ICEST 2010 23 - 26 June 2010, Ohrid, Macedonia

Image Quality of Planar InP Detector

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Abstract - **In this paper we consider InP for solid state X-ray imaging detectors in the photon energy range of medical applications. Taking into account its physical properties, we calculate spatial resolution and contrast of the pixellated InP detector and examine the effect of the fluorescence on image quality. Finally, we compare InP with materials that are currently used for medical imaging.**

Keywords **- radiation detectors, InP, fluorescence, image quality**

I. Introduction

The goal of digital radiography with efficient measurement of absorption and X-ray scattering is to provide a high quality image with good spatial and energy resolution while decreasing patient radiation dose. Semiconductors have proved to be suitable for construction of planar detectors for medical imaging. Commercially available semiconductors can be divided into two categories: elementary and compound semiconductors. Both categories have pros and cons when used for detection of X-rays and creation of medical image.

The most developed production technology is for elementary semiconductors Ge and Si [1], [2], but both of them show high resolution only while cooled. They are suitable for detection of low energy radiation, Ge because of its small gap and Si because of its high bulk leakage current.

Consequently, compound semiconductors that have wider gap and higher atomic number Z have been studied as good candidates for high-resolution detectors operating at room temperatures. When, beside detection efficiency, quality of retrieved image is also considered, one has to take into account fluorescence that degrades spatial resolution and contrast. The yield and energy of fluorescence photons increase with atomic number Z [3], so when choosing suitable semiconductor for detector construction, a compromise should be made between absorption efficiency (high Z) and decrease of Z in order to obtain good contrast and spatial resolution.

Hence, this paper deals with spatial resolution, signal-to-noise ratio (SNR) and contrast for planar pixellated detector made from InP. We compare this material with other compound semiconductors.

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II. Material supply

Performances of semiconductor detectors are limited with production technologies of quality large-area monocrystals, free of defects and with uniform electric properties.

CdTe, a dominant material for compound radiation detectors is still not available in large areas and its production is limited to a small number of companies [4]. GaAs and InP offer photon attenuation coefficients between those of Si and CdTe and are being developed principally for X-ray imaging applications. GaAs in particular has relatively mature contact technologies, but studies of SI bulk GaAs have not progressed beyond prototype demonstrators due to the inhomogeneous nature of the materials and the high concentration of the charge trapping EL2 defects which decreases $\mu\tau$ product. In recent years, there has been some progress in production of epitaxial large-area GaAs [5] with uniform electric properties and lowered defect concentration. InP and HgI_2 seem to be good candidates for high quality imaging detectors because of their atomic number and energy gap.

When X-ray photon interacts with semiconductor, it generates e-h pairs. When external electric field is applied, electrons and holes move into opposite directions causing generation of electric current. In order to achieve full charge collection on both contacts, semiconductor must have a large carrier mobility ($\mu_{e,h}$) and long trapping lifetime ($\tau_{e,h}$). It means that good charge collection efficiency requires that mean drift length $\lambda_{e,h} = \mu_{e,h}\tau_{e,h}E$ be longer than semiconductor crystal thickness. On the other hand, incident photon absorption probability increases with semiconductor's thickness, thus a compromise is usually made between high photon detection and high charge collection efficiency whereas semiconductor thickness is of the order of magnitude of carriers' mean free path. Some physical properties of semiconductors considered in this paper are given in Table I.

TABLE I
PHYSICAL PROPERTIES OF COMPOUND SEMICONDUCTORS

SC	Z	Eg (eV)	$\mu_{e,h}\tau_{e,h}$ (cm^2/V)	k_α (eV)	ω_k (%)	α_k (cm^{-1})
InP	49 15	1.30	$4,8 \times 10^{-6}$, $\leq 10^{-7}$	13	46	252
HgI_2	80 53	2.13	10^{-4} 10^{-5}	42.26	92	100
CdTe	48 52	1.44	2.0×10^{-3}, $\leq 4.010^{-4}$	74.7	86	74.7

III. Fluorescence

When a detector of active area A and thickness t absorbs in a unit of time a photon of flux Φ and energy E_0, the expected free charge N_0 to be generated equals

$$N_0 = \phi A E_0 \varepsilon^{-1}\left(1-e^{-\alpha(E_0)t}\right) \quad (1)$$

where t is semiconductor thickness, α is linear absorption coefficient and ε is energy of e-h pair formation. Because of fluorescence, total charge induced by one absorbed photon in a sphere of radius r around incident point amounts to [7]

$$N(r) = E_0 \varepsilon^{-1}\left(1-e^{-\alpha(E_0)t}\right) - \omega_k \alpha_k \varepsilon^{-1} e^{-\alpha(k_\alpha)r} \quad (2)$$

where ω_k is fluorescence yield, and $\alpha(k_\alpha)$ absorption coefficient of fluorescence photons.

The total charge induced in a sphere of radius r around the incident photon is presented in Fig. 1. InP reaches maximum efficiency of charge creation for energies around 40 keV and CdTe and HgI_2 around 50-60 keV. InP is more efficient than CdTe and HgI_2 in the energy range of medical applications.

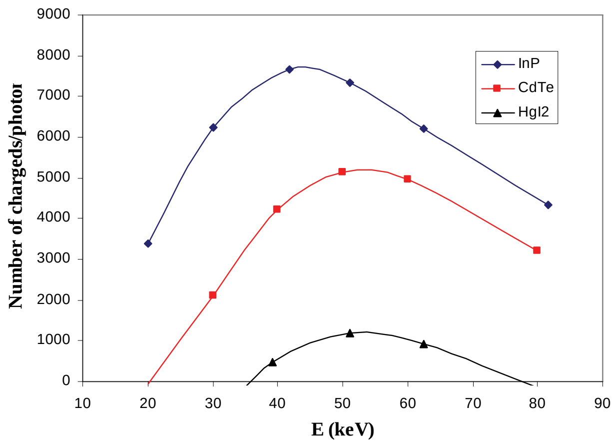

Fig.1 Number of charges per photon versus E(keV)

Considering that fluorescence around the incident photon degrades spatial resolution, we calculated distance R for which 90% of k_α photons absorption will occur. The results shown in Table II indicate that the impact of fluorescence on spatial resolution can be neglected for distances around R=90 µm for InP, but not for CdTe and HgI_2.

TABLE II
DISTANCE R FOR WHICH WE CAN NEGLECT FLUORESCENCE FOR InP, HgI_2, CdTe

Semiconductor	InP	HgI_2	CdTe
R (µm)	91	230	308

IV. Image Quality

Spatial resolution, noise and contrast determine image quality [8] and they are inter-related as follows:

$$SNR = C\sqrt{\phi A} \quad (3)$$

where SNR is signal-to-noise ratio, C contrast, Φ incident photon flux and A is object area.

Signal-to-noise ratio can be calculated as

$$SNR = \frac{N_{ON} - N_{OFF}}{\sqrt{N_{ON}}} \quad (4)$$

and contrast as

$$C = \frac{N_{ON} - N_{OFF}}{N_{ON}} \quad (5)$$

where N_{OFF} is signal inside a pixel with object, and N_{ON} is signal in adjacent radiated pixels.

In order to evaluate detector's SNR, we considered a model of pixellated detector as shown in Fig. 3, calculating signal as a number of generated e-h pairs inside the pixel, taking into account fluorescence in adjacent pixels.

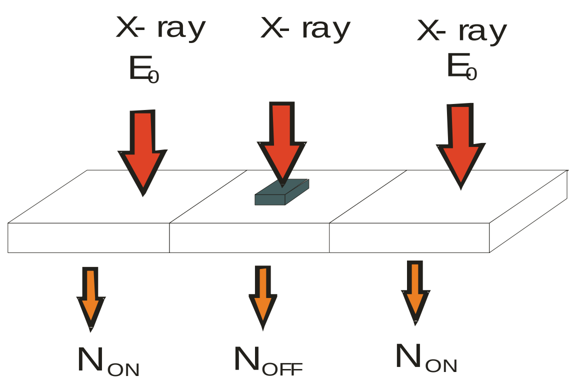

Fig 3. Detector (three pixels and object)

Fig 4. shows the results of SNR calculations depending on object size for a 100 µm pixel, photon flux ~10^8 photons/cm^2, detector thickness 200 µm and incident photon energy 30keV.

Fig 5. shows SNR as a function of incident radiation energy for a 100 µm pixel and 5µm object.

It is evident that the best SNR is achieved for InP, which means that radiation energy for CdTe and HgI_2 should be increased in order to get the same result.

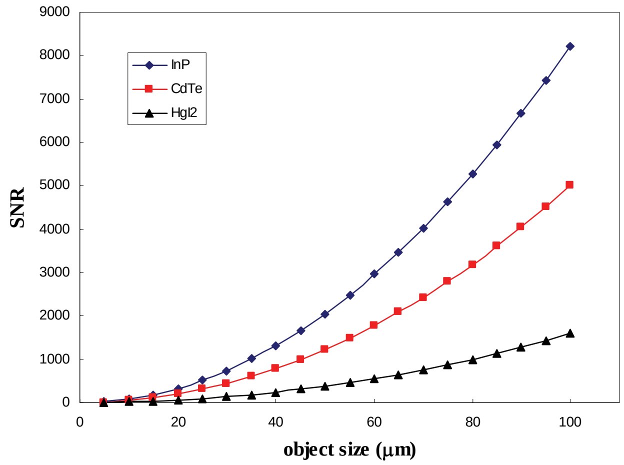

Fig 4. SNR versus object size for a 100 µm pixel and 30 keV energy for InP, CdTe and HgI_2

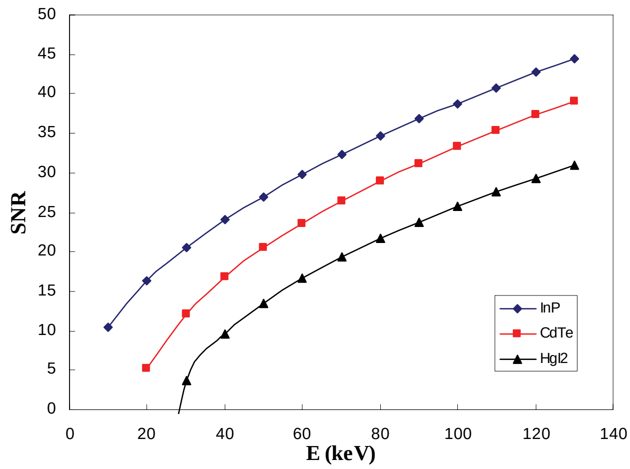

Fig 5. SNR versus incident photon energy for InP,CdTe and HgI_2

Using the same detector model, with an object smaller than a pixel that completely absorbs radiation, we have calculated contrast for different materials depending on the energy of incident X-rays.

The results show (Fig. 6) that InP has a very good contrast (about 1) for the radiation energies considered and much better contrast than CdTe for the lower radiation energies. CdTe has considerably smaller contrast for lower radiation energies, but as the incident photon energy increases, its contrast improves. Both intermediate-Z semiconductors have much better contrast than high-Z HgI_2 in the energy range of medical imaging.

As the aim of medical imaging is detection of early changes inside the tissue, i.e. detection of objects as small as possible, we calculated minimum object size that could be visible in conditions of mammographic imaging, in order to choose the best material for detector construction. During mammographic imaging, a problem is to differentiate tumorous tissue from surrounding tissue because of very similar absorption coefficients. According to Rose model [8], human eye is able to differentiate these changes if SNR is equal to 5.

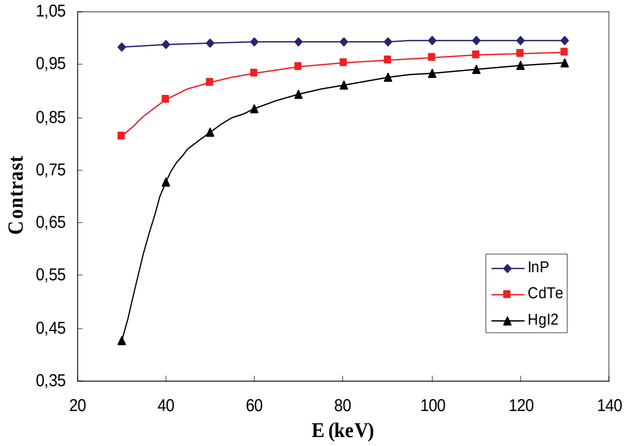

Fig 6. Contrast versus energy for InP, CdTe and HgI_2

Starting from this point and taking for E = 30 keV, absorption coefficient of tumorous tissue [9] (carcinoma) α_c =0,5 cm^{-1}, and of surrounding tissue [9] of thickness 5cm , α_t = 0,25 cm^{-1}, we have calculated minimum size of visible carcinoma in conditions typical for mammography, radiation dose of 100µGy and pixel size 150 µm .

TABLE III
MINIMUM SIZE OF VISIBLE OBJECT FOR 100 µGy RADIATION DOSE FOR InP, CdTe AND CsI

Semiconductor	InP	CdTe	HgI_2
Object size (µm)	30	47	92

The results show that minimum visible size of cubic object for radiation dose of 100µGy is 30µm for InP while for HgI_2 is 92µm. That means that we would have to drastically increase radiation dose in order to see the same size carcinoma with HgI_2 instead of InP.

Finally, we compared contrast, visible object size and radiation dose for considered materials for energy of 30 keV, and pixel size 150µm .

TABLE IV
RELATIVE VALUES OF CONTRAST, MINIMUM OBJECT SIZE AND DOSE FOR InP, CdTe AND HgI_2 FOR LOW ENERGY (30 KEV) AND PIXEL SIZE 150 µm

Relative values	InP	CdTe	HgI_2
Contrast	1	0.83	0.59
Minimum size	1	1.5	3
Dose	1	2	10

It is obvious from these results given in Table IV, that for medical imaging for low energies around 30 keV, there is a considerable reduction of radiation dose with the same contrast and minimum visible carcinoma size for InP instead of CdTe. Comparing InP and CdTe with HgI_2, it follows that intermediate-Z compound semiconductors are more suitable for X-ray medical imaging than high-Z HgI_2 because of improvement of contrast and reduction of radiation dose.

V. Conclusion

Our results show that for low X-ray energies and pixel size of 150 μm in conditions of low contrast according to Rose, minimum visible object size is 3 times smaller for InP, and a dose needed for detection of the object around 10 times smaller than for detectors based on HgI_2. HgI_2 offers good detection efficiency at higher photon energies and is suitable for hard X-ray and nuclear medicine imaging.

According to our evaluations, InP has better contrast than CdTe that is currently used for medical imaging. In order to see the same-size of carcinoma with the same contrast, dose must be 2 times higher for CdTe than for InP. Unfortunately, the detectors based on InP, which is a very soft material, are limited with production technology. The lack of rectifying contacts on SI InP produces high leakage currents in the devices, which must be reduced by cooling. Future improvements in InP detector performance depend on the development of the rectifying contacts. We can conclude that in the group of considered materials, InP seems to be the best choice for obtaining good response to X-ray radiation of energies 20-60 keV, good contrast and lower radiation dose.

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