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PHOTON DETECTORS FOR POSITRON EMISSION TOMOGRAPHY

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Abstract — The paper describes the construction and tests of 511 keV photon detectors whose performance was investigated with the aim of using such detectors in an apparatus for positron emission tomography. Monte Carlo simulation calculations are also described which contribute to the understanding the influence of the underlying physical phenomena on detector characteristics.

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Introduction — Positron Emission Tomography (PET) is a nondestructive method for imaging of the human brain. A beta plus (positron) radioactive isotope is chemically bound to a substance with specific metabolic properties. This substance is then injected into the vascular system and its distribution is measured. The knowledge of this distribution gives information about local and specific biochemical functions of the observed tissue.

The distribution of the positron emitters is determined by detection of photon pairs which are created by anihilation of positrons in the tissue. The positron usually anihilates at a distance less than about 1 mm from the decaying isotope. The so created photons have an energy of 511 keV and fly off in almost opposite directions at an angle of 180 ± 0.5 degrees. By simultaneous detection of both photons and the measurement of coordinates of hits on two opposite detector planes, one may determine a straight line near which the isotope has decayed (fig. 1). By the detection of a large number of photon pairs and determination of straight lines it is possible to reconstruct a three dimensional image of positron emitters in the tissue.

In order to minimise the radiation dose received by the patient, the detectors should have a good efficiency and subtend a large solid angle. In order to obtain a clear reconstructed image it should be possible to accurately determine the positions of photon hits (good position resolution). Random coincidences, which represent noise and thus influence image contrast, are minimised by good time resolution of the detectors and readout system.

The choice of isotope is dictated mainly by two requirements.

i) the possibility and simplicity of procedure of attaching the isotope to a biological substance with given metabolic properties. If not a natural substitute, the isotope should leave the organism after the measurement.

ii) a short half-life, which should not be much longer than the time required to make a tomographic measurement, that is not longer than about 10—20 minutes,

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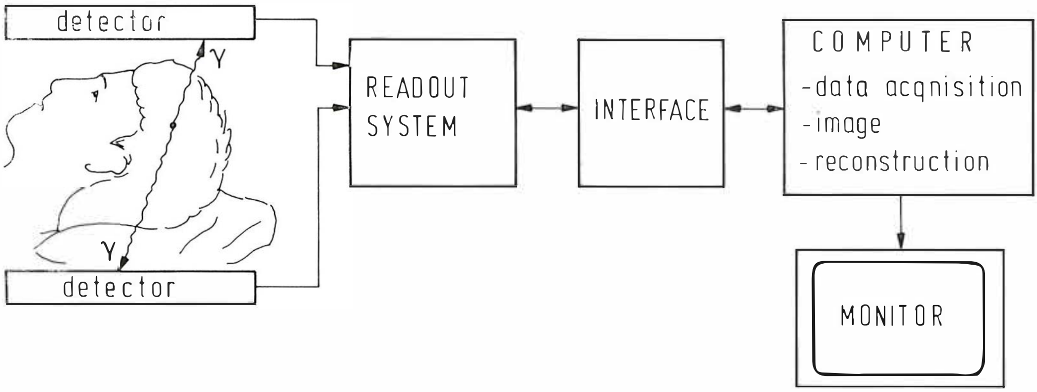

Fig. 1 — The distribution of positron emmiters is measured with the aid of two photon detectors. Data are transfered to a computer where theimage is reconstructed and displayed on the monitor

since after this it only increases the radiation dose.

Some suitable isotopes are listed in table 1.

Detectors — A multiwire proportional chamber (MWPC) equipped with a high density drift space (a photon converter) has two advantages over standardly used scintillation detectors: (i) the spatial resolution of the order of 1 mm, and (ii) a much lower cost for a larger detector area. The disadvatages are that it has lower efficiency and somewhat poorer time resolution. The lower efficiency can be compensated by larger acceptance, i.e. larger detector area.

The detector for 511 keV gamma rays consists of a photon converter, and an anode wire plane sandwiched between two cathode wire planes (fig. 2). The converter is usually made of lead or lead glass. The high atomic number and high density of lead provide high photon absorption in a relatively thin converter. A good way of manufacturing the converter had been proposed by A. P. Jeavons [1]. A sandwich is made from 0.2 mm thick lead foils and 0.1 thick sheets of fibre-glass reinforced epoxy-resin. Holes of 1 mm diameter are then drilled through it in a hexagonal pattern with a pitch of 1.155 mm. Each lead foil is connected to its own electric potential so an electric field is maintained inside the holes (fig. 3).

Incoming photons are absorbed in the converter producing free electrons, which have enough energy to escape from the converter material and penetrate onto a hole, where they ionise the gas. The secondary electrons then drift through the electric feld toward the nearest anode

Isotope	Max. e* energy	Half-life	Production source
Ga-68	1.90 MeV	68 min	daughter of Ge-68
C-11	0.98 MeV	20 min	cyclotron
N-13	1.24 MeV	10 min	cyclotron
O-15	1.68 MeV	2 min	cyclotron
F-18	0.65 MeV	110 min	reactor and cyclotron

Table 1 — Some isotopes suitable for PET

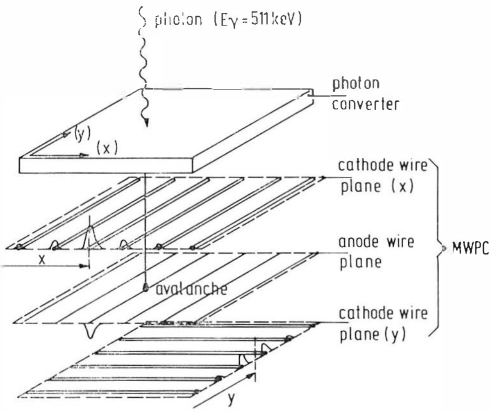

Fig. 2 — Exploded view of the detector

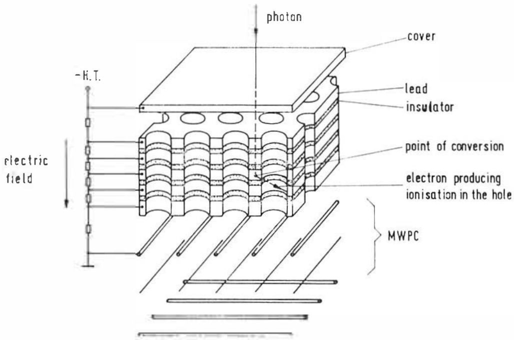

Fig. 3 — Detail of the converter

wire, where they are multiplied in an avalanche. A negative electric signal develops on the anode, and positive signals on the cathode wires. The cathode signal is spread over several wires with a maximum on the cathode wire nearest to the avalanche, as shown in fig. 2. Both x and y coordinates of the avalanche are read from the cathode planes. One cathode plane has wires parallel and the other plane perpendicular to the anode wires. The anode signal is used for the coincidence.

Typically, the anode wires are oh 20 µm diameter and are spaced by 2—3 mm, while cathode wires have a diameter of 100 µm and are spaced by 1 mm. The gap between anode and cathode planes is typically 3—5 mm.

The localisation of the photon absorption point in the direction perpendicular to the anode wires is determined by anode wire spacing, but in the anode wire direction it can be much better. The MWPC is oriented in such a way that the direction of inferior resolution is perpendicular to the tomogram slice, as the thickness of the slice is usually chosen to be several millimeters. The resolution of the detector in the direction of the superior resolution is limited by the hole diameter, i.e. typically 1 mm.

The time resolution is determined by the time secondary electrons need to drift through the converter, and it is a function of the converter thickness and the gas used. With lower converter thickness one obtains better time resolution, but lower efficiency. The problem can be solved by taking several converters, each say of 6 mm thickness, equipped with its own MWPC [1]).

The caracteristics of the detector [1] are given in table 2. If the detector is made of eight such converters, the efficiency can be about 20 %.

Dimensions	200 × 200 mm^2
Number of converters	2
Thickness of converter	6 mm
Hole diameter	1 mm
Pitch	1.155 mm
Efficiency	7.5 %
Spatial resolution	1.2 mm
Gas filling	82 % Ne + 16 % CO + + 2 % Isopropyl alcohol
Time resolution	20 ns

Table 2 — Characteristics of the detector of ref. 1

At the moment we are designing a prototype PET system. We constructed and tested two small MWPCs with an active area of 32 × 32 mm^2. Each chamber is equipped with an anode wire plane sandwiched between two cathode wire planes. The MWPCs are designed in such a way that exchanging of converters is possible without decomposing the chamber.

We also constructed a readout system based on a delay line, which is suitable due to the small number of electronic components. The cathode signals from each wire are successively fed into the delay line, where they add up into a signal whose time of propagation along the line is proportional to the position of the detected photon. The time reference for measuring the propagation time is the anode signal. Our delay line has a delay of 16 ns/nm, and a spatial resolution of 2 mm.

A converter was made from printed circuit boards. Four such boards of 1.6 mm thickness were glued together and holes of 1 mm diameter were drilled in a quadratic pattern of 1.41 mm pitch. From preliminary calculations it follows that even with this low-atomic-number material relatively good efficiency could be achieved. The converter was tested with 661 keV gamma rays from Cs-137 for which the efficiency was 2 %. By optimising the geometry, we expect to obtain about two times higher efficiency for this type of converter.

Performance of the apparatus — The over-all spatial resolution on the tomogram is determined by several factors: the MWPC resolution, the parallax error due to the converter thickness, the departure from collinearity of the two gamma rays, and the range of the positron in the tissue before anihilation.

Each of the factors was studied separately with a Monte Carlo simulation program using a 20 cm diameter water filled phantom with a point source in the centre. The apparatus consists of four 300 × 300 mm^2 MWPCs enclosing the space around the phantom. The spatial resolution of the MWPCs is take to be 1 mm. The simulation program starts with the determination of the positron anihilation point with positron range generated within a three-dimensional Gaussian distribution with $\sigma = 1$ mm. Anihilation photons are isotropically distributed within the full solid angle and with a small departure from collinearity given by a Gaussian distribution with $\sigma = 8$ mrad. If one of the gamma rays is Compton scattered in the phantom, a new direction is generated according to the Klein-Nishina differential cross-section [2]. The event is accepted if the two gamma rays hit opposite detectors.

The conversion point inside the converphoton direction. The converter and MWPC spatial resolutions are taken into account ter is generated according to the exponential probability function along the incident by assuming that the measured hit correspond to x and y-coordinates of the nearest converter hole or MWPC wire. The z-coordinate of the conversion point (the depth in the converter) is also not measured so it is taken to be at the central converter plane. This is the origin of the parallax error. The contributions of each of these effects and the overall resolution are given in table 3.

Among accepted events it was found that the relative number of events with one of the photons being scattered is over 50 % in our case. The number of random events depends on the time resolution of the detectors and on the counting rate. This was simulated by combining two gamma rays from different events. It was shown that both effects, Compton scattering in the phantom and random coincidences, result in an almost flat background distribution and thus do not influence the spatial resolution but only reduce the image contrast.

Positron range (mm)	Photon acol-linearity (mm)	Parallax error (mm)	MWPC/ converter resolution (mm)	Over-all resolution (mm)
0.84	0.96	0.52	1.0	1.7*
0.84	0.96	1.00	1.0	1.9**

Table 3 — Contributions to the spatial resolution and the over-all resolution for 4 mm thick converter* and 8 mm thick converter**

Conclusion

We have constructed and tested position sensitive detectors and a delay line readout system for detection of 511 keV photons from positron anihilation. The performance agrees with simple estimates and Monte Carlo simulation calculations of the apparatus. The next step is to connect the readout to a computer and test the entire system and image reconstruction programs in a more realistic environment.

References

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