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DESIGN CONSIDERATION FOR PET CAMERA USING SOLID SCINTILLATOR PROPORTIONAL COUNTER

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Abstract

A solid scintillator proportional counter consisting of BaF_2 - TMAE combination is well suited to the detection of 511ke V annihilation radiation for positron emission tomography (PET). Common gamma cameras for PET consists of greater number of scintillation crystals positioned in rings, readout with photomultipliers. The electronics of such systems is quite complicated and rather expensive. The position sensitive detection of scintillation light using a wire chamber is much more attractive. The SSPC combines the merits of a low pressure MWPC as a position sensitive photoelectron detector. Here we present some simple design consideration for a PET camera based on a solid scintillator proportional counter.

Key Words: Solid Scintillator Proportional Counter, PET Camera.

Introduction

The intrinsic resolution of a PET camera is determined by the positron range and the acolinearity of the two photons. These effects are studied e.g. by Derenzo et al. [1, 2, 3]. The projected point spread function of the positron range for different isotopes in water is:

^{18}F	0.23mm
^{12}C	0.39mm
^{68}Ga	1.23mm
^{82}Rb	2.6mm

These numbers give the square root of the average of the positron range projected on a plane squared. The isotopes ^{15}O and ^{13}N produce positrons with an energy intermediate between ^{11}C and ^{68}Ga, and have hence an average positron range between the values for these two last isotopes. The deviations from 180^0 in the angle between both photons for the positron annihilation in water is a nearly Gaussain distribution with an r.m.s. of 3. 5 mrad. If the distance between the two photon detectors is D, the r.m.s. of the positron measurement error induced by this is $3.5\ 10^{-3}$ D/4. The minimum value of D is of the order of 40 cm, and the contribution of this effect to the intrinsic resolution of a PET camera is hence comparable to the range effect in flour. Combining both effects we hence have for the position measurement error (r.m.s.) projected on axis.

^{18}F	0.39 mm
^{12}C	0.45 mm
^{68}Ga	0.94 mm
^{82}Rb	1.87 mm

The ultimate position resolution of a PET camera is thus given by the above numbers. The resolution is also limited by the unavoidable movement of the patient.

For tomography of e.g. the heart this will probably limit the resolution to1mm rms.

The detector degrades the resolution further due to: (1) size of the crystal or localisation error of the scintillation light in a crystal, (2) parallax eror, (3) Compton scattering in the detector. A good detector should degrade the resolution as little as possible, but equally important it should

allow a high effective counting rate and be of an affordable price.

Modern ring cameras (see Dereonzo and Budinger [2] for a review) use BGO crystals of 3 to 6 mm wide and 10 to 20 mm high. Each crystal is coupled to its photomultiplier. This allows a resolution of 1mm rms in the plane of the ring and 2 to 4mm rms axically.The resolution on the difference of the arrival times of both photons is typically 10ns. Some modern ring cameras use BaF_2 or CsF to take advantage of the improved effective counting rate made possible by the excellent timing properties of these scintillators. Time resolutions of 400ps FWHM have been reported for BaF_2.

This is the competition we face. Below we present some simple design consideration for a PET camera based on BaF_2 and photosensitive wire chambers.

2. The Parallax error

In this section we examine the constraints on the design due to the parallax error. Consider the camera represented in fig 1. It consists of two plane detectors aligned with the centre of the object to be imaged. Each plane is assembled from small crystals of length L. if the photon converts a distance l away from the average depth in the crystal the error in the image plane is given by (see fig. 1):

$$\Delta = \ell H / l$$

Averaging over H between O and H_{max} gives:

$$\sigma(\Delta) = \frac{\sqrt{2)}\, H_{max}\ \sigma(\ell)}{D\sqrt{3}}$$

The factor $\sqrt{2}$ comes from the two crystals which detect each one of the photons. If the radiation length in BaF_2 is λ and L the length of the rod, then the average conversion depth and dispersion of this conversion depth are given by.

$$<\ell> = \lambda \left[1 - \frac{L/\lambda e^{-L/l}}{1 - e^{-L/\lambda}} \right]$$

$$\sigma[\ell] = \lambda \left[1 - \frac{(L/\lambda)^2\, e^{-L/\lambda}}{(1 - e^{-L/\lambda})^2} \right]^{1/2}$$

With L = 50mm and λ = 2mm we have σ = 12.66mm. Note that this is very close to the result for $\lambda >> L$: $L / \sqrt{12}$ = 14.4mm as expected.

Consider a PET camera to study the human brain consisting of two plane 30 x 30 cm^2 detectors made out of 5 x 5 x 50 mm^3 BaF_2 bars. The detectors are 40cm apart. The detectors cannot be smaller since they should be larger than the object to be imaged in order to have an acceptable counting rate everywhere in the object. The error due to the parallax is 3.87mm rms in the centre of the image and better away from the centre. This is clearly unacceptable. Possible remedies are:

1. Increase the distance D between the detectors. This is not attractive idea. To maintain the same solid angle the size of the detector area should increase like D^2, and the cost is proportional to the detector area. The cost of such a detector will quickly become prohibitive due to the high cost of the crystals.

2. Use shorter crystals at the expense of the detection efficiency. A parallax error of 1mm required crystals which are only 10mm long. The detection efficiency is then 38%. Several effects will reduce this further, and we will probably end up with less then 25% overall efficiency.

3. Use a large number of small detector modules as shown in fig 2. The BaF_2

crystals can then be oriented such as to point to wards the centre of the image. In the centre of the ring the parallax error is than quite small but away from the centre there is another contribution to the parallax error given by (see fig. 2) :

$$\sigma(\Delta) = \sigma(\ell)\,\theta/\sqrt{8}$$

Where θ is the angle between the orientations the two crystals which detect the photons.

Consider now a ring of detector modules of 5 x 20cm^2 each. The modules are arranged in a ring as shown in figure 2 and the long side of the modules is parallel to the axis of the ring. The crystals in each detector module are 5 x 5 x 50mm^3 and are oriented towards the centre of the ring with D = 40cm. In the centre of the object parallex error is 0.64mm rms. At 10cm away from the parallax error depends on the direction of the annihilatin photons and can be up to 2mm rms. Notice also that the parallax error in the direction of the axis of the detector is 2.6 mm rsm.

The conclusion of this exercise is clear: if there is no measurement of how deep the 511KeV photon penetrates in the crystal we must use a ring or cylinder shape for the detector.

4. Time resolution and effective counting rate

The need for the best possible time resolution in PET is paramount. If the time resolution of the detector can be reduce to 1ns it results in a reduced number of accidental coincidences. The importance of this effect is illustrated by the Monte Carlo simulation results presented below. The calculation closely follows ref. 4. In this calculation we evaluate the influence of these accidental coincidences on the image quality as a function of the dose in the object to be imaged. These accidentals generate a more or less uniform background which reduces the effective counting rate. Figure 3 presents the effective counting rate Q defined as

$$Q = \frac{(\#\text{ true coincidences})^2}{(\#\text{ true coincidences}) + (\#\text{ accidental coincidences})}$$

as a function of the source activity. The dotted line is for a ring camera with solid angle of 0.054 x 4π and the solid line for rhypothetical camera with a solid angle of 0.64 x 4π. More information on this Monte Carlo simulation is given in appendix 1. In medical applications the source activity is typically in the range 0.1 to 1μ Ci / cm^3. At the lower end of this range there is only a small difference in counting rate for a difference resolution of 1ns or 10ns FWHM. At the higher end the ratio is a factor 4. This result is certainly dependent on the details of the Monte Carlo calculation but the trend should be genuine.

With BaF_2 and photomulitiers it is possible to measure the difference of the arrival times of the two photons with an accuracy of 400ps FWHM. This allows the determination of the emission point along the line of flight of the photons to be determined with a resolution of 6cm FWHM. This time of flight information reduces the mean square error (or variance) on the reconstructed positron emittor activity in the image by the ratio of the maximum time difference in the object over the time resolution FWHM [6,7,8]. The data time for a given statistical uncertainity in the image is then reduced by the same ratio. With 400ps FWHM resolution on the difference of the arrival times the data taking time for an image of the head is reduced by a factor 3 or 4.

The effect of an improved time resolution was studied experimentally by Yamamoto et al. [8] for the super PET 1. This device uses CsF scintillating crystals with a decay time of 2.5ns. A reduction of the time difference resolution from 12 to 0.5ns resulted in an increase of the effective counting rate by a factor 3.4 and 8.3 for a source activity of 0.12 or 1.45μCi / cm^3 respectively. These results confirm the previous general statement and under line the importance of a good time resolution.

5. Possible designs for a new PET camera

Because of the parallax error a PET camera base on BaF_2 and wire chambers must consist of a number of small plane detector modules assembled in a ring as illustrated in figure 2. Each module should be less than 10 x 20 cm^2 in sensitive area.

Figure 4 shows a possible detector module. The complete ring has 24 such modules. Each module contains 200 BaF_2 crystals 5 x 5 x 50mm^3. The wire chambers operate at low pressure in pure TMAE at 40^0 C. the resolution on the time difference in such a system is expected to be between 5 and 10ns FWHM, slightly better than a BGO crystals with 5 x 20mm^2 area:

- the time resolution for the BaF_2 camera is better by a factor 2.

- the resolution in the plane of the ring is worse because of compton scattering and the smaller density of BaF_2; the magnitude of this effect is difficult to evalute without a detailed study but 30%is a reasonable guess.

- the axis resolution is better by a factor 3

- the solid angle is 5 times bigger; if the BGO camera is made into a 5 ring system it has the same solid angle but will be very expensive.

- the BaF_2 cameras has almost no energy resolution; this will make the rejection of compton scattered photons less effective; again only a detailed study will allow to evaluate the importance of this effect.

The solution looks attractive. It is simple and probably competitive with current commercial ring designs. Unfortunately it does not take full advantage of the excellent timing characteristics of BaF_2.

Figure 5 shows a module which combines the advantages of wire chambers and photomultiliers. The photomultilier provides the time and energy resolution and the wire chamber identifies the crystal which is hit. A total of 256 crystals in a 8x8cm square fit on window of a standard 5 inch tube, e.g. the EMI 9823QB. The wire chamber can be at atmospheric pressure since there is no longer a need for good timing in the wire chamber. A ring can e.g. contain 12 such modules.

Conclusion

The two solutions proposed here have advantages and it is difficult to decide which one is to be preferred. The decision depends in how well each system will really perform and on the applications considered. For this we have to build a protype of each solution and discuss with the potential users

Appendix 1
Some details on the monte carlo calculations of the effective counting rate.

The phantom is a cylinder of water 10.75cm in diameter and 12cm high with a uniform activity. Compton scattering in the phantom is taken into account. The

read out dead time is 1 μs. The ring detector has an inner radius of 26cm, a crystal height of 2.8cm and is equipped with lead shields. The detection efficiency is 0.70 for Eγ > 300 KeV and 0.0 for Eγ< 300KeV. The hypothetical camera consists of 2 detectors 45 cm apart and each with an area of 43 x 43cm^2. It has no lead shields. The efficiency is 0.45 for Eγ > 140 KeV and 0.0 for Eγ < 140 KeV.

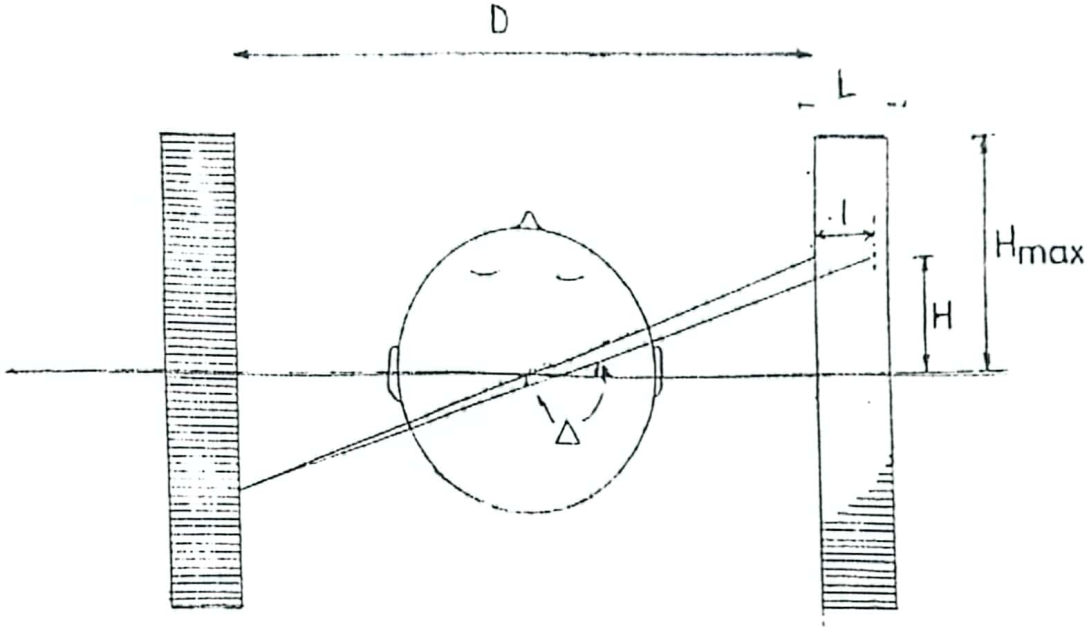


Fig: 1. The parallax error for two plane detectors geometry.

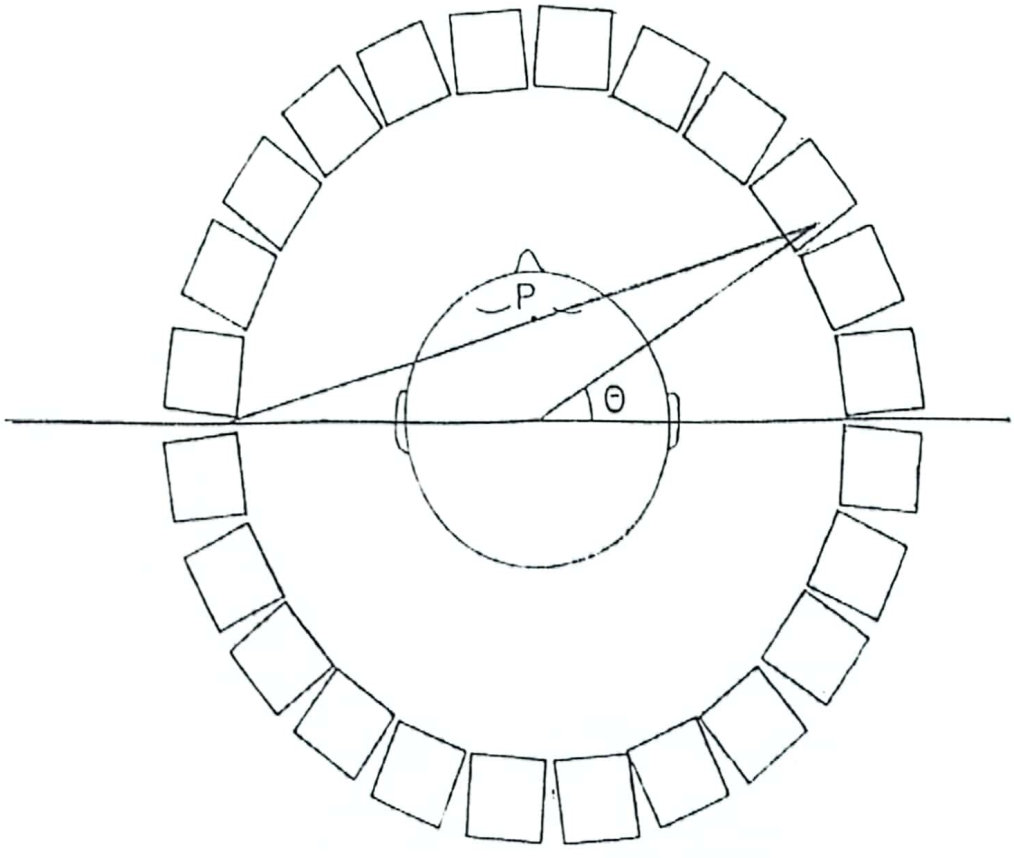


Fig: 2. PET ring using a large number of small detector modules.

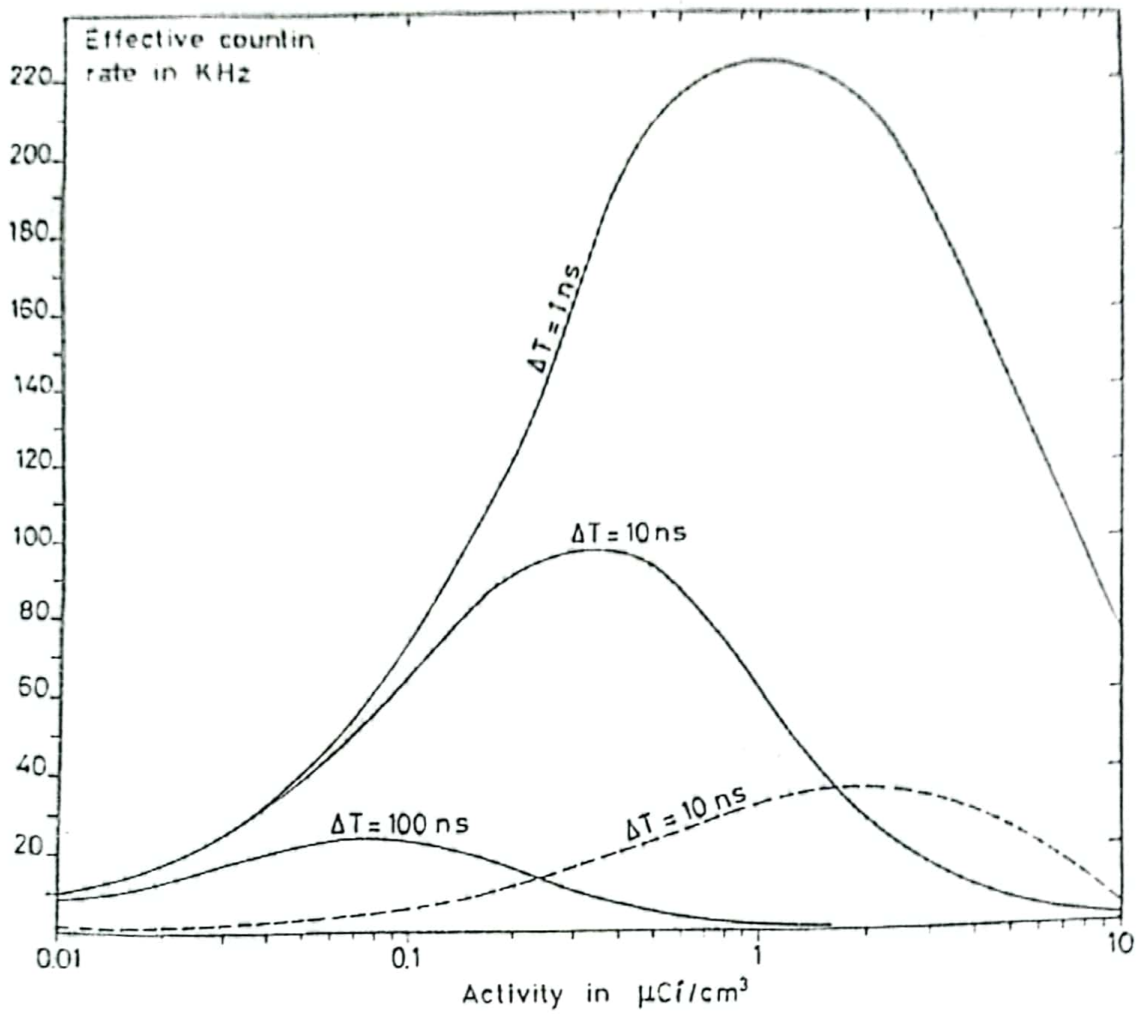


Fig: 3. The effective counting rate as the function of the source activity.

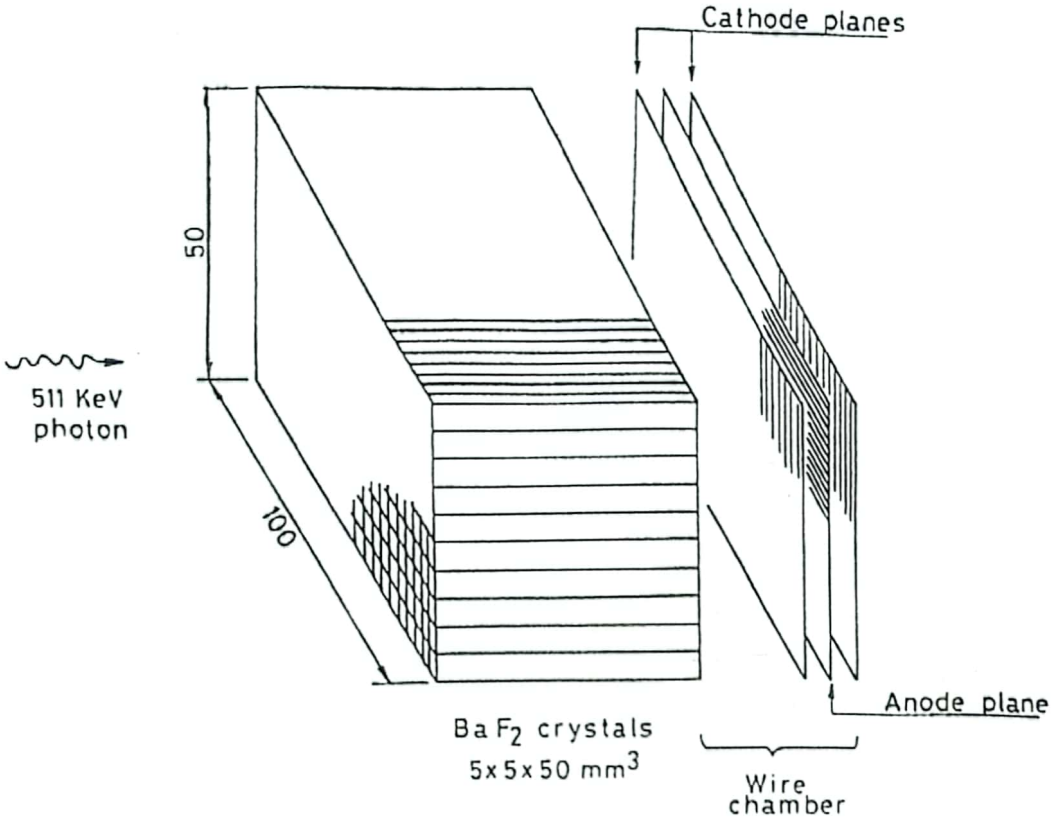


Fig: 4. A possible detector module.

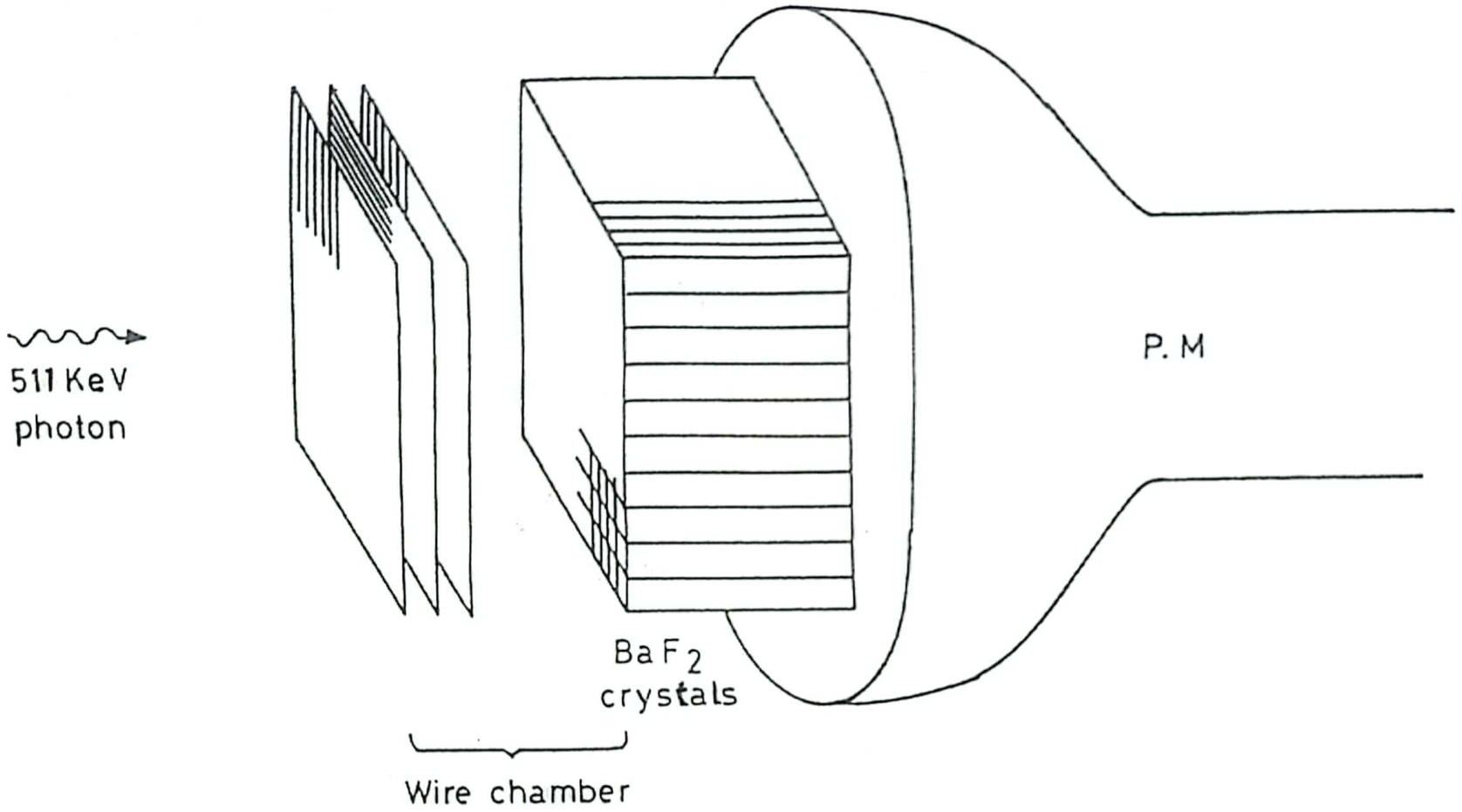


Fig: 5. Module which combines the advantages of wire chambers and photomultipliers.

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