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F. Thevenet, S. Keshmiri, M. Moussaoui, B. Huffschtmidt, J. Esteves, et al.. A heterogeneous-cavity model for miniaturized detectors of orthovoltage X-rays. *Journal of Physics: Conference Series*, 2024, 2799 (1), pp.012012. 10.1088/1742-6596/2799/1/012012 . hal-04706749

HAL Id: hal-04706749

<https://hal.science/hal-04706749v1>

Submitted on 23 Sep 2024

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To cite this article: F. Thevenet *et al* 2024 *J. Phys.: Conf. Ser.* **2799** 012012

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A heterogeneous-cavity model for miniaturized detectors of orthovoltage X-rays

F. Thevenet¹, S. Keshmiri², M. Moussaoui¹, B. Huffschmidt², J. Esteves¹, G-N. Lu¹, J-F Adam^{2,3} and P. Pittet^{1*}

¹ Université Claude Bernard Lyon 1, INL, UMR5270, CNRS, INSA Lyon, Ecole Centrale de Lyon, CPE Lyon, 69622 Villeurbanne, France

² STROBE - Synchrotron Radiation for Biomedical Research UGA Inserm UA7, Saint Martin d'Hères, 38400, France

³ Centre Hospitalier Universitaire Grenoble Alpes CHU-GA, Grenoble, 38043, France

*E-mail: patrick.pittet@univ-lyon1.fr

Abstract. To meet the needs of modelling small-sized detectors of orthovoltage X-rays, for spatially fractionated radiotherapy, we propose a model to determine responses of the detectors in cases of highly heterogeneous cavity. The model was applied to a detector whose cavity features a sparse high-Z perovskite scintillation layer embedded in an epoxy resin. Irradiations were carried out using the *Small Animal Radiation Research Platform*. Local photon and electron spectra were computed by the SpekPy toolbox and Penelope Monte Carlo simulations, respectively. The detector model considers that X-ray photons mainly interact with medium surrounding the scintillation layer, and that the output of the detector mainly results from secondary electrons in this layer. Results from model computations are compared with measurement data, showing differences less than 16% for beam energies in the 60 kVp -220 kVp range.

1. Introduction

Sub-millimeter fractionated high-dose-rate radiotherapy, such as MRT (Microbeam Radiation Therapy), uses orthovoltage X-rays and requires miniaturized detectors for dosimetry and Quality Assurance (QA). The detectors could be of different types, such as semiconductor devices operated on radio-electric effects, or scintillation structures based on radio-luminescence emission. Considering packaging of sensing elements and signal transmission schemes, a typical detector cavity – the part of the detector receiving X-ray radiation - is heterogeneous in terms of integrated materials of different components. In general, a detector cavity could integrate a signal sensing element in some cases, and a group of sensing elements in others. Predictably, the response of the detector, used to measure the average dose in a water voxel of the detector cavity volume, depends on its internal structure and material composition. As more and more new miniaturized detectors come to be employed, there has been demand for developing a model to quantify responses of heterogeneous-cavity detectors [1][2]. This is because small and large cavity theories are not applicable, and the intermediate cavity theory relies on similar media within the detector cavity, which is not the case for heterogeneous-cavity detectors. We propose here a new model to compute such responses, with the following assumptions: i) each sensing



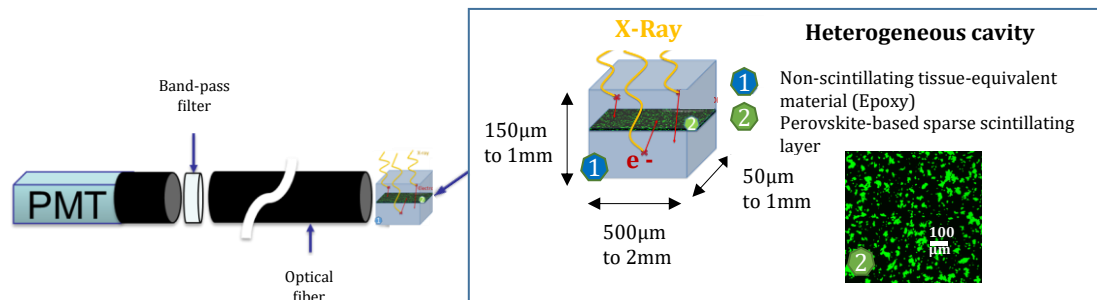


Figure 1. Schematic view of the detector with inset showing the heterogeneous cavity with a fluorescence image of micron-thick sparse scintillating layer

element in the cavity is sufficiently small that its direct interaction with the photon beam can be neglected and; ii) these elements are spaced apart (in terms of electron range) to prevent interaction between them.

As an example of model application and validation, we have studied a detector made of a micron-thick Perovskite-PDMS scintillating layer embedded in a tissue-equivalent organic material.

2. Materials and methods

2.1 Materials

The heterogeneous-cavity detector under this study consists of a micron-thick sparse $\text{CH}_5\text{N}_2\text{PbBr}_3$ Perovskite-PDMS scintillating layer embedded in a tissue-equivalent organic material (epoxy) as illustrated in figure 1. This part is optically coupled to a 1 mm plastic multimode fiber. The scintillating signal at the fiber output is filtered by a band-pass filter (543 nm/22 nm) and then detected by a photon counting PMT Module. It is noted that Cherenkov emission can be neglected since the energy threshold in the PMMA and epoxy is higher than most kV energy spectrum values [2]. Measurements on the response of the detector under a 5 mm-diameter circular radiation field were performed using the *Small Animal Radiation Research Platform* (SARRP, Xstrahl, Camberley, UK) installed at IRMaGe (SAXO platform), in Grenoble, France. The detector cavity was positioned at a reference depth of 2 cm in a RW3 phantom with 4 cm of backscattering material. Reference dose measurements were carried out at the reference depth with a calibrated PinPoint 3D ionization chamber (PTW GmbH, Germany).

2.2 Methods

With consideration of specific experimental conditions, the X-ray spectra and fluencies from the SARRP tube were modelled by using SpekPy v2.0 software toolkit [3]. Determination of the local photon and electron spectra at the reference depth were based on the X-ray normalized energy spectra. Local photon spectrum for each beam energy (depending on the kVp tube setting) noted $\varphi_{2\text{cm},\text{kVp}}(E)$ was determined by adding 2 cm of water filtration in SpekPy. The local photon fluencies were weighted by backscatter factors according to AAPM TG-61 protocol [4]. The local electron spectra noted $\Phi_{2\text{cm},\text{kVp}}(E)$ were generated using the MC code Penelope [5]. For modelling the detector, we assume that in the heterogeneous detector cavity, X-ray photons interact primarily with the epoxy material under Charge Particle Equilibrium (CPE) conditions.

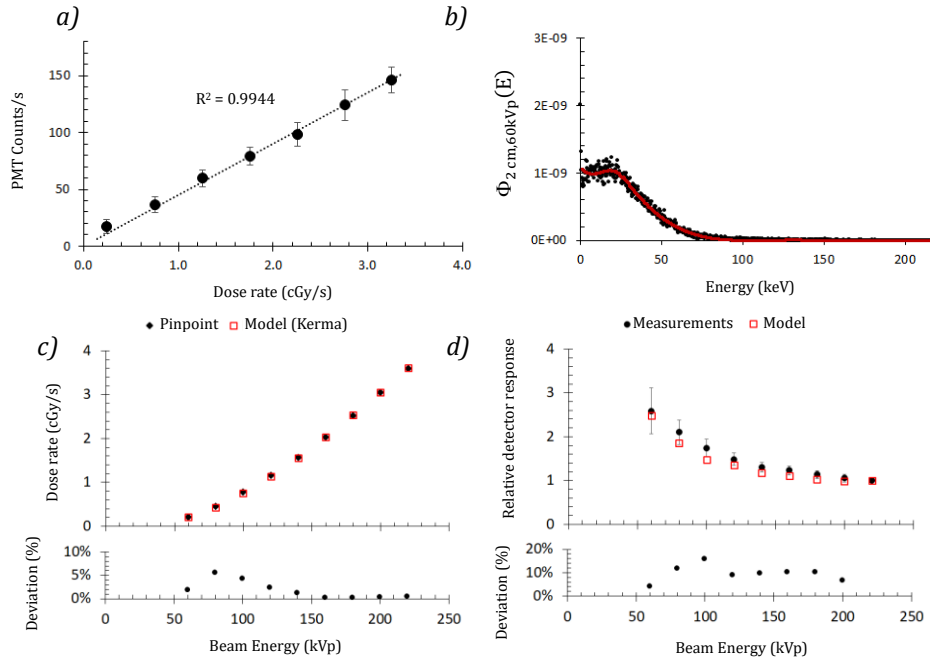


Figure 2. a) Linearity of the scintillation intensity as a function of dose at 2 cm in water for a beam energy of 220 kVp and with an additional filtration of 0.15mm of copper. Error bars correspond to $\pm 1\sigma$; b) Local electron spectra at reference depth for the 220 kVp beam energy; c) Computed kerma and measured dose at reference depth and d) measured and modelled detector response

Thus, the dose can be determined by the collision kerma in this material, which is very close to the kerma in water, $K_{w,kVp}(2\text{ cm})$. On the other hand, we also assume that the scintillation signal is mainly due to the interaction of secondary electrons with the sparse scintillation layer. Assuming that the local electron spectra is not modified by the presence of this thin layer, the scintillation signal is proportional to the restricted cema in Perovskite, $C_{Perov,kVp}(2\text{ cm})$ [6]. Based on these assumptions, the response of the detector can be determined as follows:

$$R_{kVp} = \frac{C_{Perov,kVp}(2\text{ cm})}{K_{w,kVp}(2\text{ cm})} = \frac{\left[\int_{\Delta}^{E_{max}} \Phi_{2cm,kVp}(E) \cdot \left(\frac{S_{\Delta}(E)}{\rho} \right)_{Perov} dE + TE \right]}{\int_0^{E_{max}} \varphi_{2cm,kVp}(E) \cdot \left(\frac{\mu_{en}(E)}{\rho} \right)_w \cdot E dE} \quad (1)$$

where Δ is the energy threshold below which electrons are assumed to be absorbed locally in the scintillating layer, $\frac{S_{\Delta}(E)}{\rho}$ and $\frac{\mu_{en}(E)}{\rho}$ are respectively the restricted collision stopping power and mass energy-absorption coefficients based on the data from NIST, and TE is the track-end term related to the effect of finite delta ray ranges [6].

3. Results

Figure 2.a plots measured results on the response of the detector (PMT counts per seconds) irradiated at different dose rates ranging from 0.24 cGy/s to 3.25 cGy/s. A linear trend is observed confirming a linear response with no dose rate dependence. Figure 2.b shows computed the local electron spectrum for 220 kVp beam energy, indicating more significant contributions at lower energies, which can be attributed to slowing down of Compton electrons [7]. Local photon spectra and fluencies were used to compute the kerma in water at reference depth. Figure 2.c compares the obtained kerma with measured dose using a PinPoint ionization chamber. There is a good

agreement, with differences within 6% over the range of 60 to 220 kVp, which confirms that the spectra and fluencies from the SARRP tube have been correctly modelled. Measured and modelled results on the detector response (counts/cGy) as a function of beam energy are plotted in Figure 2.d, with data points normalized to the 220 kVp beam energy. Deviation of the computed response from the measured one remains lower than 10% for [120 kVp, 220 kVp] energy range. It becomes larger (up to 16%) for [80 kVp, 100 kVp] range, which may be due to a lack of precision in spectrum modelling in this range, and/or increased measurement uncertainty at a low dose rate.

4. Discussion

The coherence between model computations and measurements is a confirmation for the model assumptions: X-ray photons interact mainly with the medium surrounding the thin scintillating layer, and the scintillation signal comes from irradiation of this layer by secondary electrons. These assumptions are applicable to other types of heterogeneous cavity detectors, such as those studied by Delage *et al.* [2]. It means that, by choosing a tissue-equivalent surrounding material, constraints on the tissue equivalence of the scintillating material can be relaxed, since the energy dependence of the stopping powers is much weaker than that of the mass energy-attenuation coefficients. This proposed model can provide an alternative approach to intermediate cavity theory without the need to choose d-parameter (a weighting factor depending on the cavity size: from unity to zero with the increase of cavity dimensions) that does not meet with consensus [1]. The presented study was focused on the response of the detector in terms of energy dependence and measurements were performed at low dose rates, whereas this detector was designed for FLASH therapies. Preliminary tests carried out under MRT conditions at ESRF (not shown here) confirmed a linear response of this detector at very high dose rates (a few hundred to a few thousand Gy/s), and thus its suitability for this application.

5. Conclusion

We have developed a heterogeneous-cavity model for small-sized detectors. It was applied to a detector based on a thin scintillating layer of Perovskite embedded in epoxy resin. The model allows computations of the response of the detector. By comparing model computations with measurements for beam energies in the 60 kVp -220 kVp range, differences remain lower than 16%, and could be narrowed by further studies such as more accurate local spectral modelling.

Acknowledgements

With financial support from ITMOCancer AVIESAN (Ioda-Master project) and from the ANR (ANR-11-LABX-0063, ANR-19-CE19-0021). Experimental developments were carried out at Nanolyon (Renatech+ network) and IRMaGe platforms.

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