High-resolution detectors for medical applications and synchrotron radiation research

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Abstract

In the present report, we summarize our experience in the development of high-resolution position sensitive gas detectors for medicine and synchrotron radiation experiments at Budker Institute of Nuclear Physics for the last years. We have designed several versions of Multistrip Ionisation Chambers with a channel pitch varying from 0.4 down to 0.1 mm. The high quantum efficiency (>65%) of these detectors allow its application in high quality diagnostic imaging. The detector with 0.1 mm strip pitch and 20 atm pressure of Xe demonstrates the best possible DQE and spatial resolution for gaseous detectors in a wide range of X-ray energies. Additionally, the initial results of feasibility study of the detector for beam position monitoring for Heavy Ion Therapy System are presented too.

Key words: X-ray detector, digital radiography,
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1. Introduction

Since the 80's of the last century development of detectors for medical radiography has been carried out at BINP. The first system (LDRD "Siberia") was designed on the basis of Multiwire Proportional Chamber (MWPC) operating in photon counting mode [1, 2]. The MWPC measured X-ray intensity distribution in horizontal direction and distribution in vertical direction was measured by mechanical scanning. The scanning method has several intrinsic advantages over conventional two-dimensional systems: large image size in scan direction, significant rejection of radiation scattered in a patient's body and a simpler detector design. Initial application at the All-Union Center of Mother and Child protection (Moscow) showed that the technology originally dedicated for nuclear physics gave the possibility to reduce the patient irradiation doses dramatically and improves diagnostics possibilities [3]. Industrial production of the installations in Russia based on the technology developed by BINP has started since 1997. More than one hundred installations based on MWPC operated in hospitals. At the beginning of the present decade, after the experience of application of MWPC in medical systems and following the increasing requirements of physicians, a new detector was introduced instead of MWPC - Multistrip Ionization Chamber (MIC) [4]. The MIC working in charge integration mode made it possible to preserve the same dose characteristics of the installation and to improve spatial resolution by a factor 1.5 (0.4 mm). Moreover, the application of the new detector solved the problems of limited counting rate capability and gas aging. More than two hundred Multistrip Ionization Chambers with 1024 registration channels were produced. During the last few years, we have designed several detectors that have different parameters depending on particular application. For example, for the installation intended for mass screening of healthy people we use the detector with the inlet aperture collecting 100% of primary beam but with the limited spatial resolution (0.26 mm). In this case, the lowest patient irradiation dose for chest radiography - 4 μSv is obtained. For comparison, the reported doses for routinely used systems are 7 – 50 μSv [5]. For general-purpose radiography where a higher resolution is required, we have designed a detector with a strip pitch of 0.2 mm. The technology of producing such detectors was transferred to the South Korean company "Advanced Digital Technology". The clinical studies of the installations manufactured there show that scanning system provides a better image quality at a lower entrance dose [6]. This system is capable to get images with a length of up to 100 cm at 21 cm/s scanning speed and its maximum throughout per day $\sim$ 700 patients.

The last modification of MIC is made for imaging at synchrotron radiation beams. This detector demonstrates the limit of gaseous technology. It has 0.1 mm strip pitch and is filled with Xe at pressure of 20 atm.

2. Design of Multistrip Ionization Chamber

MIC is a flat capacitor with a solid field electrode and the segmented signal electrode contains the strips [7] (Fig. 1). All strips are directed to the focal spot of X-ray tube. The high voltage applied to the field electrode is about +1kV. The gap between the electrodes is 2-6 mm. The charge induced by X-ray radiation is collected on the strips and its value is measured with

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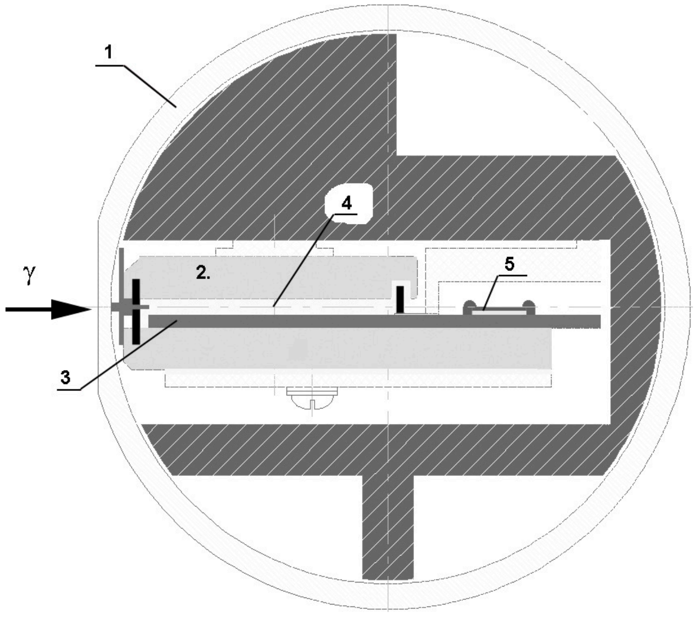

Figure 1: General layout of the detector. 1- detector box, 2 drift electrode, 3 strip PCB, 4- sensitive detector volume, 5 front-end electronics.

special electronics located inside the detector box. The front-end electronics consists of charge sensitive integrators, each of them containing 64 to 256 readout channels. Each channel has the signal capacitor connected to the strip and the control circuit that performs subsequent readout of collected charge on the capacitors. The original electronics for 1024-channel detectors was designed and produced at plant "Vostok", Novosibirsk. For high-resolution detectors, two types of commercially available integrators are used. The first chip is *XL-1 Series* multiplexer produced by PerkinElmer Inc. [8]. The second one is *ISC9717* - a low-noise 128-channel readout integrated circuit by Indigo Systems Corp. [9]. The last chip provides a better detector performance due to the possibility to operate in *Integrate While Read* mode. It allows reducing the data integration time down to ~ 0.8 ms without signal losses. For all cases the intrinsic electronics noise is less than $5000e^{-}$ RMS, this is compatible with a signal from ~ 4 x-ray quanta (60 keV).

The detector design is affected by X-ray optics that is used in the installation. In Fig. 2 two versions of X-ray optics that can be used in scanning system are shown. Both of them have certain advantages and drawbacks. The first option ("A") provides high resolution in scan direction and the highest scatter radiation rejection rate but it uses only a part of X-rays passed through a patient. The second option ("B") has the lowest patient irradiation dose due to full utilization of primary beam but has limited resolution induced by the focal spot limitations. Nevertheless, in the applications where the lowest patient irradiation dose is required (mass screening of healthy people and prenatal care) these detectors are superior.

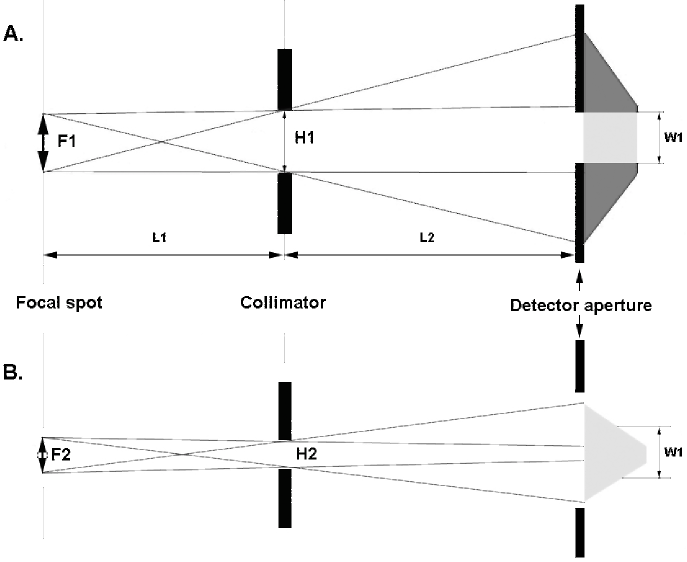

Figure 2: Two variants of X-ray optics that are used in scanning radiography and can provide the same limiting resolution. A - "Narrow beam" variant. B - "Wide beam" variant. W1 - effective beam width.

3. Detector parameters

In the following table, consolidated data for different medical detector modifications are presented.

Channels	Pitch, mm	Strip length	Gas
1024	0.4	6 cm	Kr 20 atm
1536	0.26	6 cm	Kr 25 atm
2048	0.2	3 cm	Kr 40 atm

Common technique to characterize spatial resolution of detectors is the measurement of modulation-transfer function (MTF)[10]. The part that is described by physical processes (presampled MTF) was received from the acquired edge images at the IEC-specified RQA5 beam quality [11]. For comparison, the results of simulations with GEANT 4.9.2 for the given detectors, taking into account the influence of diffusion and focal spot size, are shown in Fig. 3 and Fig. 4. The presented data clearly demonstrate that for *Xe*– and *Kr*––filled detectors GEANT4 is in a good agreement with the experiment. Some deviation at high space frequencies for MIC2560 could be referred to parallax error, because the detector is designed for a focal distance of 20 m.

4. High resolution detector for synchrotron radiation research

A high-pressure multistrip ionization chamber with 100 μm pitch of the readout strip structure was constructed for imaging at SR beam in a wide range of energies from $15keV$ to $70keV$. The chamber design is similar to the medical detectors described above. The chamber is filled with pure *Xe* at 20 atm and the length of strips is 20 mm. The chamber contains 2560 channels with a full aperture of 256 mm. For the tests MIC was exposed to X-rays from an X-ray tube with a tungsten anode.

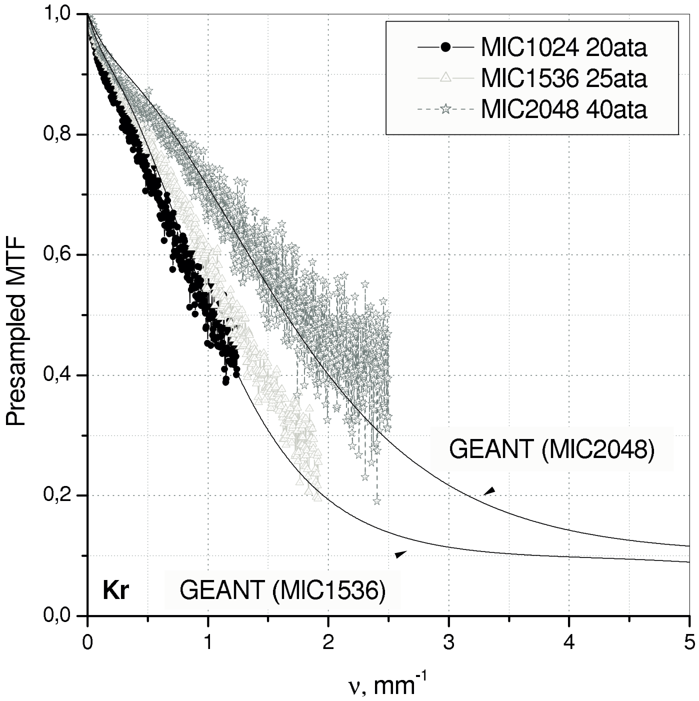

Figure 3: Presampled MTF of detectors filled with Kr and comparison with Geant 4.9.2 simulations.

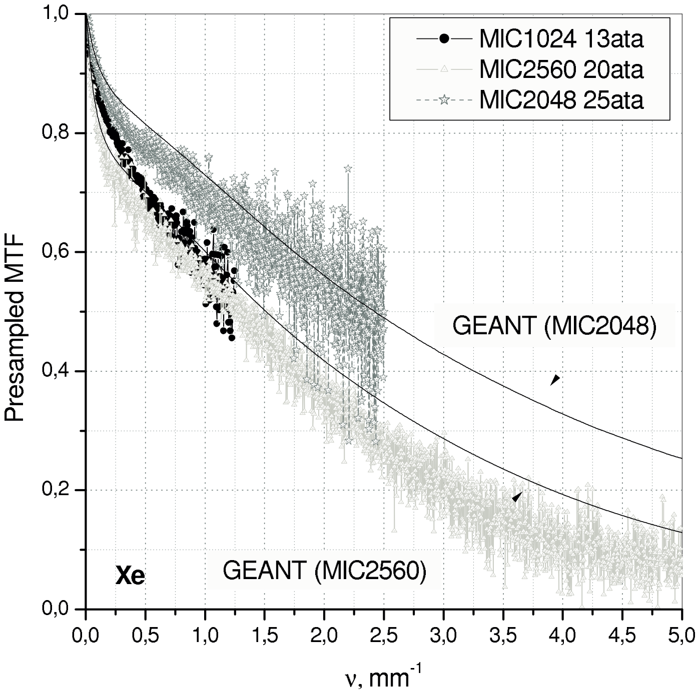

Figure 4: Presampled MTF of detectors filled with Xe and comparison with Geant 4.9.2 simulations.

DQE and spatial resolution were measured at 40kVp, 60kVp, 80kVp and 100kVp X-ray tube voltages. DQE as a function of rate for different tube voltages is shown in Fig. 5.

The drop of DQE at low fluxes is caused by the electronics noise, while the value of DQE at the plateau is determined by the fraction of photons absorbed in the sensitive volume and fluctuations of the energy deposited. We can see that DQE is always higher than 50% at photon flux above 1000 Hz/channel and increases up to 70% at 60 kVp tube voltage. This is probably because most of the X-ray spectrum in this case is just above *Xe* K-edge. Presampling MTF of the chamber measured with edge method at 76 kVp together with the result of simulation is shown in Fig. 4. The contrast at 3 lp/mm is around 20% and drops to 10% at 5 lp/mm (Nyquist frequency).

5. Prototype of beam position detector for Heavy Ion Therapy System

For the last few years the project of Heavy Ion Therapy System has been under development at Budker INP [12]. During the treatment, a tumor will be irradiated with a small high-intensity beam of ions that will move over a target volume. Therefore, the beam position and intensity have to be controlled with a high accuracy. The presence of electron cooling in the ion synchrotron provides a small size and small energy spread of the cooled beam, thus, enabling the realization of the original beam extraction scheme by small precisely dosed portions, the so-called ”Pellet” extraction. Due to high dose rate in one ”pellet”, there are several possibilities to realize beam position monitoring. One of them is a low-pressure gas ionization chamber. The possibility to operate it at a low pressure in an intensive particle beam was proved early [13]. For monitoring the beam position and its intensity in the ”Slow extraction” mode (when the beam intensity will be much lower), the detector with the same design but operating at ambient pressure will be used. Unification of the detector elements for two different extraction modes will simplify the detector service in future. To verify some technological questions a detector prototype that consists of two parallel polyimide films of 20 μm thickness covered with 0.1 μm Ti with 2 mm spacing between foils was constructed. The position-sensitive electrode containing 128 signal strips with a pitch of 1.25 mm and a work length 170 mm was produced by chemical etching (Fig. 6). The foils were glued onto a glass-epoxy frame having 170 x 170 mm window in the center. To evaluate detector response at a high dose rate, the detector was filled with Kr at ambient pressure instead of air and irradiated with x-rays. Quantum efficiency of the detector filled with Kr is $7.9 \cdot 10^{-3}$ at 60kVp, whereas for the air-filled detector it is $6.0 \cdot 10^{-5}$ only. That allowed us to estimate the detector behavior at the signal values compatible to the signals at > 100 times higher dose level in the air-filled chamber. The detector had small deviation from linear behavior up to effective dose rate 2.6 kR/min. Estimation of the detector parameters under proton-ions beams is a subject of future work.

6. Conclusion

At present, the first MIC detectors have been operating in hospitals for almost 10 years without showing any degradation of their parameters. It confirms the fact that a properly made gas detector is a reliable device. Simulations demonstrate that

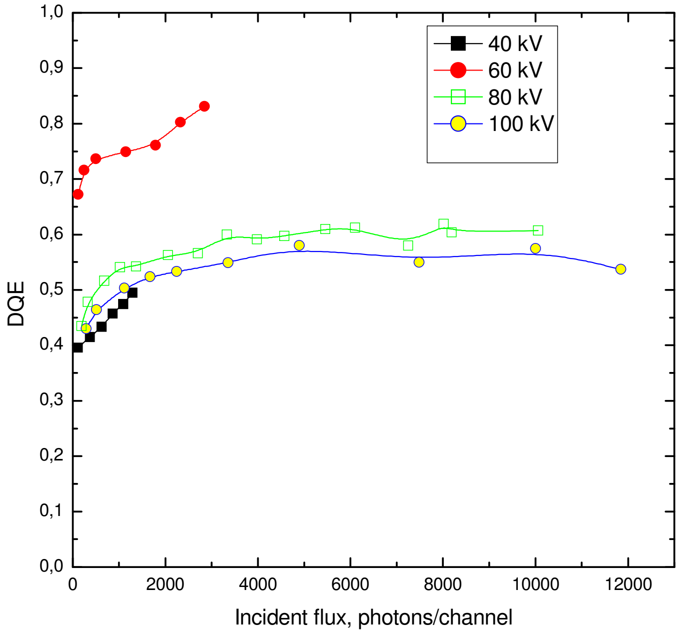

Figure 5: DQE as a function of photons rate for different tube voltages.

a further increase of pressure does not improve the spatial resolution and DQE significantly. Nevertheless for some specific applications gas detectors are superior due to its radiation hardness.

7. Acknowledgements

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8. References

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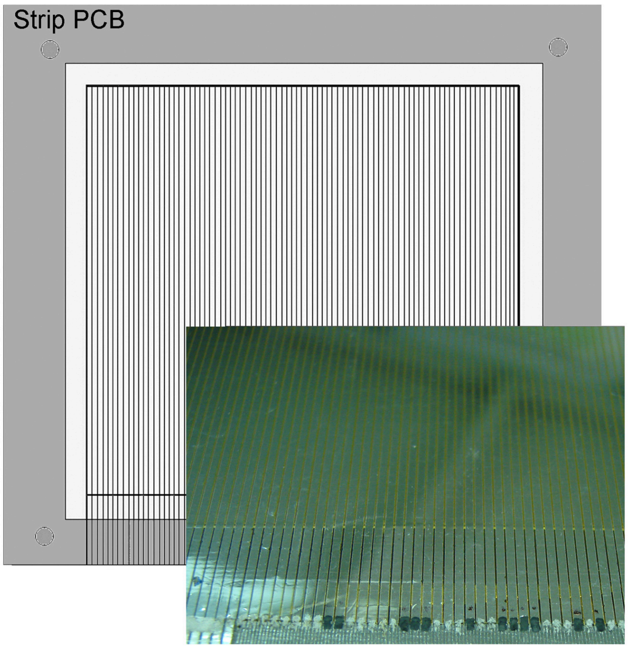

Figure 6: Signal electrode of the detector prototype for beam position monitor.

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