Nuclear Instruments and Methods in Physics Research A 732 (2013) 34–39

Contents lists available at ScienceDirect

Nuclear Instruments and Methods in Physics Research A

journal homepage: www.elsevier.com/locate/nima

Particle detector applications in medicine

Hartmut F.-W. Sadrozinski *

Santa Cruz Institute for Particle Physics, University of California, Santa Cruz, CA 95064, USA

A R T I C L E I N F O

Available online 11 June 2013

Keywords:
Medical applications
Photon detection
Computed Tomography
Proton CT

A B S T R A C T

Selected particle detectors are described which find an application in medicine and have been the topic of presentations at the 2013 Vienna Conference of Instrumentation (VCI).

1. Introduction

Particle detector applications in medicine are characterized by low energies and relatively small systems (at least until commercialization). They are still profiting from developments in High-Energy Physics and (due to the low energy), even more so in Astrophysics, although as shown in this paper, medicine-specific projects are becoming common.

This paper will concentrate on semiconductor systems; gaseous detectors in the WCC heritage have been covered e.g. in talks by V. Peskov and D. Nygren; in addition, many recent developments of scintillator materials have been reported at VCI 2013. The first section will deal with photon detectors in dosimetry and imaging, including X-ray computed tomography (CT), single-positron emission CT (SPECT), positron emission tomography (PET) and time-of-flight PET (ToF-PET). This will be followed by a discussion of detectors in hadron therapy, a topical subject with the advent of the MedAustron, and their application in interaction vertex imaging (IVI), and proton CT (pCT).

2. Instrumentation for photons

2.1. Basics of photon interactions

For photon detectors the attenuation of the photon flux $N(x)=N_oe^{-\mu x}$ is essential. The material-specific attenuation coefficient μ (in units of 1/cm), shown in Fig. 1 for selected materials, is shifted to higher energies for heavier high-*Z* materials [1]. This explains the high contrast achieved in bone-tissue X-rays. As Fig. 1 shows, in the region of interest for medical applications, 10–100 keV for X-ray radiography and CT and 500 keV for PET and SPECT, μ is strongly dependent on energy and thus influences the usefulness of different materials for photon detection. In addition, the thickness of the sensors plays a crucial role. The horizontal lines in Fig. 1, indicating the values of attenuation coefficients required to reduce the photon flux to 1/*e* for typical thickness of the materials shown, limit the use of silicon to below a few 10's keV, that of CdTe and CsI to 100 keV, and only for thick heavy crystals to about 500 keV. High-*Z* materials do not only extend the energy reach, they also shift the transition energy where the photoelectric effect has the same strength as the Compton effect to a higher value: 65 keV for Si, 150 keV for Ge, and 280 keV for CsI. While the Compton process might not provide total absorption and thus no optimal energy resolution, it can provide directional information.

2.2. Clinical dosimetry

Semiconductors are good dosimeters [2]. Silicon diodes find routinely clinical use for Q/A in X-ray therapy. They are a well-established and low-cost technology and are characterized by high sensitivity. When operated at zero bias, there is no performance degradation from leakage currents, and they can be used in *in-vivo* applications. Some of the new developments are similar to those in high-energy physics applications like the use of p-type, low-resistivity or epitaxial bulk to reduce the dose rate dependence, and pre-irradiation to reduce the sensitivity decrease with total dose.

There are ongoing efforts to use tissue-equivalent diamond sensors as dosimeters. They have small size and good dosimetric properties, and small dose rate dependence. Since natural diamonds are expensive, one development is to replace them with CVD-grown devices.

2.3. Emergency dosimetry and environmental health and safety

Remote sensing of radioactive sources in emergency situation has received increased attention with the Fukushima nuclear power plant disaster. Here the Compton camera provides both tracking and

* Tel.: +1 831 459 4670; fax: +1 831 459 5777.
E-mail address: hartmut@ucsc.edu

0168-9002/$ - see front matter
http://dx.doi.org/10.1016/j.nima.2013.05.117

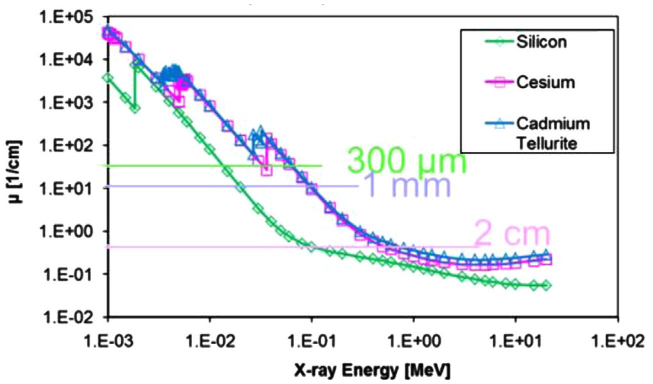

Fig. 1. Photon attenuation coefficient μ (in units of 1/cm) for selected materials. The horizontal lines indicate the values of attenuation coefficients required to reduce the photon flux by a factor $1/e$ for typical thickness of the materials shown: 300 μm for Si, 1 mm for CdTe and 2 cm for CsI [1].

spectroscopy to locate and analyze remote sources of radioactive waste X-rays in the 500 keV energy range at a distance of meters [3,4]. Since X-rays in the same energy range are used in SPECT, the same instruments can be used there.

2.4. Silicon X-ray detectors

With silicon sensors having large-scale applications as a tracking detector, the question arose if they could be of use in X-ray radiography. Fig. 1 indicates that one would need 300 μm of silicon to absorb photons of 10 keV, 1 mm for ~30 keV, and 3 cm for 500 keV. At first sight, the thickness of a typical high-resistivity Si wafer of 300 μm seems to be a barrier to the use of silicon as an X-ray detector. But by using silicon strip detectors “edge-on”, one can employ the entire length of the strips as a deep converter and make narrow sensors with pixels sized wafer thickness times pitch [5,6]. One needs to scan the limited active area of the detector with an associated narrow collimator across the patient. This has been commercialized already [7].

2.5. Positron emission tomography (PET)

PET is used to study the accumulation of radioactive tracers in specific organs. The tracer has radioactive positron decay, and the positron annihilates within a short distance with emission of a pair of 511 keV γ-photons, which are observed in coincidence by means of a ring of finely-grained crystal scintillators surrounding the patient, and the line of flight (LoF) is approximated by the line reconstructed between the two hit crystals (LoR). The overlap of the LoF inside the target volume allows the CT reconstruction. Several detector effects are deteriorating the LoF resolution: the position resolution of the X-ray detectors (pitch), the positron range, and the parallax (depth of interaction, DoI). In addition, the Compton scattering and random coincidence also worsen the signal-to-noise ratio (SNR) [8].

2.5.1. Improving spatial resolution in PET

The spatial resolution can be improved by combining the calorimeter crystals of a standard PET scanner with finely segmented silicon sensors to record the impact position of one or both photons from the annihilation pair [9]. The Si sensors are 1 mm thick and have 1×1 mm^2 pads, to be compared to the crystal pitch of a few cm. From the discussion in Section 2.1, the high-resolution Si detector will capture a relatively small portion of emitted radiation (2%), but the standard PET image can be locally improved in areas where there is a substantial probability that at least one of the photons will interact in the high-resolution silicon detector.

2.5.2. Reduction of accidentals and image improvement with ToF-PET

The target volume from which LoF intersections are accepted in the image reconstruction can generally not be constrained, which leads to substantial background from accidental coincidences. Improved localization can be achieved if the time of arrival in the crystals can be measured with high precision σ_t, such that the time difference marks a place along the LoR with a precision $\Delta D = \frac{1}{2}c\sigma_t$, where c is the speed of light [10], allowing a reduction of the size of the target volume from D to ΔD. For example, with $\sigma_t = 200$ ps, the target volume size shrinks from $D = 40$ cm to $\Delta D = 3$ cm.

The improved source localization due to time of flight (ToF) leads to an improved signal-to-noise ratio

$$SNR_{\mathrm{TOF}} = \sqrt{\frac{D}{\sigma_x}} SNR_{\mathrm{Non-TOF}}$$

and an increase of noise equivalent count (NEC)

$$NEC_{\mathrm{TOF}} = Gain_{\mathrm{TOF}} NEC_{\mathrm{Non-TOF}} = \frac{D}{\sigma_x} NEC_{\mathrm{Non-TOF}}$$

The gain in SNR and NEC as a function of time resolution is shown in Fig. 2. Improving the time resolution from the 600 ns of today's commercial PET scanners to 30 ps would allow reducing the dose by a factor 20 (!). For a given acquisition time and dose to the patient, TOF can provide better image quality and improved lesion detection, or with TOF the scan time and dose can be reduced for the same image quality.

2.5.3. PET parallax: depth of interaction (DoI)

The error in parallax due to the unknown gamma conversion depth in the crystal is being reduced by a variety of methods. One is to detect the light on both the front and back sides of a continuous crystal with Si photomultipliers (SiPM) and to determine the DoI from the asymmetry of the two signals. The expected DoI resolution is 1–2 mm [11]. Another approach is to divide the crystal in four segments and read out each with a MPPC (Fig. 3), which results in DoI resolution of below 1 mm [12].

2.5.4. AX-PET

Another solution to the parallax problem is AX-PET [13]. Long axially oriented crystals are crossed by orthogonal wavelength shifting fiber (WLS) strips, individually read out by SiPMs. This affords high resolution and high sensitivity. An evaluation with point sources, phantoms and by imaging small animals revealed a

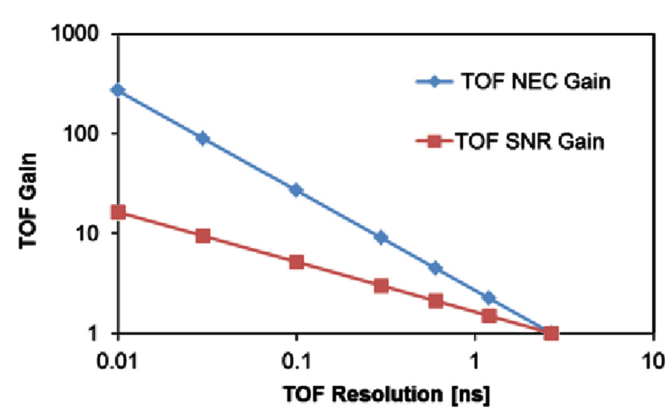

Fig. 2. Gain in SNR and NEC versus time-of-flight resolution in TOF-PET for $D = 40$ cm [10].