ИНСТИТУТ ЯДЕРНОЙ ФИЗИКИ СО АН СССР

E.A. Babichev, S.E. Baru, V.V. Gusev,
A.G. Khabakhpashev, G.M. Kolachev, G.A. Savinov,
L.I. Shekhtman, V.A. Sidorov, A.I. Volobuev

DIGITAL RADIOGRAPHIC INSTALLATION FOR MEDICAL DIAGNOSTICS

PREPRINT 89-73

НОВОСИБИРСК

Digital Radiographic Installation
for Medical Diagnostics

*E.A. Babichev, S.E. Baru, V.V. Gusev,
A.G. Khabakhpashev, G.M. Kolachev, G.A: Savinov,
L.I. Shekhtman, V.A. Sidorov, A.I. Volobuev*

Institute of Nuclear Physics
630090, Novosibirsk 90, USSR

A B S T R A C T

Digital radiographic installation of scanning type on the basis of one-coordinate multiwire proportional chamber is described. The installation reduces irradiation doses by a factor of 30—100 in comparison with standard methods of X-ray study and improves diagnostic possibilities of radiography. Digital image, obtained at the installation, has 256×256 pixels of ~1×1 mm size.

INTRODUCTION

A large step in development of techniques and diagnostic possibilities of radiography took place in 70-th with appearance of computerized tomography. Quality and informativity of images were improved, contrast sensitivity for a study of inner organs was increased. At the same time, computerized tomography, inspite of all its merits, did not force out projectional radiography. These both methods naturally divided spheres of application between each other and in some cases supplemented one another.

Success of computerized tomography and significant achievements in a field of mathematical processing of images increased an interest to digital radiography. Its merits are well known now. At the same time digital radiography can be significantly improved if indirect methods of obtaining digital image, connected to digitizing of X-ray film or analog television signal, will be replaced by a direct registration of X-radiation by appropriate detectors. Such approach gives an opportunity to diminish greatly doses of irradiation—the main source of overbackground irradiation of people, significantly widen dynamic range and exclude registration of scattered radiation.

In the Institute of Nuclear Physics in 1981—1984 digital radiographic installation of scanning type was worked out, decreasing irradiation doses by a factor of 30—100 in comparison with screen-film systems [1]. Unlike standard diagnostic apparatus this installation instead of film or EOT used for X-ray detection a multiwire proportional chamber. In 1986 a new multiwire proportional chamber was worked out, which permitted to improve spatial resolution almost by a factor of two and to increase counting rate capability by a factor of three.

EQUIPMENT AND WORK OF THE INSTALLATION

Digital radiographic installation includes standard X-ray tube, mechanical scanning system, proportional chamber and system of registration and control. Design of the installation is shown at Fig. 1. Distribution of radiation in horizontal direction is measured

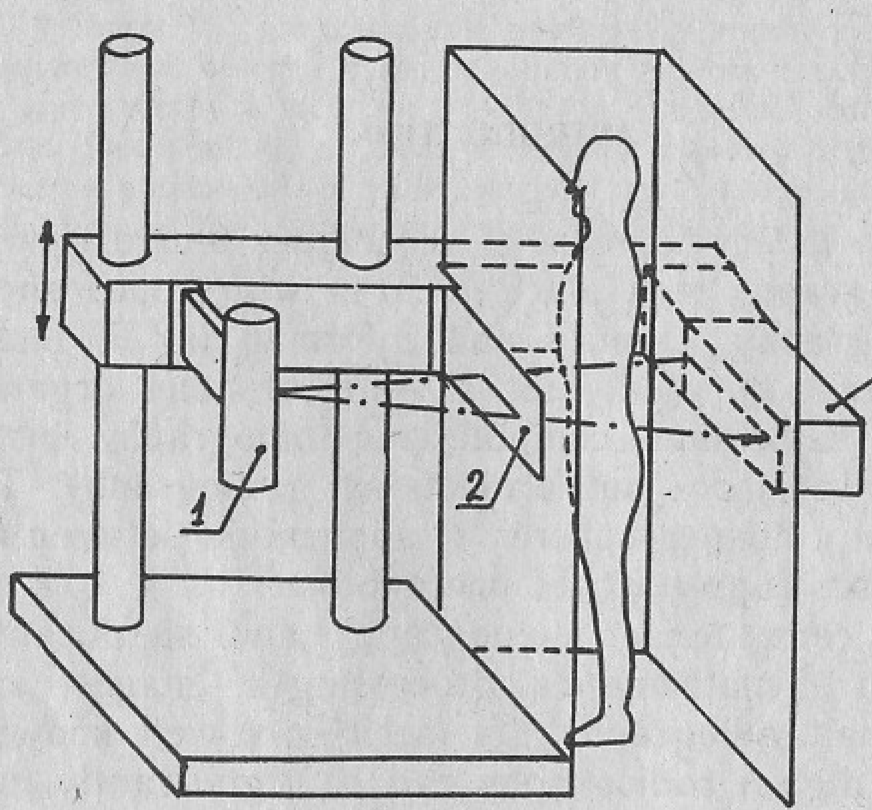

Fig. 1. Digital radiographic installation:
1—X-ray tube; *2*—slit collimator; *3*—multiwire proportional chamber.

with the help of proportional chamber, in vertical direction—by a mechanical scanning. For this purpose standard X-ray tube, slit collimator and a chamber are simultaneously and uniformly moved in a vertical direction during a shooting. Collimator with a slit of ~1 mm forms narrow fan beam, which, after penetrating through a patient's body, gets into inlet window of a proportional chamber.

Multiwire proportional chamber (MWPC) for such installation must have a channel width 1—0.5 mm, detection efficiency more than 20—30% for 60 keV energy and counting rate capability of

300—500 kHz/chan for efficiency decrease of 20%. To obtain such a counting rate a method of parallel readout of information from one-coordinate chamber must be used, and efficient length of anode wires must be sufficient to reduce space charge to admissible level. A required efficiency can be obtained, if a thickness of sensitive layer of Xe will be about several centimeters at a pressure of several atmospheres. These requirements to the length of anode wire and thickness of sensitive gas layer can be fulfilled, if a direction of quanta movement is parallel to anode wires. Up to now all MWPC, described in literature have anode wires parallel to each other. In our case at a necessary length of a chamber of 300—400 mm and a distance between X-ray tube and MWPC 1300 mm parallax at the edges of the chamber worsens spatial resolution to 4—5 mm. Even in the first installation with channel width of 2 mm resolution was reduced at the edges by a factor of ~1.5 for a length of anode wires of 30 mm.

Therefore to obtain a necessary spatial resolution a chamber with fan anode plane was worked out [2]. All anode wires are directed to the focus of X-ray tube, positioned at a distance of 1300 mm from the centre of a chamber (Fig. 2). Anode plane is

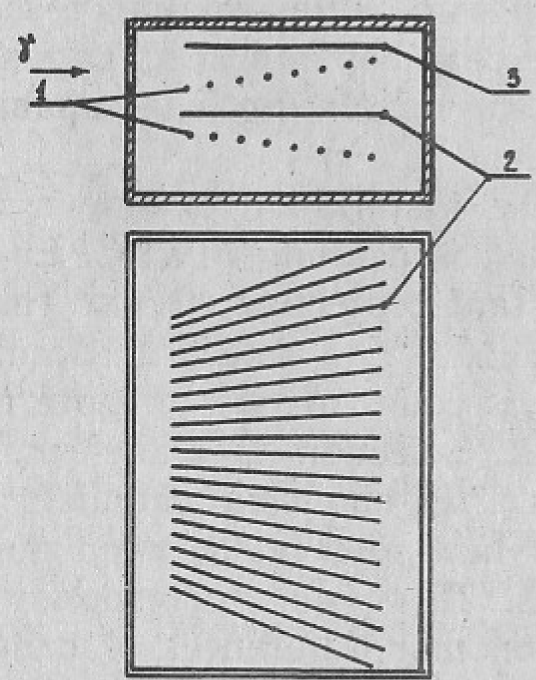

Fig. 2. Multiwire proporttional chamber:
a—side view; *b*—top view of anode plane; *1*—cathode planes; *2*—anode plane; *3*—drift electrode.

made of wires 10 microns in diameter, step of anode wires is 1.2 mm, length 50 mm. To increase counting rate capability of the chamber the distance anode-cathode is decreased to 2 mm. Cathode planes consist of wires 100 μ in diameter, stretched with a step of 1 mm. Cathode planes are inclined so to compensate change of gas gain due to nonparallelity of anode wires. Without cathodes inclination gas gain changes by 30% for the length of anode wires of 50 mm.

X-ray beam gets into the chamber between upper cathode and drift electrode, disposed above the cathode plane at a distance of ~8 mm. Primary ionization, formed in a drift gap, drifts to the upper cathode plane under the action of electric field, penetrates through it and in the vicinity of an anode wire makes electron-ion avalanche. During the movement of electrons and ions in a field of a chamber a charge is induced on the nearest anode wire (or, sometimes, two wires), which is registered by an amplifier-discriminator (AD). AD is connected to each anode wire. Signals from amplifier-discriminators come to special selecting circuits, which rejects an event, if two neighbouring wires have simultaneous hits.

MWPC is disposed in a duralumin box, filled with Xe+20%CO_2 mixture at a pressure of 3 atm. The inlet window of a chamber has a thickness of 0.5 mm.

The aggregate of pointed above features of this proportional chamber permitted to obtain equal spatial resolution of ~1 mm both in the centre of the chamber and in its edges at a constant gas gain along anode wires, high counting rate capability and good efficiency.

Block diagram of the installation is shown at Fig. 3. amplifiers-discriminators are disposed along MWPC. Each circuit-hybrid is supplied by a diod-resistant protective circuit. Thresholds of all AD are made identical ($\sim 5 \cdot 10^{-14}$C). Selecting circuits exclude events, connected to simultaneous count of two or more neighbouring channels from one quantum. Selection is made with the help of 256 anti-coincidence circuits. Neighbouring channels are considered to count simultaneously, if time interval between pulses in these channels is less than 200 ns. This value corresponds to maximum possible sum of two components: differences of drift times of primary ionisation electrons and dispersion of count moments of AD, connected to dynamic range of signals. Selection circuits are also used to give an exposure time of one line and all image from a computer .

The installation works in the following way. Vertical mechanical scanning is begun simultaneously with switching on an X-ray tube. The information, stored in scalers during an exposure time of one line, through the output unit is rewrited into intermediate CAMAC

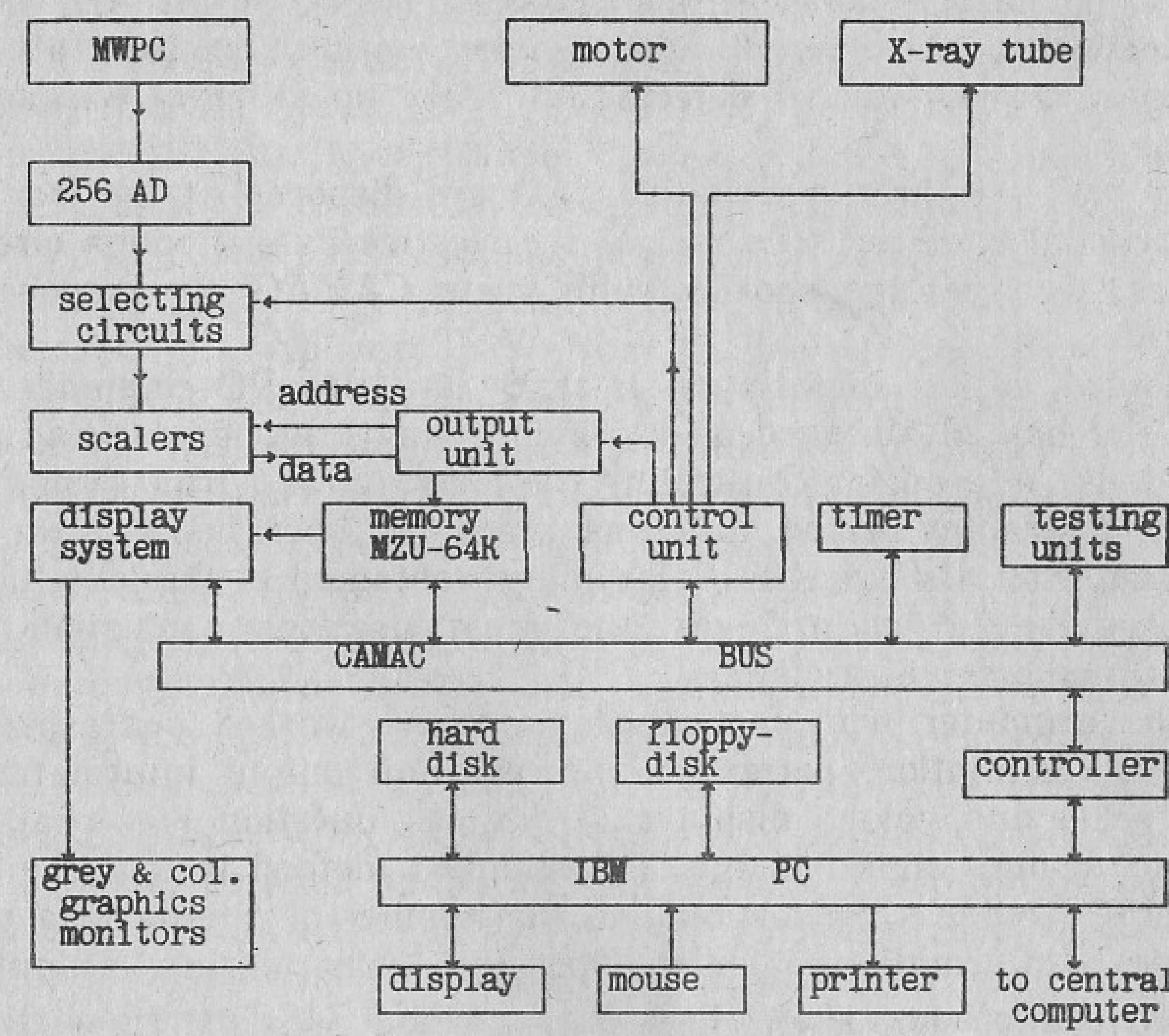

Fig. 3. Block diagram of the digital radiographic installation.

memory — MZU-64K. After the rewriting scalers begin registration of quanta in the next line. After the shooting process a so called digital image is stored in a computer memory, it is a matrix of 256×256 numbers, describing a distribution of radiation flux after a patient's body. Inspite of high image exposure time (~8 s), there are no decrease of sharpness due to small movements of patient and patient's organs, because the exposure time of one line is only 30 ms. The first unprocessed image appears on display simultaneously with scanning. Later after the necessary processing an image can be represented in the most convenient form for a doctor with the help of fast display system. To work with it image is rewritten from a computer memory, where it is disposed during processing, again to the intermediate memory.

CAMAC crate, except pointed above display system and intermediate memory, has a unit of control of X-ray apparatus, scanning mechanics and readout system, timer and testing units, with the help of which a special program checks all electronics and measures the main characteristics of the chamber (work of all AD, timing parameters of selecting circuits, proper work of scalers, all feed voltages, dependences of detector efficiency on different parameters and so on).

In the first three installations AD are disposed at the frame of proportional chamber. The rest of the electronics — selection circuits, scalers, feed voltage sources with crate CAMAC are disposed in separate unit.

Control of the installation is made by IBM PC computer. The computer has 20 Mb hard disk, 5 inch floppy drive, mouse and printer. Mouse is used for controlling the program and transforming an image at display. Hard disk has place for operation system and program, and also for 100—150 images obtained at the installation. To organize larger archives one must use communication with central computer of a clinic.

The computer with fast display system, worked out especially for this installation, permits to process and output information to grey scale and colour displays. A natural question is — what is a way to display digital image. The simplest method is to make each element brightness proportional to the number of quanta in a given channel (information is displayed this way during scanning). Simple estimations show that this method is close to optimal. Brightness of TV screen luminescence is in a definite limits proportional to amplitude of videosignal. Signal amplitude $A \sim N_{ij}$ — number of quanta, detected in the channel with coordinates of i and j. Sensitivity of an eye J_y is proportional to logarithm of brightness and therefore $J_y \sim \ln N_{ij}$.

From the other hand, an X-ray image must show the structure of inner organs, and if one proceeds from interaction of X-rays with matter, must give distribution of integral absorbtion coefficients in a plane of image. With some approximation one can say that projectional radiography must give an image of distribution of density integral along a projection in a given channel direction. As absorbtion coefficient and density has logarithmic dependence on registered number of quanta N_{ij} displaying of a digital image with pixels of N_{ij} is equal to displaying of density image of a given part of a patient's body.

In order to widen the dynamic range a transformation of matrix N_{ij} can be useful in some cases. as the first practice of phisicians working with such installations showed, sometimes it is expedient

for such purpose to make a transformation like $n_{ij}=\sqrt{N_{ij}}$ and display matrix of n_{ij}.

Special program gives doctor an opportunity to transform an image so to make it most convenient for visual analysis and thus improve diagnostic possibilities of projectional radiography. For this purpose the total range of pixel values of the matrix N_{ij} (16-bit numbers) or its arbitrary part from N_1 to N_2 (i. e. pixels which have $N_1 < N_{ij} < N_2$) is divided into 32 grades during displaying. The program permits to change arbitrary bottom N_1 and top N_2 boundaries of a chosen part of a range by movement of the mouse. Change of an image occurs practically simultaneously with movement of the mouse, it allows to find fast and distinguish at an image signs having diagnostic meaning. One can also divide image into two parts by horizontal or vertical line (position of the line is chosen by the help of the mouse) and make separate change of contrast boundaries in each part of an image. Such operation is necessary for processing images with strong differences of density and for measuring distances between points in such regions. The program permits to perform several other operations improving displayed image. In a «window» regime the total range of intensities is displayed, i. e. doctor sees all image, with simultaneous usage of a part of grades for more detailed demonstration of a chosen section. Position of this window inside the total range of densities of an image and its width can be changed with the help of mouse. This program provides also the possibility of measurement of relative density in each pixel or in arbirary region of an image; magnification of a chosen part of an image; inverse of an image; measurement of organ sizes or distances between chosen points and some other operations.

Mathematical processing of digital images — rise of sharpness, size discrimination, increase of contrast and so on, — will permit in future to improve significantly image informativity. At present time several simplest processing algorithms are realized. The program of contours underlining, which allows to improve sharpness and increase contrast of small details at an image. Another processing program makes functional transformation $n_{ij}=\sqrt{N_{ij}}$ mentioned above.

The system controlling the work of the installation includes also a special program of automatic check of good condition of the chamber and electronics. Such a check takes about 10 minutes. After the check system prints measured characteristics or those parameters which goes out of established and fixed in the program permissible deviations. Such a check is expedient to perform once a week or when errors will be found in the work of the installation.

CHARACTERISTICS OF THE INSTALLATION

The installation has the following main characteristics. The number of elements in a digital image is 256·256=65536, the value of each element is 2^{16}, element size in a plane crossing a body of a patient is 1×1 mm. Spatial resolution, efficiency and counting rate capability of the installation is defined by processes of quanta registration by the chamber, geometry of X-ray beam and dead time of electronics.

Spatial Resolution

Absorbing of an X-ray quantum by a Xe atom leads to emitting of secondary particles: photoelectrons, auger-electrons and fluorescent photones. ranges of these particles influence spatial resolution of the chamber that is resulted in some cases in working several channels simultaneously.

Simulating of absorbing processes, carried out in accordance with algorithms described in [3], showed that in Xe+20%CO_2 mixture at a pressure of 3 atm the probability of simultaneous hits in two neighbour channels exceeds the one of channels disposed at a larger distance at least by 7 times. Thus main distortions connected to double registration of X-quanta are caused principally by coincidences of neighbouring channels. To exclude them selecting circuits described above are used.

Coincidences of neighbouring channels occur mainly when photo- or auger-electron crosses a boundary between them. Primary ionisation of such events, formed, as a rule, near channel boundary, is divided into two parts. Therefore excluding of such events improves also spatial and amplitude resolution of the detector. Essencial shortcoming of the method is decrease of efficiency by about 30% at an energy of X-rays of 50—60 keV.

Spatial resolution of detectors with channel structure (an example of such detector is MWPC with pointed above method of readout) is completely described by a channel shape function—the dependence of a channel counting rate on coordinate of narrow col-

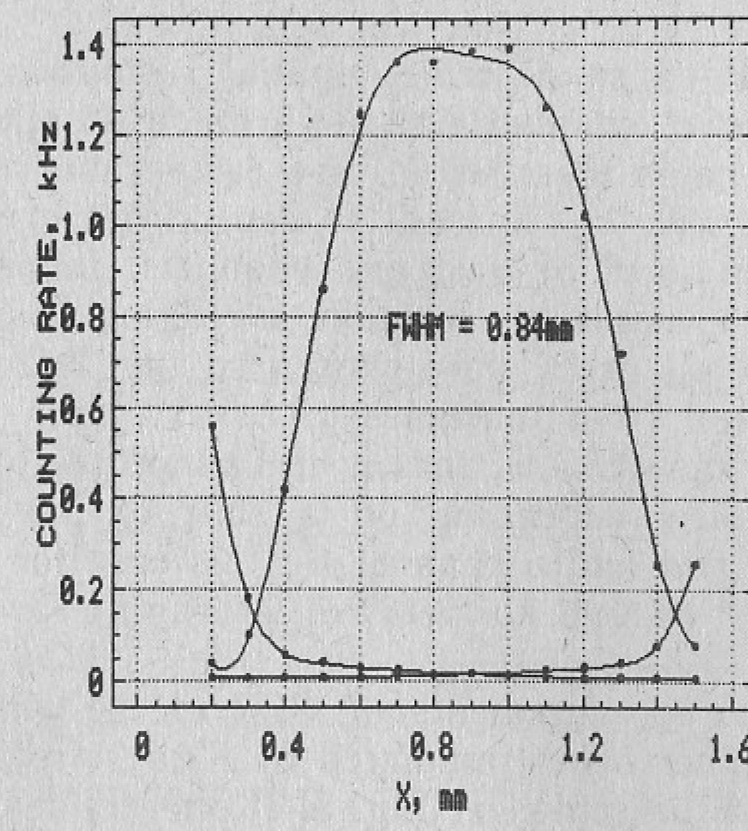

Fig. 4. Shape of a channel of the MWPC. To the right and to the left from the main curve a similar curves for the neighbouring channels are begun.

limated radiation source. At Fig. 4 a channel shape curve is represented, obtained when irradiating the chamber by an X-ray flux from a tube at 70 kV voltage with a 0.3 mm copper filter (average energy 47 keV). FWHM is 0.84 mm. This value is less than step of anode wires — 1.2 mm, as the suppression of coincidences leads mainly to excluding of events registered near channel boundary.

However for many practical purposes spatial resolution is more convenient to be described by one parameter with the help of contrast frequency response function (CFR)—by spatial frequency at a defined level of contrast. Method to define CFR is described in [3]. To measure it one must take a set of grids with periods of λ_i, made of opaque for X-rays material. At an image of such grid the dependence of counting rate on a number of channel in a line (or in a colomn, if vertical resolution is measured) has periodical form. CFR is defined as expression

$$C(\nu)=\frac{N_{\max}-N_{\min}}{N_{\max}}, \qquad (1)$$

where $\nu=1/\lambda$—spatial frequency, $N_{\min}$ an average value of those channels which has counting rate less than average rate N, N_{max} — an average value of channels with $N_i > N$.

The CFR defined so describes spatial resolution well in those cases when detector channel size σ is much smaller than a grid period λ. In our cases these values are comparable and, for example, usage of grid with $\lambda = 2\sigma$ leads to dependence of C on the phase between the boundaries of grid and channels. In order to exclude this ambiguity it is necessary to make averaging of C over a phase. For this purpose all grids were chosen so that $\lambda/2\sigma$ differed from natural values and from multiples of 0.5. At such a condition an image of a grid has «beats», that is the amplitude of oscillations is changed periodically depending on a shift of the phase between boundaries of a grid and a channel. In this case for calculation the value of C it is enough to perform averaging over a period of «beats».

The results of measurements of CFR at the centre and at the sides of the chamber being irradiated by X-rays from a tube with a 70 kV voltage are presented at Fig. 5. If one defined spatial resolution as a frequency at $C = 0.5$ it can be noticed that there are no essential differences between spatial resolution at the centre of the chamber (0.68 mm^{-1}) and at the sides of it (0.65 mm^{-1}, 0.63 mm^{-1}).

Spatial resolution in scanning direction is defined by a geometry of collimated beam, velocity of scanning and exposure time of one line of an image. Profile of the beam, i. e. dependence of X-ray flux density on a coordinate along direction perpendicular to beam plane, is defined by dimension of focus spot at anode of X-ray tube (S) and the width of main collimator slit (h_c). To obtain main parameters of the beam profile, having the shape of trapezium (Fig. 6), it is convenient to introduce the following values: the projection of focus spot on the chamber plane $S = S\frac{R - R_1}{R_1}$ and the projection of collimator $h_c = h_c\frac{R}{R_1}$. Then, the width of the beam at half height H_A is equal to greater of these values and the width of slope H_B is equal to lesser of it. With the help of the chamber collimator we can reduce additionally total width of the beam.

Vertical channel shape except of the beam profile depends also on a width of image line, i. e. multiplication of scanning velocity and channel exposure time. Connection between a shape of vertical

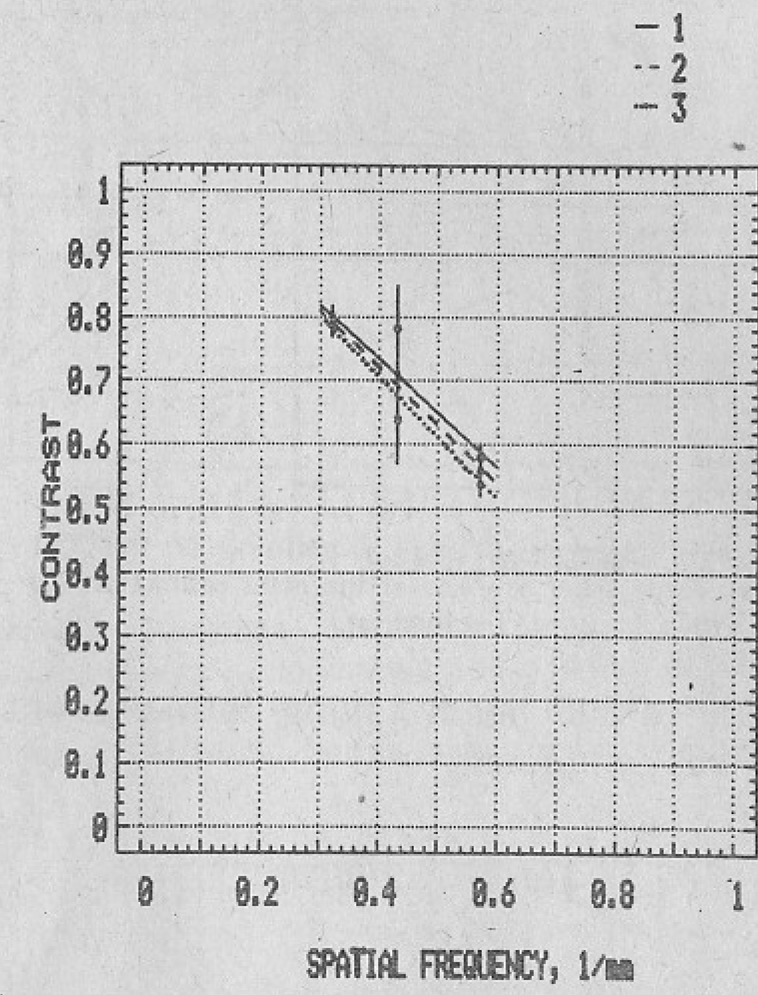

Fig. 5. Contrast-frequency response functions:
1 — at the centre of the MWPC; *2* — at the left side of (channel number ~20); *3* — at the right side (channel number ~240). Accuracy of measurements at $\nu = 0.46$ is worse than at other values, because a period of corresponding grid is about a doubled size of the MWPC channel. This makes the beats, appearing at the image of grid to have large period, which cause significant errors during calculation of C.

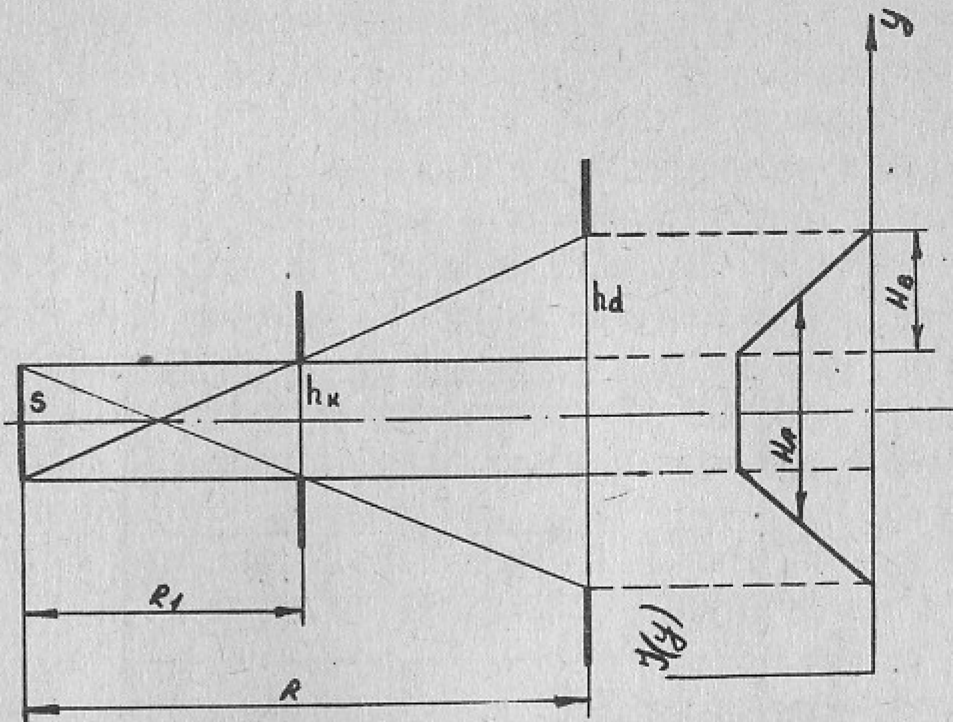

Fig. 6. Profile of collimated beam $J(y)$ depending on vertical coordinate y:
S — size of the focus of an X-ray tube; h_c — size of the main collimator; h_d — size of the chamber collimator.

channel and profile of the beam taking into account the width of image line can be easy obtained:

$$I(y) = \int_{y}^{y+\sigma} J(y)\,dy\,, \qquad (2)$$

where $J(y)$ — profile of the beam, $I(y)$ — shape of the channel. The results of calculations are convinient to represent as a set of curves with a constant value of mean-square mistake of defining a coordinate of quanta absorbtion point depending on h_d and h_c (Fig. 7). The calculation is made for $S=2$ mm, $\sigma=1$ mm, $R=1300$ mm, $R_1=650$ mm. As one can see from the figure the width of the main collimator is not practically influence the resolution. It is connected to a large focus of X-ray tube, which in fact defines the width and shape of the vertical channel. To improve spatial resolution in vertical direction it is necessary to decrease the chamber collimator. For example, when $h_d=1$ mm and $h_c=1$ mm $\sigma=0.42$ mm, that corresponds to a channel of rectangle shape with the width equal to $\sim$1.4 mm.

Decrease of chamber collimator leads to growth of X-ray tube load and increase of useless irradiating of a patient because a part of the beam penetrating through a body is not registered by the

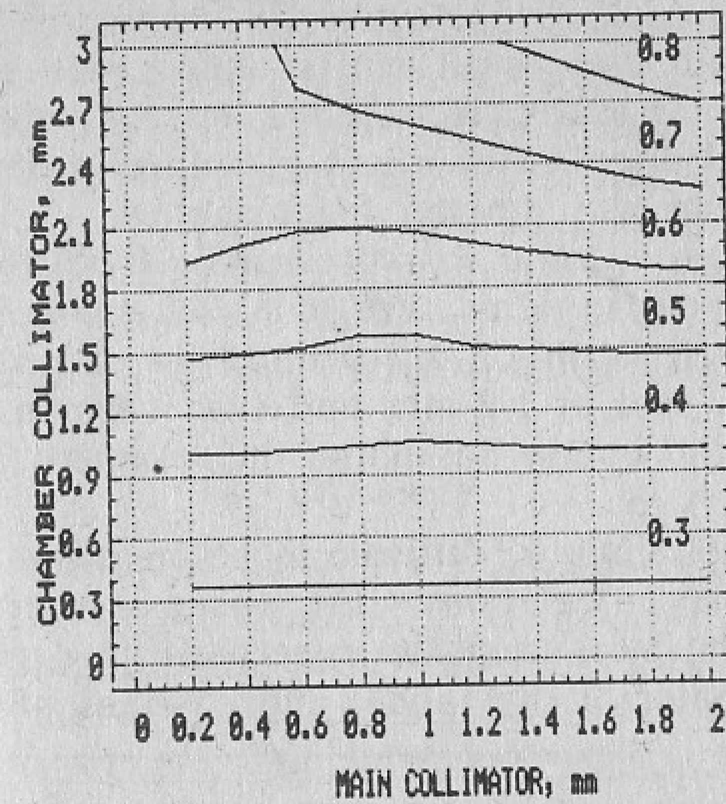

Fig. 7. Dependence of mean-square deviation of vertical coordinate of an absorbtion point σ on a width of the main collimator and a width of chamber collimator. The curves $\sigma=$const are represented on the graph, a corresponding value of σ is shown near each curve. The calculation is made for the tube with 2 mm focus.

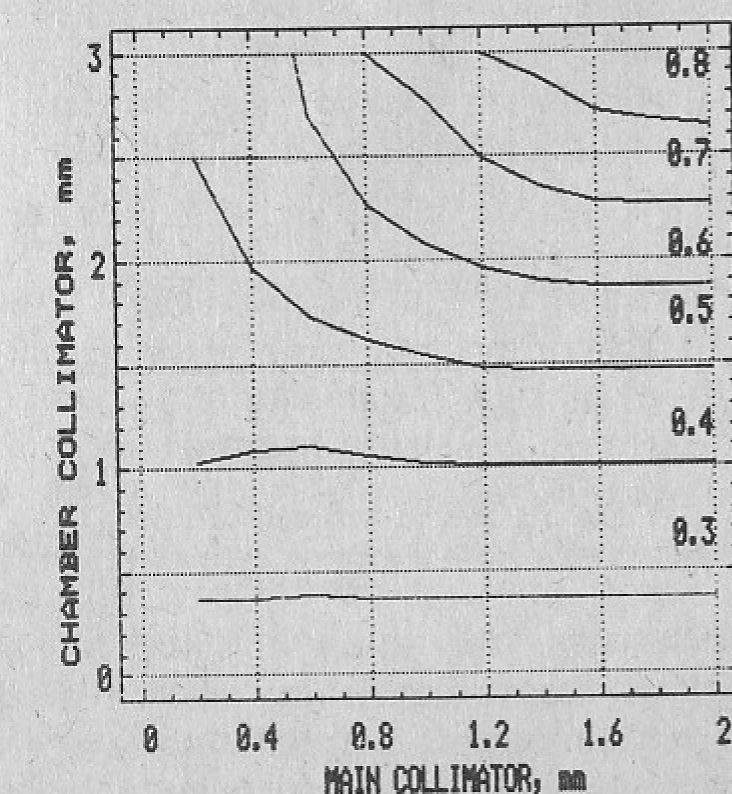

Fig. 8. the same as for Fig. 7. The calculation is made for tube focus of 1.2 mm.

chamber. For example, when chamber collimator is decreased from 3 mm to 1 mm (at $h_c=1$ mm) and counting rate is fixed, dose is increased by a factor of 2.1. In order to achieve better spatial resolution in a vertical direction at a less growth of dose it is necessary to use an X-ray tube with smaller focus spot. At Fig. 8 the diagram of σ for tube focus 1.2 mm is represented. It is easy to see that $\sigma=0.4$ mm is achieved in this case at $h_d=1$ mm and $h_c=0.6$ mm, and increase of dose is only 1.4. In future we suppose to use tubes with size of focus spot of 1.2 mm and even 0.6 mm, that will give us an opportunity to obtain additional improvement of spatial resolution.

Vertical channel shape is difficult to be measured directly, so to check the results of calculations CFR was measured with the help of set of grids. The measurements were performed with two values of chamber collimator: 1 mm and 3 mm. Values of other parameters were the following: the main collimator 1 mm, tube focus 2 mm. The results of measurments are shown at Fig. 9, curves of CFR calculated from simulated channel shapes are also represented at this graph. The results of the measurements corresponds satisfactory to the calculations.

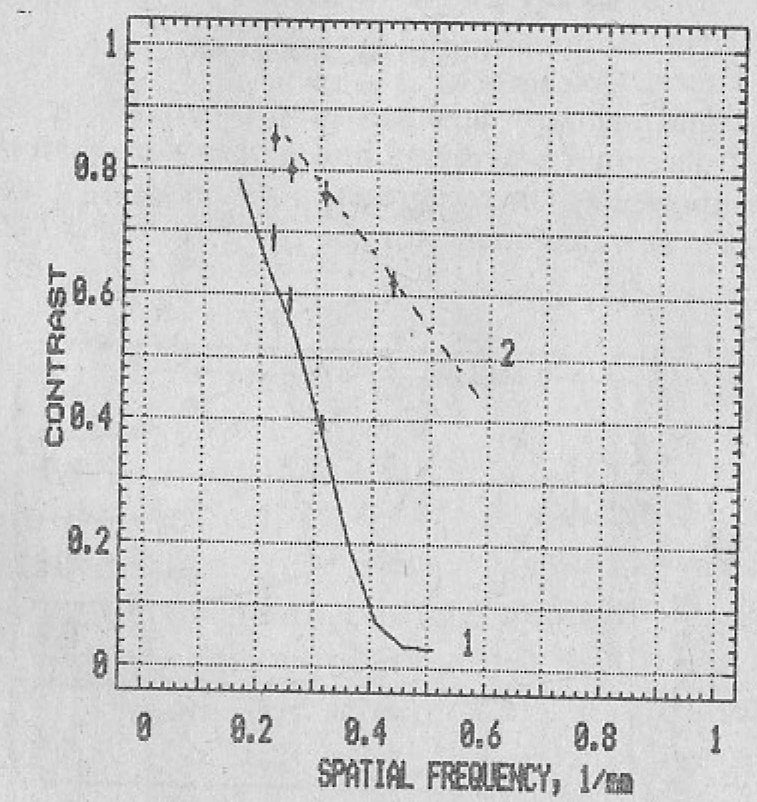

Fig. 9. CFR in vertical direction:
curve *1*—chamber collimator is 3 mm; curve *2*—chamber collimator is 1 mm.

To define spatial resolution with the help of CFR in horizontal and in vertical direction we took spatial frequency at a contrast

level of 0.5. Spatial resolution in radiography for screen-film systems and EOTs is defined at $C=0.03\div0.05$ [4]. A direct comparison of spatial frequences defined at a different C is obviously incorrect. To compare different detectors there were useful to find quantitative method of recalculation of frequences. Now we can only point out that for screen-film system spatial frequency defined at $C=0.03\div0.05$ exceeds the one defined at $C=0.5$ by 2—3 times. Approximately the same are the results of visual comparison of images obtained on films and those obtained with the help of the MWPC.

Efficiency

Quantum efficiency of the MWPC is defined by a thickness of working region $\sim$45 mm, thickness of Al inlet window—1 mm and a layer of working mixture which absorbs X-rays without detection — 27 mm (Fig. 2). The calculation made for $Xe+20\%CO_2$ mixture at 3 atm showed that at 60 keV energy efficiency is 25% that is in good accordance with a result of measurement at an energy of 57 keV (W^{181}) — 27%.

Excluding of coincidences of neighbouring channels results in dependence of a total efficiency of the detector on both the efficiency of detection of X-rays by the chamber and a part of events which cause coincidences of neighbouring channels. We measured a dependence of a number of coincidences on different factors, such as: chamber voltage, drift voltage, counting rate (at a working value of chamber voltage 2.8 kV, drift voltage 4.8 kV), X-ray tube voltage (chamber voltage and drift voltage were the same as in previous case). The measurements show that the share of coincidences can be significantly changed depending on a conditions of chamber work. For example, change of tube voltage from 70 kV to 100 kV causes increase of the number of coincidences from 30 to 42%, and growth of counting rate to 600 kHz/chan leads to decrease of the share of coincidences from 30 to 22% (at a tube voltage of 70 kV). The working voltage of the chamber, drift electrode and X-ray tube is not changed during the process of shooting. Counting rate of the chamber can be changed significantly in a different part of an image, and therefore the number of coincidences can be changed also. However as it will be shown below, it does not lead to distortions of an image because changes of efficiency caused by different share of coincidences in a different parts of an image are compensated by another effects.

Counting Rate Capability

A counting rate capability of the MWPC depends on space charge accumulated in a working volume at high rates [5, 6], and dead time of the electronics.

Errors of the electronics depend on a dead time of the amplifiers-discriminators and a dead time of selection circuit. In our case the values of these times are approximately equal to each other and make up $\sim$200 ns. The calculation of electronics errors can be done with the help of well known formulas [2]. For example, for a counting rate of 500 kHz/chan this calculation gives 36% of errors.

Space charge, filling the chamber between upper and lower cathode planes at high rates, screens a field near anode wires and causes therefore reduction of gas gain and efficiency of the chamber. Anode-cathode gap of the chamber most essentially influences the space charge effect [6]. This distance defines a thickness of space charge which reduces the field of anode wires. That was the reason to choose the gap of 2 mm in our chamber. The results of measurements of dependences of efficiency on counting rate for the chambers with anode-cathode gap of 2 mm, 3 mm and 4 mm are represented at Fig. 10. We will call by the rate capability the counting rate at which the efficiency of the chamber decreases by 20%. The rate capability of the chambers with the gaps of 2, 3 and 4 mm equals correspondingly 600, 450 and 230 kHz/chan.

The main characteristic of the counting rate capability of the chamber and registration electronics, measured at a working chamber voltage and X-ray energy (Fig. 10, curve *1*), shows that the rate capability obtained is higher than it follows from the calculations of electronics errors. This result can be explaned taking into account the decrease of gas amplification with growth of space charge due to the increase of counting rate, and therefore a part of coincidences, which at low N_{ij} were excluded by a selecting circuits now would be registered as a single events (they exceeded a threshold of registration only in one of two neighbouring channels). Such a compensation of a loss of single events counting rate due to registration of a part of coincidences permits to obtain the detector rate capability of 600 kHz/chan in spite of great number of electronic errors.

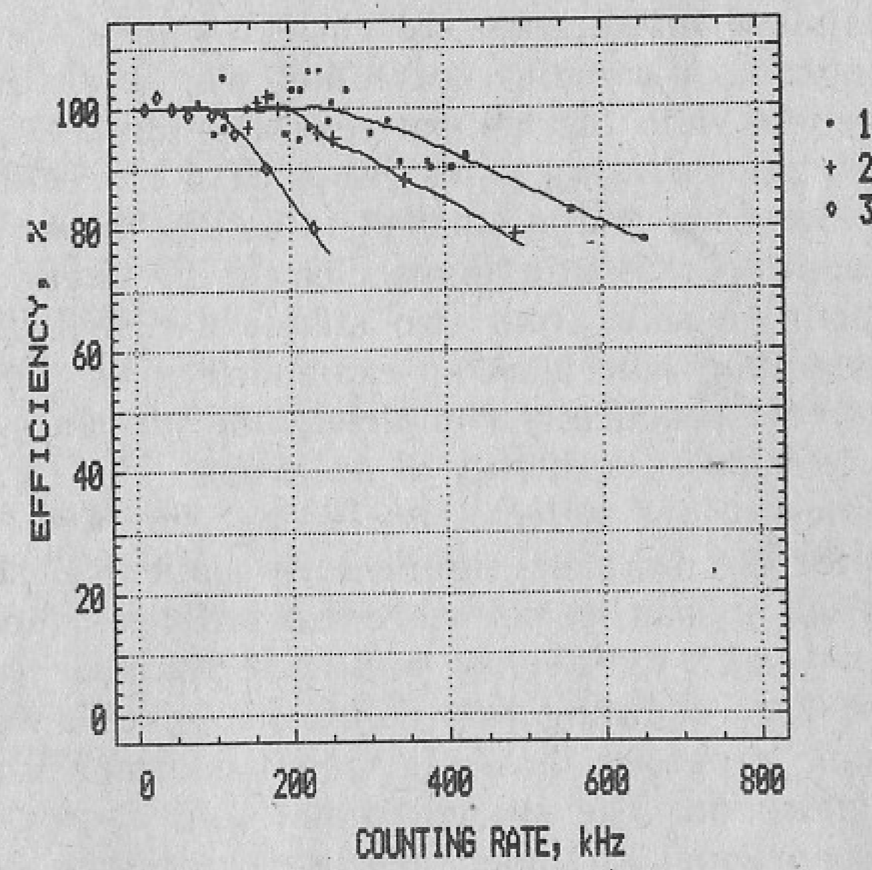

Fig. 10. Dependence of the detector efficiency on count rate: the gap between anode and cathode $l=2$ mm (*1*); $l=3$ mm (*2*); $l=4$mm (*3*).

Uniformity, Dynamic Range, Scattered Radiation, Doses

Nonuniformity of the efficiency of the MWPC channels depends on differences of thresholds of discriminators and accuracy of the anode wires step. The defects of chamber making are stable and can be corrected with the help of standard distribution of chamber counting rate measured at a uniform irradiation. The differences of the electronics thresholds can be changed in time due to the system warming. The measurements of stability of the installation [2] showed, that a time of system warming is about 1 hour, after that changes of nonuniformity with time become neglegibly small.

Except the factors enumerated above uniformity of channels of the detector depends also on chamber counting rate. Decrease of gas amplification at high rates leads to change of spectrum density near threshold of signal amplitudes. Therefore the differences of channel efficiencies are changed in spite of stable deviations of AD thresholds. For example, when counting rate changes from 100 to 300 kHz/chan mean-square nonuniformity of the detector is increased from 1.6 to 2.5%. So an image with high differences of counting rate in a different parts can not be efficiently corrected with the help of one standard distribution. To correct nonuniformity of such images the program of complex correction was worked out, which performs corrections with the help of several standard distributions obtained at different counting rates. At such a correction normalizing coefficient for each image element is defined separately taking into account counting rate in a given element by linear approximation of corresponding sells from two standard distributions, which have average counting rate close to examining pixel. This program permits to eliminate practically completely the influence of nonuniformity of the detector on a quality of an image.

High efficiency of the detector in comparison with screen-film systems permitted to decrease significantly patient doses. Besides that the installation has some more advantages. Measuring of radiation intensity with the help of multiwire chamber, having zero background and high counting rate capability, gives an opportunity to widen dynamic range of the installation in comparison to other methods of registration. The slit collimator and narrow inlet window of the proportional chamber practically exclude detection of radiation, scattered in a patient body. These both features of digital radiographic installation of scanning type improve image contrast and permit a doctor to be limited to one image to study both soft and dense tissues.

When discussing a question about irradiation doses one must take into account accumulation factor and reduction of doses for inner organs in comparison with surface dose. It is necessary to notice that these both factors are equal for standard radiography (screen-film systems, image intensifiers and so on) and for slit method of scanning digital installation. As we are interested mainly in a comparison of doses with ordinary radiography, accumulation factor can not be taken into account for these calculations.

Surface patient irradiation dose at a slit scanning method depends on the time of irradiating an arbitrary fixed point at a body surface, turned to X-ray tube. This time $T=\frac{h_c R_2}{V R_1}$, where h_c is the width of collimator slit, V is scanning velocity, R_1 and R_2 are the distances between the tube focus and collimator and surface of patient body correspondingly. In our case at $h_c=1$ mm, $V=34$ mm/s, $R_1=650$ mm and $R_2=1000$ mm the time of irradiation is $T=45.2$ ms. If one know tube voltage and current and thickness of X-ray filter used to obtain an image, irradiation dose

can be obtained from tables of Ref. [7] or can be calculated with the help of empirical formulas. Images of chest of good quality were obtained by the installation at a surface dose of 3 mR. Images of stomach were obtained at a dose of 18 mR. The results of measurements performed with the help of dosimeter DRG3-04 differ from calculated values less than 15%. The measurements were performed without a fantom, and so the result does not take into account the accumulation factor.

Calculated doses at a front and side images of pregnant women, obtained, as it was described above are equal to 40 mR and 80 mR correspondingly. Irradiation dose to obtain front image of pregnant woman was measured jointly with specialists of Moscow Scientific-Research Roentgen-Radiological Institute and it was equal to 40 ± 10 mR. To obtain the same image on film of RM-1 with EUI-1 screen and 10:1 anti-scatter grid the dose was about 1700 ± 30 mR. So the digital installation for a front image in pelvimetry decreases doses by 40 times [8].

METHODS OF FURTHER IMPROVEMENT OF THE DETECTOR CHARACTERISTICS

The first installations had MWPC with 256 channels and an image was 314×256 mm size (in the chamber plane). For several problems, for example, to study organs of chest, it is necessary to have an image larger. For this purpose a future installations will have a chamber with 320 channels, image size of 392×256 mm with a number of pixels of $320\times256\approx82000$. Obviously, a vertical dimension of an image can be increased to the necessary value (the restriction is permitted tube power at increasing exposure time when X-raying especially «thick» objects and necessity of allocating a large computer memory).

Futhermore, these installations will have selecting circuits and scalers directly on the chamber and all this electronics will be made of special hybrid circuits. This reconstruction will permit to exclude hundreds of wires connecting AD with registration unit and significantly improve reliability of the installation.

Digital radiographic installation must define optimum voltage and current to obtain image of good quality at a maximum sensitivity to a small deviations of density. Performed calculations show that if several calibration measurements have been made, it is possible to obtain a distribution of X-rays along one line in the middle of an image for a standard voltage and small tube current and to define with the help of computer an optimum regime for an image. The program of calculations must take into account limit of tube power. In the near future we suppose to check this method of defining a regime.

Multiwire proportional chambers with fan anode plane and nonparallel anode and cathode planes were successfully used for the first three digital radiographic installations. Such a chambers give an opportunity to obtain spatial resolution of $\sim$1 mm at an object, that is quite sufficient for the majority of problems in radiographic diagnostics. At the same time it would be useful to improve spatial resolution of the chamber for a several X-ray studies. Formally, the decrease of anode wires step to 0.5—0.6 mm would be sufficient for this purpose. However, practically, this way is not perspective. To make a chamber with such step one need to decrease diameter of anode wires or to work at higher voltages. Simultaneously it will be necessary to improve the accuracy of anode plane manufacturing. Thus, these methods give the chamber with worse characteristics and, that is the most important, less reliable.

Another more attractive method to improve spatial resolution is to use events, connected to coincidences in the neighbouring channels, which at present time are rejected by the selecting circuit. For this purpose it is necessary to register separately the events with hit of single anode wire and those which have two neighbouring wires operated simultaneously. As the value of electron range is comparable with the step of anode wires, about a half of all events gives such coincidences. As the results of measurements, mentioned above, show, at a pressure of 3 atm a part of coincidences is about 30—40% for the tube voltages of 70—100 kV. Geometrically, X-ray quanta, absorbed in the region between anode wires, cause coincidences, quanta, absorbed near an anode wire are registered as single events.

The total number of channels for such method of readout is increased by a factor of two, approximately the same is the improvement of spatial resolution.

To check the operation of such readout system we made measurements of the shape of single event channel and the shape of coincident event channel. For this purpose selecting circuits were disconnected from a group of anode wires, and the circuit, simulating

described method of readout, was assembled. Efficiencies of coincident event channels and single event channels were equalized by a decrease of pressure inside the MWPC to 2 atm. The results are shown at Fig. 11. For a channel of coincidences the FWHM is

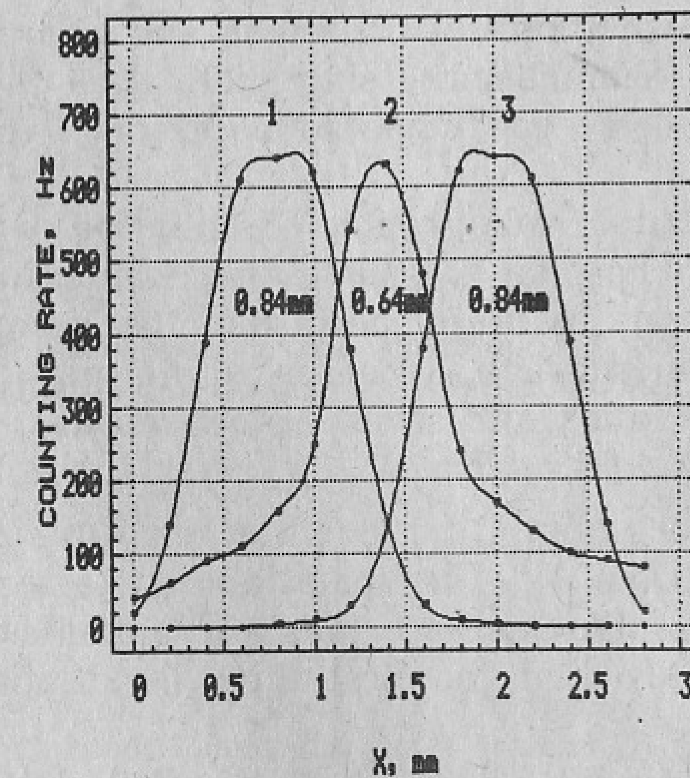

Fig. 11. the curves of channel shape with modified selection circuits: *1, 3*—single event channels; *2*—coincident event channel.

0.64 mm, for single event channel this value is 0.84 mm. This result confirms a supposition of an opportunity to improve spatial resolution for such method of readout without a reconstruction of chamber and associated electronics (AD). Furthermore, the usage of coincident events significuntly increase the total efficiency of the detector.

To improve spatial resolution along another coordinate it is necessary to diminish correspondingly the width of the slit collimator and an exposure time of an image row. In this case the number of pixels will be increased by a factor of four. Element size will be diminished to 0.5$\times$0.5 mm (at an object), the number of pixels will be 640$\times$512$\approx$330000.

CONCLUSIONS

During the four years of operation of the first digital installation in the all-union scientific research centre for maternity and child protection more than a thousand of pregnant women were studied. The pelvimetry in a front and side projections were performed.

The method of division of an image into two parts with separate regulation of contrast was very useful here. The dimensions between a chosen points are calculated taking into account two projections and estimation of area for pelvis and child head cross-sections are performed. As a result of great number of studies the connection between the correlations of the areas and a probability of birth complications was established. Using these data and the results of measurements a doctor can come to necessary decision with more reliability [9].

The second digital radiographic installation was given to the Novosibirsk region hospital in 1987. The chair of radiology of the Novosibirsk Medical Institute performs the comparison of this installation and fluorograph, research of normal and pathological state of lungs ventilation and several other works.

The third installation with horizontal position of patient works in the INP from the beginning of 1988. It is used for studying the characteristics of new MWPC. This installation is also used by a polyclinic department of the Institute to study patients. The total number of patients studyed at the first three installations exceeds one and a half thousand.

In conclusion several words about another digital installations for radiography.

In the advertisment of «American Science and Engineering» digital installation named «Microdose» is described, which has both coordinates scanning mechanically. Such approach simplifies requirements to a detector but makes mechanics more complex and for other equal conditions demands the increase of X-ray tube power by two orders. According to our knowledge, this installation is not produced by industry now.

«Toshiba» worked out installation which has a special phosphor with memory instead of film. After exposure a cassete with phosphor is carried to a special apparatus which scans a phosphor plate by a laser beam, reads and digitizes information. The installation has spatial resolution of 0.2×0.2 mm and wide dynamic range. Irradiation doses are reduced by a factor of 10 [10], though according to some data it can be smaller. Together with these significant merits the installation has several shortcomings. It is not exclude scattered radiation decreasing contrast. Image appears only after transportation a cassete into the readout device and digitizing. Precise mechanical device is necessary for scanning by a laser beam. High cost of the installation (about one million dollars) is obviously connected to this factor. «Siemens» began to produce the installation of the same type.

Authors express gratitude to G.M. Parkhomchuk, who worked out the first image processing programs, to V.A. Jakovlev and O.W. Tekutiev for help in the work. Authors are grateful to doctors P.I. Denisov, W.P. Shurigin and Ju.B. Jurchenko for the study of diagnostic possibilities of the digital installation and useful discussions.

REFERENCES

1. *S.E. Baru, A.G. Khabakhpashev, I.R. Makarov, G.A. Savinov, L.I. Shekhtman, V.A. Sidorov.* Digital X-Ray Imaging Installation for Medical Diagnostics, Nucl. Instr. and Meth., A238 (1985) 165.
2. *S.E. Baru, A.G. Khabakhpashev, L.I. Shekhtman.* Multiwire Proportional Chamber for Digital Radiographic Installation, Preprint INP 89-39, Novosibirsk, 1989.
3. *J.E. Bateman, M.W. Waters, R.E. Jones.* Spatial Resolution in a Xenon Filled MWPC X-Ray Imaging Detector — a Computing Physics Approach. — RL-75-140. Detector Physics Group, High Energy Physics Division, Rutherford Laboratory (1975).
4. *N.N. Blinov, E.M. Jukov, E.B. Koslovsky, A.I. Mazurov.* Television Methods of Processing of X-Ray and Gamma Images. — Moscow: Energoizdat, 1982.
5. *R.W. Hendricks.* Space Charge Effects in Proportional Counters. — ReV. Scl. Instr., 40, 9 (1969) 1216.
6. *E. Mathieson.* Dependence of Gain on Count Rate, Due to Space Charge in Coaxial and Multiwire Proportional Chambers. — Nucl. Instr. and Meth., A249 (1986) 413.
7. Rentgenotekhnika, Ref. Book Edited by V.V. Kluev, Moscow: Mashinostroenie, 1980, b.1, 63.
8. *A.I. Volobuev, P.I. Denisov et al.* Irradiation Doses During Studies at the Digital Radiographic Installation. — Vestnik Rentgenologii i Radiologii, 1, 1989, 59.
9. *A.I. Volobuev, P.I. Denisov et. al.* Usage of the Digital Radiographic Installation in Obstetrical Clinic. — Submitted to «Akusherstvo i Ginekologija».
10. *R. Schittenhelm.* Imaging Systems for Digital Radiography: Present Status and Future Prospects. — Electromedica 54 (1986) N 2.

E.A. Babichev, S.E. Baru, V.V. Gusev,
A.G. Khabakhpashev, G.M. Kolachev, G.A. Savinov,
L.I. Shekhtman, V.A. Sidorov, A.I. Volobuev.

**Digital Radiographic Installation
for Medical Diagnostics**

Е.А. Бабичев, С.Е. Бару, А.И. Волобуев,
В.В. Гусев, G.M. Kolachev, Г.А. Савинов,
В.А. Сидоров, А.Г. Хабахпашев, Л.И. Шехтман

**Цифровая рентгенографическая установка
для медицинской диагностики**

Ответственный за выпуск С.Г.Попов

Работа поступила 21 августа 1989 г.
Подписано в печать 28.08 1989 г. МН 10353.
Формат бумаги 60×90 1/16 Объем 1,4 печ.л., 1,1 уч.-изд.л.
Тираж 290 экз. Бесплатно. Заказ № 74.

Набрано в автоматизированной системе на базе фотонаборного автомата ФА1000 и ЭВМ «Электроника» и отпечатано на ротапринте Института ядерной физики СО АН СССР,
Новосибирск, 630090, пр. академика Лаврентьева, 11.