The effect of K x-ray photons on nuclear medical detectors

P. Liaparinos[1], I. Kandarakis[2], D. Cavouras[2], G. Kontaxakis[3], A. Santos[3], G. Panayiotakis[1]

[1] Department of Medical Physics, Medical school, University of Patras, 265 00 Patras, Greece

[2] Department of Medical Instruments Technology, Technological Educational Institution of Athens, Ag .Spyridonos Street, Aigaleo, 122 10 Athens, Greece

[3] Department of Electronic Engineering, Technical University of Madrid, E.T.S.I.T., Ciudad Universitaria s/n, E-28040 Madrid, Spain

Abstract

The purpose of this study was to examine the effect of K x-ray radiation on the performance of scintillator crystals applied in nuclear medicine detectors. The K x-ray photons are intrinsically produced within the scintillating material, when a photoelectric effect takes place and may either be reabsorbed or may escape the scintillator. In both cases, the scintillator´s performance may be affected resulting in either spatial or energy resolution degradation. By developing a Monte Carlo simulation program, the transportation of K-characteristic radiation within the most commonly used scintillator materials, was examined. For this purpose, the most dominant gamma ray interactions (elastic and inelastic scattering and photoelectric absorption) within the scintillator mass were taken into account. The investigation was carried out by considering a monoenergetic pencil beam geometry of energy 140keV. Results showed that the scintillator's emission efficiency may be reduced from K x-rays production and emission affecting spatial or energy resolution.

1. Introduction

The performance of nuclear medicine imaging detectors, i.e. system spatial resolution, system sensitivity and counting efficiency, may be significantly affected by the characteristics of the scintillator, employed to convert incident radiation into emitted light [1-4]. Properties of primary importance to be taken into account in developing new scintillator materials are the following: photon detection efficiency, high light yield, fast response, good linearity, minimal afterglow, easy growth and low cost. It has been previously shown that $Bi_4Ge_3O_{12}$ (BGO), Lu_2SiO_5 (LSO), Gd_2SiO_5 (GSO) are commercially available scintillators exhibiting high scintillating efficiency by combining many of the aforementioned requirements [5-9]. On the other hand, BGO is a considerably slow scintillator, which is also characterized by high refractive index. High refractive index increases light reflection at the crystal-photomultiplier interface. Thus a large amount of light produced by the crystal is not converted into an electrical pulse (low light output). LSO is a scintillator of high cost that contains intrinsic radioactivity because of the lutetium (^{176}Lu, 2.6%). This radioactivity, in case of PET scanners, increases the background counts. Finally GSO is a scintillator of relative low effective atomic number which exhibits considerably low light output [7-9]. The probability of a K-fluorescence photon generation depends on the atomic number of the element as well on the energy of the incident photon [10]. In the energy range of nuclear medicine applications, where relatively high energy photon detection may be required, scintillators containing one or two heavy (high-Z) elements are commonly used (e.g. Lu element in LSO, Gd element in GSO, Bi and Ge elements in BGO) [1]. In these detectors, the produced K-characteristic x-rays, which are emitted isotropically, may carry a considerable fraction of the initially absorbed radiation. This occurs by either escape or be reabsorbed within the scintillator´s volume in different sites. In first case, the system's counting efficiency is decreased while in other case K-x-rays may degrade the scintillator´s spatial resolution. Many studies, based on either theoretical or Monte Carlo models, investigated the effect of K x-rays in x-ray imaging detectors [10-14]. However, the effect of K x-rays photon production and emission in the performance of scintillators employed in gamma-ray imaging applications has not yet been systematically investigated.

	BGO	LSO	GSO
Density (g/cm^3)	7.13	7.4	6.7
$\rho Z^4 eff$	75	66	59
K-edge (keV)	90.83	63.44	50.30
Light yield (photons/ MeV)	9000	27000	8000
Decay time (ns)	300	40	60
Emission peak (nm)	480	420	430
Index of refraction	2.15	1.82	1.85

Table 1. Physical and scintillating properties of BGO, LSO and GSO

In the present study, a computational program, based on Monte Carlo methods [15-17], was developed in MATLAB platform, in order to investigate the contribution of the K x-rays to the performance of nuclear medicine scintillators. Modelling was carried out by simulating the dominant gamma ray interactions and by taking into account the probability of K characteristic radiation generation with respect to the production of K-Auger electrons. In case of K x-rays production, a miscounting in comparison to the case of K-Auger electron generation, may occur. For this reason, the effect of K-fluorescence radiation generation was studied and the fraction of the K-fluorescence x-ray photons escaping the crystal was estimated. The study was focused on BGO, LSO and GSO scintillators whose physical properties are listed in Table 1 [1,2,5,9].

2. Materials and methods

The simulation of the K-characteristic radiation production and propagation was carried out for the state-of-the art BGO, LSO and GSO scintillator crystals. Crystal dimensions were chosen to be 18x18 mm^2 determining the material surface with corresponding thickness ranging from 0.5 up to 2 mm. The incident radiation beam was assumed to be monochromatic with photon energy equal to 140keV. For each individual simulation, photons were assumed to strike the scintillator surface following pencil beam geometry.

The Monte Carlo code was based on the description of the principal radiation transport and comprised the following basic stages:

A/ Determination of the free path length [18].

B/ Calculation of the interaction site via the free path length and the direction of the gamma-ray photon.

C/ Determination of the interaction type in the interacting element of the material compound.

Photoelectric effect [19], inelastic scattering [20] and elastic scattering [18] were the three possible interactions that were taken into account. The type of interaction was determined by the relative probabilities of occurrence for each interaction. These probabilities were calculated by the mass attenuation coefficient of the material and the partial interaction coefficients μ_{PHOT} , μ_{INC} , μ_{COH} as given bellow:

$$p_{PHOT} = \frac{\mu_{PHOT}}{\mu},\ p_{INC} = \frac{\mu_{INC}}{\mu},\ p_{COH} = \frac{\mu_{COH}}{\mu} \qquad (1)$$

In the case of photoelectric effect, photoelectric absorption takes place in the K-shell of the interacting element. K-fluorescence photons are produced according to the corresponding K-fluorescent yield. At the end of each simulation, the quantum detection efficiency (QDE) of the scintillator was estimated. QDE was defined as the ratio of the gamma ray photons interacting with the crystal over the incident gamma ray photons. This ratio expresses the ability of the scintillator to detect the primary radiation. In the present study, only photons totally absorbed within the crystal were considered in the QDE calculations. All scattered gamma ray photons or intrinsic K x-rays escaping the crystal were not included. In addition the counting efficiency of the system was evaluated. Efficiency which corresponds to the number of counts that the system measure per the total number of counts that all the incident gamma ray photons could produce via only photoelectric absorption (the total number of counts equals to the number of incident gamma photons). This efficiency expresses the ability of the system to provide sufficient number of counts and assumes that the number of the measured counts corresponds to those photons that deposited all their energy within the scintillator. The possible decrease in this efficiency was estimated by evaluating the loss of counts due to the amount of the K x-ray photons that finally escape the scintillating crystal. In this case, counts of lower amplitude are produced, which normally are rejected by the pulse-height analyzer.

3. Results and discussion

The variation of the QDE with respect to the scintillator thickness, for 140 keV incident monoenergetic gamma photons is shown in Fig. 1. Scintillator thickness was allowed to vary from 0 up to 2 mm. Results correspond to three different scintillator materials, BGO, LSO and GSO. BGO scintillator was found to have better absorption properties, according to the higher values of QDE, in the whole range of thickness. At the maximum value of scintillator thickness (i.e.. 2mm), the QDE was estimated approximately 90%, 83% and 71% for BGO, LSO and GSO, respectively. This deals with the presence of higher number of photoelectric events and therefore higher number of produced K x-rays.

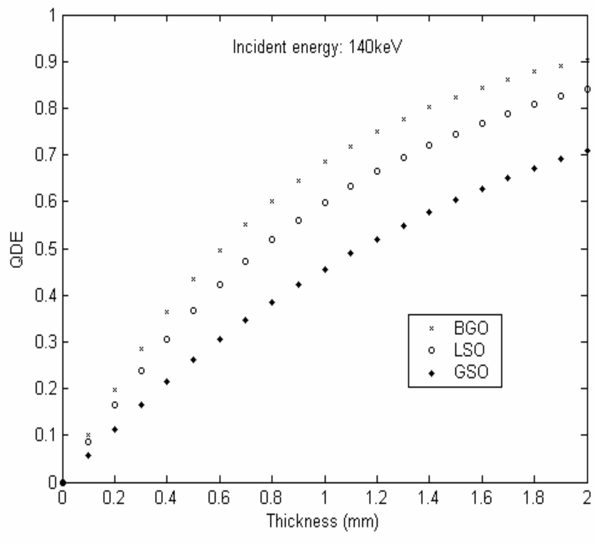

Figure 1. Variation of QDE with respect to crystal thickness for incident gamma ray photons of 140keV in three different scintillator materials: BGO, LSO and GSO.

The variation of the fraction of generated K-photons escaping the crystal with respect to scintillator thickness is shown in Fig. 2. According to the results, for BGO scintillating material, the portion of the escaped K-photons varies from 20% up to 50% for thicknesses 0.5mm and 2mm, respectively. These portions in nuclear medical applications (e.g. 140 keV) may produce a miscounting. This is due to the considerable high energy of the produced K x-ray that escapes the scintillator volume.

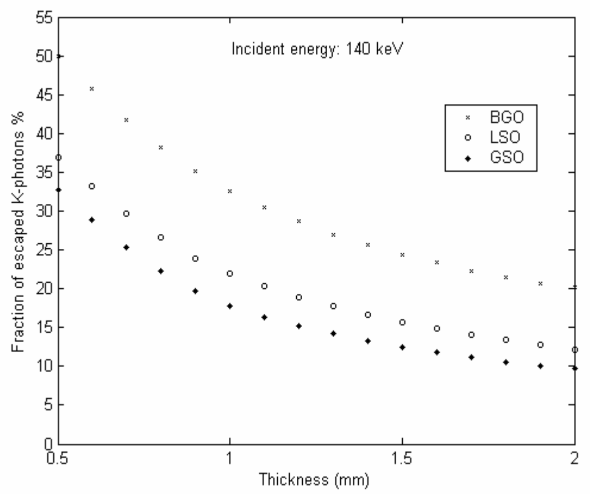

Figure 2. Variation of the fraction of escaped K-photons with respect to screen thickness, for incident radiation energy 140keV in three different scintillator materials: BGO, LSO and GSO

However, a relatively large amount of K x-rays may be reabsorbed within the crystal. Since K-characteristic radiation generation follows an isotropic spatial distribution [11], (the K-photon energy is dispersed backward, forward and laterally in the detector), a reduction of the spatial resolution may occur.

Both escape and re-absorption of the K- fluorescence x-ray photons lead to output detector signal (ratio of light pulses over incident gamma-photons) degradation. A numerical evaluation of possible loss of counts is shown in Table 2.

Beam energy: 140keV	Thickness (1mm)		
	BGO	LSO	GSO
Maximum count loss %	16.92	9.95	6.46

Table 2. Calculation of the maximum count loss (%) for BGO, LSO, GSO crystals applying incident photons of energy 140keV

Results are given for incident photon energy of 140 keV and for crystal thickness of 1 mm. Assuming a pulse-height analyzer of narrow width, the maximum value of count loss was estimated by calculating: i) the number of gamma photons that do only one photoelectric event ii) the number of gamma photons that do a photoelectric event after a scatter event and iii) the number of the K x-rays produced after one photoelectric event (case i) and escape the scintillating material. According to Table 2, BGO showed to present the maximum portion of count loss, which reaches approximately the value of 16.92%. On the other hand, LSO and GSO scintillators, showed lower ability to detect radiation, and by allowing lower portion of K-photons to escape, a lower value of the maximum count loss arises, approximately 10% and 6.5%, respectively. The number of K-photons produced in all scintillating materials was found almost the same (approximately 76%). Taking into account that BGO scintillating material showed higher probability to absorb the initial gamma rays (68%) as well the higher portion of escaped K photons per produced K photons (33.16%), these parameters seem to contribute more to higher amounts of count loss.

4. Conclusion

A simulation code, based on Monte Carlo methods, was developed and used in order to examine the effect of K x-ray photons on three state-of-the art scintillators applied in nuclear medicine detectors. Depending on the results, K-photons may affect the scintillator performance and therefore the quality of the final diagnostic image. This effect was found to be more dominant at the crystals containing high-Z elements (e. g. BGO).

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