QUALITY ASSURANCE IN TOTAL BODY IRRADIATION BY IN VIVO DOSIMETRY

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SUMMARY

After establishing accuracy requirements for in vivo dosimetry in total body irradiation, the technical and systematical limitations of the commonly used detectors are summarized. The potential of this tool for quality assurance is demonstrated by the results of in vivo measurements with two different photon energies.

INTRODUCTION

Although in vivo dosimetry plays a minor role in the whole field of radiation therapy, it has a distinctive importance in total body irradiation (TBI). After an inquiry at European radiooncologic centres [Quast,1987] a majority of them uses in vivo dosimetry for different purposes in TBI. Some groups use in vivo dosimetry for localization, dose distribution calculation and modification during fractionized TBI treatments instead of prior localisation, basic dosimetry and treatment planning. Most groups however prefer it for treatment verification or quality assurance in a broader sense.

An important reason for the use of in vivo dosimetry in TBI is the complex and irregular target volume, which renders a reliable dose calculation based on simple basic phantom measurements more difficult. Furthermore some authors [Christ et al, 1991] point out that in TBI *a high dose is delivered in a relative short time*. Therefore it is usually not possible to correct faulty procedures.

ACCURACY REQUIREMENTS

A first orientation concerning the required accuracy of in vivo dosimetry in TBI is provided by generally accepted or recommended dose variations in the irradiated target volume. After the International Commission on Radiation Units and Measurements (ICRU) [ICRU, 1993] the heterogeneity of the dose distribution within the planning target volume should be kept within +7% and -5% of prescribed dose. This is a rather rigorous demand regarding the irregular target volume in TBI. It also imposes a rather high quality requirement on in vivo measurements, for instance a precision of 1.7% corresponding to standard deviation is necessary to detect a dose deficit of 5% in the target volume on the 95% confidence level.

The American Association of Physicists in Medicine (AAPM) [AAPM, 1986] answers pragmatically the question of accuracy. Starting from the ICRU [ICRU, 1976] recommended overall accuracy in dose delivery of ±5% the AAPM suggested to relax this accuracy lewel, if the prescribed dose is well below the onset of normal tissue dose is limited locally. AAPM gives no hints which confidence level is considered [AAPM, 1986].

Recently Sánchez-Doblado et al. [Sachez-Doblado et al, 1995] publisched guidelines for TBI which are the result of a joint working party of the Project Managing Group of the „Treatment of Hematological Malignancies by Bone Marrow Transplantation from Volumteer Donors" project sponsored by the European Union and a dosimetry task group of the European Late Effects Project Group (EULEP) and the European group for Blood and Marrow Transplantation (EBMT). These guidelines recommend an accuracy as good as possible, preferably below ±5% (95%confidence level) for TBI dosimetry. The recommended dose homogeneity in the transverse section containing the dose specification point i. e. at mid abdomen at height of umbilicus as well as along patient's midline should be within ±10%.

Summarizing these considerations an overall accuracy in the range of ±3% to ±5% (I standard deviation) is necessary for in vivo dosimetry in TBI. We should keep in mind that

this accuracy is insufficent for the evaluation of dose and dose distribution according to the ICRU's recommendations on the 95% confidence level. If not stated otherwise, errors will be given as I standard deviation in the following sections.

MEASUREMENTS, PRACTICAL ASPECTS AND LIMITATIONS

The most commonly used detectors for in vivo dosimetry are ionization chambers, semiconductor diodes and thermoluminescent dosimeters (TLD's). Several authors [Quast, 1982; AAPM, 1986; Aukett, 1991; Heukelom et al, 1991; Sanchez-Doblado et al, 1995] summarized and disscused the problems with detectors for in vivo dosimetry. Ionization chambers generally have unwieldy cables and need a high voltage. Open chambers are problematic, because body temperatures influence the readings and the instantaneous effect is difficult to ases. A temperature increase of the chamber from 22°C to 37°C requires a change of the temperature correction factor by 5%. These difficulties are overcome for closed chambers, although an influence of air adsorption on the inner chamber walls theoretically persists. A careful evaluation of the effects caused by irradiation of the cable and coupling and monitoring of blank signals are strongly recommended in any case, especially when exit doses with low signal levels shall be recorded.

Semiconductor diodes have smaller cables, which facilitates their handling but these detectors show other disadvantages. They tend to be energy dependent and furthenmore the response of diodes is dependent on the total dose received by the diode. Also there is an effect of temperature and dose per pulse on the diode's reading. Nevertheless, after an evaluation of these effects and establishing related correction factors, these detectors can be used cuccesfully for in vivo dosimetry as several authors have demonstrated [Aukett, 1991; Heukelom et al, 1991; Sanchez-Doblado,1995]

TLD's have the advantage of being very small and furthermore, they don't need any cables. It is relatively easy to fix these detectors at many points on the patient for entrance and exit dose measurements. The well known supralinearity of TLD's is of minor importance for in vivo dosimetry in TBI, if LiF chips are used and the doses to be recorded are smaller than about 1.5 Gy. The energy response should be considered, especially in the case of large ^{60}Co irradiation fields. The predominant disadvantage, however, is the time required for readout and dose determination and the delay of results.

Besides these inherent and more technical limitations of the different dosimeters, some systematical restrictions of in vivo dosimetry still remain. One convenient method for establishing not directly accessible midplane doses employs entrance and exit dose measurements. But only in the case of small children and higher x ray energies these doses can be calculated by simple averaging of the measured values. If patients are larger, corrections taking account for the depth dose profile(s) must be applied.

The possible lack of scatter at the exit surface is of additional concern [AAPM, 1986; Gagnon and Horton, 1979; Lambert et al, 1983; Podgorsak et al, 1985]. It depends on radiation energy, irradiation geometry, especially the distance of the patient's exit surface to the wall or the floor of the irradiation room is important. Additionally the detectors by theirselves providing different scatter conditions influence the dose evaluation. Typically the corrections are in the range of 2% to 5%, when the exit surface is not immediately backed by a scattering medium, e.g. the couch or the floor of the irradiation room.

Entrance dose measurements are not burdenet by such systematical difficulties, because in TBI a spoiler in front of the patient provides normally dose build-up for the skin, which belongs to the target volume. Care should be taken for the distance between spoiler and skin. Enrarging this distance results in partial loss of dose build-up.

Despite all these problems in vivo dosimetry provides valuable perceptions of dose and dose distribution in TBI. The main reason is that, in contrast to normal radiation therapy, the complex target volume and the scatter conditions can only roughly be approximated by phantom measurements. Moreover only now commercially available computer planning systems with advanced algorithms seem to be able to calculate reliable dose distributions in TBI.

Table 1 summarizes some results of in vivo dosimetry in TBI at our institution. They were obtained from 1987 to 1990 for ^{60}Co and since November 1993 for 10 MV-X. The numbers represent the mean relative deviations of measured values versus calculated values in the thorax for the lateral irradiation fields. During this irradiation part the patient lies supine and his upper arm is placed laterally before the lung dose reduction. If the upper arm's cross section is too small, additional bolus material is used.

Table 1
Mean relative deviations of measured values versus calculated values in the thorax. Abbreviations: e: entrance, ex: exit, IC: ionization chamber, m: in the middle, behind the arm, l: in the lower part, behind the bolus. Errors represent I standart deviation of the collective.

Location	Dosimeter	^{60}Co	Number	10 MV-X	Number
upper arm, e	TLD	-0,2 ± 2,0	31	-0,9 ± 1,4	22
thorax, e	m TLD	-6,6 ± 7,1	28	-5,4 ± 3,9	22
	IC			-4,6 ± 3,1	24
	l TLD	-4,3 ± 4,8	16	+2,0 ± 4,5	22
thorax, ex	m TLD	-19 ± 19	30	-12,8 ± 7,9	22
	IC			-12,0 ± 5,5	24
	l TLD	-21 ± 13	16	-9,2 ± 7,5	22
upper arm, ex	TLD	-13 ± 16	31	-7,0 ± 8,4	22

The dose calculations relay on depth dose curves, profiles, and absolut dosimetry recorded under TBI conditions. The patient's anatomy is evaluated by a series of CT scans and topograms. Summing up the results of in vivo measurements, the measured entrance doses show only small, in clinical practice negligable deviations from the calculated values. On the beam exit these deviations become rather large. They are larger for ^{60}Co than for 10 MV-X, reflecting the energy dependent distributions of scattered radiation. The smaller standard deviations of the higher energy mean values are in close agreement with this finding.

CONCLUSIONS

In vivo measurements yield valuable information in TBI. They enable the medical physicist to minimize dose deviations from the prescribed values, if technical and systematical limitations are accounted for. An accurate control of the radiation dose delivery is an important prerequisite for evaluating clinical trials and last but not least for the individual patient treatments.

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