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I.Kovar*, A.G.Molokanov, G.V.Mytsin, F.Spurny*,
R.Wagner*, V.P.Zorin

DOSIMETRY OF MEDICAL PROTON BEAMS AT THE JINR PHASOTRON IN DUBNA

*Institute of Radiation Dosimetry, Praque

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1. INTRODUCTION

The proton therapy has a number of important advantages over conventional radiation therapy with γ-rays and electrons. Realization of these advantages requires higher precision of proton beam dosimetry and dose field and target overlapping. Various types of accelerators used for proton therapy with different energy, time structure and spatial beam distributions lead to various dosimetry methods.

In view of rapidly increasing number of accelerators used for radiation therapy with heavy charged particles international dosimetry protocols were elaborated for uniformity in beams calibrations. There exist "Protocol for Heavy Charged Particle Beam Dosimetry" written by Task Group of the AAPM (American Association of Physicists in Medicine) [1] and "Code of Practice for Clinical Proton Dosimetry" elaborated by ECHED working group (European Clinical Heavy Particles Dosimetry Group) [2]. These protocols make recommendations for the determination of absorbed dose to tissue and for measurement accuracy.

A six-compartment clinico-physical facility for radiation therapy with proton, negative pion and neutron beams was realized at the JINR phasotron after its conversion. The clinico-physical facility consists of several medical channels: three therapeutic proton beams with energies from 100 to 660 MeV; a negative pion beam with energy up to 80 MeV; a therapeutic neutron beam with the mean energy about 350 MeV and a therapeutic γ-unit with the ^{60}Co source.

This paper describes a method for determination of the dose rate absorbed by tissue for medical proton beams on a basis of clinical dosimeter calibration with the ^{60}Co γ-source, the main parameters of detectors used for measurements of spatial dose distributions, results of ion recombination correction factors in air thimble ionization chambers measurements.

2. MEASUREMENTS OF ABSORBED DOSE RATE FOR THERAPEUTIC PROTON BEAMS AT JINR PHASOTRON IN DUBNA

For absorbed dose rate measurements of therapeutic proton beams we use the KD-27012 clinical dosimeters with thimble air ionization

chambers VAK-251 (volume 50 mm^3) and VAK-253 (volume 1.5 cm^3) with air-equivalent walls.

Dosimeter calibration was made with the ^{60}Co source of the therapeutic γ-unit, placed in one of the cabins of our clinico-physical facility. The ^{60}Co source was calibrated against the primary standard of the Prague Institute of Radiation Dosimetry. The accuracy of the γ-unit calibration is 1.3 % (one standard deviation) [3].

Using the γ-unit as a calibrated stand for ionization chambers of our clinical dosimeters was described in [4] and practically coincided with recommendations from [2].

For the KD-27012 dosimeter we obtained the exposure calibration factor for the ^{60}Co source N_x: [R/reading]. The air kerma calibration factor in the ^{60}Co field N_k [Gy/reading] may be obtained from the exposure calibration factor N_x.

The absorbed dose calibration factor for the proton beams A_{cal} may be obtained from the calibration of the ionization chamber in the ^{60}Co source using the relations:

$$A_{cal} = N_k * C_p = N_x * (W_{air}/e)_\gamma/(1-g) * C_p \ .$$

The proton conversion factor C_p may be obtained using the formulae from [2]:

$$C_p = A_{wall} * [(\bar{S}/\rho)^{tissue}_{air}]_p * k \ ,$$

where

$$k = (1-g) * \frac{(W_{air}/e)_p}{(W_{air}/e)_\gamma} * \frac{[(\bar{\mu}_{en}/\rho)^{wall}_{air}]_\gamma}{[(L/\rho)^{wall}_{air}]_\gamma} .$$

For our dosimeter we use the following coefficients from [2] and [4]:

$A_{wall} = 0.99 \pm 0.01$ - the factor that takes into account the absorption and scattering of γ-rays produced in the ionization chamber walls;

$[(\bar{S}/\rho)^{tissue}_{air}]_p$ - the ratio of the mass stopping powers of tissue to air for the proton beam - may be obtained from [6]. For proton energies 50 - 1000 MeV this ratio varies very little and may be used as a constant value - 1.136 ± 0.016 (1.4 %);

$(1 - g)$ - correction to bremsstrahlung in air in the ^{60}Co beam ($g = 0.003$);

$(W_{air}/e)_p = 35.2eV$ (4.0%) $(W_{air}/e)_\gamma = 33.97eV$ (0.2%) - the average energy required to produce an ion pair per unit charge in dry air for the γ-rays and protons [2];

$[(\bar{\mu}_{en}/\rho)^{wall}_{air}]_\gamma = 1.0227$ (1%) - the ratio of the mean mass energy absorption coefficients of chamber wall material to air for γ-rays for our ionization chambers from [4];

$[(\bar{L}/\rho)^{wall}_{air}]_\gamma = 1.0076$ (1%) - the ratio of the mean restricted collision mass stopping powers of chamber wall material to air for γ-rays [4].

For the VAK-253 chamber $k = 1.049$ (as compared with $k = 0.993$ for A-150 TE plastic[2]), the proton conversion factor from γ-rays $C_p = 1,18$ and the calibrated conversion factor from dosimeter scale in "roentgen/min" units to absorbed dose rate in "Gy/min" is $A_{cal} = 0.0104$ [Gy/min /"roentgen/min"]. So, for the KD-27012 dosimeter, calibrated at the ^{60}Co γ-source in exposure dose units the reading "100 roentgen/min" corresponds to 1.04 Gy/min. The full estimated uncertainty for our dosimeter calibration is 4,7 % (one standard deviation).

3. DETECTORS FOR SPATIAL DOSE DISTRIBUTION MEASUREMENT

The miniature semiconductor silicon detectors are usually used for spatial dose distribution measurements of our beams. The small size of the sensitive volume makes it possible to measure spatial distributions for beams with high dose gradients. The sensibility of the silicon detectors is 10^4 times higher than that of an ionization chamber with the same volume (the energy for electron-hole pair production is ≈ 10 times smaller than that for production of an ion pair in the air and the density of Si is 10^3 higher than the density of air). The pulse from silicon detectors is shorter than that from the air ionization chamber and this allows one to use them with computer controlled systems.

In our measurements we use lithium-drifted silicon detectors with a volume of several mm^3 and commercially available diodes KD-208

and KD-209. The sensitive volume of these diodes has a form of 1 mm diameter by 0.2 mm thick disc. Such a small detector size permits one to use them for measurement of narrow beams with diameters up to several mm.

But the silicon detectors cannot be used for absolute dose measurements because their sensibility decreases during irradiation because of radiation damages of the silicon crystal and its temperature instability. For absolute dosimetric measurements diamond detectors may be used after their calibration.

The diamond detector consists of a diamond crystal with sensitive volume from 1 to 5 mm^3, 0.1 - 0.4 mm thick, a preamplifier placed near the crystal and a current measuring device. The diamond detector has high sensitivity, high spatial accuracy owing to its small size, high radiation firmness and good tissue equivalence [7,8].

The spatial dose distribution measurements for proton beams may be distorted because of energy dependence of sensitivity of different types of detectors to protons.

But for protons, despite the fact that the values of dE/dx strongly change with energy, their detector material-to-tissue ratios in a wide range of energies vary very little (Fig.1) and these dependences do not lead to significant distortion of spatial dose distributions. The distortion may be visible only with silicon detectors for low energy protons, i.e. in Bragg peak region. For a modified Bragg peak and for beams with a wide shape of the Bragg peak these distortions must be insignificant.

For high levels of dose rate measurements with ionization chambers there may be the distortion caused by ion recombinations in gas within the chamber volume. Fig.2 shows the depth-dose distributions measured with silicon, diamond detectors and with an air ionization chamber for two levels of dose rate. All distributions were normalized at the point with minimal depth of water. Whereas for slow dose rate these distributions practically coincide, for high level of dose rate they have some discrepancy caused by ion recombination.

The spatial dose rate distribution measurements are computer-controlled. The device "ISODOSEGRAPH" [9] is intended for automatic measurements of dose field in the water bath or in air. The electromechanical system can move a small detector in three perpendicular di-

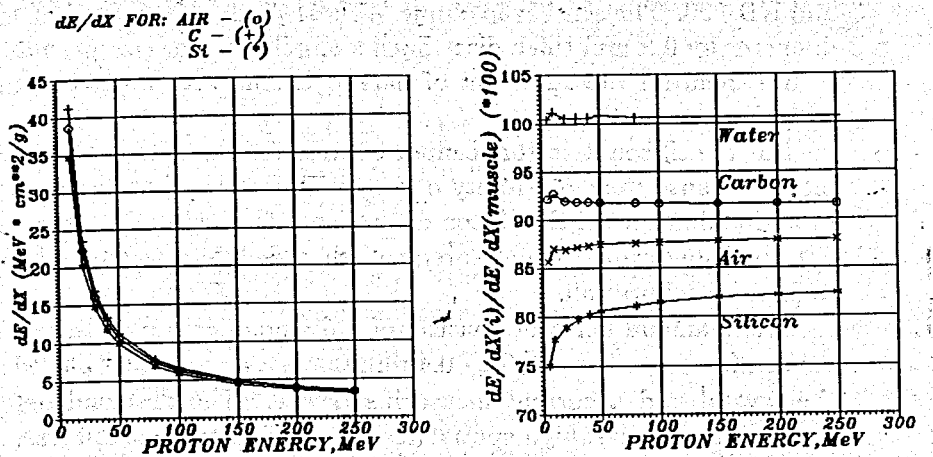

Fig.1. dE/dx for various materials from [6] and their ratio to dE/dx for biological tissues

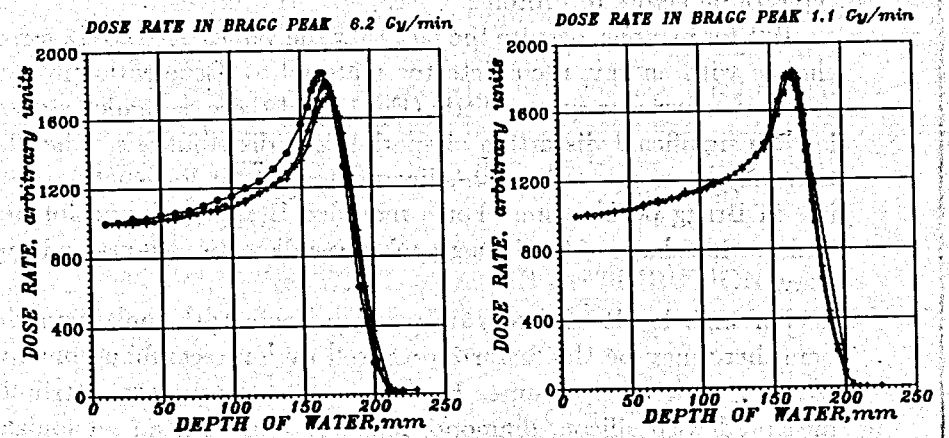

Fig.2. Depth-dose distributions for proton beams measured with various detectors (+ - Air ionization chamber; * - Silicon detector; o - Diamond)

rections. The accuracy of each coordinate measuring is 0.1 mm. The result of measurement is a beam profile or isodose maps.

Another system for beam profile measurements is a diode line scanner [10]. It consists of 28 silicon diodes placed in line and is intended for measuring a beam profile in air.

Recently we have constructed a system from silicon diodes of cross-like configuration. This system allows dose field measurements in horizontal and vertical plates simultaneously.

4.CORRECTION FOR ION RECOMBINATION IN AIR THIMBLE IONIZATION CHAMBERS

The need for saturation corrections for an ionization chamber in proton beam measurements is caused by its higher value for beams with pulse time structure than for continuous radiation. The time structure of the phasotron external proton beam consists of 40 μs long pulses with a frequency of 250 Hz, the calibration γ-field has a continuous value of the dose rate. Different time structure of the proton beam and the calibrated γ-field may lead to errors in chamber calibrations.

The charge collection efficiency f (*f = charge collected / charge produced*) (or saturation coefficient $K_{sat} = 1/f$) depends on the configuration of the chamber, nature of the gas within, the rate of charge production within the effective volume and the electric field strength. A number of papers [11 - 15] described experience of charge collection for different types of chamber and for different types of radiation.

Extrapolation methods are usually used for the determination of the collection efficiency f. One can obtain f from plots of the current from the chamber versus the voltage by extrapolating the curves to the ordinate to represent infinite applied polarizing voltage.

For continuous radiation the collection efficiency of ionization chambers may be written as:

$$f = \frac{1}{K_{sat}} = 1 - \frac{sq}{V^2} ,$$

where q - is the volume rate of charge formation in the chamber,
V - is the collection voltage,
s,t - are the parameters determined from the configuration of the chamber and the nature of the gas within.

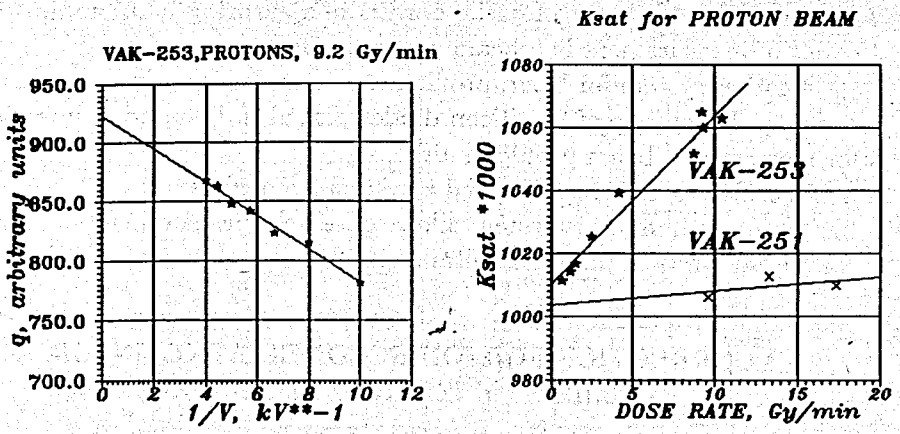

Fig.3. The example of charge collected q vs reciprocal of the polarizing voltage 1/V (left) and K_{sat} vs dose rate for proton beam (right)

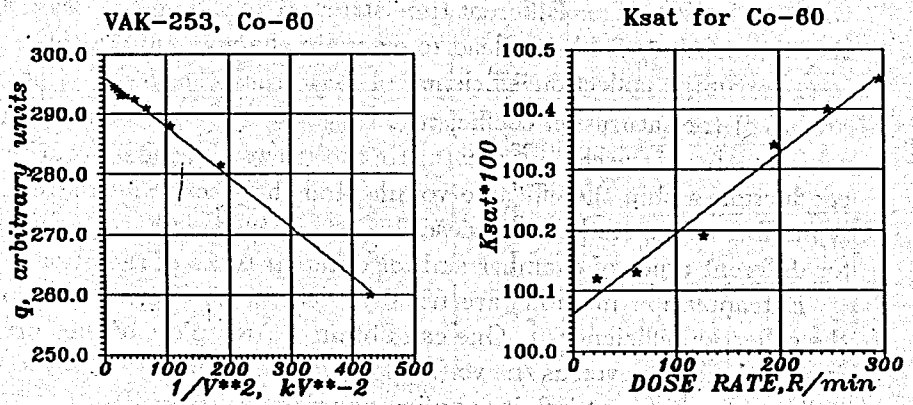

Fig.4. The example of charge collected q vs reciprocal of the polarizing voltage square $1/V^2$ (left) and K_{sat} vs dose rate for ^{60}Co γ-source (right)

The collection efficiency of ionization chambers for pulsed radiation is:

$$f = \frac{Vln(1+\frac{tr}{V})}{tr},$$

where r is the charge produced per unit of volume per pulse [11].

For the values V when $\frac{tr}{V} \ll 1$ this formula may be transformed to

$$f = 1 - \frac{tr}{2V}.$$

This approximation gives an accuracy of 5% for $f \geq 0.8$ and 0.5% for $f \geq 0.95$ [11].

By plotting the current from the chamber versus the reciprocal of the voltage $\frac{1}{V}$ for pulsed radiation and the reciprocal of the voltage square $\frac{1}{V^2}$ for continuous radiation, the true ionization current may be obtained by extrapolation of the curves to infinite applied polarizing voltage.

The quantities K_{sat} for air thimble ionization chambers VAK-251 and VAK-253 were measured with the ^{60}Co γ-source at the therapeutic γ-unit and with a proton beam with mean energy of 200 MeV and in the Bragg peak for different dose rates. Ionization current was measured by the VA-J-18 dosimeter, which allows one to decrease collection voltage from standard U = 250 V down to U = 50 V. The ionization chambers were placed in air on the proton beam axis. The proton beam was monitored by a parallel plate transmission ionization chamber located at the entrance of the proton beam to the treatment room.

The values of K_{sat} were calculated from extrapolation of plots of the current from the ionization chamber versus the reciprocal of the voltage to 0 ($\frac{1}{V} \longrightarrow 0$) for pulsed beams and versus the reciprocal of the voltage square to 0 ($\frac{1}{V^2} \longrightarrow 0$) for continuous radiation.

Typical dependences for pulsed and continuous radiation are plotted in Figs. 3 and 4 (left). The solid lines in these pictures are a linear regression (least square fit) of those data from which we can determine the values of K_{sat} . The dependences of K_{sat} versus dose rate for pulsed and continuous radiation are plotted in Figs. 3 and 4 (right).

The results of K_{sat} measurement for the VAK-253 ionization chamber with continuous radiation are in good agreement with the data for the same chamber from [15].

From the comparison of the values of K_{sat} for pulsed and continuous radiation we can see that for pulsed beams from the JINR phasotron recombination is approximately ten times higher than that for continuous radiation. Thus, a correction for ion recombination for proton beams must be taken into account.

It is found that the error of JINR phasotron proton beams dosimetry is about 5%. This accuracy meets the international requirements for the therapeutic proton beams.

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