# … of a Novel PET Imaging System, Based on Resistive-Plate Chambers

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# Development of a Novel PET Imaging System, Based on Resistive-Plate Chambers (RPC)

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**Abstract.** The Resistive Plate Chambers (RPC) are charged-particle detectors with excellent spatial and time resolution. Transforming them into gamma-quanta detectors opens the way towards their application as a basic element of a hybrid imaging system, which combines Positron Emission Tomography (PET) with Magnetic Resonance Imaging (MRI). We present results from the optimization of the RPC construction by means of GEANT4 simulations. Several different detector designs and converter materials are investigated to meet the objectives for a prospective RPCPET detector: maximal electron yield for 511 *KeV* photons and reduced efficiency for registration of lower-energy scattered photons. The efficiency of a multi-gap RPC detector is studied.

Keywords: resistive plate chambers, optimization, positron emission tomography
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## INTRODUCTION

Positron Emission Tomography (PET) is a nuclear-medicine imaging technique for registration of the whole-body distribution of positron-emitting biomarkers [1, 2]. Different positron emitting forms of common elements with a short half-life, such as $^{11}C$ (~20 *min*), $^{13}N$ (~10 *min*), $^{15}O$ (~2 *min*), and $^{18}F$ (~110 *min*), are used to mark biologically active molecules participating in the investigated process. When injected, these radionuclides or radiopharmaceuticals undergo $\beta+$ decay; the emitted positrons pass some minimal distance in the body before an annihilation with a body electron takes place, resulting in the production of pairs of 511 *KeV* $\gamma$-quanta emitted simultaneously at nearly 180°. The main advantage of PET is its high sensitivity in identifying tissue damages and functional disorders at an early stage; long before anatomical structure changes or disease symptoms come to being.

The poor spatial resolution of the PET image is compensated by the complementary information, usually obtained by a consequent Computerized-Tomography (CT) scan. The problem is that there are physical limitations in the PET image reconstruction accuracy, Fig.1. First, there are random coincidences, when two photons from different annihilation events are detected "simultaneously". The number of the random coincidences is proportional to the detector time window (the time period in which two registered photons are considered originating from the same annihilation event), thus it is determined by the detector time resolution. Next, the photon scattering in the human body may cause a false determination of the line on which the annihilation takes place. This error is proportional to the detector sensitivity to photons with energies lower than 511 *KeV*, therefore the effective suppression of these scattered photons is crucial. Finally, the so called parallax error heavily depends on the detector spatial resolution as it accounts for the final size of the individual detector elements (the detector "point").

In the present PET scanners the photons are usually detected by scintillation detectors. The high cost of the scintillation crystals hampers construction of detectors with large field of view, thus requiring longer examination and higher radionuclide doses. Another serious problem of the scintillator-based PET scanners are the obstacles to the MRI/PET (Magnetic-Resonance Imaging) melding, the latter being considered by many experts to become a major breakthrough in clinical practice [3]. The importance of this diagnostic method and the objective physical and technological difficulties boost the research activities in the field of PET detectors. Most of the efforts in the

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development of PET detector technologies are focused on scintillator-based detectors, as described above, wherein various alternatives continue to be considered, among them wire chambers and solid-state devices for very high spatial resolution applications.

We focus on another alternative – the development of a PET scanner for human-body imaging, based on Resistive Plate Chambers (RPC)[4]. These are gaseous parallel-plate charged-particle detectors with high resistivity ($10^{10}$–$10^{11}$ *Ωcm*), widely used in large-scale high energy physics experiments. RPC's main advantages are their excellent time and spatial resolution and the ability to work in strong magnetic fields (as those in MRI), thus they would provide an improvement in the image accuracy, but also better prerequisites for a multi-modality MRI/PET single-device scanner.

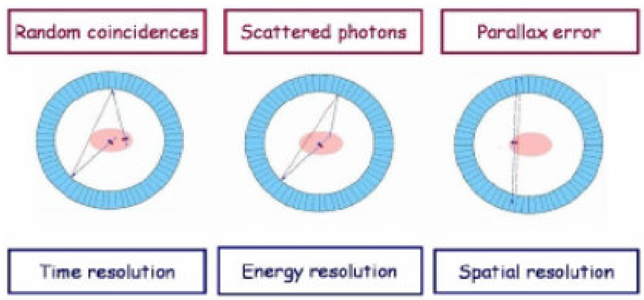


**FIGURE 1.** Error sources in PET image reconstruction.

A modification of the RPCs has been suggested in Refs. [5, 6], opening the way for their application in PET systems, that is – as gamma quanta detectors. The idea is to choose appropriate materials for one of the electrodes, thus transforming it into a gamma-to-electron converter. The schematic RPCPET design is shown on Fig. 2. In the detector, the gamma quanta from the positron annihilation interact with the converter medium and the ejected electrons pass to the gas gap, where they induce ionization avalanches, which trigger the detector signal.

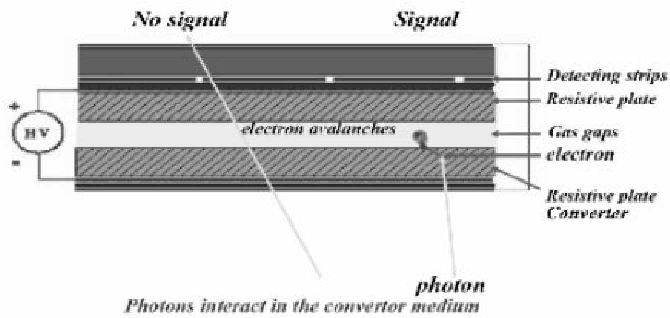


**FIGURE 2.** General construction of RPCs as PET detectors.

Based on [7], a small-animal PET system with timing RPC technology has been built and tested [8]. The tests confirmed the expectations from the simulations, the system revealing a space resolution of 0.6 *mm* FWHM without optimization of detector parameters. Progress in the improvement of the time resolution (of the order of tens of ps) has been reported in Refs. [9, 10]. Possibilities for an effective suppression of the scattered low-energy photons (1.5–2 times lower sensitivity to 300 *KeV* photons as compared to the 511 *KeV* ones) were further discussed in Refs. [11, 12].

In the present paper we perform model analysis of the RPC properties in the context of the PET purposes by means of GEANT4 based simulations [13, 14]. We target at the optimization of the RPC construction for maximal efficiency of registration of 511 *KeV* photons and essential suppression of their sensitivity to scattered lower-energy photons.

## METHODS

The investigations encompass the whole chain from the annihilation of the positron in the body, through the conversion of the created photons into electrons and to the optimization of the electron yield in the gas. For a rough simulation of the processes of interest in the human body, the latter was represented as a homogeneous parallelepiped of size 40 x 40 x 150 $cm^3$, density of 1.01 $g/cm^3$, and the following contents: O – 61.4%; C – 22.9%; H – 10.0%; N – 2.6%; Ca – 1.4%; P – 1.1%; K – 0.2%; S – 0.2%; Na – 0.1%; Cl – 0.1%. The photon propagation was considered as starting from the centre of the volume.

For the simulations, the GEANT4 package has been used (version Geant4-09-01-patch-02, 9-May-2008). The calculations were based on GEANT4 physical models for particle interactions at low energies. The approximation of the simulation data was performed with CINT/ROOT interpreter for C/C++ [15].

## RESULTS AND DISCUSSION

**Photon propagation in the human body**. Two physical processes define the interactions of the photons with the human body and the converter medium of the PET detector:

- Compton scattering – the interaction cross-section is proportional to the atomic number (Z) of the converter material and anti-proportional to the energy of the photons;
- Photo-effect – the interaction cross-section grows with the fifth power of the atomic number ($Z^5$) of the material and also anti-proportional to the energy of the photons.

The simulation of the 511 *KeV* photon propagation in the human body shows that approximately 38% of the photons are absorbed in the body. From those that pass through it and possibly reach the detector, about 82% are scattered (have energies lower than 511 *KeV*), and only the remaining 18% (some 11% from the initial amount) are of interest from PET perspective. The energy spectrum of the outgoing photons is shown on Fig.3.

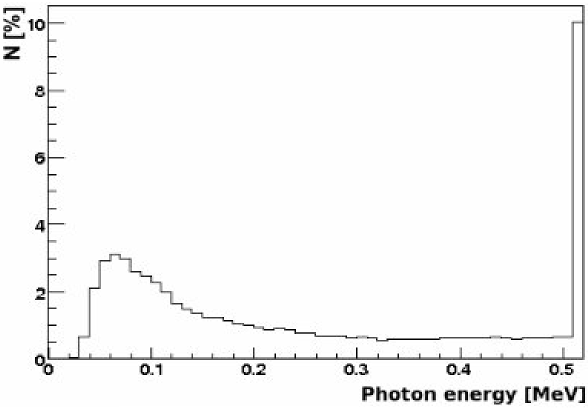


**FIGURE 3.** Energy spectrum of the photons, leaving the human body.

**Electron yield in the gas.** The RPC efficiency is determined by the electron yield in the gas gap, given by the number of photons, for which at least one interaction within the converter has lead to the ejection of an electron into the gas gap. Electron yield depends on two processes – photon interactions in the converter, and electron propagation through the converter to the gas. The electron distribution in the converter is described by

$$\frac{dN}{dx} = kN_g - sN \tag{1}$$

where $x$ is the depth, $k$ is a photon interaction coefficient, $N_g$ is the number of photons at depth $x$, $s$ is an electron interaction coefficient.

For thin converters, when $N_g$ can be considered as a constant, the solution of Eq. (1) is:

$$N = a_s\left(1 - e^{-x/b}\right), \tag{2}$$

where $a_s$ is the maximal electron yield in the gas ($a_s = N_0k/s$, $N_0$ is the initial number of photons), so the electron yield increases with $x$ till some maximal value, when saturation occurs.

In the case of larger converter thickness, photon-beam attenuation takes place and the solution of Eq. (1) reads instead:

$$N = a_l\left(e^{-x/c} - e^{-x/b}\right) \tag{3}$$

where $a_l$ is the maximal electron yield in the gas ($a_l = N_0k/(s-1/c)$); $c$ is a coefficient which accounts for the photon beam attenuation. In this case, with the increasing of $x$ the electron yield decreases after the maximum.

The cross-sections for photon interactions increase with $Z$, so in the simulations elements with high $Z$ between 74 and 83 – W, Pt, Au, Pb, and Bi – were tried as converters[1].

**Electron yield: GC design.** The direct contact between the converter and the gas (Fig. 4) apparently facilitates the propagation of the emitted electrons into the gas gap. The GC design includes a 300 $\mu m$ gas gap, the gas mixture being composed of 85% $C_2H_2F_4$, 5% i-$C_4H_{10}$, and 10% $SF_6$, 2 *mm* glass plate, with an electric field of 100 *kV/cm* applied.

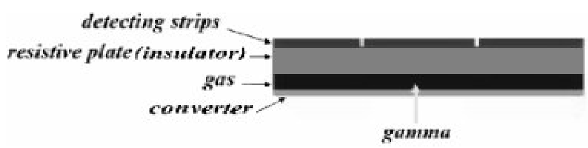


**FIGURE 4.** RPCPET: GC (gas – converter) design
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[1] The electron yield tends asymptotically to a maximum, Eq. (2); in the analysis we refer to the converter thickness at which 95% of the maximal value is reached.