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V.Kostjuchenko, D. Nichiporov

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A REWIEW OF INSTRUMENTS AND METHODS.

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1. Peculiarities of the ITEP proton clinical beam and special means of clinical dosimetry.

In this report we will concentrate only on the beam parameters which influenced considerably on the choice of means and methods of clinical dosimetry and beam monitoring.

ITEP synchrotrone proton beam has a well pronounced pulse structure: spills ~100 ns long follow with ~3 s intervals. Maximum spill intensity of ~10^{18} prot/s and flux densities of ~10^{15} - 10^{16} prot/(cm^{2}s) are characteristic values. As a rule, other machines used for clinical applications operate at intensities 10^5 - 10^9 times lower than that of ours. Extremely high pulse intensity of the beam determined the choice of a monitor to control dose in a treatment session and the choice of a standard dosimetry system. Another circumstance that had to be taken into account, was a considerable electromagnetic noise. Because of these two difficulties, at the time when the concept of our clinical dosimetry has been forming, we had to decline the traditional ionization method. Semiconductors and photographic films could not easily be employed, either.

2. Absolute dosimetry: a method of induced activity.

In 1960-s, when it became necessary to measure the absorbed dose to tissue with an accuracy sufficient for clinical application, calorimeters had not

been developed to a proper extent. Gamma-calibrated ionization chambers could not be used, either, because of specific parameters of the ITEP beam. Owing to these objective reasons and because of the fact, that it was physicists who challenged the problem, a method of induced activity by the $^{12}C(p;pn)^{11}C$ reaction was chosen to measure the number of particles in the beam. It has become a standard dosimetry system at ITEP.

This choice came from the following considerations,

- the measurement technique was well-known and wide-spread;
- the excitation function of the reaction was available from literature [1,2] and had been claimed to have an uncertainty not more than ±5% in all range of interest (i.e. for proton energies from 40 to 200 MeV);
- ^{11}C is a pure positron emitter and this fact allowed us to design an induced activity measuring instrument. It employs the method of β-γ coincidences, and its design eliminates the necessity to determine accurately the counting efficiencies for either of counting channels;
- a taget in which activity is induced, is a solid piece made of scintillation polystyrene and, therefore, it is not subjected to electromagnetic noise. The size of the target may be varied widely providing greater convenience of work;
- proton stopping power, $^{dE}/_{dx}$ is known as a function of proton energy, with a good accuracy [3,4], while the energy of particles can be determined accurately by their residual range.

Taking all this into account, one had a real possibility to work out a method of the determination of absorbed dose to tissue, which would be based on the measurement of the number of particles using the induced activity

technique. The overall error was estimated to be ±7% [5] and was considered to be sufficiently small to start patient treatments.

3. Beam monitor: a Current Monitor.

Spill duration being very short, the beam current is sufficient to use a method of magnetic induction to measure the number of particles. Hence, a current monitor has been developed to monitor the beam. It proved to be a simple and quite reliable instrument which operates on the principles of a current transformer. Electromotive force generated in the winding of CM by the proton beam is proportional to the number of particles having traversed the monitor's measuring aperture [6]. Current monitor can be independently calibrated against a metrologically tested pulse generator with a wire put through the monitor's aperture to simulate the beam. Being accompanied by a precision nanosecond pulse generator, the current monitor can probably compete with a Faraday cup (FC), which is a traditional means of beam monitoring.

Now the design of the current monitor's reading circuit is improved so as to monitor the beam just downstream of the final (patient) collimator [7]. The error of relative measurements is estimated to be ±3%. Absolute calibration of CM is performed by the standard method of induced activity.

4. Phantom dosimetry: dose distribution measurements.

To irradiate the target properly, one has to know dose distribution created in a tissue-equivalent phantom by the beam. Most often, lucite and water are used as phantom materials. Obviously, a good field detector should have a high spatial resolution and linearity of response to the absorbed dose to tissue.

4.1. Photographic technique.

Today, most of the measurements are performed with a photografic film detector [8]. A photo-densitometric method comprising 3-D computer reconstruction of dose distribution has been developed. Non-linearity of emulsions' energy response is corrected with the results of the measurements taken with a wide-aperture parallel plate air-filled ionization chamber.

4.2. Silicon detectors.

Our cooperation with Gustaf Werner Institute (Sweden) resulted in the development of a p-Si semiconductor dosimeter capable of working in beams with a high spill dose rate [9]. The semiconductor technique is quite helpful in many cases. However, low radiation stability of Si along with the time structure of the ITEP beam, limit potentialities of the method. Additionally, non-tissue-equivalence of Si must distort the results of depth-dose distriburion measurements in the Bragg peak region. So, much hopes are pinned on a natural diamond-based detector which must be free from the drawbacks inherent in silicon dosimeter.

4.3. Ionization chambers.

Ionization method in dosimetry has been developed at ITEP in a special way. According to Boag's theory, for an ionization chamber to be efficient in a beam like the one at ITEP, it has to operate at the ultimate tension of the electric field. In this case the chamber must have a linear response to the absorbed dose, and no corrections in the working range should be applied. We use air-filled parallel-plate ionization chamber with electrodes made of aluminum-coated Mylar. The chamber has an aperture big enough to overlap the whole beam; the gap between the electrodes is 1 mm wide. The chamber

is destined to operate in pulse beams, measuring not the charge collected on the electrodes, but a value proportional to it.

Scientific cooperation with the Institute of High-Energy Physics has yielded ionization chambers of different design. These are a multi-strip parallel plate ionization chamber and a small-volume parallel plate chamber with the walls made of tissue-equivalent plastic.

The multi-strip chamber has glass walls, each having a series of sixteen 5-mm wide aluminum strips coated upon them. The series of strips are arranged orthogonally to each other. The chamber may, therefore, serve as a profilometer, providing information on horizontal and vertical distributions. An integrated output signal can also be measured. Hence, the chamber can also be used as a second monitor of the beam additionally to the current monitor.

The small-volume ionization chamber has electrodes 5 mm in diameter and a gap of ~0.1 mm between them. Constancy of volume is one of the advantages of the chamber. The chamber's design permits the instrument to operate in a water phantom.

The prototypes of these two models of ionization chambers are currently passing various tests in the proton beam. Later the chambers will hopefully be included into a range of field instruments used for the purposes of clinical dosimetry at ITEP.

4.4. Thermoluminescent dosimetry.

We have some experience in using of TLDs for dosimetry intercomparison; several series of such experiments were carried out in the USSR at ITEP, Central Scientific Institute of X-rays and Radiology, and the Institute for

Nuclear Research (both St.Petersburg) [10] and at the Gustaf Werner Institute. Paul Todd, during the RBE measurements at ITEP in 1977, performed a test measurement using TLDs [11]. Dosimeters were supplied and the results were read by the Department of Radiation Medicine of the Massachusetts General Hospital.

At present, however, ITEP does not possess of any of TLD equipment.

4.5. Photoluminescent dosimetry.

The main advantage of photoluminescent dosimeters (PLD) [12] is that information may be read for many times without erasure and that it can be preserved for years. Therefore, PLD may be used as witness dosimeters and they may be arranged in a kind of a dosimetry "library". One of the drawbacks of conventional PLDs is rather big dimentions (12x12 mm, 4 mm thick).

5. Beam delivery techniques and concepts of dose calculation.

At present, dose distributions are formed in a trivial way: after the beam is scattered, the central part of it is collimated to obtain a uniform lateral distribution. A method of beam delivery subsequently used depends on a specific patient treatment technique.

A variety of clinical treatment techniques is used at our three-room facility. Applications vary in complexity (and, therefore, in time of accomplishment) from a trivial single-field application to the Bragg peak irradiation of intracranial targets using converging beams of different diameters. To provide a

substantial number of patient treatments per day we have to find a compromise between opposite approaches:

- to increase the number of patients treated one has to use a set of pre-designed field applications;
- to obtain a better alignment of a field application to a specific target each patient should be treated individually.

In fact, we use both of these concepts and their combination in our practice: in common cases pre-designed field applications are used, while patients with exotic targets are treated individually. In intermediate cases calculations are needed to predict the result.

Let us consider two general, but characteristic, beam delivery techniques employed in irradiation of intracranial targets.

5.1. Narrow beam "through-out" irradiation of intracranial targets.

The target is irradiated convergently with a beam 5 to 15 mm in diameter. The number of converging beam directions may vary from 16 to 25 [13,14]. The variety of these directions is implemented by the rotation of a stereotaxic device. The energy of the proton beam at the centre of convergence is ~150 MeV (this value is considered in the text below). Having been diverged by the scattering targets with a total thickness of ~1 g/cm2 the beam is shaped by the primary and patient collimators. Dose delivered to the centre of convergence is determined analytically:

$$D = \sum_{i=1}^{n} D_i ,$$

where n is the number of beam directions and

$$Di = 1.6 \cdot 10^{-8} \cdot \frac{F}{S} \cdot {}^{dE}/_{dX} \text{ , rad,} \qquad (1)$$

where ${}^{dE}/_{dX}$ is the proton mass stopping power for E_p = 156 MeV (proton energy is obtained from tables [3], provided the initial proton energy E_0 is 200 MeV and the depth of the target is known); S is a 50% - isodose area of a single beam at the centre of convergence, i.e. at 75 mm depth in tissue; F is the number of particles measured with the Current Monitor just downstream of the patient collimator.

Analysis of the formula (1) shows, that the method of dose calculation contains certain assumptions and inaccuracies. The patient treatment technique was developed 20 years ago, and though subsequently it has been criticized, no correction has been made yet.

5.2. Bragg peak irradiation of intracranial targets.

Modified depth-dose distributions (spread-out Bragg peaks) of a single beam are formed by ridge filters of original design [15]. Proton range adjust is performed by a variable thickness water range degrader. The beam 15 to 30 mm in diameter is delivered to the target from several directions [16]. Diameter of the beam, beam directions, their number and relative contribution are determined proceeding from the size and shape of the target. Dose delivered to the target is calculated according to the following formulae.

$$Di = 1.6 \cdot 10^{-8} \cdot \frac{F}{S} i \cdot {}^{dE}/_{dX} \text{ , rad,} \qquad (2)$$

where Di, Fi, ${}^{dE}/_{dX}$, and S are determined just downstream of the patient collimator. In this case dose at the maximum of the field is,

$$D_{max} = \frac{\sum_{i=1}^{n} D_i}{\sum_{i=1}^{n} P_i} \cdot H_{max}, \qquad (3)$$

where n is the number of beam directions, Pi is the fluence from i direction related to the minimum fluence of all directions (Pi ≥ 1), Hmax is the relative height of dose distribution at the maximum of the dose field related to the dose at zero depth of the variable water degrader (provided that P=1). Calculations are performed for homogeneous medium.

This method of dose calculation is free from the drawbacks inherent in the "through-out" calculation method. All parameters in the formula (2) are determined for the same geometrical point; in the formula (3) Pi accounts for the relative variability of proton fluence from different directions, and Hmax accounts both for the Bragg curve relative increase (for a single beam) from the point of beam monitoring to the point of the target's location, and for the geometrical parameters of beams' composition at a given depth.

This patient treatment technique is most preferable for the beginning of the unification of our dosimetry protocols in accordance with the recommendations of the "European Code of Practice..." [17] and of AAPM Report No.16 [18].

6. Conclusion.

Today there is an understanding that cooperation between different proton clinical facilities cannot progress without a thorough analysis of results. The proper analysis, in its turn, necessitates coordinated methods of measurements.

For this purpose, we at ITEP see it necessary to join one of the international dosimetry protocols [17,18] that have been worked out specifically for proton clinical facilities and summarize the experience gained to date in the field. We will have to fulfill the recommendations of the protocols regarding the methods of beam monitoring and the accuracy of dose measurements. In particular, the methods of absolute dose measurements using the induced activity method must be increased up to ±5%, and the devices of two different types must be used to monitor the extracted beam.

Our recent progress in ionization chambers design, as well as the preliminary results of all-Russian dosimetry intercomparisons (featuring ITEP, Dubna and Gatchina facilities) provide the perspectives of achieving these goals.

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