RPC: from High Energy Physics to Positron Emission Tomography

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Abstract: A low cost gas-based charged particle detector, the Resistive Plate Counter (RPC) intensively used in fixed target and collider high energy experiments, is proposed as basic detector for Positron Emission Tomography. The performance of RPCs in terms of intrinsic space and time resolution and electronic pulse height response, makes it possible to transform standard RPCs into photon detectors and therefore to compensate for the photon sensitivity of scintillating crystals, when the efficiency of the complex crystal + photomultiplier is turned into standard quantum efficiency (q.e). Prototype multigap glass RPCs were developed which optimize $\gamma$ detection efficiency and thus might substitute the traditional scintillators setups.

1. Introduction.

Gas avalanche detectors, employed in particle physics research, have several applications in life sciences. In medical diagnostics, gaseous detectors are currently employed in digital X-ray radiography[1a] and angiography[1b]. High-pressure gas ionization chambers have been used, e.g., X ray sensors in Computerized Tomography (CT). Since they have a rather limited sensitivity to energetic photons, methods were found to couple position-sensitive gas avalanche detectors to solid gamma converters, as in gamma cameras equipped with thick metal-grid converters[1c]. Such devices were routinely applied to medical inspection. Other applications of gaseous detectors as diagnostic tools can be found in biomedicine, in monitoring radiotherapy[1d], in radiation dosimetry, etc.

One of the most interesting new fields where their unique characteristic could be exploited is the Positron Emission Tomography (PET). Recently, gaseous detectors were employed in a high-resolution 3D small-animal Positron Emission Tomography (PET) imaging[2a]. Small-animal PET cameras were also developed, where UV photons from BaF2 crystals are detected in wire chambers operated with a photosensitive (TMAE ) gas[2b].

PET is a radiotracer imaging technique in which tracer compounds labeled with positron emitting radio-nuclides are injected into the object under investigation. After a short path length[3] the positron annihilates with an electron of the medium emitting simultaneously two (almost) anti-parallel 511 KeV photons. The detection of both photons (whose coupling is made through a check on their time of flight) identifies the occurrence of an annihilation along the chord connecting the detection points. Since the $\beta^+$ emitters are linked to some physiologic substrate such as glucose or oxygen, mapping the density of the positron sources after a while gives a measure of the rate of activity inside the human body. These tracer compounds are used to track biomedical and physiological processes, with applications ranging from the early detection of cancer to neurophysiology studies.

The development of image reconstruction techniques for the detection of very small fluctuations in
density (e.g., tumors at the early stages of their formation) requires both high-resolution, high-rate detectors as starting points. During the last 30 years much effort was made to develop scintillator-based PET detectors and, short of a great technological breakthrough, it seems that their limits have almost been reached. Research in this field focuses mainly on the growth of crystals with large light output, good space and time resolution. However, the possible outcome of this kind of research seems to be limited, while advanced gas-based detectors might be a promising alternative path to follow as long as they may provide excellent space ad time resolutions.

In this paper we describe a novel feasibility study to build a Resistive Plate Chamber (RPC) PET detector prototype with high efficiency, high space resolution and high time resolution and competitive overall performances. All the results shown were produced with the Geant simulation toolkit (version 4.7.1), the CLHEP libraries (version 1.9.2) and compiled with gcc 3.3.3 under Slackware Linux 10.2. Most of them are compared to experimental results obtained with a prototype multi-gap glass RPC of limited dimensions, exploited to the purpose of showing the maximum obtainable gamma detection efficiency and the general performances.

2. The Resistive Plate Chambers (RPC)
The RPC[4] is a gaseous detector that can identify charged particles with high time (1 ns down to 50 ps) and high space (down to 30 μm ) resolution. The basic RPC unit is one gas gap (SG) between two resistive plates (in the range $10^{10} - 10^{12} \Omega \text{cm}$) made of high-pressure plastic laminates (HPL), separated by insulator spacers. The primary ionization electrons, created by the passage of a charged particle into the gas, are multiplied into an avalanche by a high, uniform electric field of typically 4.5 kV/mm. The signal is readout via capacitive coupling by means of metallic strips on the external side of the electrodes (see Fig. 1a). The RPCs have a simple mechanical structure, use no wires and are easy to manufacture. Two such units can then be assembled together to form a Double Gap (DG) structure with common readout strips in between (Fig. 1b).

![Fig 1](image)

The electric field can be independently applied to the SG units so as to have three different operation modes according to the needs of the experimenters. The signal is capacitive inductive detected by the copper strips facing one or both electrodes. The signal of the DG structure is an analogue OR of the two separate SG signals. A DG can reach efficiencies close to 95-98% (Fig. 1c). Depending on the applied electric field the detector can work in avalanche or streamer mode, the two being distinguished by a different value of the generated charge inside the gas gap. Typical values for the induced charge are few picocoulombs in a RPC operated in avalanche mode while are a factor 50 higher for a streamer operated one. The latter has been mainly used in cosmic rays as well as in high-energy physics experiments. Avalanche mode can be used in applications where the detection of high particle fluxes is required. In this paper we assume an RPC operation in avalanche mode.
RPC using time resolutions in the picosecond region requires a new conceptual assembling method introduced in 1996[5] and called “multigap” (MRPC), in which more plates are inserted into the gas generating several gaps of smaller width within the detector. The conceptual improvement of the MRPC design is the introduction of electrically floating resistive electrodes segmenting the full gas volume into independent gas gaps some hundreds microns wide.

Fig. 2a shows the space resolution obtained in reconstructing a radioactive source with 2 opposite sets of 16 SG hybrid glass RPC[6a]; fig. 2b shows the contributions from different terms[6a]. Fig. 2c shows a typical time resolution (around 50 ps) in time of flight measurements with glass MRPCs[6b].

Fig. 2 a) space resolution of 16 SG hybrid glass RPCs[6a]; b) its decomposition[6a]; c) MRPC time of flight resolution[6b].

2. Scintillators vs RPCs.
It is relevant to compare the working principles of two hypothetical PET detectors exploited using classical crystal detectors and/or RPC gaseous detectors.

The limitation of most present PET imaging scanners is largely due to the high price of the crystal-based detectors. The scintillating crystals, due to the very strict requirements on their characteristics must be blessed with a very high detection efficiency, large light output and good time and energy resolutions. Thus, the major limit in the design of a scintillator-based PET detector is the strong dependence of its working parameters and on the physical features of every single piece of them. Usually crystals are no more than 25 mm thick in order to have a good space resolution (or their parallax and interaction depth uncertainties would be too large). They must have also both a good light output to compensate for the limited efficiency of the coupling to the photomultipliers and a good time and energy resolution to cut noise. Unfortunately, growing such highly specific crystals is still a very expensive process: BGO (Bi$_4$Ge$_3$O$_{12}$) crystals 25 mm thick (“commonly” used for PET detectors) cost some 100 $/cm^2$ odd, while more exotic crystals can have even a higher price. Even with such expensive crystals, the space resolution of the detector is limited by the parallax effect.

Furthermore, to reduce the photon rate far from the normal to the crystal surface, these detectors are commonly built in a cylindrical setup [90 cm eter, 30 cm field of view (FOV), for a total-body detector] where the patient is to lay still along the axial line, i.e. along the FOV. To limit the parallax incidence the crystals are separated by thin metal layers usually made of lead or tungsten (called inter-plane septa) whose depth can be modified to tune the efficiency for the rejection of non perpendicular photons. The septa can improve image contrast reducing the amount of scattered gammas up to 10 – 15 % of the total counts acquired[7] but the FOV reduces the geometrical acceptance of the detector. Depending upon the depth of the septa, the scans are said to be either 3D (almost no septa between crystals) or 2D
(fully separated crystal planes). As shown in Fig. 3 the signal to noise ratio greatly improves with a harsh cut (<45°) on the acceptance angle till it reaches a maximum; then it drops when the cut strongly reduces the data. The maximum of the function and the rise in its gradient depend heavily on the FOV. In any case the q.e of the system (scintillation efficiency of the crystal times the PMT photocathode efficiency) is around 13% for an NaI(Tl) crystal[8].

Search for new crystals has been extensive in the literature[9]; the main characteristics to be considered being: physical properties (mainly hygroscopy), interaction mean free path, light output and decay time. These parameters for tellurium activated sodium iodine (NaI), bismuth germanate (BGO) and lutetium ortosilicate (LSO) are summarized in Table I. Data show that q.e. is ~2% for BGO and ~11% for LSO.

Table I: main parameters of common scintillator materials used in PET scanners.

|                  | NaI(Tl) | BGO | LSO |
|------------------|---------|-----|-----|
| Density (g/cm³)  | 3.7     | 7.1 | 7.4 |
| Mean free path (cm) | 2.9     | 1.1 | 1.2 |
| Light output [NaI(Tl)=100] | 100 | 15 | 75 |
| Decay time (ns)  | 230     | 300 | 40  |
| Hygroscopic      | YES     | NO  | NO  |

On the contrary, first of all, an RPC costs of the order of some $100/m² odd, cost of electronics and power supply being similar in the two detectors. An RPC directly detects charged particles, needs no readout PMTs and is not particularly affected by parallax effects. Standard RPC are built as large as 3m x 3m; the signals produced by any charged particle crossing the gas gap induce into the readout strips a very fast pulse which can be as short as about 4-5 ns. Specific readout electronics[6c] has been able to locate the position of the avalanche with an accuracy of order few hundred microns. The fast response is a reason why RPC are extensively used[5b] for time of flight measurements in high energy experiments. For a PET RPC the crucial issue to be tackled is to maximize the q.e. of the detector, i.e., the overall percentage of detected 511 KeV γ's relative to the number of incoming γ's. The goal is pursued by adopting a MRPC and by inserting thin glass plates converting γ's into electrons, Compton effect being far the most important contribution (photoelectric effect is down by order of magnitude and pair production is forbidden).

3. Simulations

In principle a proper MRPC could substitute a crystal setup in a PET scanner. Here we intend to simulate some examples and to compare the output to the performance of some built prototype.

The “total” γ conversion probability is the product of the interaction probability of the γ in a single target layer of material (glass or lead glass) time the probability of letting the produced electrons into the gas gap above a give threshold setup. Electromagnetic interaction of γ with matter is very well known, can be numerically calculated and introduced in appropriate simulations. For sake of comparison to experimental data, the simulations[9e] have been done for 511 keV γ's, as well as for 661keV γ's, the γ energy of a 137Cs source. Fig.s 4 show the γ sensitivity εe (probability of photoelectrons produced inside and exiting a plate) vs. thickness.

[fig. 4. a) γ sensitivity εe vs. thickness for HPL plates at 661 keV γ's (upper curve) and 511 keV γ's (lower curve); b) same for glass plates; c) γ sensitivity εe for a MRPC with 20 lead glass plates.

Since any electron entering the gap will start an avalanche in the electric field, the γ sensitivities εe -above our threshold setup- in fig. 4a reach a stable maximum value at a thickness of about 400
microns for a single HPL “target” layer. Fig. 4b shows the same plots for glass plates, where the plateau is reached at about 200 μm; fig. 4c shows, as an example, that εe for 20 glass layers can reach the value of εe for BGO and ~20-25% of εe for LSO, comparable to the εe of BGO crystal based detectors. It will be a matter of convenience to choose the number of plates to be inserted into a MRPC, since we believe that some factor can be recovered in the fitting process by the better MRPC space and time resolutions.

4. Performance of MRPC prototypes

A number of MRPC prototypes have been built and tested. Investigations are in progress on attempts to dope HPL electrodes with high Z materials on high Z chemical mixtures to properly coat the RPC electrodes, however the results were not good since the dark current of the RPC was consistently too high to encourage continuing the experimentation. An attempt is being made to build electrodes and plates with high Z insulating new materials such as ceramic xerogels or similar compounds.

The most promising possibility at present seems to be assembling glass MRPCs maximizing the γ conversion in all their components (plates and electrodes).

In this paper we concentrate our attention on the performance of 2 MRPC using glass plates and electrodes; the only plate thicknesses available on the market are 0.15 and 0.4 mm. Common features are: a- high voltage (glued graphite tape) glass electrodes; b- 5 gas gaps spaced by means of 0.3 mm diameter nylon fishing line; c- detector enclosed in gas tight aluminium case filled with a C2H2F4 92.5%, SF6 2.5%; iso-C4H10 5% gas mixture. Prototype MG-I: 4 glass plates 0.15 mm thick, 1 mm thick electrodes. Prototype MG-II: 4 glass plates as well as electrodes 0.4 mm thick. Fig.s 5 show a detailed view of the prototype 2cm x 5 cm in size (fig. 5a); the readout strips used (fig. 5b), and a visualization of simulated reconstructed tracks (fig. 5c).

![Image](image_url)

Fig. 5. a- glass MRPC prototype being assembled; b- 2 mm readout strips facing the detector; c- visualization of simulated tracks in aluminum case.

The MRPCs were exposed to a 137Cs, 5 μCi, point-like source. The major results are shown in fig.s 6. Fig. 6a shows the background subtracted counting rate vs. applied HV for a 2.5 mV threshold, while fig 6b compares simulated and real count rates vs. threshold, normalized to the 137Cs source activity. Fig. 6c shows the effect of electrode thickness on MRPC efficiency (thinner electrodes work at lower HV values although the efficiency seems to stay around 90-95%). This particular issue deserves further detailed investigation. It can be noted that the simulation programs adequately reproduce the behaviour of the real data for 661 keV γ-rays (fig. 6b) and there are no reasons to believe that the same should not occur for 511 keV γ from e+e- annihilation not available for direct experimentation.

5. Conclusions

In conclusion the overall performance of a glass MRPC can reach limits competitive with those of crystal-based PET scanners at a lower cost. The simulations are well compatible with the real data and the performance of thin glass MRPCs is well under control. The image reconstruction is an issue in principle independent from the specific technique employed. The signals readout from an RPC might be simpler to handle, being free from parallax effects; the coincidence of 2 opposite detectors could be faster and the time window shorter.
Fig. 6. a- background subtracted counting rates vs. applied HV (2.5 mV threshold); b- simulated and real performance vs. thresholds (data Δ: at 2.5; 3.5 ; 4.5 mV); c- MRPC efficiency using electrodes 0.4 mm and 1mm thick, respectively.

Whether or not this technology could be employed for a full body PET tomography rather than for small animal tomography is still to be proven. However, the MRPC approach seems to be rather promising and makes it worthwhile to continue investigations in this direction.

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References
[1] a- E.A. Babichev et al.: Nucl. Instr. and Meth. A419 (1998) 290; b- M. Lohmann et al.: Nucl. Instr. Meth. A419 (1998) 276; c- A. Jeavons et al.: Nucl. Instr. Meth. 124 (1975) 491; d- A. Jeavons et al.: IEEE Trans. Nucl. Sci. NS-46 (1999) 468.
[2] a- A. Brahme et al.: Nucl. Instr. Meth. A454 (2000) 136; b- P. Bruyndonckx et al.: ibidem A392 (1997) 407.
[3] S.E. Derenzo: Positron Annihilation (Ed. The Japan Inst. of Metals, 1979).
[4] R. Santonico and R Cardarelli: Nucl. Instr. Meth. 187 (1981) 377.
[5] Cerron Zeballos et al.: Nucl. Instr. and Meth. A374 (1996) 132.
[6] a- A. Blanco et al.: Resolution studies on a small animal RPC/PET prototype, Proc. VIIth Int. Workshop on RPC and Related Detectors, (Ed. World Sci., Singapore, 2005) to appear; b- G. Scioli et al.: The MRPC detector for the Alice time of flight, ibid.; c- R. Cardarelli et al.: Spatial resolution in RPCs, ibid.
[7] F. H. Fahey: Radiol. Clin. N. Am. 39 (2001) 919.
[8] see for instance G.F. Knoll: Radiation Detection and Measurement (Ed. J. Wiley & Son., New York, III ed., 2000) p.334.
[9] a; B. Bendriem et al.: Rev de l’ACOMEN (1999) vol. 5 n.2; b- W.W. Moses et al.: Trans. Nucl. Sci. NS-39 (1992) 1190; c- ] F. W. K. Firk: Nucl. Instr. Meth. A297 (1990) 532; d- Monica M. Necchi: Sci. Acta 21 (2005) 94;e- G. Sani: Sci. Acta 21 (2005) 133.