{l_e(a+s)}\right]^2\left[\frac{a}{a+s}\right]^2 , \tag{1}$$

And each parameter used in here displayed in the figure (2).

From this equation we can estimate and compare to the simulation result showing in the figure (7). From the figure (7), even if the tendency that increasing the collimator heights decreases their geometric efficiency was equal each other, there is a somewhat differences except short collimator height.

E. Spatial resolution

The spatial resolution for both the theoretical formulation and the simulation were done also with the same conditions used in the geometric efficiency. The spatial resolution for parallel hole collimator with square hole was given as follows;

$$d_{FWHM} = \left\{ \left[\frac{a(l+z+b)}{l} \right]^2 + d_i^{\,2} \right\}^{1/2} . \qquad (2)$$

where d_i^2, intrinsic spatial resolution, was assumed 3mm because of fixed crystal area of 3mm. The figure (8) shows the tendency also that increasing collimator heights increases their spatial resolution.

F. Trade-Off between geometric efficiency and spatial resolution

With the above results from the geometric efficiency and the spatial resolution, both are conflicting parameters in the collimator design. The former requires short collimator height, while the latter requires the opposite. According to the purpose of collimator for diagnosis of breast tumor, it is useful to choose suitable collimators for expected tumor size through the Trade-Off as shown in figure (9).

G. Accuracy and background effects

From the simulations and their Trade-Off, it is recommendable to use the high sensitivity collimator having short height to detect small cancer with early. However, we should also consider whether it is possible to detect small size tumor with the recommended high sensitivity collimator. In practical field considering the background count from non-tumor volume, there is restriction to use recommended collimators. To give more detail information to diagnosis, we calculated detectable counts per second according to the tumor size and compared with the background count. The calculation based on simulation with derived parameter values is given in the equation (3).

$$C = \sum e_{photopeak} \times RBT \times Time \times Volume , \qquad (3)$$

where $\sum e_{photopeak}$ is the summation of geometric efficiency of total photopeak counts using simulation. The RBT(Radioactivity in Breast Tissue per volume) is assumed 14800 cps, equal to 400 nCi/cm^3. [5], The Time is the exposure time for diagnosis, and the Volume is the various tumor volume. The background count per volume was assumed as 22.5 [5] from non-tumor volume. Figure (10) shows its restriction according to the background count in natural logarithmic scale. In this figure, to detect small size tumor it is recommendable to use as short as possible. And, these results showed that there is a limit of tumor size to be detected for each collimator with a finite height because of background count

V. Conclusion and Further study

From the above result we propose that the optimum crystal height is 5 ~ 6mm and the top and the side surface condition should be painted 0.98RC and polished 0.98RC respectively. Under these conditions, the light transmission is about 94 ~ 94.5%.

In the case of septal thickness of 0.25mm, hole size of 3mm×3mm, collimator height of 10 ~ 20mm for high sensitivity, 20 ~ 40mm for all-purpose, and 40 ~ 60mm for high resolution are generally recommended. But the optimum height for each purpose should be chosen by considering the tumor size and depth.

The optical bonding material refractive index of 1.8 gives the best light transmission rate and the effect of thickness is negligible.

The calculated geometric efficiency and the spatial resolution may be used as a reference for the design of collimator, but the background level in practical use condition is limitation factor in selection of best collimator particularly early detection of small tumor.

In this study we just estimated light transmission through the geometric efficiency and the spatial resolution. The reason to find out the best or suitable parameter values of both the collimator and the crystal are to give maximum light for desirable signal of electronic circuits because solid-state detectors such as photodiode even with preamplifier are low charge amplification gain compared to general scintimammography system based on the PMT. Therefore, in order to get the image from the scintimammography system high quality photodiode with low noise and high signal is indispensable. For future work, it is necessary to study light absorption mechanism and signal generation mechanism in the photosensor, signal amplification in amplifier, and overall noise analysis.

VI. Acknowledgement

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VII. References

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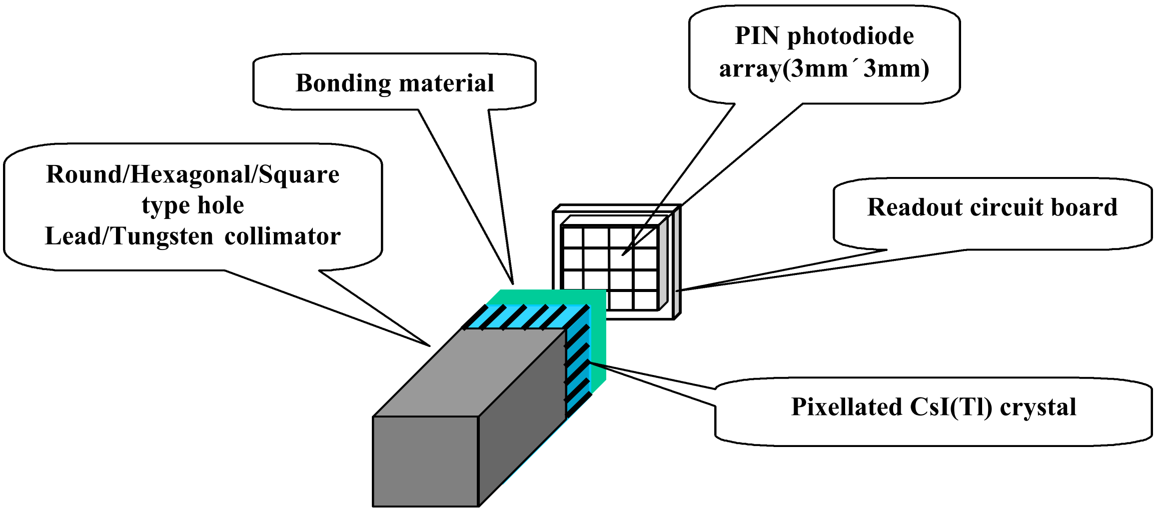

Figure 1. Compact scintimammography system for breast cancer

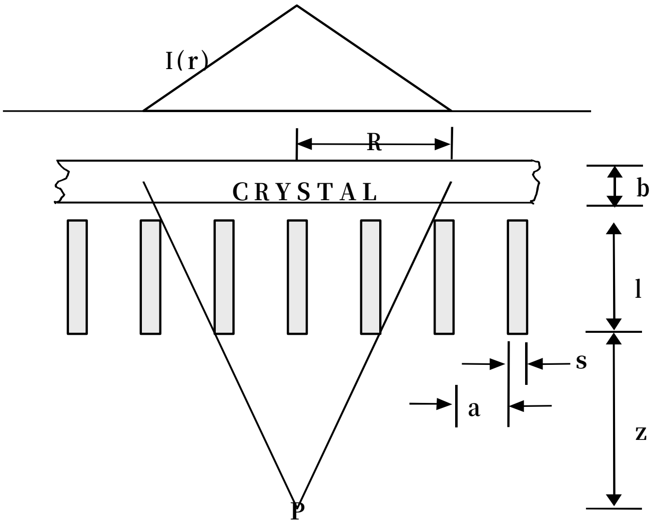

Figure 2. Schematic diagram of collimator, crystal, and source position

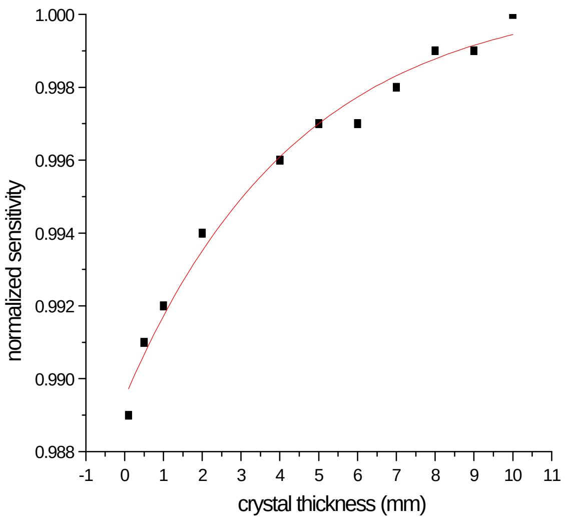

Figure 3. Normalized deposition energy or sensitivity depending on the crystal thickness

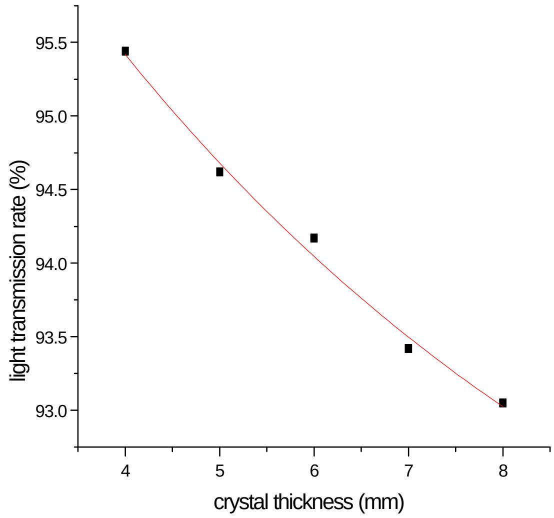

Figure 4. Light transmission rate or efficiency depending on the crystal thickness

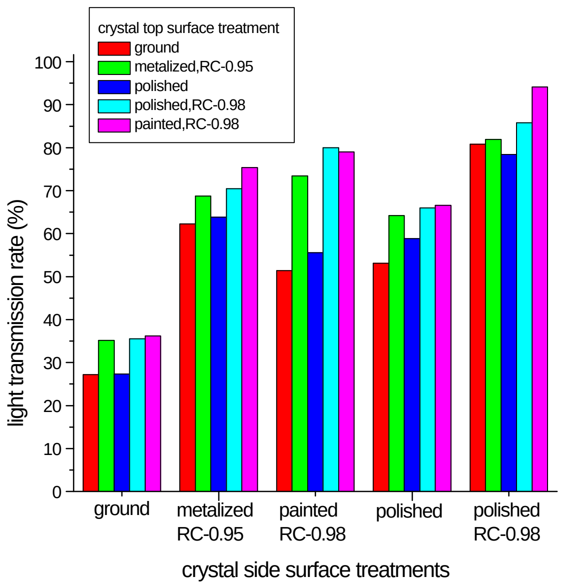

Figure 5. Light transmission rate or efficiency depending on the treatments for crystal surface of top and side

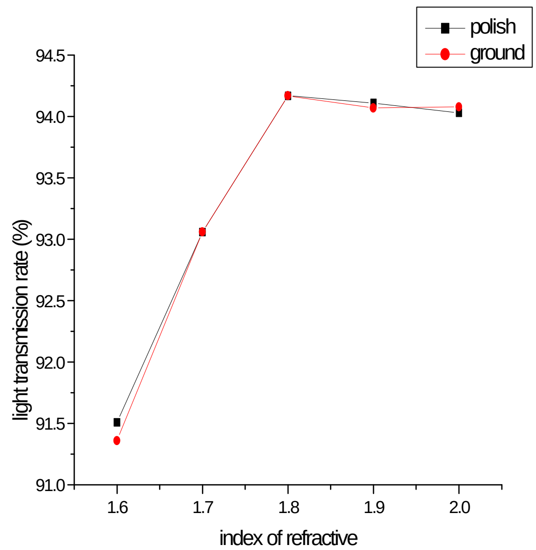

Figure 6. Light transmission rate or efficiency depending on the bonding material

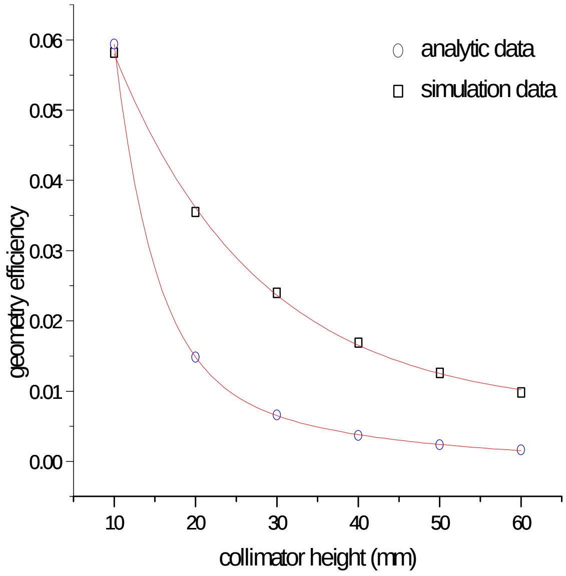

Figure 7. Geometric efficiency depending on the collimator heights

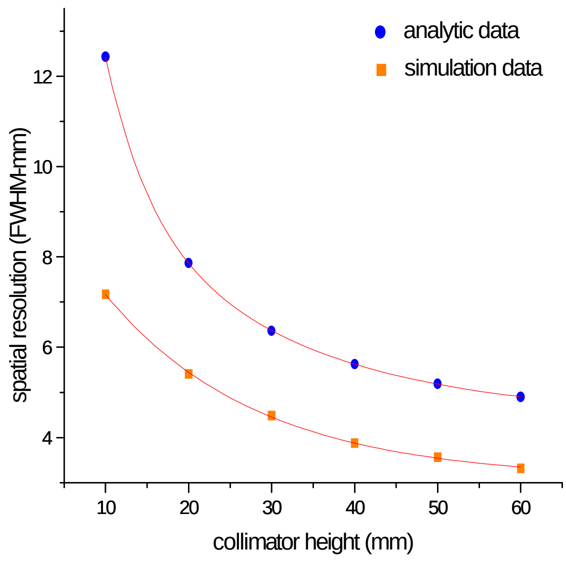

Figure 8. Spatial resolution depending on the collimator heights

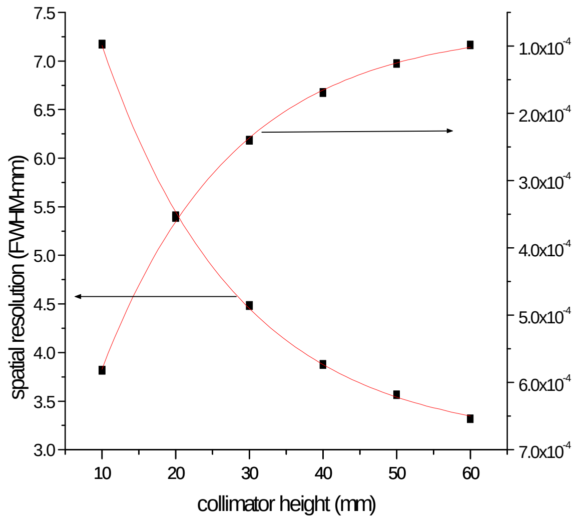

Figure 9. Trade-Off between spatial resolution and geometric efficiency

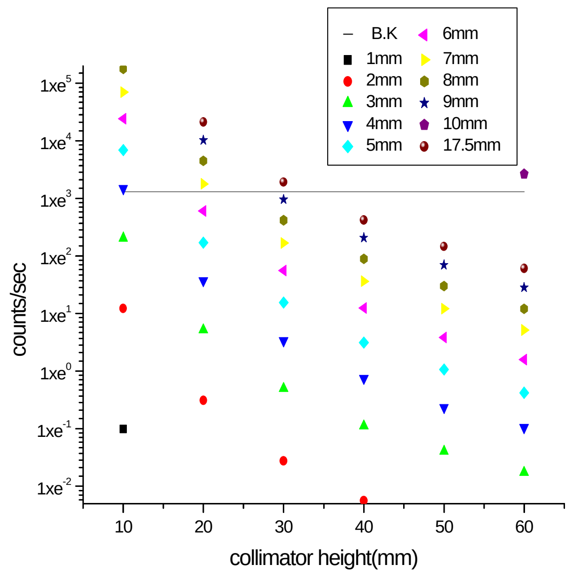

Figure 10. Counts limit depending on the collimator heights and tumor sizes