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Yu.V.Zanevsky

COMPARISON OF GASEOUS AND SEMICONDUCTOR DETECTORS FOR MEDICAL IMAGING

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1. INTRODUCTION

The majority of the detectors used for medical imaging can be subdivided into two main types — counting and integrating ones. The counting group includes gaseous, microstrip silicon and scintillation detectors; the integrating group consists of CCD-matrices, photosensitive diode array, amorphous silicon detectors, gaseous ionization chambers, image plates and X-ray films.

The counting detectors provide an extended dynamic range ~ 10^6 while being limited by the maximum counting rate. The integrated detectors are devoid of this limitation and ensure high space resolution, but possess rather limited dynamic range. According to imaging technique, the detectors can be further classified into two-dimensional (2D) and one-dimensional (1D) ones. To obtain a 2D-image in the latter case one performs an object's scanning. The main parameters of radiation detectors are space resolution, efficiency, dynamic range, maximum counting rate, stability and radiation hardness.

2. GASEOUS POSITION-SENSITIVE DETECTORS

In the multitude of the existing and developing gaseous position-sensitive detectors one can specify a few main devices for medical imaging.

2.1. Conventional Multiwire Proportional Chambers

In such chambers the primary electrons, generated in gas by an ionizing particle, are miltiplied by an electric fields in the close vicinity near a thin anode wire. The counting information can be readout in one of the following ways: parallel readout with simple amplifiers-discriminators on each of anode wires or cathode strips; delay lines to perform space-time correlation; measurement of the center of gravity of the charge induced on cathode strips, etc.

Proportional chambers are successfully used in biomedical application [1]. In the 2D-detectors based on proportional chambers the information readout is often performed by means of delay lines. Such systems ensure the maximum counting rate up to 10^6 events/s supported by both low cost and high reliability [2]. The substantially more expensive 2D-systems with parallel readout allow the maximum counting rate to be increased up to 10^7 events/s [3].

2.2. High Rate Gaseous Detectors

A comparatively high maximum counting rate can be obtained by means of gaseous 1D-detectors, where each signal wire is used as an individual pulse counter. In the low gas amplification mode of operation the detector can provide the maximum rates up to ~ 10^9 events/s [4]. With a parallel electronic readout system the space resolution can achieve ~ 200 μm.

2.3. Microstrip Gaseous Detectors

These detectors fabricated by means of microlithograpy can eliminate the main drawback of conventional gaseous wire detectors, in particular those associated with maximum counting rated and space resolution [5]. Figure 1 shows that a microstrip detector is composed of thin metallic anode and cathode strips etched on a glass substratum. The upper electrode defines the sensitive

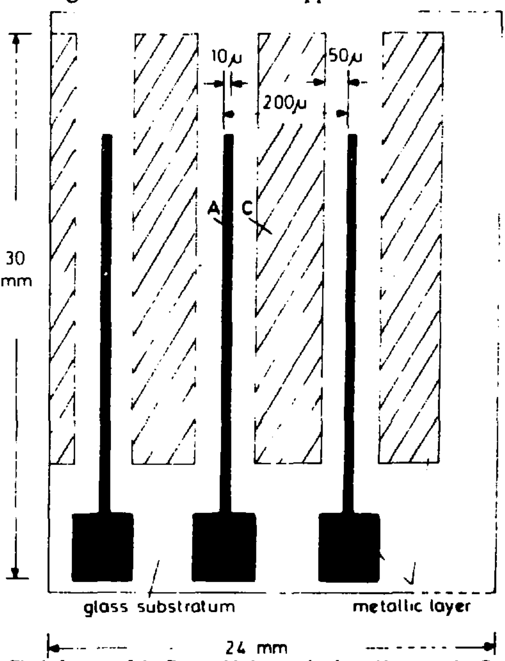

Fig.1. Layout of the first multistip anode plane (A — anode, C — cathode)

detector's volume. Electrons, generated within the gaseous volume, drift to anode strips whereas in a conventional wire chambers an avalanche is formed. The potential is applied on the backplane to prevent an ion accumulation on the isolating support. The gas amplification ratio is about 10^4 and the achieved space resolution is ~ 100 μm. The second coordinate can be readout from the backplane. Maximum counting rate of the detector is ~ 5×10^5 events/s/mm^2.

3. SEMICONDUCTOR POSITION-SENSITIVE DETECTORS

The need for improving the space resolution in experimental physics intensified the interest for segmented semiconductor silicon detectors. One of the main problems here is the accessability of the low cost electronic components for processing information from such multielement systems.

3.1. Charge-Coupled Devices (CCD-Matrices)

The modern 2D CCD-matrices are optical sensors fabricated by integrated technology and composed of a large number of ~ 20 × 20 μm pixels with the total area of 1–2 cm^2. The readout of accumulated charge is performed successively, by means of its transfer from cell to cell, to a built-in amplifier. A typical readout time of some tens ms, in case of more strict requirement for noise level, can deterioriate to 10 s. The relatively small area of the CCD-matrix and the low detection efficiency of photon with energy 30—60 keV limit its use for medical imaging. Therefore the CCD-matrices are preferred to be used for both registration and readout of information from the Image Intensifiers with a built-in scintillator [6].

This scintillator transforms the X-ray image received via a light-guide. The diameter of sensitive window of the device is 230 mm. The output 20 mm window is optically coupled with a CCD-matrix comprising 1024 × 1024 pixels. The space resolution on the input window is ~ 200 × 200 μm. A dynamic range of this integrating system is ~ 10^4.

3.2. Microstrip Silicon Detectors

Such detectors possess a fully depleted layer with an anode structure segmented into isolated elements (e.g., 1000 diodes with a 5 cm × 10 μm size stepped through 50 μ). The thickness of detecting layer is 300 μm [7]. The diodes operated as independent detectors at room temperature. As these are counting detectors, each strip is connected to an amplifier per every 100 μm; the resulting signals are written into memory via appropriate logic units. The detector advantages include a high position resolution and a counting rate of ~ 10^4 events/s/mm^2. The main drawback is a low detection efficiency for 30—50 keV X-rays.

3.3. Amorphous Silicon Detectors

The amorphous silicon (a-Si:H) can be precipitated at temperature of ~250°C onto large area substrates. It is specified by a high photoconductivity, semiconducting properties and a high radiation hardness, thus being a very attractive candidate for producing large area pixel detectors. The schematic of the detector based on the amorphous silicon technique is shown in Fig.2 [8]. Each photodiode (sensor) consists of a lower metallic contact and a $p-i-n$ structure made from amorphous silicon. This contact is coated with a transparent metallic film made from indium-tin oxide which can be in turn coated by scintillator. These sensors form a regular 2D-structure. Each sensor is connected to horizontal readout lines through a-Si:H field effect transistor (FET). The gate electrodes of each FET are connected to vertical control lines. The sensors need only n-control lines and n-gating lines. The recently developed detector samples contain 256 × 256 pixels each 100 × 100 μm.

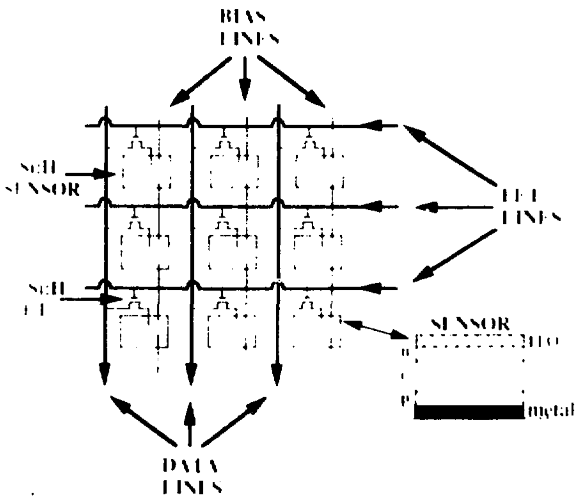

Fig.2. Schematic diagram of the layout of a amorphous silicon photodiode array with construction of an individual p-i-n sensor

4. APPLICATION OF SEMICONDUCTOR AND GASEOUS DETECTORS FOR MEDICAL IMAGING

Below there will be presented some devices for biomedical application based on semiconductor and gaseous detectors.

4.1. Scanning Diode Array System for Digital Radiography

The setup for digital chest radiography has been based on the array of 1024 photodiodes (each 0.5 mm in lenght) connected to a gadolinium oxysulphide scintillator [9]. The scanning time is 4.5 s. «Fore» and «aft» slits (0.5 mm and 1.0 mm wide,respectively) collimate the X-ray beam.

Fig.3. The result of using image subtraction with two tube voltage to eliminate bone leaving only the soft-tissue components of the patient

The detection efficiency for X-rays is about 25% with radiation dose 0.25 mGy. The readout has been performed with 12-bit ADC. Final images have been presented on a TV-monitor as a 1024×1024 matrix. This system has been used for radiography at two different X-ray energies. Figure 3 shows an image [10] resulting from the subtraction of the images obtained at two voltages of the X-ray tube to study patient's soft-tissues.

Another linear scanning system with gaseous multiwire detector was designed for the study of pregnant women at low doses [11].

4.2. 2D-Device Based on Silicon Microstrip Detector for Digital Radiography

Collaboration RADIN [12] develops a device for digital radiography based on microstrip silicon detectors. This device will be composed of separate silicon two-coordinate detectors with a 200 μm step between adjacent strips. A few layers of silicon detectors will be used to increase detection efficiency. The electronic readout system is parallel. A good image contrast ratio and an acceptable exposure time require $\sim 10^8$ photons/s to be registered by the device. The main problem arises from the high event rate to be handled by the electronics in order to decrease the random coincidence rate down to some reasonable limit.

4.3. Gaseous Detector for Noninvasive Subtraction Angiography

For noninvasive coronary angiography with synchrotron radiation there has been developed a gaseous X-ray detector based on strip ionization chamber. Two liner detectors separated by a common drift electrode are placed inside a common gaseous volume (Fig.4) [13]. Each upper and lower readout electrode is subdivided into 256 strips. The Frish grids to ensure position independent signals on the strip electrodes are placed at a distance of 800 μ from these latter. The detector gas Xe/CO_2 mixture at a pressure of 10 atm is used. The object is scanned at the 0.5 mm/ms rate by two flat monochromatic X-ray beams ($E \sim 33$ keV) with a small shift in energy. The resulting image contains 256×256 pixels of 0.5×0.4 mm. The readout is performed in a parallel mode. Each strip is connected to preamplifier, FADC and serial-to-parallel counter. To ensure an acceptable signal to noise ratio requires the beam intensity of $\sim 10^{11}$ protons/s/cm^2. The detection efficiency of the system is $\sim 80\%$, the dynamic range $\sim 4 \times 10^3$ and the position resolution ~ 0.5 mm. With the iodine concentration of ~ 40 mg/ml the achieved contrast ratio is better than 0.5%.

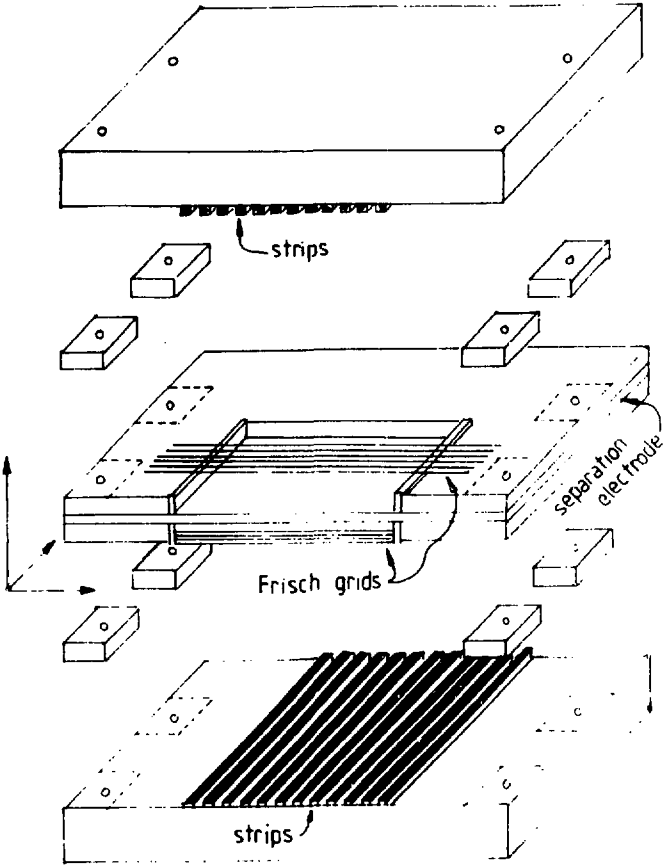

Fig.4. Setup of the detector

4.4. Gaseous Detectors for Digital Autoradiography

2D-detector for digital autoradiography based on a multiwire chamber has been developed at the JINR [14]. The 20 × 20 cm^2 samples labelled with tracers ^{32}P, ^{14}C, ^{35}S and ^{125}I can be placed over the sensitive window of the detector. The samples labelled by tritium tracer can be placed for analysis into

Fig.5. The wire chamber radiograph of a rat kidney obtained about a hundred times faster than using conventional means

detector gas volume. Information readout has been performed via delay lines. The spatial resolution is ~1.5 mm with the sensitivity 1000 times better compared to photographic film. The next important development step has included the combine of a multistage gaseous chamber with an optical readout of the light emitted by electron avalanche. That has been realized by means of an Image Intensifier and a CCD-matrix [15]. Figure 5 shows a radiograph image of a rat kidney cut labelled by tritium tracer. The kidney ducts of ~ 50μm diameter can be easily seen.

5. CONCLUSION

The following table contains main parameters of the gaseous and semiconductor detectors for medical imaging.

The main drawback of the gaseous detectors is due to the maximum counting rate limit. It is conceivable that the gaseous microstrip detectors will improve this parameter. A substantial improvement of this parameter is provided by the parallel readout system.

The semiconductor detectors seem to become the next step in developing of new generation 2D-detectors with the total pixel number up to 10^7 and pixel size of $200 \times 200 \mu m$. The VLSI readout chip could be deposited at the same time as the readout electronics in this technology.

Table

Parameter	Gaseous Detectors	Semiconductor Detectors
Spatial resolution	~ 100 μm	100 μm
Detection efficiency	> 50 %	50 % (with scintillator)
Rate capability	~ 10^9 events/s	no limination
Dynamic range	no limitation	~ 10^4
Radiation hardness	~ 10 MR	~ 10 MR

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