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Particle Detector Applications in Medicine

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Abstract

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).

Keywords

Medical applications; Photon detection; Computed Tomography; Proton CT

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; the 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

3. Basics of photon interactions

For photon detectors the attenuation of the photon flux $N(x) = N_0 e^{-\mu x}$ 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, respectively, 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

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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, respectively. While the Compton process might not provide total absorption and thus no optimal energy resolution, it can provide directional information.

3.1 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.

3.2 Emergency dosimetry and environmental health & 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 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.

3.3 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, respectively. 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].

3.4 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, Compton scattering and random coincidence also worsen the signal-to-noise ratio (SNR) [8].

i. 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 $1 \times 1 \text{ mm}^2$ pads, to be compared to the crystal pitch of a few cm. From the discussion of Sect. 2a, 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.

ii. 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 Δt , such that the time difference marks a place along the LoR with a precision $\Delta D = \frac{1}{2} \cdot c \cdot \Delta t$, with c the speed of light [10], allowing a reduction of the size of the target volume from D to ΔD . For example, with $\Delta t = 200$ 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

$$SNR_{TOF} = \sqrt{\frac{D}{\sigma}} \cdot SNR_{Non-TOF}$$

and an increase of noise equivalent count (NEC)

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

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.

iii. 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 side 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 4 segments and read out each with a MPPC (Fig. 3), which results in DoI resolution of below 1 mm [12].

iv. 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 position resolution of 1.6 mm FWHM. An extension of this approach is ToF – PET with AX-PET, reading out the crystals on both sides with a very fast digital SiPM, which has an excellent coincidence time resolution of 225 ps FWHM even with 100 mm long LYSO crystals, constant over the entire field-of-view.

v. ToF-PET with Cherenkov light: Cherenkov light from electrons in the photon conversions can be used for ToF-PET. A high-Z crystal increases the speed of the electron,

although the number of photons is small (<10). Using a MCP-PMT (prototypes for Belle II - TOP supplied by Hamamatsu Photonics) with intrinsic resolution ~ 20 ps, the time resolution obtained with such a system was 116 ps FWHM, with 50 ps from DoI uncertainty, and the single-detector efficiency reached was 4.3% [14].

3.5 Progress in photon detectors—Advances in photon detection are supported by the development of Silicon Photomultipliers (SiPM) [15] or Multi-Pixel Photon Counters (MPPC) [16], solid state devices which are a good match for various photon detectors, because of their feature size, speed, power, low bias voltages, ruggedness, and insensitivity to magnetic fields. One trend is to integrate them to large monolithic arrays which can read out large-area continuous crystals [16]. Improvement in performance is derived from the successful operation at fairly large overvoltage without excessive noise or after-pulsing. Resolution in energy and coincidence timing as well as the photon detection efficiency (PDE) are improved [17], sometimes by on-device filtering of the signal, leaving cross-talk a remaining problem [18]. The coincidence time resolution is now in the range 120 – 140 ps [18, 19].

3.6 Chromatic X-ray counting—Due to the energy dependence of the X-ray attenuation (Fig. 1), a measurement of the X-ray energy can increase the contrast. For very fast counting experiments with direct-conversion semiconductors this can be achieved by pseudo-analog recording with two counters per pixel with two thresholds [20]. A CdTe pixel sensor of large area ($30.96 \times 24.98 \text{ mm}^2 \times 0.65 \text{ mm}$) and a pixel pitch of $60 \mu\text{m}$ is operated at 400–500 V bias. This allows reading of two ‘colors’ (2 thresholds, 2 counters) or, alternatively, almost dead-time free counting in one counter while reading the other one. The dead-time corrected pixel rate capability is 10^6 counts/pixel/s with a global rate capability of 2.4×10^{11} counts/s per chip. A position resolution of 11 LP/mm at 50% MTF has been achieved, with a detection efficiency of 100 % at 10 keV, 98% at 50 keV, and 45% at 100 keV, respectively.

3.7 Multi-modal: PET/CT and PET/MR—There is a trend to combine different imaging modalities: PET needs CT or MRI data to locate and delineate the tumor and correct for attenuation [21, 22]. Use of PET in conjunction with MRI has the added advantage that the range of the positrons is greatly reduced in the large magnetic field.

4. Instrumentation for hadrons

Hadron therapy with either protons or ions like carbon is a highly precise treatment modality due to the Bragg peak, which delivers a large dose at the end of the range with fast distal fall-off. This inherent precision is compromised by uncertainties in localizing the patient and/or the beam, which is addressed by Interaction Vertex Imaging (IVI), and the uncertainty in the stopping power distribution in the patient, which can be addressed by proton CT (pCT).

4.1 Interaction Vertex Imaging IVI

The large kinetic energy of the carbon nuclei allows the secondaries from beam interactions to have sufficient energies such that they can be detected after passing through the phantom/patient. Both single and coincident particles can then be used to reconstruct the beam properties. As the simulations of the longitudinal (depth) profile of the vertices shown in Fig. 4 demonstrate, the yield from the most interesting target region, the Bragg peak, is very small. One deduces its location from the “inflection point” in the profile of reconstructed vertices.

Using a Compton camera to detect gamma’s and reconstruct their directions, the location of the vertex can be determined with no range or multiple Coulomb scattering (MCS)

limitations [24]. For charged particles, range and multiple scattering are an issue, and thin, highly segmented sensors are required [25]. This applies to irradiation volume identification (IVI) in proton therapy as well; the reconstruction efficiency is small, but during treatment the flux is high. A location precision of the order of mm has been demonstrated [26].

4.2 Proton CT (pCT)

Proton therapy and treatment planning requires the knowledge of the stopping power in the patient, so that the Bragg peak can be located within the tumor. X-ray CT has been shown to give insufficiently accurate relative stopping power (RSP) maps due to the complexity of the phantoms or uncertainty in converting Hounsfield values to RSP. This need for better RSP maps can be satisfied with proton CT, where one is to reconstruct a 3D map of the stopping power within the patient, using protons (instead of X-rays) in transmission.

i. Proton CT concept—In pCT, a low intensity cone beam of protons traverses the patient/phantom, and for each proton the position and direction at both entry and exit and the energy loss is measured. In a rotational scan the integrated stopping power is determined for every view by a measurement of the energy loss, which translates into water-equivalent path length (WEPL). Thus a pCT instrument consists of a tracker and a “WEPL” detector.

Minimal proton loss and high detection efficiency make this a low-dose imaging modality [27]. In distinction to the absorption of X-rays, where the attenuation factor can vary by a factor 10 between bone and soft tissue, proton radiography delivers low contrast, since the RSP differs by only 50% between bone and soft tissue. A variety of detector technologies are used in the different pCT instruments for both the tracker and the energy detector as shown in Table 1.

ii. pCT Challenges—The upgrade of the present pCT scanner [31] addresses several issues which impact the performance of the scanner, shown in a model in Fig. 5 [33].

Multiple Coulomb scattering: Low-energy protons undergo appreciable multiple Coulomb scattering (MCS) in the phantom and thus the proton path inside the patient/phantom is not straight. To avoid blurred images, the path of **every** proton has to be measured before and after the phantom and its path inside the patient reconstructed.

From deflection and displacement, the most likely path MLP can be calculated [34]

Beam test measurements have confirmed that the path within the phantom can be reconstructed with high resolution silicon sensors to better than 500 μm [35].

Proton data rate: Since one needs 100 protons/1mm voxel and 180 views, the required proton rate is $\sim 7 \cdot 10^8$ protons [33]. An acceptable 6 min scan will require a proton rate of 2 MHz, requiring a fast data acquisition (DAQ) system which can read out the silicon tracker and a fast WEPL detector at this rate [36].

Image reconstruction: To reconstruct images with $> 10^7$ voxels using $\sim 10^9$ protons is NOT trivial even with parallel processing. Our reconstruction code is running on GPU's in anticipation of the high data rates [37].

WEPL detector: The measurement of the proton's WEPL in the phantom is a critical parameter. The inherent low contrast of pCT requires uniform response of the WEPL detector. Range straggling amounts to about $\sim 1\%$ of the range, i.e. ~ 3 mm for 200 MeV protons. The hodoscopic CsI calorimeter of the existing scanner [31] is both too slow and in need of tedious calibrations to ensure the needed uniformity [38]. A range counter with 4

mm scintillator plates with direct MCPP readout showed good SNR, with signals 3–5 times that of 3 mm plates and WLSF readout [39]. The baseline is a novel multi-stage scintillator (MSS) counter, consisting of three $10 \times 10 \times 40 \text{ cm}^3$ polystyrene blocks with light guides and 3 photomultiplier (PM) readout [40]. The WEPL calibration of each stage is done with polystyrene degraders [31]. The use of plastic scintillator supports the required high data rate. A comparison of the WEPL resolution of the different WEPL detectors is shown in Fig. 6, indicating that the resolution of both the range counter and the MSS counter are close to the theoretical straggling limit for all phantom thicknesses.

Future WEPL detector: 4-D Ultra-Fast Si Detectors (UFSD): As mentioned above, the straggling of protons in the calorimetric WEPL detector limits the WEPL resolution. We are working on very fast silicon detectors, which would replace the separate back telescope and the calorimetric WEPL counter by a combined tracker-ToF system consisting only of fast, light-weight tracking planes. The UFSD are thin silicon sensors which achieve good SNR for a time resolution of 10 ps by employing internal gain based on modest charge multiplication [41]. Fig. 7 shows the predicted WEPL resolution of such a system with different ToF length. With a length of 1 m, the WEPL resolution will be below the range straggling limit.

Large area Si-Tracker: The coverage of the large scanner area requires tiling of 4 silicon sensors, since their size is limited by the 6" wafer size. Commercial sensors have ~1mm inactive edges which create image artifacts if sensors are butted against each other, or if they are overlapping. We are using S-C-P (scribe + cleave + passivate) treated silicon sensors with "slim edges" [42], where the dead area at the edge is cut away to within 50 μm of the guard ring.

4.3 Hand radiography with protons

One of the most famous pictures in medical history is the first X-ray image taken in 1895 by W.C. Röntgen of his wife Berta's hand, denoted "Hand mit Ringen" [43]. We duplicated this with our new imaging modality pCT by imaging a hand phantom with 200 MeV protons at the Loma Linda University synchrotron, using the existing pCT scanner [31]. Fig. 8 shows a map of the stopping power: note the varying thickness of the hand and the clear structural details. In addition, we show the map of the scattering angle, exhibiting very similar features [44]. We are working on combining the two maps in order to extract more information of the phantom material content beyond the stopping power.

5. Conclusions

At the VCI 2013 two sessions on medical instrumentation and a large number of posters were an indication of the growing interest in this field. With heritage from HEP and Astrophysics, medical applications are supported by active targeted R&D efforts.

New scintillators are being developed, as well as impressive application-specific modes for read-out. The advent of SiPM/MPPC opens new applications.

The increasing focus on multimodality with PET/CT and PET/MR, combining high-resolution structural delineation and localization with the organ specific information, recalls the success of the multi-wavelength campaigns in Astrophysics.

Instrumentation in support of hadron therapy is pursued both with interaction vertexing and proton CT to reduce the uncertainty in treatment planning and delivery and will find application at MedAustron [45]. Another focus is directed towards precision time

measurements, elevating the 4-th dimension to equal status, and providing a tool for remote sensing.

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Figure 1.

Photon attenuation coefficient μ (in units of $1/\text{cm}$) 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: $300 \mu\text{m}$ for Si, 1 mm for CdTe and 2 cm for CsI [1].

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

Figure 3.
Longitudinally segmented crystal with MPPC readout to improve DoI resolution [12].

Figure 4. Simulation of depth profile of generated and detected vertices and of the dose in carbon interaction [23].

Figure 5. Model of the upgrade UCSC/LLU head scanner with (from left) the front Si-tracker telescope, the phantom, the rear Si-tracker telescope, and the scintillator energy (WEPL) detector.

Figure 6. Measured WEPL resolution for different degrader thickness for the WEPL detector choices: the existing CsI calorimeter [31], a range counter with 4 mm plates [39], one MSS block measured “head-on” labeled “calorimeter” [40], and a 3-stage MSS [40].

Figure 7. Projected WEPL resolution using a 4D-UFSD ToF system with 10 ps time resolution for different ToF path lengths [41].

Figure 8. Radiograph of a hand phantom taken with 200 MeV protons. Left: Stopping power map; right: multiple scattering map [44].

Table 1

Tracker types and their respective position resolutions and energy detector technologies used in different pCT experiments.

Experiment	Tracker	Energy Detector
TERA/CERN [28][29]	GEM $\sim 100\ \mu\text{m}$	Range (3mm) + WLSF + MPPC
Firenze/LNS [30]	SI SSD $80\ \mu\text{m}$	Fast crystal calorimeter + P.D.
LLU/UCSC/NIU [31]	Si SSD $80\ \mu\text{m}$	CsI + P.D.
NIU/FNAL [32]	SciFi + MPCC $0.3\text{-}0.5\ \mu\text{m}$	Range (3mm) + WLSF + MPCC
UCSC/LLU [33]	Si SSD $80\ \mu\text{m}$ "Slim edges"	Polystyrene Calorimeter + PMT