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Radiation dose in radiotherapy from prescription to delivery

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FOREWORD

Cancer incidence is increasing in developed as well as in developing countries. However, since in some advanced countries the cure rate is increasing faster than the cancer incidence rate, the cancer mortality rate is no longer increasing in such countries. The increased cure rate can be attributed to early diagnosis and improved therapy. On the other hand, until recently, in some parts of the world - particularly in developing countries - cancer control and therapy programmes have had relatively low priority. The reason is the great need to control communicable diseases. Today a rapidly increasing number of these diseases are under control. Thus, cancer may be expected to become a prominent problem and this will result in public pressure for higher priorities on cancer care. The creation of adequate treatment facilities and the training of the necessary personnel will take time, in some cases 10 to 15 years.

In some relatively advanced developing countries radiation therapy is applied in about 50% of all detected cancer cases. Approximately half of these treatments have curative intent. Surgery and radiotherapy applied individually or combined result in the cure of about 40% of all patients. The application of chemotherapy alone has curative effects only on a small percentage of the cancer patients. Moreover, palliative radiotherapy is often excellent in providing prolonged life and increased life quality for patients with incurable cancer.

It is encouraging to note that the results achieved by radiation therapy show continuous improvement. This can be traced back to a number of developments: increased knowledge regarding tumour and normal tissue response to radiation, early diagnosis with improved tumour localisation, improved dosimetry and dose planning. The introduction of modern equipment (CT-scanners, ^{60}Co-units, linear accelerators, computerised treatment planning systems, etc.) has been crucial in these developments and makes possible a more accurate target delineation, better treatment planning resulting in irradiation of the Planning Target Volume (PTV) with a highly uniform dose and, simultaneously, a reduction in dose to healthy tissues outside the PTV.

Experience shows that high quality radiotherapy can only be achieved if it is conducted by a skilled team working closely together with good communication between various categories of staff. The team must consist of radiation oncologists, radiation physicists and radiographers. It is also shown that dose prescribed and dose delivered have to agree within +/- 5% in the PTV to achieve a controlled cure rate without excessive complications to normal tissue. Due to the increasing demands for high accuracy in dose delivery, one of the goals of the seminar was to deal with all the different steps in treatment procedure from the decision of treatment strategy to the quality assurance of the treatments.

In some advanced developing countries - especially in Latin America - the trend now is to move from ^{60}Co-units to linear accelerators. Absolute doses as well as dose distributions from the latter type of therapy machines vary and can easily be altered through service actions or faulty parts. Therefore, seminars and training courses covering all aspects of quality control in radiotherapy and dosimetry are of great importance and should be held regionally or nationally on a regular basis.

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CONTENTS

I. ACCURACY REQUIREMENTS IN RADIOTHERAPY

TUMOR AND NORMAL TISSUE RESPONSES TO FRACTIONATED NON-UNIFORM DOSE DELIVERY

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Abstract

The volume dependence of the radiation response of a tumor is straight forward to quantify because it depends primarily on the eradication of all its clonogenic cells. A tumor therefore has a parallel organization as any surviving clonogen in principle can repopulate the tumor. The difficulty with the response of the tumor is instead to know the density and sensitivity distribution of the most resistant clonogenic cells. The increase in the 50% tumor control dose and the decrease in the maximum normalized slope of the dose response relation, γ, in presence of small compartments of resistant tumor cells have therefore been quantified to describe their influence on the dose response relation. Injury to normal tissue is a much more complex and gradual process. It depends on earlier effects induced long before depletion of the differentiated and clonogenic cells that in addition may have a complex structural and functional organization. The volume dependence of the dose response relation of normal tissues is therefore described here by the relative seriality, s, of the infrastructure of the organ. The model can also be generalized to describe the response of heterogeneous tissues to non uniform dose distributions. The new model is compared with clinical and experimental data on normal tissue response, and shows good agreement both with regard to the shape of dose response relation and the volume dependence of the isoeffect dose. The response of tumors and normal tissues are quantified for arbitrary dose fractionations using the linear quadratic cell survival parameters α and β. The parameters of the dose response relation are derived both for a constant dose per fraction and a constant number of dose fractions, thus in the latter case accounting also for non uniform dose delivery.

1. INTRODUCTION

The major lines of progress of radiation therapy has during the last decade been: the development of three dimensional (3D) diagnostic methods; accurate 3D treatment planning techniques; and new accelerator and isotope devices for precise 3D dose delivery. Our understanding of the development and spread of tumors has also increased through the identification of oncogenes and the development of tracer techniques. The importance of fractionated radiotherapy has also become understood, due to an improved knowledge of the time dependence of repair processes in malignant and normal tissues. Equally important for the development of modern radiotherapy is the modeling of radiation responses. This knowledge is fundamental for an accurate evaluation of treatment response, and for the determination of that optimal dose distribution which will eradicate a given tumor with minimal adverse reactions in normal tissue [1, 2, 3].

Basically there are two different ways to approach the prescription of the absorbed dose for a given tumor location. If the tumor type is well known and not too radiation resistant the dose level is given mainly by its sensitivity and size. However, for most

tumors the situation is more complex. There may be a resistant population of tumor cells. The prescribed dose level is then largely determined by the acceptable level of complications in the surrounding normal tissues. Morphologically, a tissue can be modelled as a complex of serial and parallel structures. The most trivial cases of first order are purely serial or parallel structures as shown in Fig 1 a and b respectively. In this work the simplest non trivial second order texture of parallel serial structures have been used as described by the relative seriality ($s = m/(m \cdot n) = 1/n$, *cf.* Fig. 1c & Eq.(16) below) to describe the volume dependence of normal tissues. However, most organs have an organization of such structures to even higher degrees of complexity as illustrated by the kidney in Fig. 1d.

For tumors the volume dependence of the radiation response is fairly simple to handle as it basically depends on the eradication of all clonogenic tumor cells. From a structural point of view the tumor is a parallel tissue because all clonogens have to be eliminated to control the tumor. Instead the difficulty with the tumor is to know the

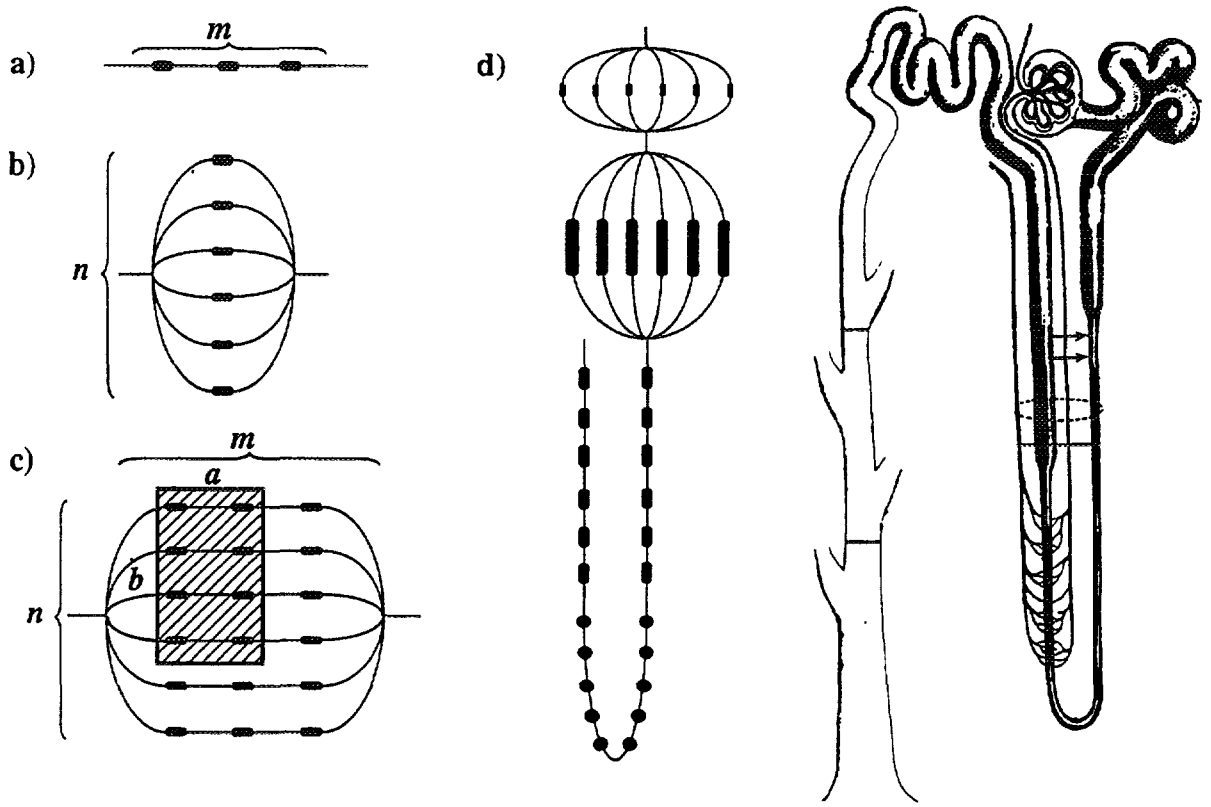


Figure 1 Schematic examples of tissue organization structures in the parallel-serial model. A serial string of subunits as described by equation (9) is shown in a), the parallel structure, equation (10), in b) and finally the serial-parallel structure in c) is a more realistic organ model. The properties of the organ are now controlled by the parameters n and m, a and b as indicated in the figure, where $0 \le a \le 1$, $0 \le b \le 1$ are the relative portions of an organ that is irradiated (equation 11). An example of the parallel-serial model applied to a functional subunit of kidney, a nephron, is shown in d). The first parallel structure is the capillary system inside the glomerular capsule, followed by the capsule itself and the limbs and Henle's loop. These are the functional subunits of a kidney as described by equations (9)-(11).

density and radiation sensitivity of the most resistant clonogenic cells. Today it is well known that they may determine the result of therapy. It has recently been shown that even if they only make up as little as 10^{-4} - 10^{-5} of the clonogens their distribution will determine the shape of the optimal dose distribution and the required dose level for tumor eradication[1, 2].

Normal tissues, on the other hand, cannot be well described by a purely parallel structure, and injury is induced before the depletion of all stem cells. Models for normal tissue reactions should therefore consider various degrees of injury to substructures of higher complexity. The developed model is generalized to fractionated non uniform dose delivery and finally compared to clinical and experimental dose response data.

2. THEORY

2.1. Classical dose response relationships

The dose response relation for tumors and normal tissues has been described by a variety of mathematical functions. To allow a precise comparison (Fig. 2) we will write the three most common of them in a standard format using only the 50% response dose, D_{50}, and the maximum value of the normalized dose response

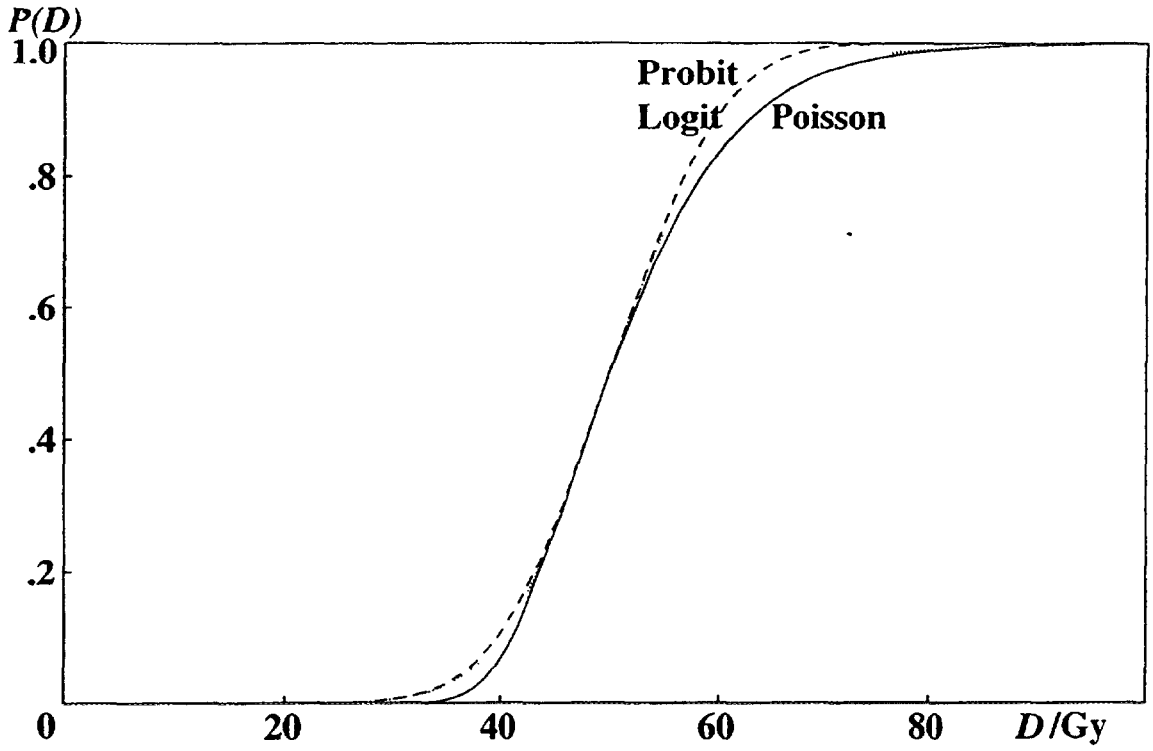


Figure 2 The three different dose-response curves based on equations (1)-(3). The solid curve is the Logit, the dashed curve is the Probit and the dotted curve is the Poisson expression. In this case D_{50} = 50 Gy and γ = 2·5.

gradient, γ, as descriptors. The Probit, Logit and Poisson models respectively then take the forms:

$$P(D) = \frac{1}{2}\left[1 - \mathrm{Erf}\left[\sqrt{\pi}\gamma\,(1 - \frac{D}{D_{50}})\right]\right] \quad (1)$$

$$P(D) = \left[1 + (\frac{D_{50}}{D})^{4\gamma}\right]^{-1} \quad (2)$$

$$P(D) = 2^{-e^{[e\gamma\,(1 - D/D_{50})]}} \quad (3)$$

Only Eq.(3) has a strict radiobiological background since it is based on the Poisson statistical model of cell kill. The Probit and Logit models may be used to approximate the shape of the radiobiologically more relevant Poisson model. The Probit model is simple to use to estimate the influence of dosimetric and biological uncertainties [4, 5, 6, 7]. The Probit and Logit models are mathematically easier to use when analyzing a large clinical material, but less desirable as these models have no biological base and merely approximates the sigmoidal curve shape. For example is the maximum value of the normalized dose response gradient, $\gamma = D\ dP/dD$, just above D_{50} for the Probit model, precisely at D_{50} for the Logit model, but just above the 37% probability level for the Poisson model, see Fig. 2. Furthermore, the curve shape at low and high effect probabilities, which are important for the normal tissues and tumors, respectively, may deviate considerably.

2.2. Volume dependence of dose response models

The effect of a homogeneous dose delivery to an entire tumor or organ at risk may be approximated by Eqs.(1-3). In radiation therapy, only a fraction of each organ will normally be irradiated at a high dose, a situation not directly covered by these equations. To calculate the injury caused by partial irradiation, the organ could be organized in a structure of sensitive subunits. For tumors the classical model is to assume a uniform and parallel structure so the probability to control a fraction of the whole tumor volume, $v = V/V_{ref}$, of known response $P(1)$ for the reference volume V_{ref} ($v = 1$) is given by:

$$P(v) = [P(1)]^{v} \quad (4)$$

For the Poisson model, Eq.(3), the dose and volume dependent expression becomes especially simple:

$$P(D,v) = 2^{-e^{[\,e\gamma(1 - D/D_{50}) + \ln v\,]}} \quad (5a)$$

This expression may be rewritten to express the volume dependence of the γ and D_{50} values:

$$P(D,v) = 2^{-e^{\left[e\gamma_{v}\,(1 - D/D_{50,v})\right]}} \quad (6)$$

where

$$\gamma_v = \gamma\,(1+\frac{\ln v}{e\gamma}) \tag{7}$$

and

$$D_{50,v} = D_{50}(1+\frac{\ln v}{e\gamma}) \tag{8}$$

The normalized dose response gradient and the dose giving 50 percent control probability increases with the logarithm of the volume or the number of tumor cells [5, 6]. The above relations, derived for uniform tumors, may be generalized to model the volume dependence also for normal tissues or structured tumors by inserting a constant k in front of the logarithmic terms in Eqs.(5a), (7) and (8). For normal tissue Eq.(5a) then becomes

$$P(D,v) = 2^{-e^{\left[e\gamma\left(1-D/D_{50}\right)+k\,\ln v\right]}} \tag{5b}$$

The constant k will be equal to unity for uniform tumors but will generally have a negative value for normal tissues. This is a mathematical way handling the decreased risk of causing injury when a smaller volume of normal tissue is irradiated as it is radiobiologically comparable to increasing the effective clonogen number N_0 and γ ($\gamma \approx \ln N_0/e$, [5]). Data on D_{50}, γ and k for some representative tissues are given in Table I by fitting to recent clinical data from Emami *et al.* [8]. By fitting Eq.(5b) to clinical tumor data [1] k values that are less than unity (0.50 - 0.55 for their data) are often obtained indicating that the density of clonogenic cells is non uniform over the tumor volume. For simplicity this model is called the k -model below.

A more relevant description of an organ is obtained by dividing it into morphological or functional subunits. Modeling an organ as comprised by subunits has been done by several workers. Morphologically, the elementary compartment can be structurally well defined or undefined. The term "functional subunit" (FSU, [9) is suited for structurally well defined tissue compartments like the nephron, but for structurally quasi homogeneous tissues like the skin, the "tissue rescuing unit" (TRU, [10]) or the "regenerative unit" (RU, [11]) is a more valid descriptor. When defining a functional subunit one should keep in mind that the target cells of an organ is not only the functional cells but the tissue regenerating cells may be even more important. The division of an organ into functional subunits is therefore quite complex if the centers of function and regeneration do not coincide. Tissue regeneration can follow one of two different pathways: either 1) through proliferation and diffusion of progenitor cells that differentiate to functionally mature cells. This is the most frequent mode of regeneration and takes place in for example epithelial and hematopoietic stem cells; or 2) through proliferation of differentiated, functional cells until tissue damage is repaired. Typical cell types belonging to the latter group are hepatocytes and endothelial cells. Wheldon *et al.* [12] called these different modes of cell renewal as H-type (hierarchical), and F-type (flexible), respectively, and discussed their response to radiation. In most tissues the regenerative or functional subunits all contain both functional and progenitor cells for both modes of regeneration. Hence, the observation that permanent tissue damage arise from injury to regenerating cells is also consistent with the presence of subunits.

The subunits are arranged structurally to give the functional properties of an organ. Figure 1a shows an organ containing m subunits in series. The response probability of

the entire organ P made up of purely serial subunits depends on the local response of its subunits P_i, described by Eq.(3), according to

$$P = 1 - \prod_{i=1}^{m} (1 - P_i) \tag{9}$$

If the organ is structured as n subunits in parallel, (Fig. 1b), the response P is instead given by

$$P = \prod_{j=1}^{n} P_j \tag{10}$$

Many organs have a of serial, parallel and/or cross-linked organization of their subunits to a varying degree of complexity (Fig. 1c and d). The simplest non trivial second order structure is a $n \cdot m$ matrix of parallel and serial subunits (Fig. 1c), giving the following response

$$P = \prod_{j=1}^{n} \left[1 - \prod_{i=1}^{m} (1 - P_{ij}) \right] \tag{11}$$

When the sensitivity of all subunits is assumed to be identical and the absorbed dose distribution is homogeneous, $P_{ij} \equiv P_{\Delta}$, Eq.(11) becomes:

$$P = \left[1 - (1 - P_{\Delta})^{m} \right]^{n} \tag{12}$$

giving

$$P_{\Delta} = 1 - (1 - P^{1/n})^{1/m} \tag{13}$$

The probability of inducing injury to a fraction $a \cdot b$ of the whole organ P_{ab} may now be obtained from

$$P_{ab} = \left[1 - (1 - P_{\Delta})^{a \cdot m} \right]^{b \cdot n} = \left[1 - (1 - P^{1/n})^{a} \right]^{b \cdot n} \tag{14}$$

for the composed serial-parallel tissue, where $0 \le a \le 1, 0 \le b \le 1$, are the relative fractions of the parallel and serial tissue subunits being irradiated.

Such a model could be used as a first approximation of a kidney even if it is known to have a more complex organization of subunits. The parenchyma of one kidney comprises 10^6 parallel venal tubules, or nephrons. Each nephron consists of a series of functional sub-units, all with a number of regenerative units arranged in parallel. The endothelial cells of the capillary system of the capsule are shown as the first parallel structure in Fig. 1d, followed by the epithelial cells of the glomerular capsule, descending limb, Henle's loop and ascending limb. The nephron will have a relatively parallel behavior, and a large part of the 10^3 cells in a nephron will have to be sterilized in order to destroy the subunit [13]. Ultimately, each organ would be divided into its elementary functional units and the probability of eradication of an individual subunit could be calculated.

However, when numerically calculating the response, the elementary compartment will not be a functional subunit, but a volume element or voxel. Such a subunit is not

equivalent to an FSU and the response of a voxel is a purely computational entity, used to determine the integral response of the organ for a given dose distribution as described by the model, and it's response cannot be interpreted simply as a local probability of injury. For structurally undefined tissues like the skin and neural tissues this is particularly so, because the survival of a single subunit is affected also by other factors than DNA repair mechanisms and the regenerative capacity, for instance cell diffusion from neighboring volumes. A refined analysis should include all causes of variations in the number of functional cells in the irradiated volume, also second order effects as cell diffusion and the transformation of differentiated cells to an altered state of differentiation, for example from astrocytes and oligodentrocytes. A critical volume model that takes diffusion into account have been developed by Yaes and Kalend [14]. However, in the first approximation cell diffusion is taken into account in the present model as the size of the subunits are for some tissues influenced by the effective diffusion distance of for example progenitor cells.

No tissue is purely serial or parallel, but a large group of normal tissues are preferable parallel [9, 11] as the liver, the lungs and all tumors. In the serial extreme there are the spinal cord and esophagus. In the next section a generalized description of tissue organization is introduced that represent a further simplification in that it does not require a detailed knowledge of the geometrical arrangement of subunits.

2.3. Response for heterogeneous tissues and dose distributions

2.3.1. Normal tissues

The above described models for tissue response allow the calculation of the response of one sub-compartment P_D when the structure and response of the entire organ is known for uniform irradiation. For the purpose of finding a descriptor of treatment success which will allow us to select the optimal beam configuration and irradiation geometry (*cf.* sec. 2.4 and [2]), the calculation of the probability of benefit and injury has to be performed in the patient's anatomy with heterogeneous tissues and dose distributions. The calculation must also be reversible, *i.e.* when the response probability distribution P_{ij} of all the subunits is known, how can the integral response of the entire organ P be calculated?

One of the computational problems experienced then is that a completely serial tissue as described by Eq.(9) gives $P = 1$ if $P_i = 1$ for any subunit i. Similarly the parallel organ gives $P = 0$ if $P_j = 0$ for any subunit j. The strong influence of a single hot or cold spot respectively, is not clinically realistic for most normal tissues. Possible exceptions are extremely homogeneously serial tissues. This problem is decreased and almost eliminated if a structure combining serial and parallel subunits is used as described by Eqs.(15-16) below.

However, an inconvenience with this, or any, structured tissue model is that the response is critically dependent on the geometrical alignment of the organ structures relative to the incident beam. Irradiating a certain percentage of one serial compartment will not result in the same probability of damage as irradiating a corresponding fraction of the parallel compartments. Physiologically, this is reasonable but it demands considerably improved knowledge about the organ fine structure and behavior to make it useful in therapy planning. Furthermore, the exact location of the various structures has to be known in relation to the therapy beam. This will be a major obstacle in ordinary therapy planning as it is performed today, even though it may be needed for a precise description of certain heterogeneous organs. It should

therefore be much more preferable if the response could be based on an organ model without a too complex internal geometrical structure, but with a correct integral and partial volume response.

A special degenerated case of the parallel-serial model described above is obtained by defining a parameter, s, which essentially describe the relative seriality of the tissue with the additional simplifying assumption that $b = 1$. The response P_v of a subunit v of a tissue can then be determined using Eq.(14) if the relative seriality of the tissue is known:

$$s = \frac{m}{n \cdot m} = \frac{1}{n}$$

$$P_v = (1 - (1 - P^s)^v)^{1/s} \tag{15}$$

where P is the response calculated from Eq.(3) and v as before is the relative volume fraction of the entire organ. This definition of a subunit makes non-zero response values possible, even for partly non-irradiated parallel organs. Inverting this equation and assuming a homogeneous dose distribution to all functional sub-units with survival P_v at dose D leads to

$$P = (1 - (1 - P_v^s)^{1/v})^{1/s} \tag{16}$$

If we here regard s as a generalized parameter describing the relative seriality, independent of irradiation geometry, then Eq.(16) fulfills the requirements stated above since it represents a structureless model with decreased influence of local extreme low or high dose values.

The next step is now to express the response of the entire organ when the dose distribution is non uniform and the cell density is homogeneous. The fractional volume of a volume element or voxel, $D_{v'}$ will be equal to one over the number of voxels, M, of the organ in the diagnostic image, and we obtain ($P = (1 - (1 - (P_{\Delta v})^s)^M)^{1/s}$ which may be generalized to a non-uniform dose distribution by:

$$P = \left[1 - \prod_{i=1}^{M} \left[1 - \left(P_{\Delta v}(D_i) \right)^s \right] \right]^{1/s} \tag{17}$$

By expressing $P_{\Delta v}(D_i)$ in terms of P for the whole organ, using Eq. (15), this may be rewritten as:

$$P = \left[1 - \prod_{i=1}^{M} \left[1 - P(D_i)^s \right]^{\Delta v} \right]^{1/s} \tag{18}$$

where the response of the entire organ P to a non uniform dose distribution now can be described as a function of the response of the whole organ for the dose D_i in each compartment i. The sensitivity of normal tissues is generally not heterogeneous within an organ, so even though the model can include heterogeneity, this is generally not required for normal tissues. For simplicity the present model, Eqs.(15-18), will be referred to as the s-model below.

2.3.2. Tumors

Modelling the response of heterogeneous tumors to non-uniform dose delivery is rather simple as all tumors *a priori* have a purely parallel structure [6]. The clinical problem is to obtain sufficient knowledge of the heterogeneities with regard to the effective density $n_e(\vec{r})$ and sensitivity $D_e(\vec{r})$, (effective D_0)of the clonogenic tumor cells. If these distributions are known, *eg.* by SPECT or PET studies and predictive assays, it is relatively straight forward to express the probability to control the tumor. The mean number of surviving clonogens is given by the volume integral:

$$N_s = \int n_e(\vec{r})\, e^{-D(\vec{r})/D_e(\vec{r})}\, d^3r \tag{19}$$

Under the assumption that Poisson statistics holds ([5, 6] the probability to control the tumor thus becomes:

$$P(D(\vec{r})) = e^{-N_s} = e^{-\int n_e(\vec{r})\, e^{-D(\vec{r})/D_e(\vec{r})}\, d^3r} \tag{20}$$

The relative steepness γ of the dose response relation may now be derived from its definition

$$\gamma(D) = D\frac{\partial P}{\partial D} = P(D) \cdot \int n_e(\vec{r}) \frac{D(\vec{r})}{D_e(\vec{r})}\, e^{-D(\vec{r})/D_e(\vec{r})}\, d^3r \tag{21}$$

where $P(D)$ is given by Eq. (20). To get a feeling for the implications of these equations we will apply them on the simplest possible heterogeneous tumor consisting only of two different cell populations, the first of normal radiation sensitivity ($D_e = D_1 \approx 2.5$ Gy) and the second radiation resistant (eg. hypoxic cells with $D_2 \approx 7.5$ Gy, at an OER ≈ 3) and a uniform delivered dose D. Eqs. (20) and (21) then reduce to

$$P_{1+2} = e^{-(N_1 e^{-D/D_1} + N_2 e^{-D/D_2})} \tag{22}$$

and

$$\gamma_{1+2}(D) = P_{1+2}\left(\frac{N_1}{D_1}\, e^{-D/D_1} + \frac{N_2}{D_2}\, e^{-D/D_2}\right) D \tag{23}$$

where the corresponding quantities for each cell population alone becomes:

$$P_1 = e^{-N_1 e^{-D/D_1}} \tag{24}$$

$$P_2 = e^{-N_2 e^{-D/D_2}} \tag{25}$$

$$\gamma_1(D) = e^{-N_1 e^{-D/D_1}}\, \frac{N_1}{D_1} e^{-D/D_1}\, D \tag{26}$$

$$\gamma_2(D) = e^{-N_2 e^{-D/D_2}}\, \frac{N_2}{D_2} e^{-D/D_2}\, D \tag{27}$$

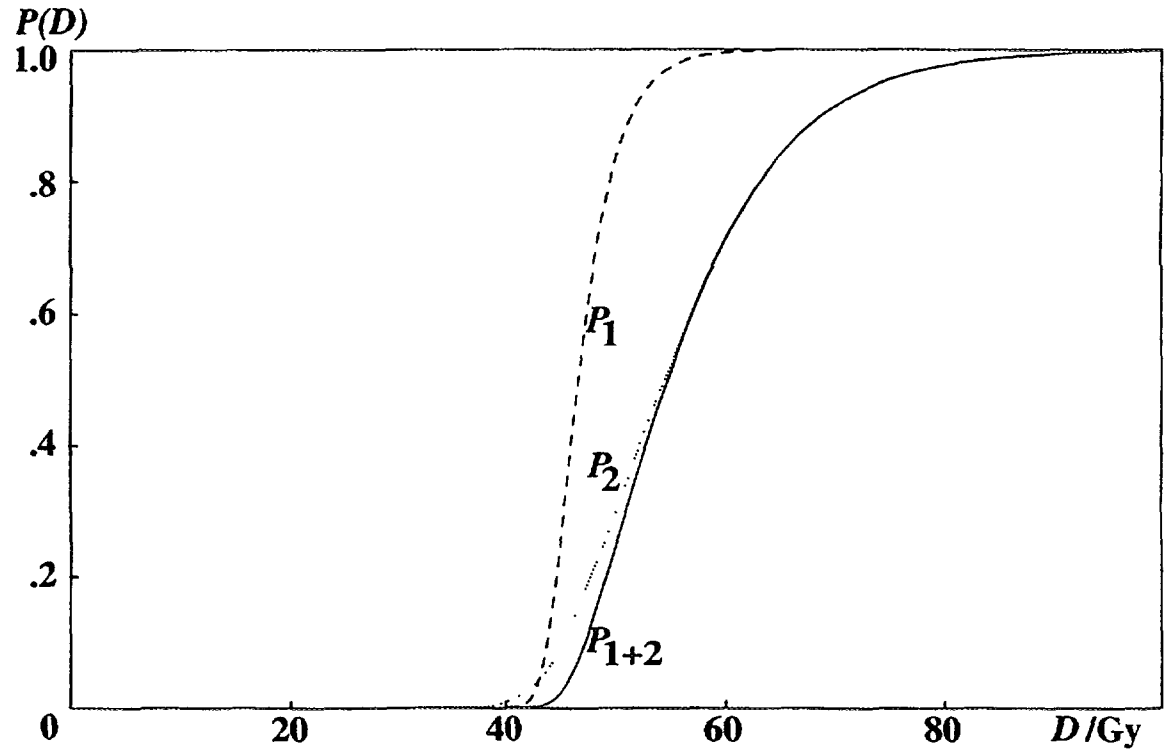


Figure 3 The dose-response relationship for a heterogeneous tumor P_{1+2} consisting of 10^8 'normal' tumor cells with D_0 = 2·5 Gy and 10^3 hypoxic cells with D_0 = 7·5 Gy (OER = 3) with a dose response according to curves P_1 and P_2 respectively. It is seen that the few resistant hypoxic cells lower the slope of the dose-response relationship to a value almost as low as that of the most resistant cell fractions.

Using these later notations Eqs.(22) and (23) may be reduced to:

$$P_{1+2} = P_1 \cdot P_2 \tag{28}$$

$$\gamma_{1+2}(D) = P_2 \gamma_1(D) + P_1 \gamma_2(D) \tag{29}$$

As mentioned above the maximum relative steepness of the dose response relations may be approximated by:

$$\gamma_1 \approx \ln N_1 / e \tag{30}$$

$$\gamma_2 \approx \ln N_2 / e \tag{31}$$

Similarly, the corresponding D_{50} values according to Eqs.(24 and 25) become:

$$D_{50,1} \approx D_1 \ln N_1 \tag{32}$$

$$D_{50,2} \approx D_2 \ln N_2 \tag{33}$$

From the last two relations it is clear that if there are even a few resistant cells in comparison to normal cells, their larger D_e value may cause a similar or even higher D_{50} value. If N_1 is equal to 10^8 and N_2 is only 10^3 then $D_{50,1} = 47$ Gy and $D_{50,2} = 55$ Gy but $\gamma_1 \approx 6.8$ and $\gamma_2 \approx 2.5$. Thus the few resistant cells will dominate the tumor control as their D_{50} value is higher and according to Eq.(28) thus $P_{1+2} \approx P_2$ as illustrated by the curves in Fig. 3. And according to Eq.(29) their low γ value will strongly influence the heterogeneous tumor even if the 100.000 times larger number of normal cells have a much steeper response as P_1 is close to unity and P_2 is small over the steep portion of the dose response curve.

Obviously there are a large number possible explanations for the shallow dose response curve observed clinically for many tumors. A number of workers have suggested a more or less random spread in sensitivity around the mean value [4, 5, 15, 16] to cause the shallow clinical dose responses observed ($\gamma \approx 2$ - 5 instead of $\gamma \approx 6$ - 7 for a homogeneous tumor, [17, 18]). The above analysis indicates that a very small compartment of resistant cells may cause an even shallower dose response than a wide spread around a mean value. The resistant cells could be repair efficient cells [1, 19, 20] or hypoxic cells remaining after reoxygenation, even if before treatment the fraction of hypoxic cells could have been quite large [21, 22]. From our present knowledge of the clinical role of hypoxic cells [23] it is likely that they may play a major role in causing a shallow dose response for some tumors even if their D_{50} values are not influenced so much as seen in Fig. 3.

2.4. Complication free tumor control

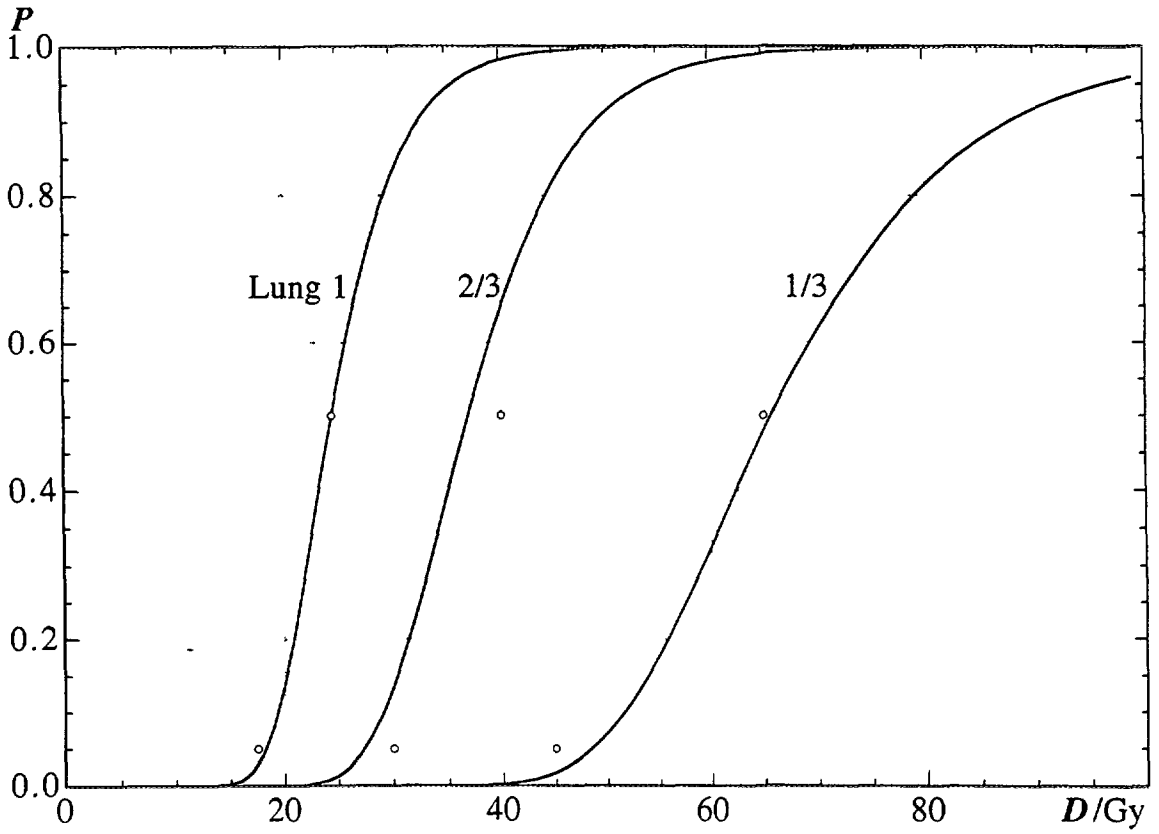
In order to judge the clinical merits of a given dose distribution it is important to be able to compare its advantages in terms of tumor control with its disadvantages in the form of normal tissue complications. In the general case this is difficult since the radiation effects in different tissues generally are incommensurable entities. However, for fatal normal tissue injuries that cannot be salvaged by surgery, a strict comparison is possible as this end-point is as undesirable as an irresectable tumor recurrence. If we define P_B as the probability of getting benefit from the treatment (*i.e.* tumor control) and P_I as the probability of causing injury to normal tissues then the general expression for the probability of achieving complication free tumor control P_+, is given by:

$$P_+ = P_B - P_B \cap P_I \tag{34}$$

In the general case a fraction δ of the patients are statistically independent [1], and P_+ may be approximated by:

$$P+ = P_B - P_I + \delta P_I(1 - P_B) \tag{35}$$

The parameter δ ($\approx$ 0,2) specifies the fraction of patients where benefit and injury are statistically independent end-points. Here P_B could be taken from Eqs. (20 or 22) and P_I from Eq.(18). If several organs at risk are present then their total probability of injury can be calculated using Eq.(9), so that total injury results if only one organ is fatally affected [24]. For less fatal normal tissue endpoints a reduction operator may be applied on their individual P_I to consider their influence on the treatment outcome.


P
1.0
0.8
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Lung 1
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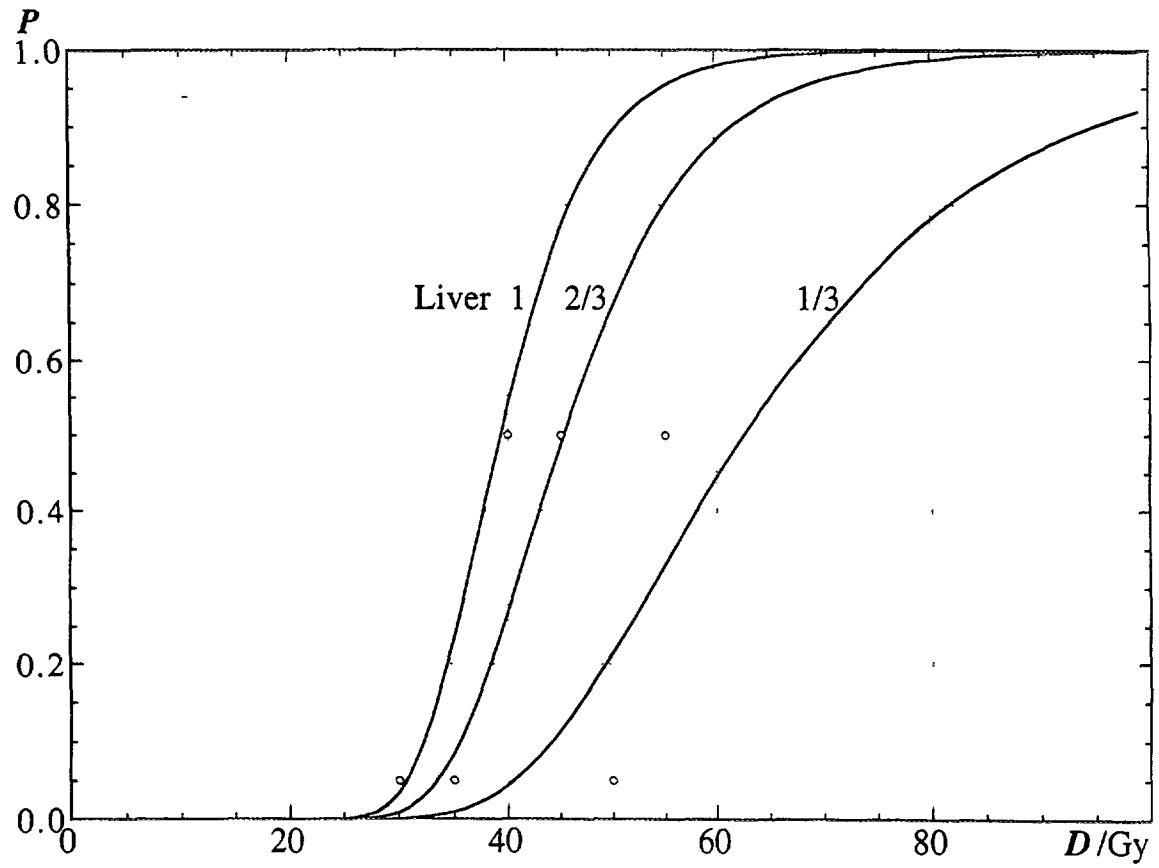
D /Gy

(a)


P
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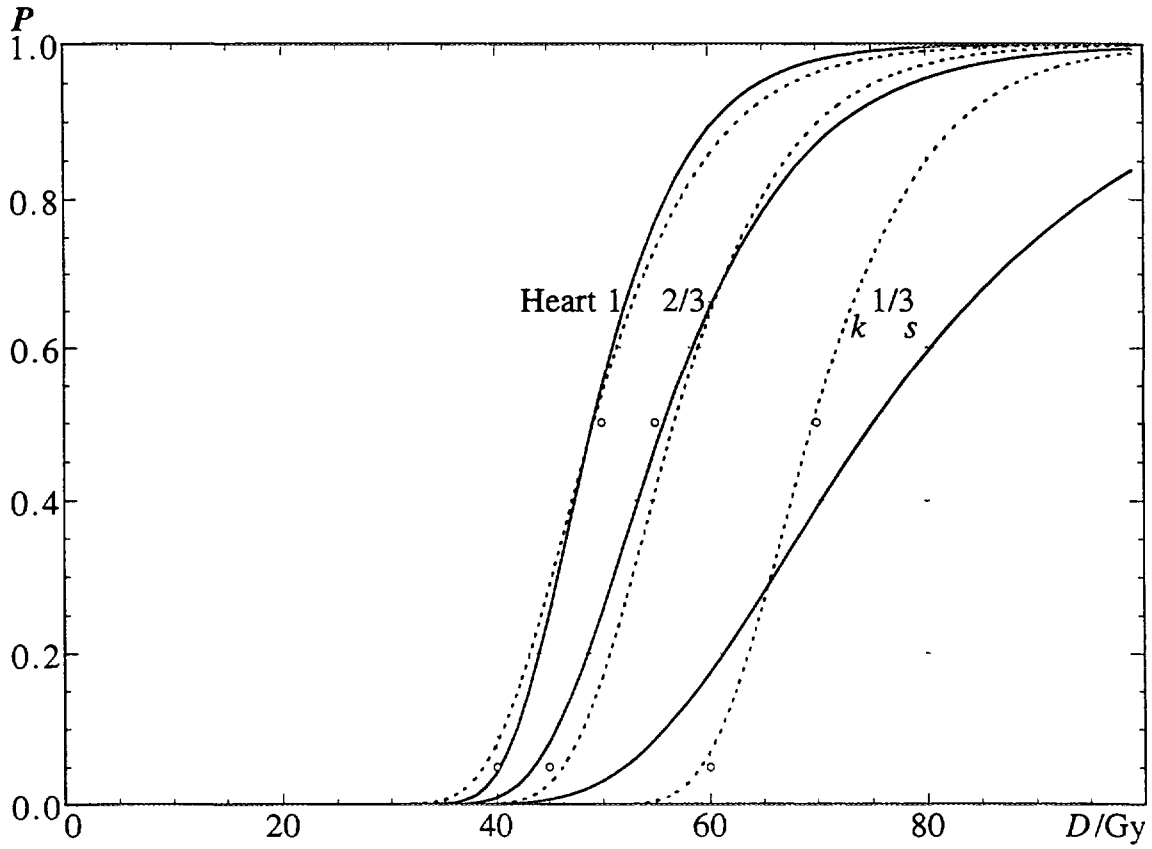
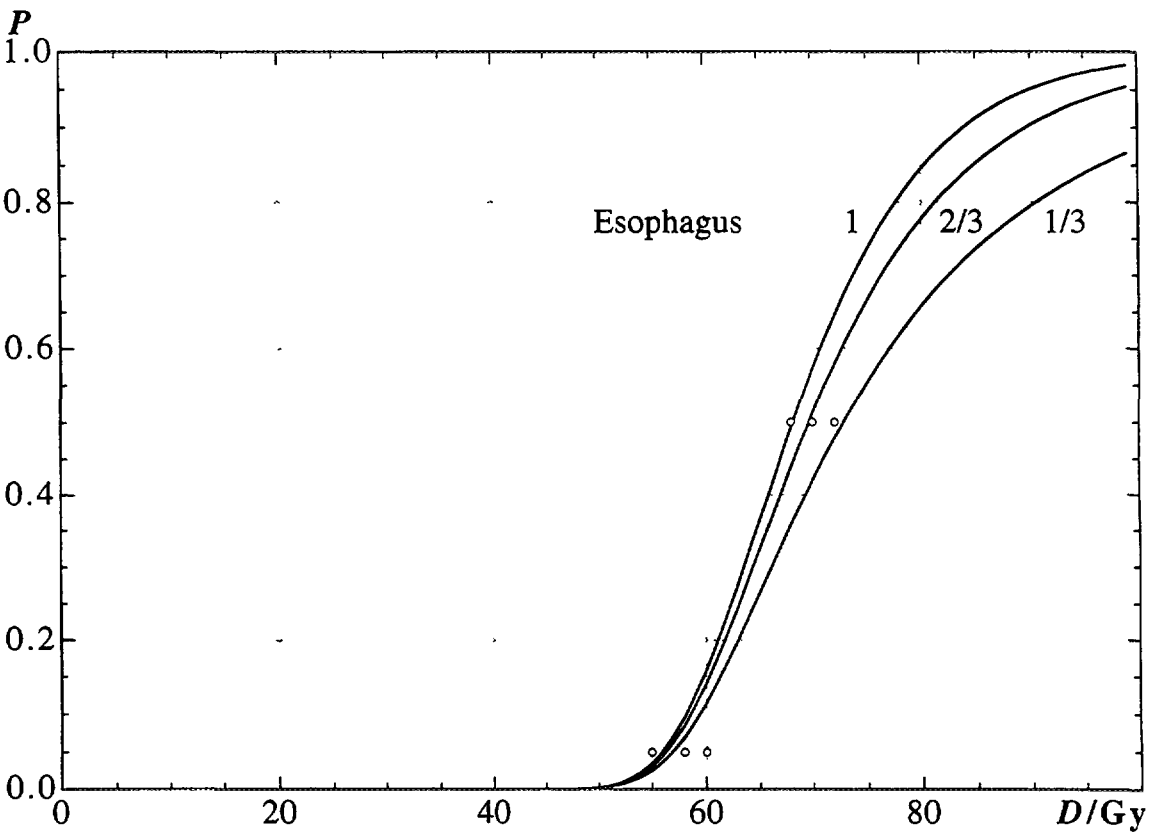
(b)



(c)



(d)

Figure 4 The volume-dependent dose-response functions compared with clinical data. Here the *s*-model has been applied to the NCI data set for: (a) lung, (b) liver, (c) heart, (d) oesophagus. The solid lines in (c) represent the *s*-model and the dashed lines are the fit with the *k*-model.

3. COMPARISON WITH CLINICAL DATA

The dose response relation using the "*s*-model" in Eq.(16) has been applied to the recent NCI clinical data set for normal tissue complications [8]. The set includes many organs in the human body, and contains dose values for 5 and 50% probability of injury when irradiating 1/3, 2/3 and 3/3 of a given organ. A comparison of D_{50}, γ and s values is made for several organs in the abdominal region. The data set is based more on clinical experience than solid empirical investigations, therefore both the probability of complication at a specified dose level and the irradiated volume are uncertain. With a suitable selection of parameters and number of unknowns it will always be possible to fit a model to the available data. The aim must be to find a model that is both simple and able to predict dose volume responses from a limited set of clinical data.

Fig. 4 shows that Eq. (16) conforms quite well to clinical data, without loss of simplicity. The curves represent a least-square fit for variables D_{50}, γ and s. In all cases a higher weight has been applied to the values for the irradiation of the entire organ, because of their expected higher accuracy. Basically, normal tissues are parallel, which is seen in Fig. 4 a-d where the fits for lung, heart, liver and esophagus are shown. The uncertainty of the clinical data [8] is sometimes quite large and the observed deviations may be acceptable. Esophagus would intuitively be expected to behave in a serial manner. The *s*-values for this organ in humans are even exceeding one, indicating that each subunit in the serial chain contains less than one regenerating unit. Extreme seriality leads to an negligible volume dependence, as seen from Fig. 4 d, where the effect of irradiating 100% of the esophagus is similar to that of the 33% irradiation. Parallel tissue has a strong volume dependence and the uncertainties in the volumes irradiated are probably quite large. The *k*-model from Eq.(5 b) has also been applied to the clinical data and the resulting *k*-values are listed in Table I. The *k*-model conforms quite well to the data for those organs where the γ value increases slowly with decreasing irradiated volume. Similar to the method of Lyman [25] and Burman [26] the shape of the dose response curve is essentially retained when the irradiated volume is increasing. An example of this is seen in Fig. 4 c for the heart. The *k*-model, however, is not based on the functionality and structure of the organ. It can only with difficulty be generalized to heterogeneous dose distributions, and is strictly applicable only to tumors. The clinical data is specified as a given volume fraction irradiated to a homogeneous dose level, when the rest of the organ receives zero dose. Clearly, this is an unphysical assumption, and in the future there is a need for a more refined analysis of treatment plans and complication probabilities with heterogeneous dose distributions. This could be achieved by using 3D treatment plans, accurate biological models and accurate diagnostic information.

4. DISCUSSION AND CONCLUSIONS

The radiobiological properties of tumors and normal tissues are gradually becoming better understood. Considering the uncertainty of available clinical data the volume dependence of the dose response models presented here are quite realistic. As expected, a strong serial behavior is shown by organs like the spinal cord and esophagus, and a marked parallelity is shown by the liver and the lungs. Despite this clear tendency, more reliable clinical information on the response of normal tissues is still needed. There is need for a more refined analysis concerning the quality of life of the patient and the remaining functionality of an injured organ when it is partially injured. Total eradication of one lung or one kidney can still lead to patient survival,

and thus a 100% probability of injury to one of two paired organ can be allowed if it decreases the dose burden to other critical structures, and increases the tumor control probability. Ultimately, the difficulty will of course be to quantify the impact on the quality of life and this evaluation should be an integral part of the normal follow-up of the patient.

When the integral functionality of the organs after irradiation and the associated influence on the tumor control probability have been calculated, they could be used to find a scalar measure that quantifies how useful a given treatment plan is. One such measure is the complication free tumor control P_+, as given by Eqs.(34, 25). The aim of treatment planning will then be to maximize that measure of treatment success [2, 3]. The evaluation of different treatment plans by the use of often quite complex dose volume histograms is a complex and often subjective method. With accurate radiobiological models the selection of the best treatment plan is simply reduced to the comparison of scalar probabilities such as P_+ values. The outcome of such a treatment optimization will depend on the accuracy of the response parameters of tumors and normal tissues and on the delineation of the target volume. Data on dose response relationships have been gathered by several workers and the implementation of more refined radiobiological models in treatment planning will also facilitate and promote the collection of such data. The information available today is accurate enough for many diseases to be clinically useful in treatment optimization. The development of improved radiobiological dose response models and tumor imaging techniques will be of paramount importance for the future improvement of radiation therapy.

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CONVERTING DOSE DISTRIBUTIONS INTO TUMOUR CONTROL PROBABILITY

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Abstract

The endpoints in radiotherapy that are truly of relevance are not dose distributions but the *probability of local control*, sometimes known as the *Tumour Control Probability* (TCP) and the *Probability of Normal Tissue Complications* (NTCP). A model for the estimation of TCP based on simple radiobiological considerations is described. It is shown that incorporation of inter-patient heterogeneity into the radiosensitivity parameter α through s_α can result in a clinically realistic slope for the dose-response curve. The model is applied to inhomogeneous target dose distributions in order to demonstrate the relationship between dose uniformity and s_α. The consequences of varying clonogenic density are also explored. Finally the model is applied to the target-volume DVHs for patients in a clinical trial of conformal pelvic radiotherapy; the effect of dose inhomogeneities on distributions of TCP are shown as well as the potential benefits of customizing the target dose according to normal-tissue DVHs.

1. INTRODUCTION

Physicists working in Radiotherapy spend a lot of their time measuring doses in phantoms and then calculating the dose distributions in patients due to a particular arrangement of beams. This is because the radiotherapist prescribes the treatment in terms of a (uniform) dose to the *target volume* accompanied by some sort of constraint on the dose to one or more *organs at risk*. However, the endpoints in radiotherapy that are truly of relevance are not dose distributions but the *probability of local control*, sometimes known as the *Tumour Control Probability* (TCP) and the *Probability of Normal Tissue Complications* (NTCP). The aim of the radiotherapist is to maximise the TCP while the NTCP remains below some "acceptable" (usually very low) level.

This lecture will deal with the biological modelling of Tumour Control Probability in terms of the *spatial* distribution of the absorbed dose within the patient but not the *temporal* distribution i.e. the difference between different fractionation scheme. It should be noted that the reference list is provided primarily as an aid to further reading rather than an attempt to acknowledge the originators of the various concepts and ideas.

Some of the reasons why a model for TCP is desirable are listed below (the references cited are not intended to be exhaustive):

- Dose distributions in 3-D are inherently very complex and some way of assimilating this vast amount of information is needed [1,2]
- As a means of quantifying treatment plan comparisons [1,2]
- As a way of estimating the effect of non-uniformities in the tumour dose distribution [3]
- Enables one to make estimates of the effect of dose and patient position uncertainties on therapy outcome [4-6]
- Optimization is beginning to be done in terms of TCP and NTCP [7,8]

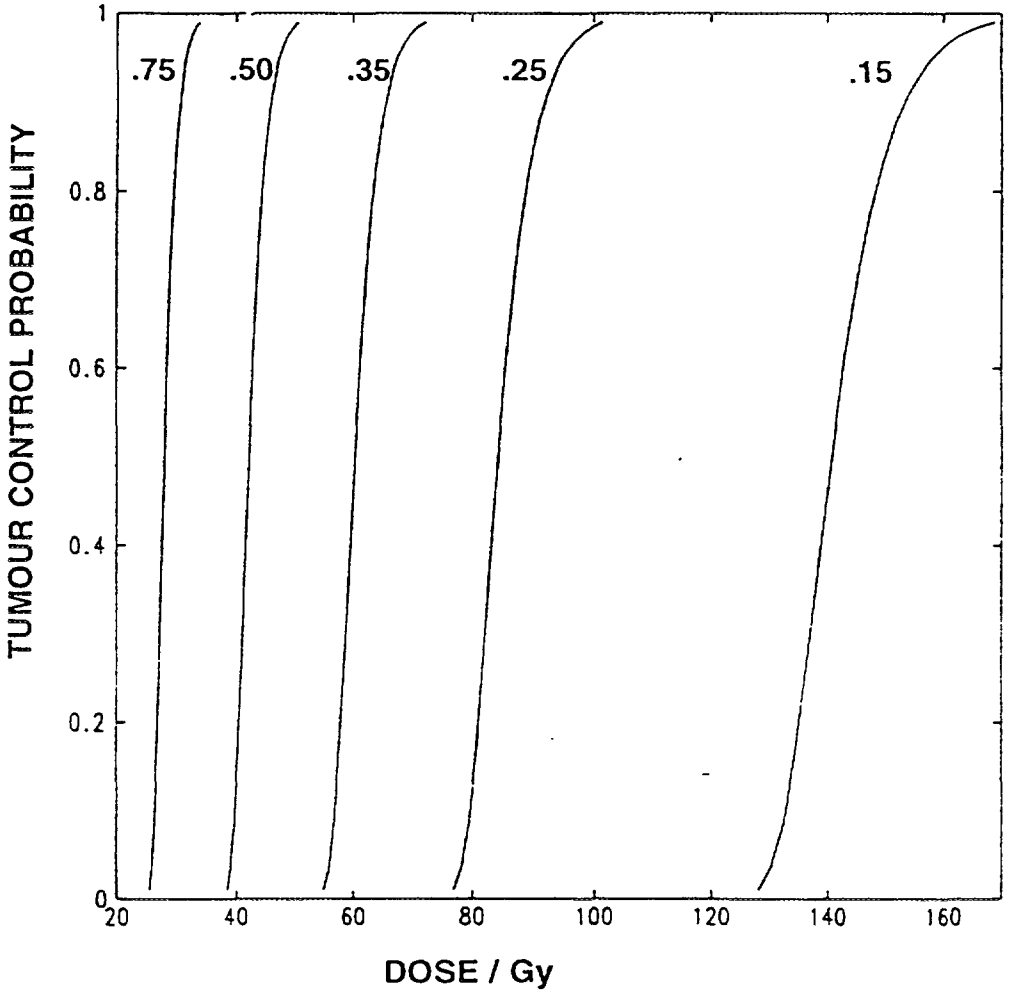


Figure 1. *TCP curves from Equ. 4 for clinically realistic values of $N_o = 10^9$ and a ranging from 0.1 to 1.0 Gy^{-1}. Note the unrealistically steep slopes.*

- As a way of using the results of biological assays for α(tumour) etc. [9-11]
- As the only (?) way (short of doing a full clinical trial) of quantifying the benefits of *improvements* in dose distributions through new technology e.g. Multileaf Collimators (MLCs), 3-D planning etc. [12]
- As an aid to clarity of thought in external-beam radiotherapy [13]

2. A SIMPLE BIOLOGICALLY BASED TCP MODEL

2.1. General

It is well-known that the so-called Dose-Response curve has a sigmoid shape e.g. [3]. Several authors have fitted mathematical functions to this curve. However, it is not easy to see how changes in basic parameters such as tumour cell radiosensitivity, inhomogeneities in the dose distribution, variation in tumour volume and in clonogenic cell density etc. can be accommodated by empirical curve-fitting approaches. In the case of Tumour Control, in contrast to that for Complications in Normal Tissues e.g. [14], it is possible to develop a model starting from the response of cells to

radiation. Nimierko and Goitein [15] have recently described such a model, which is basically identical to the one developed here [12,16]. It is this latter model which is described below and used in the rest of the analysis.

2.1.1. Basic Cell-Survival Curves

Numerous radiobiological experiments have demonstrated beyond doubt that the killing of cells by radiation can be described by an expression of the form

$$S = \exp\left(-\alpha D - \beta D^2\right) \tag{1}$$

where S is the surviving fraction after a (uniform) dose D of radiation to a population of cells. The parameters a and b characterise the initial slope and degree of curvature, respectively, of the *survival* curve. This is known as the Linear-Quadratic or LQ model of cell killing [10,17-19].

When the irradiation is fractionated as in external-beam radiotherapy (Fig. 2: full curve), for the almost universally adopted 2-Gy (per day) fraction scheme the effective slope of the survival curve is very nearly given by the value of a alone. Thus one can write:

$$N_s \approx N_o \exp\,[-\alpha D] \tag{2}$$

where N_0 is initial number and N_s the surviving number of clonogenic cells, assumed here to be irradiated uniformly and to have uniform radiosensitivity a (Gy^{-1}). Note it would be not be difficult to reinstate the b term; Nimierko and Goitein [15] include it in their model.

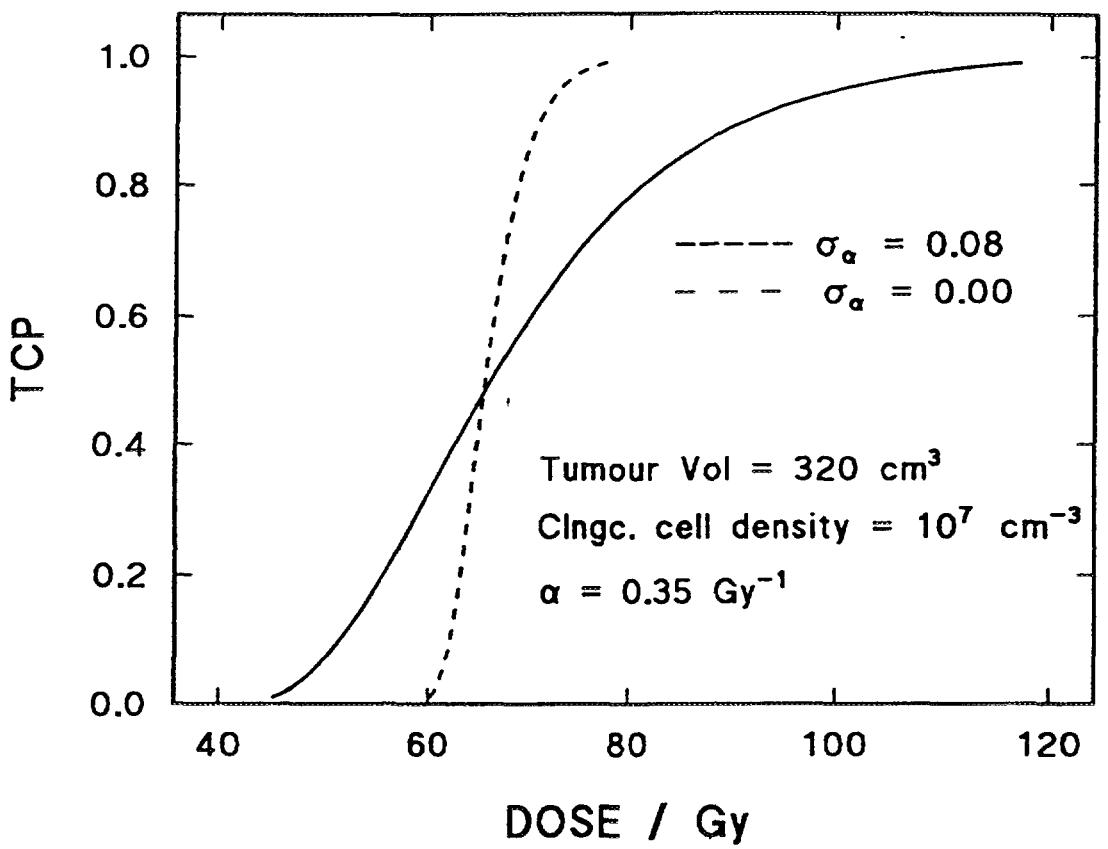


Figure 2. *Tumour Control Probability (TCP) as a function of target dose, derived from Eqs 5 and 6, with a = 0.35, $r_{cl} = 10^7$ for a 320 cc volume, for $s_a = 0.0$ and for the clinically more realistic $s_a = 0.08$ (adapted from [12]).*

2.1.2. *The Poisson Statistics Result*

The next step is to incorporate the endpoint i.e. the eradication of the tumour into the model. There is considerable radiobiological evidence for the statement that a tumour is only "dead" when every single clonogenic cell (i.e. cells with the potential for uncontrolled division) has been eliminated. Thus the quantity that we require is the *Probability that No Single Clonogenic Cell Survives*; equating this with Tumour Control Probability (TCP), we then exploit the Poisson Statistics Relation e.g. [20]:

$$TCP = \exp[-N_s] \tag{3}$$

If we now substitute Equ. 2 for N_s into Equ. 3 we arrive at

$$TCP = \exp\left[-N_o \exp(-\alpha D)\right] \tag{4}$$

A plot of this expression for TCP as a function of Dose D produce the well-known sigmoidal curve. Using a realistic value for the number of initial clonogenic cells N_o of the order of 10^9 [19] and realistic values of α from 0.1 to 1 Gy^{-1} (e.g. [10]) one obtains the family of curves shown in Figure 1.

2.1.3. *Inter-patient variability in radiosensitivity*

Theoretical models must be compared to clinical data wherever possible. There exist, in the literature, a number of *Local Control vs Tumour Dose* studies e.g. [21-25]. Despite the limitations associated with such data i.e. uncertainties in the dosimetry, inadequate patient numbers, imprecise clinical definition of Local Control etc. there are almost no Tumour Control vs Dose curves with slopes anything like as steep as the ones in Figure 1. This has led several investigators to favour an empirical model to fit these clinical Dose-Response curves e.g. [3,8].

Various hypotheses have been advanced over the years to explain the shallowness of the clinically observed dose-local control curves e.g. [26]. The explanation that is currently thought to be the most likely one is *inter-patient heterogeneity* in the intrinsic radiosensitivity of the tumour cells i.e. the α values [4,10,13]. This is in contrast to the possible heterogeneity of radiosensitivity of the clonogenic cells within any one patient's tumour. Hypoxia, long considered to be a major cause of failure to achieve local control in radiotherapy, is not now thought to play such an important role [17,19]. Thus explanations based on the effect of a small hypoxic, and therefore radioresistant, fraction of cells have fallen into disfavour e.g. [27]. A very interesting analysis of the clinical Dose-Response data has recently been published by Brenner [28]. He demonstrated that it was possible to explain the wide variations in the dose required to achieve local control for a number of different lesions solely in terms of variations in α from one lesion type to another and the variation in the number of clonogenic cells, the assumed proportional to the volume of the lesion. The *bottom line* of this study was that one did **not** need to invoke any assumptions on e.g. the variation in the hypoxic cell fraction with tumour size. Thus the Brenner analysis lends support to models for TCP based on only two parameters, intrinsic tumour-cell radiosensitivity α and clonogenic cell density, that vary with tumour type. However, Brenner did not build into his analysis any inter-patient variation in radiosensitivity and as a consequence the number of clonogenic cells N_o required to fit the clinical data came out as unrealistically small.

The TCP model described here, which I believe to be soundly based on meaningful radiobiological parameters, explicitly incorporates inter-patient variation by assuming that a is distributed normally amongst the patient population, with standard deviation s_a [12,16]. As one increases the value of s_a the slope of the dose-response curve decreases. One way of thinking about this is to regard the resulting curve as the sum of the curves for different a values in Fig. 3. There is thus a group of patients with low a values who will never be *cured* (TCP = zero), another group with high a values who will always be *cured*, and a group with intermediate a values for whom the term *stochastic fraction* has been coined as the outcome for these patients is literally a matter of chance [29].

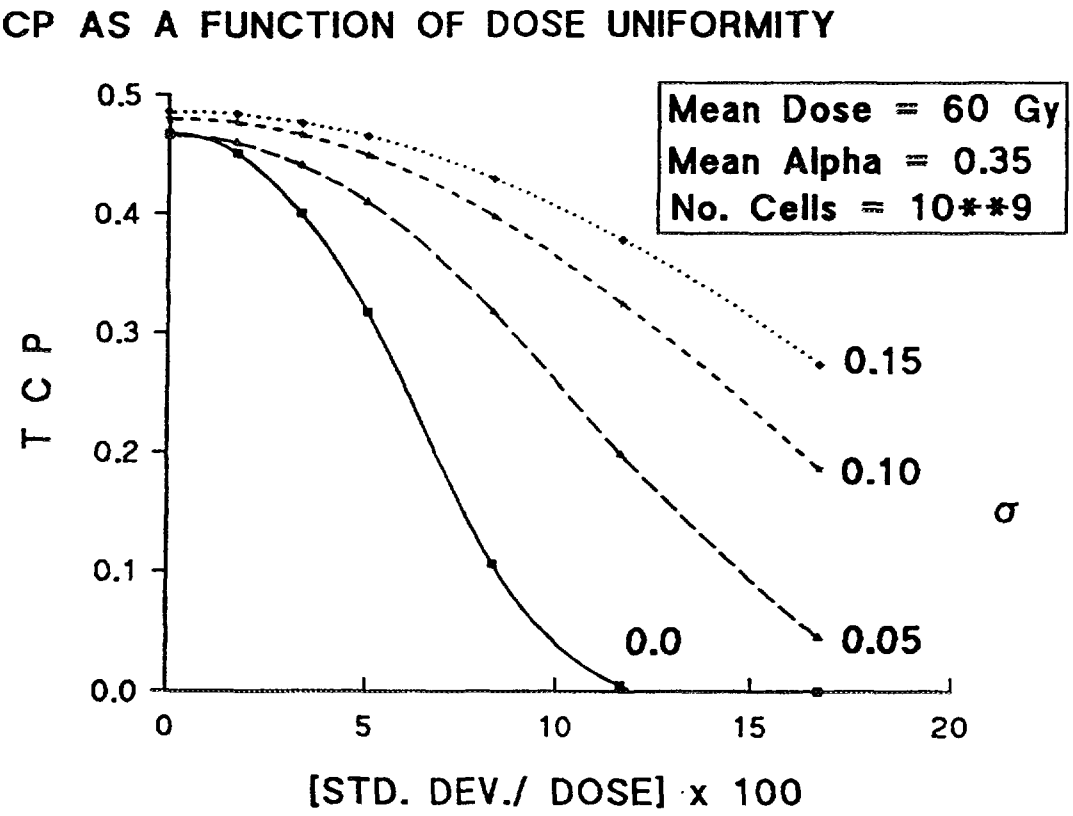


Figure 3. *The effect on TCP of non-uniformity in the target dose distribution, expressed as s_D/D, for different values of the inter-patient radiosensitivity variation parameter s_a; the mean target dose is 60 Gy, the mean a = 0.35 (corresponding to bladder tumours - [10]) and $N_o = 10^9$.*

2.1.4. Application to clinical dose-response data

The present model has been applied to the case of bladder tumours; through the irradiation of human bladder tumour cells grown *in vitro* Deacon et al [10] determined a mean a value of 0.35 Gy^{-1} which was adopted. Local control vs dose curves were then computed from

$$\overline{TCP(D)} = \sum_{i=1}^{K} g_i TCP(\alpha, D, N_o) \tag{5}$$

where $TCP(a,D,N_o)$ is given by Equ. 4 and a fraction g_i of the patients have $a = a_i$ such that

$$g_i \propto \exp\left[-(\alpha_i - \bar{\alpha})^2 / 2\sigma_\alpha^2\right] \tag{6}$$

and $Sg_i = 1$. The initial number of clonogenic cells, N_o, has been estimated from the product r_{cl} x V_{tgt} with the clonogenic cell density taken to be 10^7 [19] and the mean value of the target volume V_{tgt} = 320 cm^3; this latter value was derived from an analysis of the actual target volumes, as outlined on CT, of patients entered into the ongoing Royal Marsden clinical trial of conformal pelvic radiotherapy [30]. The two curves in Figure 2 have been calculated using the above data. In both cases the TCP at the actual clinical dose used in this hospital, 64 Gy (32 x 2-Gy fractions), come out at just below 0.5. This is consistent with clinical findings and lends some confidence to the model. The value of s_a = 0.08 was arrived at by adjusting s_a until the curve "fitted" the dose-response data in [22]; thus this is an entirely empirical value. It has been used in subsequent analyses e.g. [16].

2.1.5. *Inhomogeneous Dose Distributions*

The model developed thus far has assumed that all cells receive exactly the same dose. In radiotherapy practice this will never be case. Thus some way is needed to incorporate dose distributions into the TCP model. The data that is required is the number of clonogenic cells $N_{o,i}$ that receive a dose D_i. This is most conveniently obtained from a *Dose-Volume Histogram* or DVH (e.g. [1,31]) generated by the planning computer. Strictly what is required is the *differential* dose-volume distribution, dV/dD from which the more familiar cumulative DVH is calculated.

Thus one generalises Equ. 2 to:

$$N_s = \sum_{i=1}^{n} N_{o,i} \exp\left[-\alpha D_i\right] \tag{7}$$

where the summation is carried out over the n bins in the DVH. This expression should be also used in Equ. 5 in order to take account of the effect of both dose inhomogeneities and inter-patient a variability [16].

Strictly speaking the relevant DVH is not that for the *target* but instead that for the *tumour* volume (GTV in ICRU 50 terminology [32]) i.e. one should not include the margin added to account for patient movement. However, it is currently not possible to be more precise about such issues. A more serious limitation is probably the implicit assumption that the clonogenic cell density r_{cl} is constant right out to the edges of the tumour or target volume. This is discussed in more detail below.

Brahme [3] applied a TCP model, which differed only slightly from the one described here, to the question of the effect on TCP of both inhomogeneities in the target dose distribution and also uncertainties in the absolute absorbed dose determination. A similar exercise has been carried out here for the particular case of the bladder tumour data. The dose inhomogeneity in the target volume, consisting of 10^9 clonogenic cells of uniform radiosensitivity, was assumed to follow a *normal* distribution i.e. $N_{o,i}$ was varied *normally* as a function of D_i, with a variable s_D/D (in Equ. 7). The results of this exercise are shown in Figure 3. The calculation was carried out separately for different values of s_a. The mean dose was set to 60 Gy, which is close to what is employed clinically (see above).

For a group of patients with tumours of exactly the same radiosensitivity i.e. $s_a = 0.0$ even small inhomogeneities in dose have a disastrous effect on the TCP; this corresponds to the very steep dose-response curves in Figs. 1 and 2. More *realistic* values of s_a e.g. 0.10 result in a much less dramatic reduction in TCP as the dose inhomogeneity is increased. The message of this study is that the appreciable inter-patient variability in radiosensitivity indicated clinically for many types of tumours considerably reduces the consequences of even moderate deviations from target dose uniformity. The corollary of this is the conclusion reached by Brahme [3] that for certain classes of tumours with steep dose-response slopes, notably in the larynx (the normalised dose gradient $g > 4$) only very small uncertainties in the absolute dose determination can be tolerated.

2.1.6. *Variation in clonogenic cell density*

If the model as described thus far is applied to the DVH of the target volume (the PTV in ICRU50 terminology [32]) then implicitly the assumption is made that the clonogenic cell density is constant over the whole of the PTV i.e. one calculates the number of clonogenic cells at dose D_i, $N_{o,i}$, in Equ. 7 from the product of $V_{o,i}$ and r_{cl}. However, the PTV actually involves a margin for microscopic spread plus a second margin for geometrically inaccuracies. Thus clearly the assumption of constant r_{cl} is quite unrealistic. Figure 4 illustrates this point.

Whilst there is presently no clinical data on exactly how the cell density does vary throughout a radiotherapy target volume one can use the model to assess the effect that such variations might have on the predicted TCP. One way of looking at this is to calculate the change in dose D that corresponds to a change in cell density r_{cl} when one requires that the TCP remains unchanged for a given volume element of cells. Taking the clonogenic cell density at the tumour centre, say, to be $r(0)$ and the

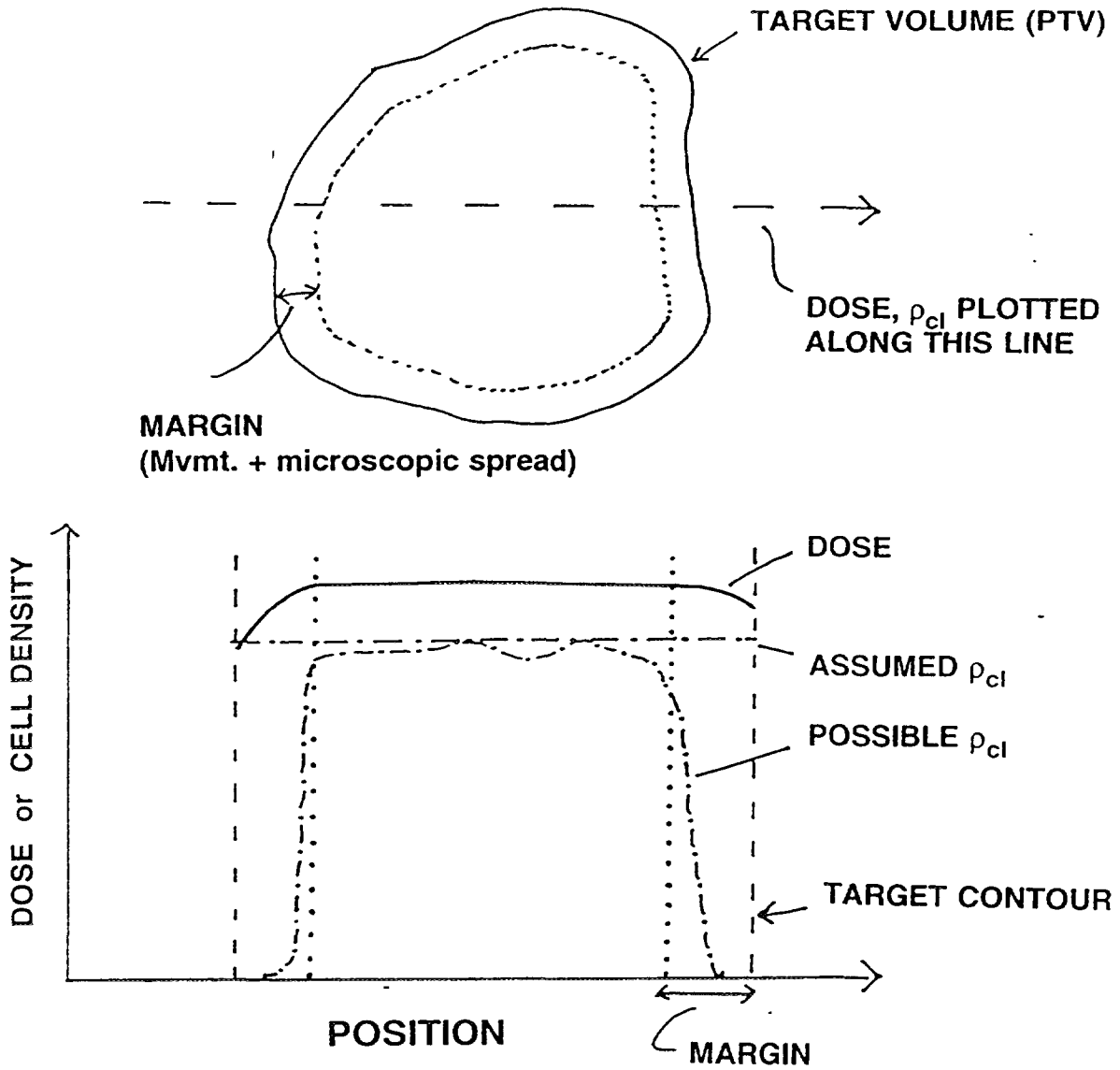


Figure 4. *Schematic drawing illustrating the problem of the variation of clonogenic cell density at the edge of the PTV; a hypothetical dose and cell density profile through the centre of the PTV is shown.*

corresponding quantity at some position r to be $r(r)$, then it is straightforward to show that the change in dose at r to yield the same TCP for the same size of volume element is given by

$$-\Delta D = \frac{1}{\alpha} \log_e \left[\frac{\rho(0)}{\rho(r)} \right] \tag{8}$$

where we see that the product aDD is proportional to the logarithm of the ratio of cell densities. Figure 5 gives the dose change for three different values of radiosensitivity a.

Webb and Nahum [16] have attempted to address this problem by extending the present TCP model to account for variations in r_{cl} as a function of position in the tumour. Figure 6 is taken from their paper. This is an attempt to illustrate the practical consequences of Equ. 8 on a tumour where the decrease in clonogenic cell density follows the (entirely hypothetical) full curve in Figure 6. The main message is that for a considerable decrease in r_{cl} the *allowable* dose decrease is very modest. Similar conclusions were drawn in [33].

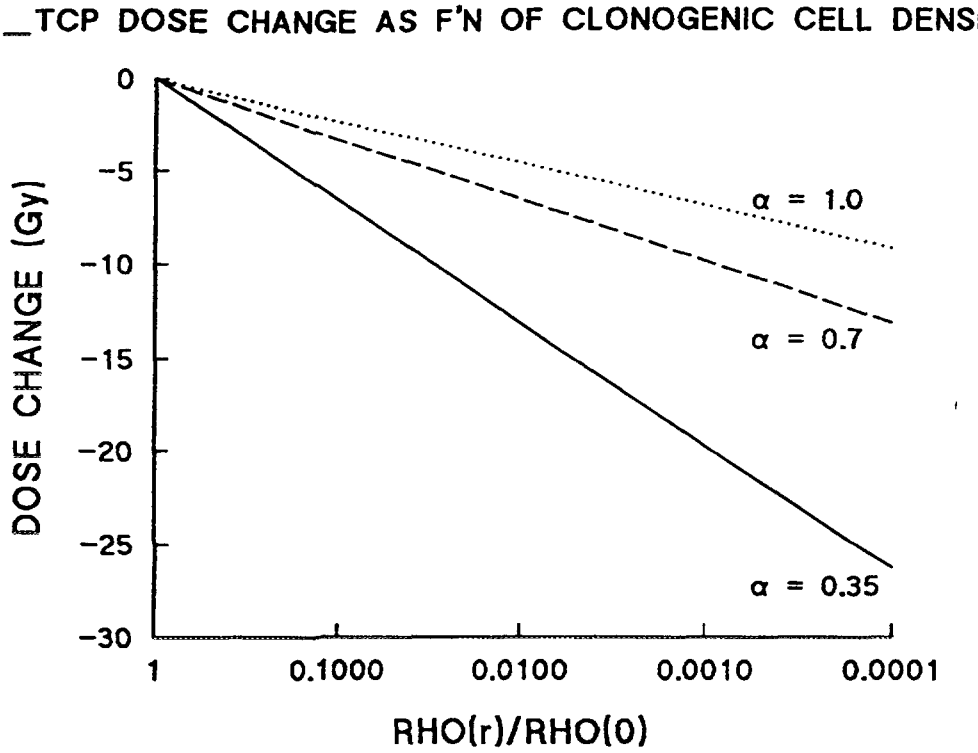


Figure 5. *The change in dose as a function of the ratio of clonogenic cell densities for a constant TCP in equal volume elements, according to Equ.8.*

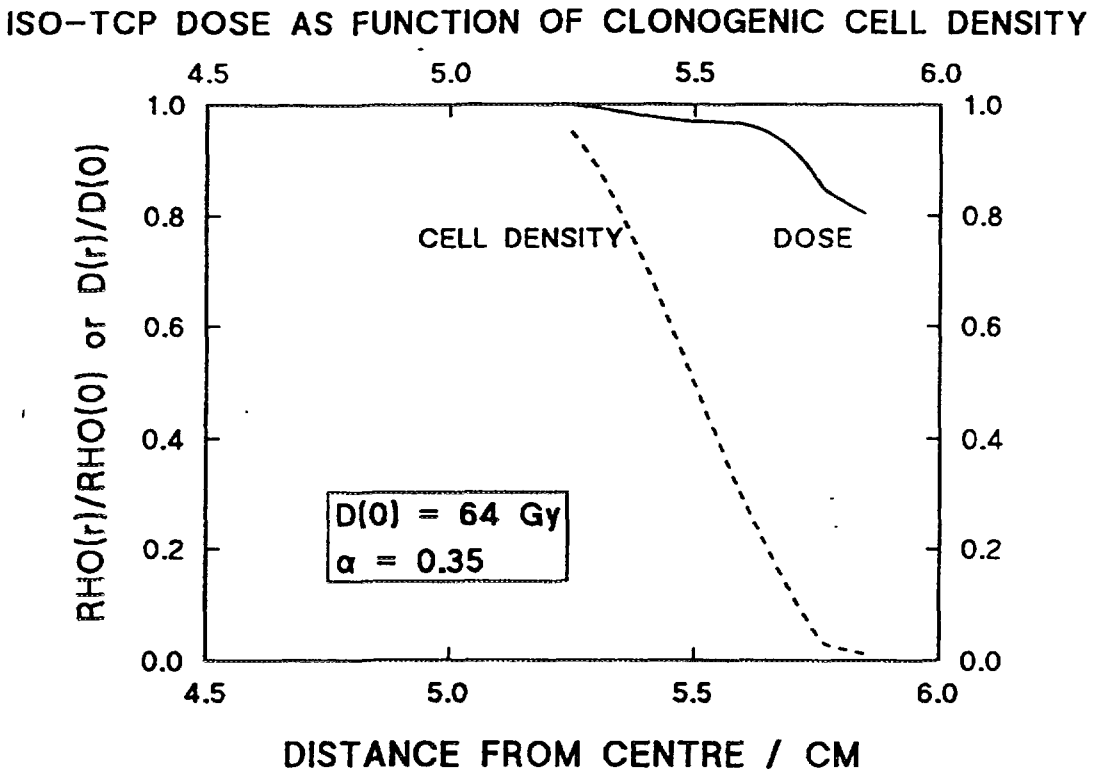


Figure 6. *The variation in the dose corresponding to the iso-TCP condition for a tumour with a variation in clonogenic cell density given by the full curve (reproduced with permission from [16]).*

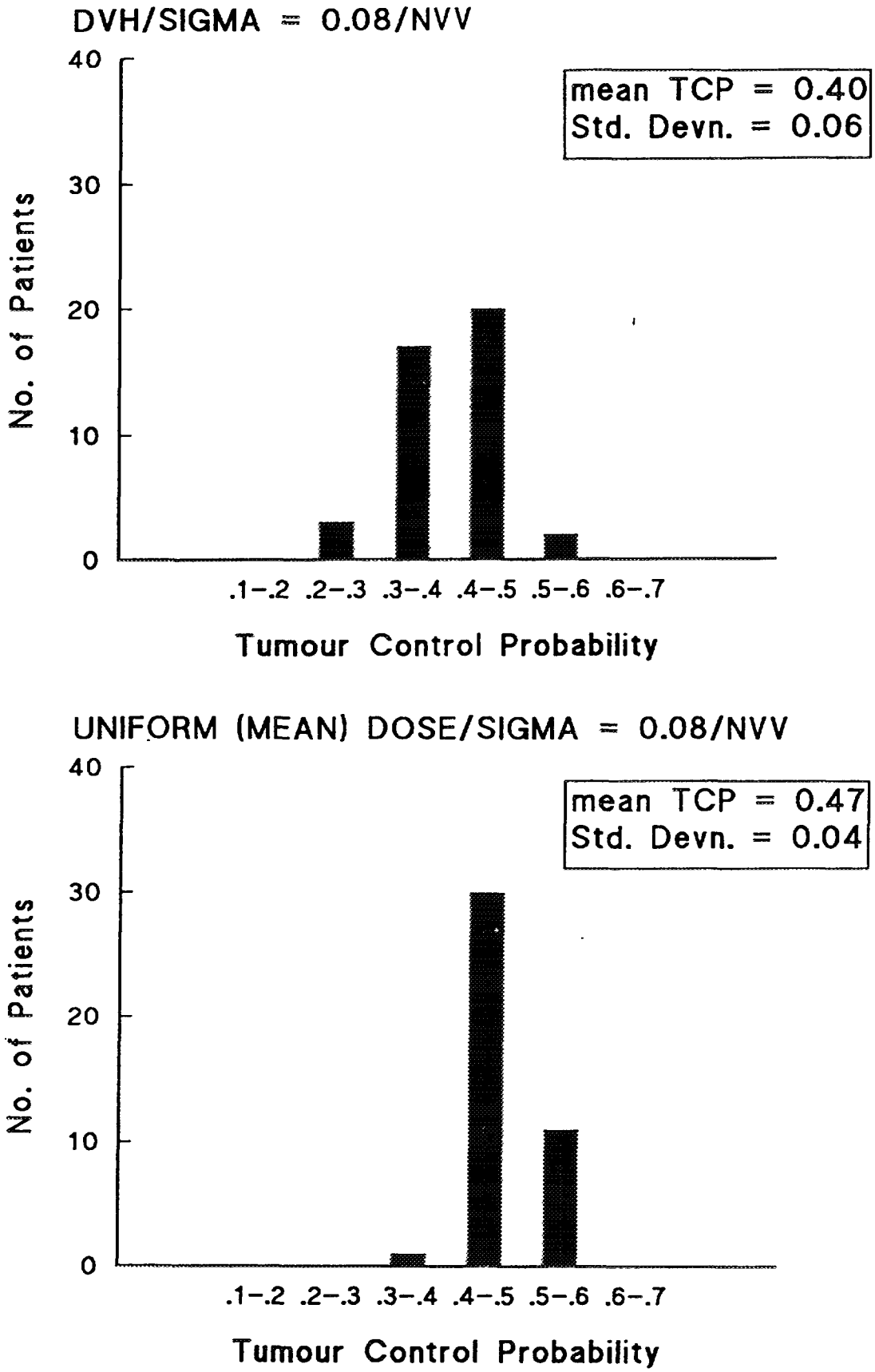
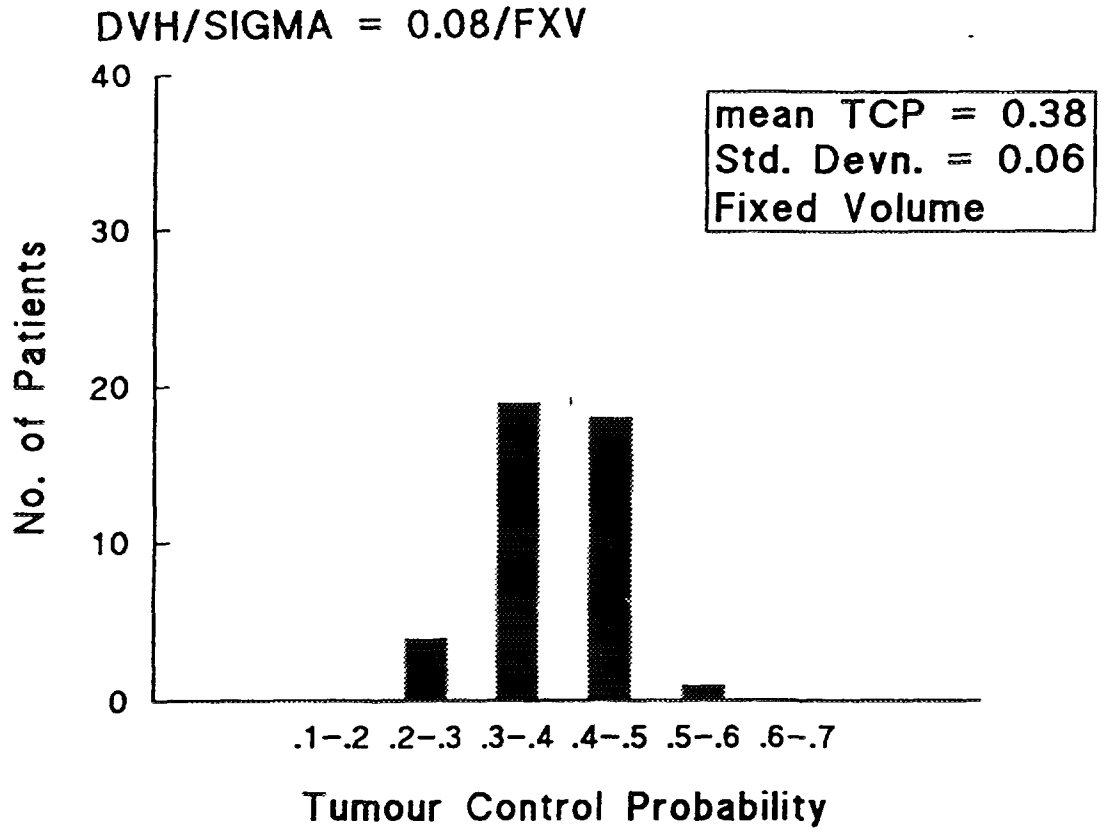


Figure 7 a,b. *Distributions of TCP for a subset of treatment plans taken from the RMH pelvic trial. The ones shown here correspond to the conventional arm and plans which have no non-zero volumes in dose bins below 70% of the isocentre dose.*

a - DVH from the PTV, natural volume variation (NVV)

b - Uniform mean dose, NVV.



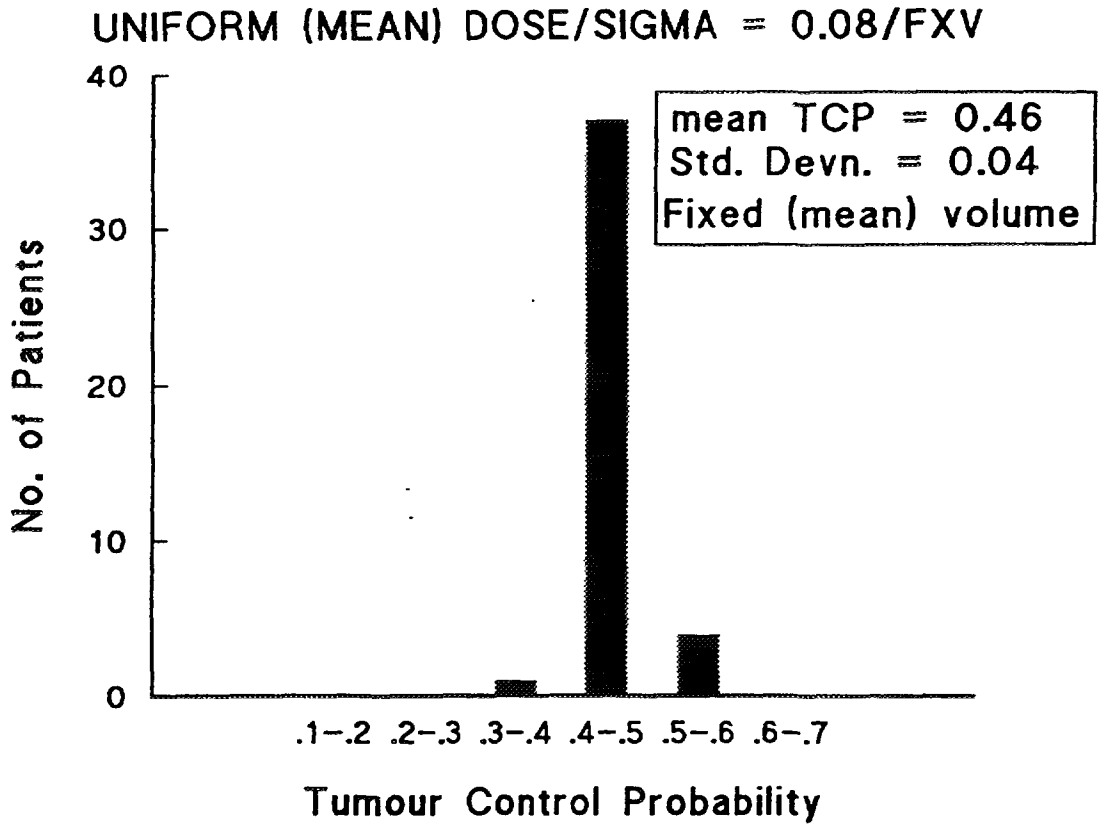


Figure 8 a,b. *Distributions of TCP for a subset of treatment plans taken from the RMH pelvic trial. The ones shown here correspond to the conventional arm and plans which have no non-zero volumes in dose bins below 70% of the isocentre dose.*
a - DVH from the PTV, fixed mean volume (FXV)
b - Uniform mean dose, FXV.

3. THE TCP MODEL APPLIED TO A LARGE BODY OF PATIENT DATA

The Royal Marsden pelvic trial [30,34] has furnished DVH data for over 200 treatment plans. The TCP model described here has been applied to the DVHs for the target volumes (PTVs). Originally TCP calculations were carried out for values of a = 0.35 but for s_a equal to 0.0 and 0.08 (see earlier section); the dose at isocentre was assumed to be 64 Gy in all cases. As expected the clinically unrealistic s_a = 0.0 calculations yielded a low average value for TCP and a very broad spread in values.

The s_a = 0.08 calculations, which correspond to a more realistic dose-response i.e. less steep dose-response relationship, also yielded a surprisingly broad spread in TCP values, with some plans having TCP values below 0.2, whereas the mean value of around 0.4. This was subsequently traced to errors in the DVH calculation by the treatment planning system. There were some plans which had non-zero values even in dose bins below 70% of the isocentre dose, effectively corresponding to severe "cold spots" i.e. underdosage in the PTV. Another aspect of this problem is the position of these "cold spots". If one assumes that these were located close to the edge of the PTV then in reality there would be unlikely to be any clonogenic cells involved. However, as discussed earlier, the model as applied here assumes a uniform density of clonogenic cells everywhere in the PTV. Clearly then, the resulting TCP values will represent some kind of most pessimistic estimate.

In an attempt to get around the above problem, an alternative set of calculations were carried out in which it was assumed that the dose was **uniform** in the PTV, with the value of the single dose given by the mean dose in the PTV. These calculations yielded a much smaller spread in the TCP. One can expect that the "truth" lies somewhere in-between these two extreme calculations.

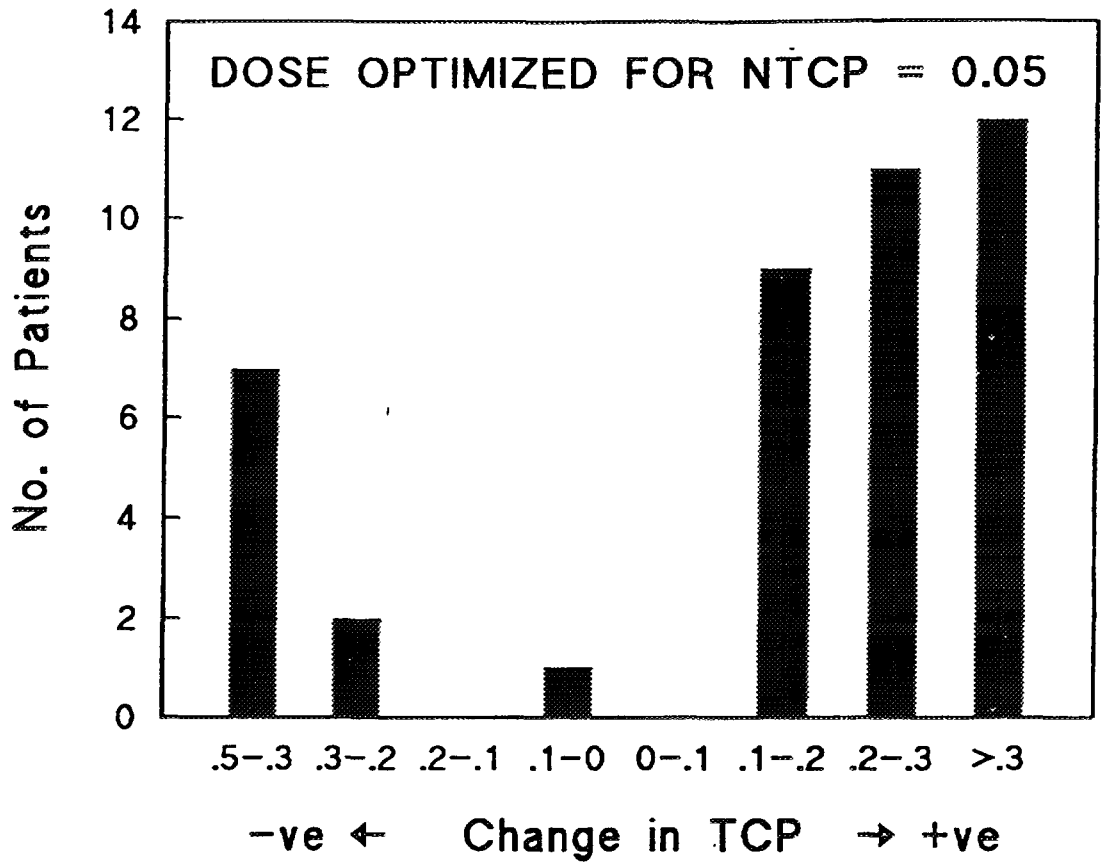


Figure 9. *A frequency distribution of the changes in TCP that would result from customizing the target dose such that there was a 5% complication probability for each of 51 patients (adapted from [12]).*

A final "tweaking" of the parameters has been done in order to investigate the effect of the range tumour volumes (actually the range in PTVs) on the spread in TCPs. For the patients in the RMH pelvic trial these range from 100 to 1680 cc with a mean value of 360 cc and a standard deviation of 277cc. Does the TCP distribution become narrower if all the PTVS are forced to be the same volume? As Figure 8 shows, in fact the tumour volume has almost no effect at all on the standard deviation of the TCP.

4. A FUTURE PROSPECT

Nahum and Tait [12] carried out a modelling study on 51 patients treated conformally in the Royal Marsden clinical trial for pelvic radiotherapy (about equal numbers of bladder and prostate tumours). Firstly it was assumed that all patients were treated with 64 Gy (in 2-Gy fractions). Rectal and small-bowel (late) complications were estimated using the Kutcher-Burman model for NTCP [14] together with the Burman et col [35] values of TD_{50} etc. A very broad spectrum of NTCP values resulted, which was entirely due to the spectrum of normal-tissue DVHs found for this group of patients. Then the prescribed dose for each patient was adjusted until the NTCP equalled 5%, thus yielding now a spectrum of doses. Figure 9 shows the changes in TCP that would have resulted compared to the standard 64-Gy prescription; the increases in TCP easily outweigh the decreases. Could customized dose prescription be part of the optimized radiotherapy of the future?

5. CONCLUDING REMARKS

The TCP model described here is mathematically very simple and yet it is a reasonably complete description of the process of tumour eradication by irradiation, apart from proliferation effects. The main limitations in its use are the lack of clinical data on radiosensitivity and clonogenic cell density. Thus the absolute values of TCP predicted must be treated with caution. Nevertheless this and similar models can be used to gain insight into the effect on TCP of certain features of dose distributions such as inhomogeneities [36], patient movement [37] and of differences in patient radiosensitivities. It is to be hoped that the biological assays for the latter currently under development [9,11] will provide data to enable treatments to be individualized both biologically as well as physically.

ACKNOWLEDGEMENTS

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DEFINITION OF TREATMENT GEOMETRY IN RADIATION THERAPY

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Abstract

When accurate systems for quality assurance and treatment optimization are employed, a precise system for fixation and dosimetric and portal verification are as important as a continued and standardized code of practice for dosimetry and patient follow-up, including registration of tumor responses and acute and late normal tissue reactions. To improve the accuracy of existing dose response relations in order to improve future therapy the treatment geometry and dose delivery concepts have to be accurately defined and uniformly employed. A Nordic working group was set up in 1991 (by Nordic Association of Clinical Physics) to standardize the concepts and quantities used during the whole radiotherapy process in the Nordic countries. Now the group is finalizing its report "Specification of Dose Delivery in Radiation Therapy". The report emphasizes that the treatment geometry shall be consistent with the geometry used during the diagnostic work up. The patient fixation is of importance early in the diagnostic phase to ensure that the same reference points and patient position will be used both during the diagnostic work up, simulation and treatment execution. Reference Coordinate System of the patient is a concept based on defined anatomic reference points. This Patient Reference System is a local system which has validity for the tissues, organs and volumes defined during radiotherapy. The reference points of the Patient Reference System should in turn be used for beam set-up. The treatment geometry is then defined by using different concepts describing tissues which are mobile in the Patient Reference System, and finally, volumes which are fixed in this coordinate system. A Set-up Margin has to be considered for movements of the volumes defined in the Reference Coordinate System of the Patient in relation to the radiation beam. The Set-up Margin is dependent on the treatment technique and it is needed in the treatment planning procedure to ensure that the prescribed dose to the Target Volume is delivered.

1. INTRODUCTION

The performance of external beam radiation therapy accelerators and other radiation therapy devices has been developed considerably during the last two decades with regard to the quality and precision of the beams through new target and filter designs, improved stability of the accelerators, increased flexibility in beam collimation systems and compensation techniques and improved dosimetric and geometric treatment verification methods. New powerful 3 dimensinal diagnostic equipment has been developed starting from computed tomography to magnetic resonance imaging and spectroscopy. Simultaneously computerized treatment planning and dose delivery optimization methods have been developed considerably. Partly due to the inaccuracies in basic definitions one of the weak links in this development has been the way we define our target volumes and specify and prescribe the dose delivery.

An investigation among the Nordic radiotherapy centres in 1991 confirmed that inconsistent use of dose and volume concepts is jeopardizing the high standard of radiation therapy [1]. A Nordic Working group was set up by NACP to standardize the concepts and quantities used throughout the whole radiation therapy process. Now the group is finalizing its report "Specification of Dose Delivery in Radiation Therapy" [2]. One of the main subjects is the definition of treatment geometry in radiation therapy. The aim has been to recommend the use of concepts based on recent scientific developement in the field of radiation therapy which are needed for the developement of daily clinical practice.

The principal aim the draft report is to treat the situation at clinics with state of the art equipment and procedures. For obvious reasons the NACP report is also written primarily with the fairly uniform equipment situation in the Nordic countries in mind. However, it is our firm belief that once general high quality procedures have been developed for advanced equipment they can also be transferred and adopted to more traditional equipment once the basic underlying principles have been developed. Many of the definitions introduced have obvious counterparts in classical radiation therapy procedures, even though they are not always coinciding with all established methods since some new proposals have had to be made for new irradiation techniques. These proposals have evolved in discussions among radiotherapists and physicists in the Nordic countries during the last five years. Of course they have also been considerably influenced by discussions with, and work of (ICRU 50 [3]) the international radiation therapy community.

2. DEFINITION OF REFERENCE POINTS AND TREATMENT GEOMETRY

In the following the methods of defining the treatment geometry suggested by the NACP-draft report are presented.

2.1. Reference points and alignment markings

2.1.1. Reference coordinate system of the patient

The concept of a reference coordinate system of the patient is based on defined anatomic reference points. The patient reference system is local system which has validity for the tissues, organs and volumes defined during radiotherapy. There does not exit a general patient reference system, since the human body is not rigid. The patient reference system is defined with one of the anatomic reference points and the other reference points (or markings on the skin) are for orientation of the system and alignment of the patient. This is illustrated in Fig 1 for a cervix cancer patient. Note that alignment markings for patient set-up can not always be firmly connected to the reference points such as the symphysis or the sternal notch. The coordinate system with external reference points should therefore preferably be used for beam set-up in order to have a "more rigid" relation to the target volume than the more uncertain skin mark often used in radiotherapy. The reference points are preferably defined already during the diagnostic stage so that they can be used for a coherent set up on all imaging modalities and on the treatment machines. The reference points will then be able to work as markers for image matching and fusion and to form an accurate integrated diagnostic data set as a base for the planning procedure. During dose planning the reference points are used for the location and definition of the isocenter relative to the reference point as defined and indicated on the dose plan.

The tissues, organs and volumes delineated for radiation therapy planning should be defined in relation to the reference point of the patient coordinate system. These reference points should in turn be used for beam set-up. The aim of using reference points is that the definition of the target volume and beam set-up refers to one and the same local coordinate system. In clinical practice the reference point should be located on the surface of a bony structure, or on the skin close to bony structures. The reference point should: 1) be possible to visualize on simulator or verification films and beams-eye-views plots, 2) be as rigid as possible in relation to the target tissues, 3) be located as close as possible to the target tissues in order to minimize beam set-up errors due to patient misalignment. Unfortunately, a rigid connection between the target tissues and the reference point is rare situation. This implies that an anatomical margin has to be added to the target tissues to account for the movements of the tumor in relation to the reference points when specifying the target volume.

2.1.2. Internal reference points

The internal reference points are located inside the body. An internal reference point is used for beam set-up on the simulator, before the first treatment. The internal reference point should thus be selected so that it can be seen on diagnostic radiographs at the simulator. This makes it possible to have a very accurate beam set-up at the simulator, but also to take simulator films to which treatment unit verification films can be compared.

2.1.3. External reference points

External reference points are palpable or visible and located on the surface of the body or on the surface of fixation devices that fit closely to the exterior of the body (e.g. facemasks and shells). The external reference point may be palpable bony structure, a skin marking or an alignment tattoo preferably where the skin is tight over a bony structure as for example on sternum. The external reference point is used for beam set-up both at the simulator and the treatment unit. The external reference points are normally palpable bony structures which are easy to find. For the extremities this will typically be at the end of the large bones. For the trunc points on the pelvic bones can be used for the lower part, and the sternum for the upper part. For the head the lower point of the mandibula and upper point of the nose can be used for the sagittal plane, and the ears for the lateral points.

2.1.4. External reference systems

The external reference points on the surface of fixation devices may be developed in the form of a local stereotactic system and define an external reference coordinate system in which the target volume is described and defined. Several stereotactic systems has been used for the head. Reproducible systems for the abdomen has also been used. It is important that the same external reference system is used at the CT, at the simulator and at the treatment unit. The coordinates of the isocenter can then be defined in the external reference system during dose planning. Similarly, the internal and external anatomic reference points define the local coordinate system of the patient in which the target volumes and organs at risk should be delineated.

2.2. Treatment geometry

The following definitions are made so general that they pertain both to curative and palliative treatments. For curative therapy the terms target tissues or target cells can be replaced by the clonogenic tumor cells as normal tissues are not generally the target for the treatment. However, the target volume often has to contain normal tissues or ensure a curative dose to all tumor cells. For a postoperative treatment no gross tumor may be left, and the definition of the target tissues consists of the remaining microscopic disease. In some regions, e.g. in the head, the target tissues may be delineated partly by osseous barries, partly by surfaces, or on clinical grounds include areas with known probability of metastases. Thus, by the target tissues is understood the tumorous tissues with a sufficiently high probability of tumor cell spread to be considered for radiation therapy.

DEFINITIONS:

Gross Tumor

The Gross Tumor consists of solid demonstrable malignant tissues in the patient (see Fig. 1) and it is often mobile in the local coordinate system of the patient.

Verified Disease

The Verified Disease includes all demonstrable macroscopic malignant tissues in the patient (see Fig. 1). The verified disease includes all the Gross Tumor and verified nodes and it is often mobile in relation to the local coordinate system of the patient.

Presumed Microscopic Disease

The presumed Microscopic Disease (see Fig. 1) contains or has a high risk of containing clonogenic malignant cells to be eradicated. It is often mobile in the local coordinate system of the patient.

Target Tissues

The Target Tissues contain all verified and/or presumed disease to be treated to a prescribed time-dose pattern (see Fig. 1). To be more precise in radical radiotherapy the target cells are the clonogenic tumor cells of the gross tumor and associated microscopic disease. The Target Tissues are often mobile in the local coordinate system of the patient. The Target Tissues, when treating non malignant disease, will include benign tissues to be treated for example with a palliative intent.

Anatomic Margin

The Anatomic Margin is a margin around the Target Tissues to account for expected movements and/or changes of shape and size of those tissues or cell structures in relation to the reference points in the patient. The anatomic margin also contains possible uncertainties in microscopic spread. The outer boundary of the Anatomic Margin specifies a fixed volume in the local coordinate system of the patient.

Target Volume

The Target Volume is an anatomically defined volume fixed in the coordinate system of the patient, which contains or has a high risk of containing tissues or cells to be treated to a prescribed time-dose pattern. This volume is defined and enclosed by the outer boundary of the Anatomic Margin. The Target Volume is therefore specified in relation to internal and external anatomic reference points (see Fig. 1) which preferably should be rigidly related to each other through bony structures.

Organs at Risk

The Organs at Risk are normal tissues whose presence influence treatment planning and/or dose prescription. Like for the target volume, the location of the organs at risk should be defined in relation to the anatomical reference points.

Treated Tissues

The Treated Tissues are tissues enclosed by an isodose surface in the cumulated dose distribution in the patient being representative for tumor eradication or palliation (e.g. 0,95 x prescribed dose).

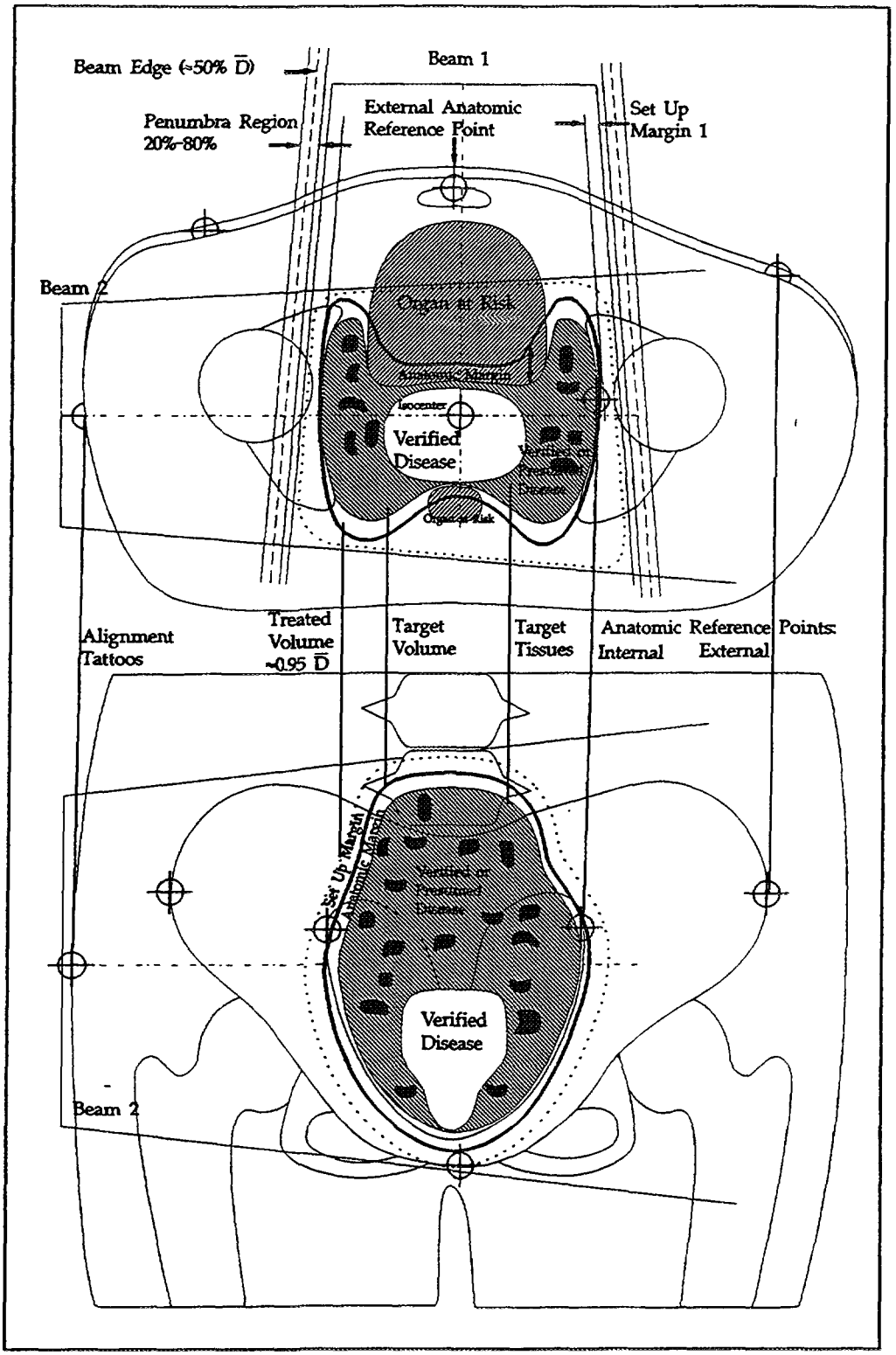

Fig 1. Illustration of the different anatomic reference points, margins, and volume concepts for an advanced cervix tumor. The internal anatomic reference points are essential for simulations and portal verification whereas the external reference points are primarily intended to improve the precision in patient and radiation beam set-up.

Irradiated Tissues

The Irradiated Tissues are tissues which receive a dose that is considered significant in relation to the biological effects of normal tissues.

Cold Region

A Cold Region is a volume inside the Target Volume, which receives a dose lower than the prescribed dose to the Target Volume. For quantification purposes it is recommended that the dose quoted should be the mean value in a volume of 0.1 cm^3 or less.

Hot Region

A Hot Region is a volume, which receives a dose larger than the prescribed dose in the Target Volume. For quantification purposes it is recommended that the dose quoted should be the mean value in a volume of 2 cm^3 or larger. A Hot Region outside the target volume is often called a Hot Spot.

Set-up Margin

The Set-up Margin is a margin for movements of the Target Volume or Organs at Risk in relation to the radiation beam. The Set-up Margin is dependent on the treatment technique and it is needed in the treatment planning procedure for example to ensure that the prescribed dose to the Target Volume is delivered and the dose to healthy normal tissues is as low as possible. This margin has to account for uncertainties in 1) patient positioning (interfractional movements), 2) movements of the patient during each treatment fraction (intrafractional movements), 3) dose planning and treatment technique in general and 4) treatment unit performance characteristics.

3. THE RECOMMENDED USE OF THE CONCEPTS

To deliver the right dose distribution to the target tissues would be no great problem provided there were 1) no uncertainly in microscopic tumor spread, 2) no positional uncertainties due to motions of internal tissues, 3) no uncertainty in the alignment of the patient with the therapy beams and finally, 4) no uncertainty in the delivered dose distributions. All these four categories of uncertainties decrease the probability of achieving complication free control of the tumor growth, especially if they are not accounted for in the planning procedure.

Obviously, the best thing would be if one could eliminate as far as possible the positional and set-up uncertainties by good fixation techniques, possibly combined with synchronization of the irradiation with breathing or other internal motions. However, the uncertainty in microscopic spread is very hard to eliminate both due to patient individual patterns of spread and due to the finite resolution of the diagnostic methods. In the first approximation one would think is does not matter much whether the uncertainty in the location of the target tissues is due to uncertainties in microscopic tumor spread, organ motions, or radiation beam set-up. However, accurate patient set up requires the use of external reference points and thus separation of internal (organ motions or microscopic spread) and external (set up) uncertainties.

When the target volumes and organs at risk have been accurately delineated relative to the reference points the dose delivery technique has to be considered. If there is no reason to expect different sensitivities for the verified gross disease and its presumed microscopic extension and the tumor cell densities are not too different, a single target volume and a uniform dose delivery may be sufficient.

However, if the verified gross disease may contain more resistant cell compartments such as hypoxic tumor cells, and/or if the density of tumor clonogens is considerably lower in the sub clinical region, different dose levels may generally be desirable. The definition of two or more distinct target volumes is then called for and the most suitable dose level for each must be specified.

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DOSIMETRIC PRECISION REQUIREMENTS AND QUANTITIES FOR CHARACTERIZING THE RESPONSE OF TUMORS AND NORMAL TISSUES

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Abstract

Based on simple radiobiological models the effect of the distribution of absorbed dose in therapy beams on the radiation response of tumor and normal tissue volumes are investigated. Under the assumption that the dose variation in the treated volume is small it is shown that the response of the tissue to radiation is determined mainly by the mean dose to the tumor or normal tissue volume in question. Quantitative expressions are also given for the increased probability of normal tissue complications and the decreased probability of tumor control as a function of increasing dose variations around the mean dose level to these tissues. When the dose variations are large the minimum tumor dose (to cm^3 size volumes) will generally be better related to tumor control and the highest dose to significant portions of normal tissue correlates best to complications. In order not to lose more than one out of 20 curable patients (95% of highest possible treatment outcome) the required accuracy in the dose distribution delivered to the target volume should be 2.5% (1σ) for a mean dose response gradient γ in the range 2 - 3. For more steeply responding tumors and normal tissues even stricter requirements may be desirable.

1. RADIOBIOLOGICAL MODEL

1.1. Statistics of tissue damage

The classical theories for the killing of a cell assume that single or multiple hits are necessary in one or several targets in this cell. The targets are often considered to be located in the cell nucleus. To achieve a fair agreement between experimental results and theory rather complex multi hit multi target combinations have to be used. Furthermore, it is also known today that the curvature of the dose response relation is mainly caused by repair processes and to a lesser extent to target structure. For this reason the mathematically more simple semi empirical linear quadratic expression has been used more extensively during recent years for comparison with experimental data.

However, when the survival of a certain organ or a tumor is considered the single hit multi target model may still be of interest as illustrated schematically in Fig. 1. The N targets are now interpreted as the functional sub units or clonogenic cells making up the organ or the tumor, instead of the various subtargets in the cell nucleus previously assumed to cause cell death when being hit. For simplicity it is also assumed that each clonogenic cell is inactivated when an ionizing particle hits the sensitive area of the cell, as quantified by the inactivation cross-section σ_0 and illustrated in Fig. 1 (see also Eq. 3 below). The incident radiation beam causing the cell inactivation is described by its fluence of ionizing particles, Φ. In the case of electron and photon beams this is the electron fluence, and it can be related to absorbed dose to the organ by multiplying the

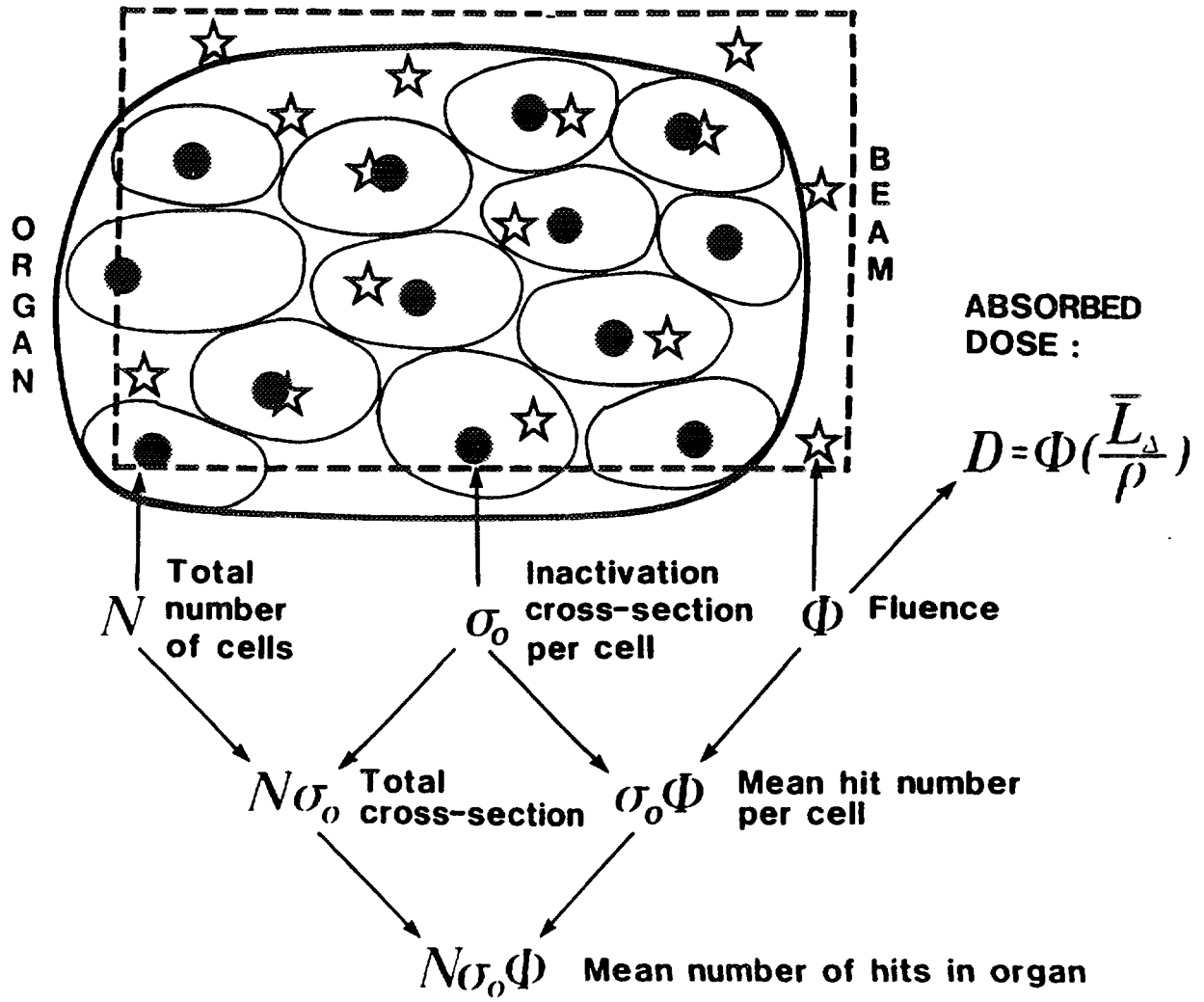

Fig. 1 The model used to calculate the response of an organ consisting of a large number of individual cells (N) each with an inactivation cross section (σ_0/cm^2 per cell), when exposed to a radiation beam which is specified by its fluence (Φ/particles cm^{-2}).

fluence with the mean restricted collision mass stopping power L_Δ/ρ for the electron spectrum. in question. At a dose level of 1 Gy the fluence is typically $3 \cdot 10^9$ electrons per cm^2, assuming a mean stopping power typical for high energy electrons and photons of about 2 MeV cm^{2}g^{-1}. Similarly, the density of ionizing events may be calculated based on a mean inactivation event size of about 60 eV in unit density material. At 1 Gy the density becomes 10^{14} events per cm^3. This corresponds to a mean distance between events of about 0.2 μm in agreement with the fluence value just calculated.

When the inactivation cross section and the initial number of cells are known the mean cell survival as a function of absorbed dose or fluence can be calculated somewhat in analogy with the nuclear reaction probability in an irradiated medium. Since the probability for a given cell to be hit by a given particle is extremely small but the beam consists of a huge number of particles, the mean hit number is finite and the problem is ideally suited for Poisson statistics [1, 2]. Thus the mean number of inactivated clonogenic cells, $d\bar{N}$, due to a fluence increase, $d\Phi$, is given by:

$$d\bar{N} = -\sigma_0 \bar{N} \, d\Phi \qquad (1)$$

This is the differential equation resulting in the traditional single hit exponential cell survival:

$$\overline{N} = -N_0 e^{-\sigma_0 \Phi} = N_0 e^{-D/D_0} \tag{2}$$

where $\overline{N}$ is the mean number of surviving cells and N_0 is the initial value. The relation between D_0 and the inactivation cross-section, σ_0, using $\overline{D} = \Phi \overline{L}_\Delta / \rho$ is given by

$$D_0 = \frac{1}{\sigma_0}\left(\frac{\overline{L}_\Delta}{\rho}\right) \tag{3}$$

A more detailed discussion on the dependence of σ_0 on radiation quality is given by Zaider and Rossi [3]. The inactivation cross-section for each cell is very small and the fluence of particles very large so that the product of them, the mean hit number per cell, $\sigma_0 \Phi$ (*cf.* Fig. 1) is finite. Therefore, Poisson statistics can be applied to estimate the probability of having precisely v hits:

$$P_h(v) = \frac{e^{-\sigma_0 \Phi}(\sigma_0 \Phi)^v}{v!} \tag{4}$$

Thus the probability for no hits, $P_h(0)$, or the probability that a given cell survives, P_S, becomes:

$$P_S = P_h(0) = e^{-\sigma_0 \Phi} = e^{-D/D_0} \tag{5}$$

The probability that a single cell is killed (i.e. one or more hits) is therefore $P_e = 1 - P_h(0)$. Provided the killing of a cell is statistically independent of what happens to every other cell, the probability that a tissue or tumor consisting of N_0 cells is completely eradicated by killing all its N_0 clonogenic cells is given by the conditional as expressed by

$$P_e = \left(1 - e^{-D/D_0}\right)^{N_0} \tag{6}$$

The survival probability for a tissue consisting of N_0 cells is thus $1 - P_e$ which is recognized as being mathematically similar to the traditional single hit multi target survival curve equation. If we again apply Poisson statistics a very useful alternative expression for the probability of eradication of a given organ can be derived from the probability of having precisely v surviving cells:

$$P_S(v) = \frac{e^{-N} N^v}{v!} \tag{7}$$

where N is the mean number of surviving cells as given for example by Eq.(2). From this expression the probability of no survivals, $P_S(0)$, or the eradication probability, P_e, becomes simply:

$$P_e = e^{-N} = e^{-N_0 e^{-D/D_0}} = \left(e^{-e^{-D/D_0}}\right)^{N_0} \tag{8}$$

This expression is very closely related to Eq.(6) particularly at high dose levels such as on over the sigmoidal part of the dose response curve. Eq.(8) always gives a slightly larger value than Eq.(6) as can be demonstrated by power expansion of Eqs.(6) and (8) for dose values both larger and smaller than D_0. However, the more basic Eq.(6) should in principle be more accurate, but the difference is never clinically significant for large values of N_0. Due to the greater mathematical simplicity, Eq. 8 has been extensively used over the years to accurately describe the shape of the dose response relation[1, 4].

1.2. Cell survival curve

For a given cell population in vitro the relationship between the surviving cell fraction, S, and the absorbed dose deposited in a single irradiation is described by the cell survival curve (cf. Fig. 2). The above equations are all derived in the approximation of the single event cell kill without consideration of repair mechanisms. The simplest way to generalize the above equations to take such processes into account in an approximate manner is to replace D_0 by D_e the effective D_0 [4, 5]. An even better fitting of experimental cell survival data over a wider range of doses is obtained if the simple exponential cell survival $P_s = e^{-D/D_0}$ in Eq.(5) is replaced by a more precise expression in equations like (6) and (8). The shape of the cell survival curve taking the effect of repair processes into account is generally very well described by an equation of the type :

$$S(D) = e^{-\left(\alpha D + \beta D^2\right)} \tag{9}$$

The coefficient α of the linear term in absorbed dose determines the slope of the survival curve at small doses. The coefficient β of the quadratic term is a measure of the shape of the shoulder of the survival curve. A collection of survival parameter data for human tissues have been published by Thames and Hendry [6]. An approximate but more simple cell survival curve model is given by an expression of the type

$$S(D) = f e^{-D/D_0} \tag{10}$$

where f is the extrapolated cell fraction at zero dose assuming the slope of the almost linear part of the cell survival curve on a logarithmic scale to be constant. D_0 is the absorbed dose that reduces the proportion of surviving clonogenic cells to e^{-1} around the mean dose of interest $\bar{D}$ (see Fig. 2). This relation has the advantage of being more accurate than Eq.(5) but it is still linear in dose and therefore valid also for fractionated non uniform dose delivery with the total dose as sole variable. The parameters D_0 and f can be related to α and β through simple relations if, for the dose range of interest, the logarithm of the cell survival around the mean dose, $\bar{D}$ can be approximated by a straight line. The shape of the survival curve as determined by the above factors will in the first approximation describe the response of a cell system or organ to fractionated irradiation.

1.3. Dose response relation

Based on the shape of the cell survival curve after a single irradiation in vitro the response of a tumor or an organ to multiple irradiations in vivo can be estimated. A detailed analysis is quite complex as it has to consider the capacity of the resting cells to enter the cell cycle, the growth delay of the different phases of the cell cycle, the efficiency of repair processes and influence of oxygenation and nutritional factors, the condition of the vascular system and the radiation modality used (e.g. particle type, energy and dose rate). For the present purposes only the gross radiation effects are included in order not to complicate the mathematics.

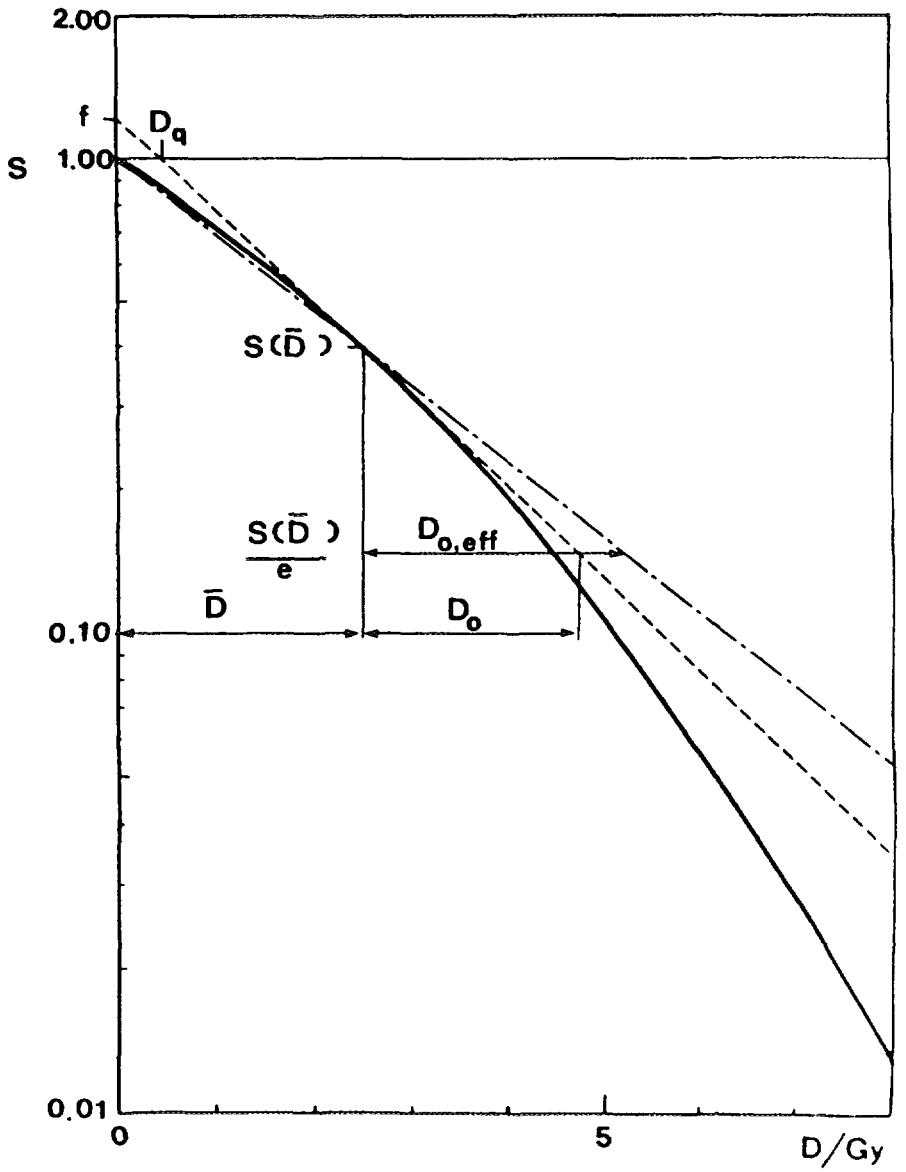

Fig. 2 Comparison of different cell survival curve models. The solid line curve is the linear quadratic relation whereas the dashed line corresponds to a tangent and the dash dotted a secant through the mean dose per fraction.

Using the linear quadratic model to describe cell survival after a total dose D the probability to control a tumor is given by:

$$P(D) = e^{-N_0 e^{-\alpha D - \beta D^2}} \tag{11}$$

where N_0 is the initial number of clonogens. This model makes it possible to accurately predict the effect of dose fractionation by replacing D by the dose per fraction, D/n , by writing:

$$P(D) = e^{-N_0\left[e^{-\alpha \frac{D}{n} - \beta \left(\frac{D}{n}\right)^2}\right]^n} = e^{-N_0 e^{-\alpha D - \beta \frac{D^2}{n}}} \tag{12}$$

To illustrate the effect of the fractionation, the tumor control given by Eq.(12) is shown in Fig. 3 as a function of the total dose and the dose per fraction. In this diagram it is seen how the total dose required to control a tumor is reduced as the dose per fraction is increased. When the dose per fraction is decreased and the number of fractions is increased the response surface becomes very smooth with minimal effects of individual dose fractions. The general curve shape in Fig. 3 is therefore determined by the effective mean cell kill per dose fraction. The cell survival can over a clinically relevant dose per fraction interval around $\overline{D}/n$ according to Eq.(10) be accurately approximated by:

$$S = e^{-\alpha\frac{D}{n}-\beta\frac{D}{n}^2} \approx f e^{-\frac{D}{nD_e}} \tag{13}$$

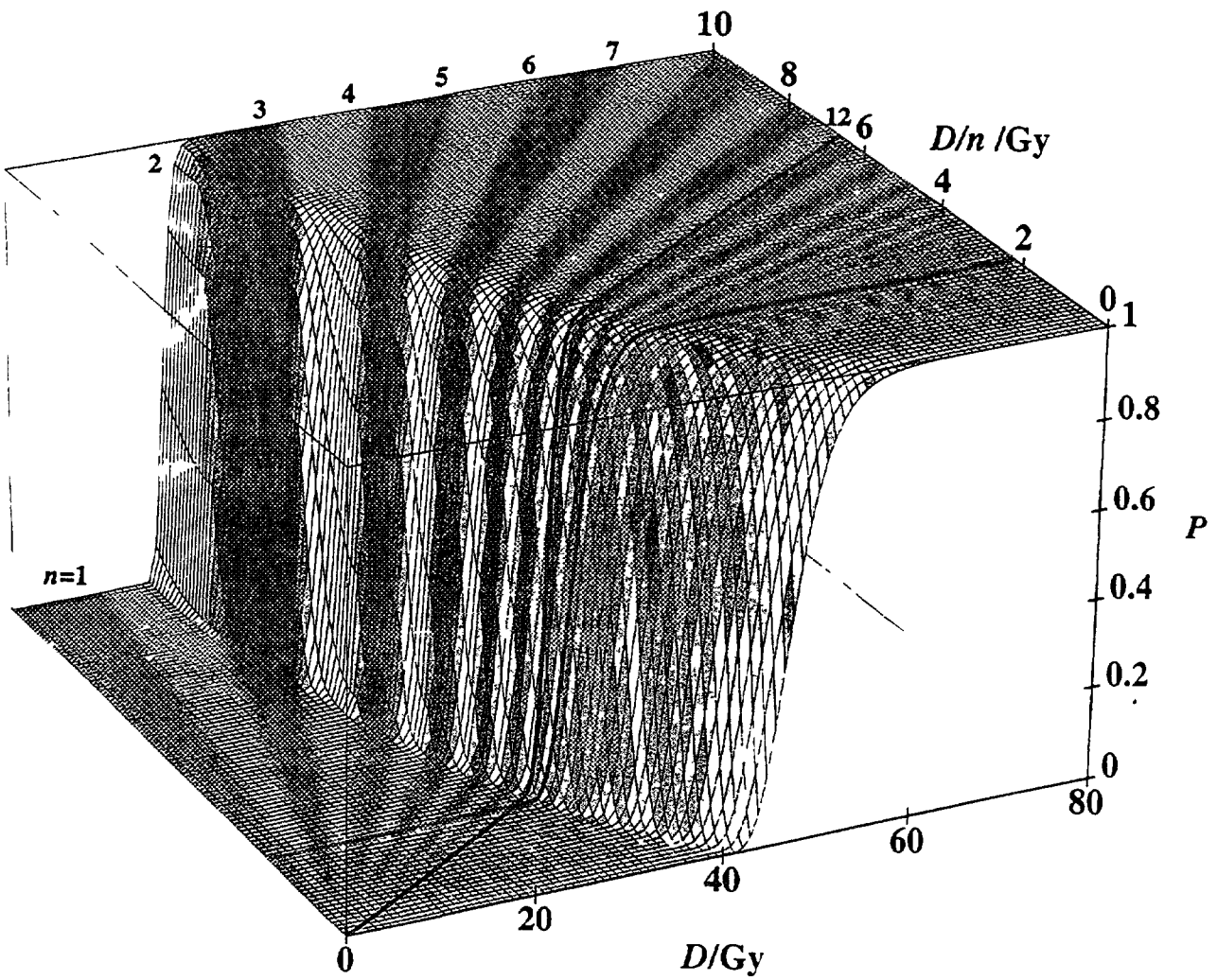

Fig. 3 The variation of the probability to control a tumor consisting of 10^8 clonogenic cells with the total dose and the dose per fraction at an α/β = 5 Gy. Odd dose fractions are shaded to illustrate the effect of fractionation. It is seen that as the dose per fraction decreases the isoeffect dose increases. The solid parallel and oblique lines correspond to a fixed dose per fraction of 2 Gy and a fixed number of fractions (12), respectively.

In order to make both the cell survival and the slope of the dose response relation equal at $D = \overline{D}$, the effective extrapolation number, f, and the effective D_0, D_e, are given by:

$$f = e^{\beta\left(\frac{\overline{D}}{n}\right)^2} \tag{14}$$

$$D_e = \frac{1}{(\alpha + 2\beta\frac{\overline{D}}{n})} \tag{15}$$

Here $\overline{D}$ is the mean dose in the organ in question and α and β are the cell survival parameters of the linear quadratic model. The advantage of the above transform is that it will hold quite well also for non uniform dose delivery as discussed by Brahme [4] since it is linear in the total dose, D. This is illustrated by the two intersections in the dose response surface in Fig. 3. The dose response curve cut out by a vertical plane parallel to the total dose axis correspond to a constant dose of just above 2 Gy per fraction. However, the oblique cut corresponds to a fixed number of dose fractions delivered to the target and has the important property of being practically independent of dose distribution fluctuations in the dose range of ±25 per cent around $\overline{D}$ (*cf.* Fig. 2 and [4]). Using the above notation, Eq.(12) may be simplified according to

$$P(D) = e^{-N_0\left(f\, e D/nD_e\right)^n} = e^{-f^n N_0 e^{-D/D_e}} = e^{-e^{\ln N_e - D/D_e}} \tag{16}$$

where for simplicity the effective clonogen number, $f^n N_0$, is denoted N_e. Eq.(16) is a simple and useful form of the dose response relation for tumor control. It can be generalized to describe the radiation effect in any tissue by expressing the variables N_e and D_e in terms of the normalized dose response gradient, γ, and the dose causing 50% probability of effect, D_{50}. If it is assumed that the number of fractions is constant the dose response gradient at the inflection point becomes simply:

$$\left(\frac{dP}{dD}\right)_n = \frac{1}{eD_0} \tag{17a}$$

In Fig. 4 a constant dose per fraction was assumed instead. The corresponding dose response gradient is slightly lower as given by

$$\left(\frac{dP}{dD}\right)_D = \frac{1}{eD_0} - \frac{n \ln f}{eD} \tag{17b}$$

Thus, at the steepest part of the dose response curve a dose increase of D_0 increases the tumor control by about 37 per cent. However, the dose increase should preferably be measured relative to the total dose when effects of dosimetric uncertainties are investigated. A more important quantity with regard to precision requirements in radiation therapy is therefore the normalized dose response gradient, γ, defined by:

$$\gamma = D\frac{dP}{dD} \tag{18}$$

This parameter is a dimensionless number which describes how large a change in tumor control probability is to be expected for a given relative increase in absorbed dose. In fact a

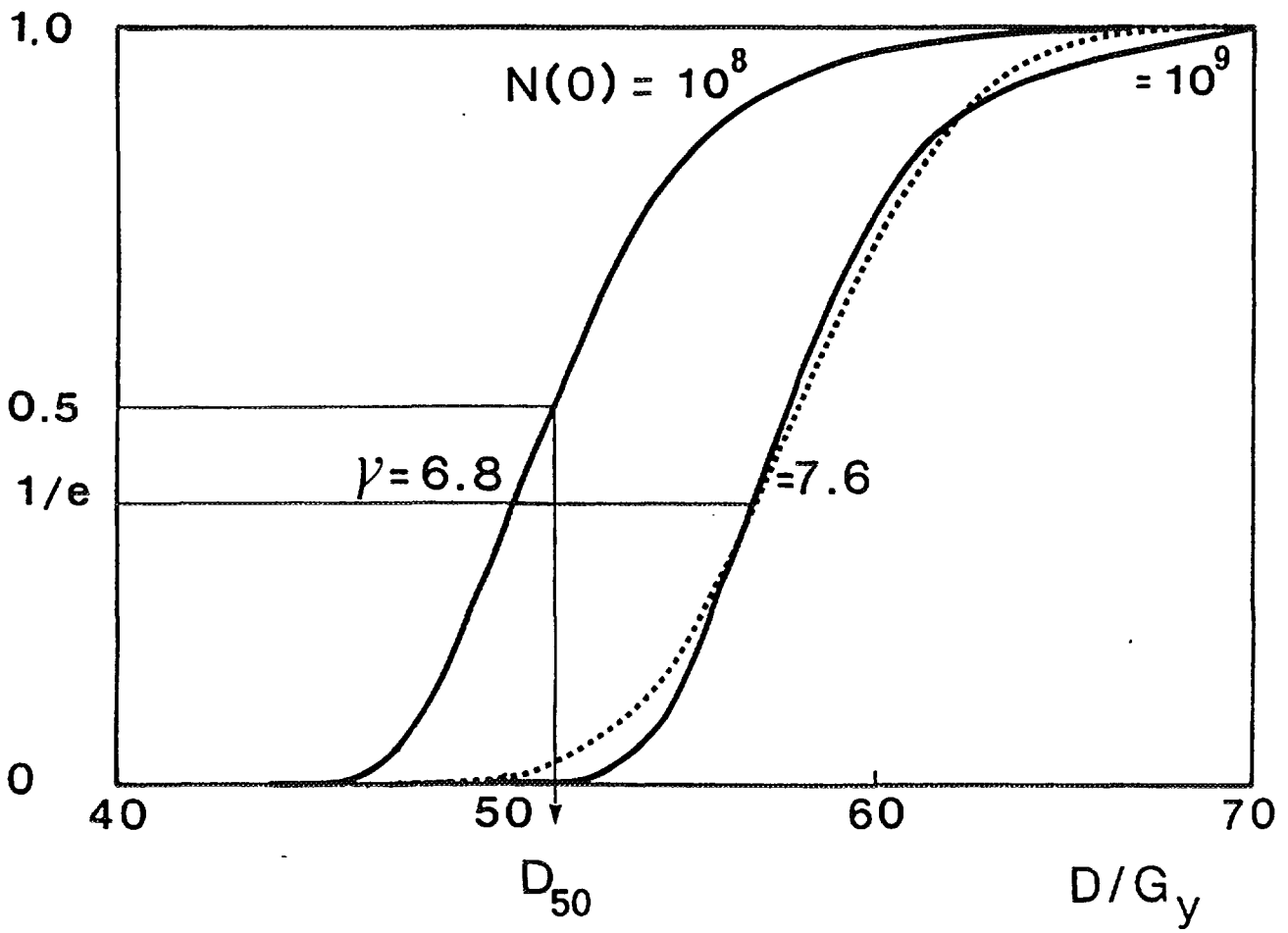

Fig. 4 The shape of the dose response relation given by Eq. (9) for 10^8 and 10^9 tumor cells. The dotted line is a cumulative normal probability curve fitted by eye to the larger cell population.

dose increase of one per cent on the linear part of the dose response curve will result in an increase in tumor control probability of precisely γ per cent. Under the assumption that the number of fractions is constant and that only the dose per fraction is changed the normalized response gradient at the inflection point, γ_n, becomes:

$$\gamma_n = \frac{\ln N_e + n \ln f}{e} \tag{19a}$$

If instead the dose per fraction is kept constant the normalized response gradient is reduced simply to:

$$\gamma_D = \frac{\ln N_e}{e} \tag{19b}$$

These expressions show, under the present simplifying assumptions, that for uniform tumors the normalized gradient increases logarithmically with tumor size. It is also seen that a fixed number of dose fractions results in a steeper dose response curve as also illustrated in Fig. 3.

If we replace D_e by D_{50} the uppermost exponent of Eq.(16) has to be equal to ln(ln(2)) when D is equal to D_{50}, thus Eq.(16) may be rewritten as

$$P(D) = e^{-e^{\left[\ln N_e - \ln(\ln(2))\right]\left(1 - D/D_{50}\right) + \ln(\ln(2))}} = 2^{-e^{\left[\gamma e - \ln(\ln(2))\right]\ (1\ -\ D/D_{50})}} \tag{20}$$

Even if this equation was derived for tumor control it is in this general form with γ and D_{50} as only clinical variables, also applicable to normal tissue injury, as discussed in detail by Källman *et al.* [7]. From Eqs.(14) and (15) combined with Eqs.(19b) and (20), respectively, it is possible to derive how γ and D_{50} will vary with the number of fractions, n:

$$\gamma \approx \frac{\ln N_e + \beta \bar{D}^2/n}{e} \tag{21}$$

$$D_{50} \approx \frac{\ln N_e}{\alpha + 2\beta \bar{D}/n} \tag{22}$$

From Eq.(21) it is clear that the value of γ may increase by a few units when going from a constant dose per fraction to a constant number of fractions (*cf.* Fig. 3). It is also clear how the D_{50} value increases with increasing number of fractions or decreasing dose per fraction. Unfortunately, the increased γ value with a constant fraction number will not generally make the therapeutic window wider even though the same effect will influence both tumors and normal tissues. This is because the difference between the D_{50} values will decrease simultaneously.

2. INFLUENCE OF ABSORBED DOSE DISTRIBUTION ON LOCAL TUMOR CONTROL AND NORMAL TISSUE COMPLICATIONS

2.1. Fundamental considerations

In the preceding section it has been assumed that the absorbed dose distribution was perfectly uniform over the entire tissue volume. In practice the dose distribution is seldom uniform and different parts of an organ or a target volume may also have different sensitivities. Variations in local sensitivity and delivered absorbed dose distribution will influence the tissue effect in a similar way during fractionated therapy since local variations in the surviving fraction after each treatment are multiplied to give the net effect of a complete series of treatments.

The dependence of the probability of a radiation effect on a change in the absorbed dose level can in the uniform case easily be expressed under assumption that the dose variation falls on a sufficiently linear portion of the dose response curve. The increase in effect probability due to a dose increase ΔD may then be approximated by the first few terms of the Taylor expansion:

$$P(D + \Delta D) = P(D) + \frac{dP}{dD}\Delta D = P(D) + \gamma \frac{\Delta D}{D} \tag{23}$$

where γ is the dose response gradient defined in Eq.(18).

Let's first look at a tumor. Assume that the initial clonogenic tumor cells, N, can be divided in two populations of cells, N_a and N_b, of equal sensitivity, irradiated to somewhat different dose levels, D_a and D_b, respectively. The control probability of each cell population is then given by an expression of the type:

$$P_a = e^{-N_a \prod_{i=1}^{n} S\left(D_{a,i}\right)} \tag{24}$$

where $S(D_{a,i})$ is the survival after each dose fraction, $D_{a,i}$, to the population a, and with $\sum_{i=1}^{n} D_{a,i} = D_a$ (compare Eq.16). The probability to control the total cell population is given by the conditional probability, P, that b is controlled when a is known to be controlled. If, as a first approximation, it is assumed that the actions on a and b are statistically independent, P is expressed by:

$$P = P_a \cdot P_b \tag{25}$$

which may be rewritten:

$$P_t = e^{-\left[N_a \prod_{i=1}^{n} S(D_{a,i}) + N_b \prod_{i=1}^{n} S(D_{b,i})\right]} \tag{26}$$

This expression is in agreement with Eqs.(8) and (16) as the terms inside the square brackets give the expected mean number of surviving cells. If, for a moment, it is assumed that the dose delivery is the same for both populations Eq.(24) may be rewritten:

$$P_a = P^{\frac{N_a}{N}} = P^{\frac{m_a}{m}} \tag{27}$$

where the last step is based on the assumption that the cell mass is constant throughout both cell populations of masses m_a and m_b, respectively ($m=m_a+m_b$). It is seen that this expression is consistent with Eqs.(25) and (26) because $m_a/m+m_b/m=1$. This is an explicit expression showing, in general agreement with Fig. 4, that the dose response curve is shifted to higher control rates and lower doses as the number of cells, the tumor mass, or the volume decreases. Under assumption of a constant tissue density a related expression was used by Goitein [8] and Schultheiss *et coll.* [9].

2.2. The effect of over- and under dosage

The increase of the control probability of the whole tumor when a fraction Dm of the tumor mass m is receiving an excessive dose ΔD can now be expressed by combining Eqs.(23) and (27)

$$P(D+\Delta D, \Delta m) = \left(P(D) + \gamma \frac{\Delta D}{D}\right)^{\Delta m/m} P(D)^{(1-(\Delta m/m))} \tag{28}$$

which may be rewritten:

$$P(D+\Delta D,\Delta m)=\left(1+\gamma\frac{\Delta D}{DP(D)}\right)^{\Delta m/m}P(D)\,, \tag{29}$$

If it is now assumed that $\Delta D/D$ and $\Delta m/m$ are so small that a power expansion is valid the change in control probability may be approximated by the first order term given by:

$$\Delta P\approx\gamma\frac{\Delta D}{D}\cdot\frac{\Delta m}{m}=\gamma\frac{\Delta\bar{\varepsilon}}{\bar{\varepsilon}} \tag{30}$$

which clearly illustrates how dose and mass errors combine as an integral dose error in the first approximation. The second equality is based on the definition of the related quantity: the mean energy imparted, $\bar{\varepsilon}$, [10]. In the first approximation it is thus the relative changes from the desired mean energy imparted in the tumor volume that alter the control probability. This result has important consequences for dose specification in radiation therapy. If the dose variations are not too large the mean absorbed dose to the tumor volume should be closely related to the therapeutic effect. The mean absorbed dose is defined by the expression:

$$\bar{D}=\frac{\bar{\varepsilon}}{m}=\frac{\int D(\boldsymbol{r})\mathrm{d}m}{\int \mathrm{d}m}=\frac{\int D(\boldsymbol{r})\rho(\boldsymbol{r})\mathrm{d}V}{\int \rho(\boldsymbol{r})\mathrm{d}V} \tag{31}$$

where $D(\boldsymbol{r})$ and $r(\boldsymbol{r})$ are the absorbed dose and density of the tumor at position $\boldsymbol{r}$ and $\mathrm{d}V$ is the differential volume element. The use of $\bar{D}$ will make the increased and decreased cell killing in hot and cold areas, respectively, compensate each other in the first approximation. This is so because in the hot part of the tumor volume the error in mean energy imparted, $\Delta\bar{\varepsilon}_+$, is larger than zero whereas in the cold portion $\Delta\bar{\varepsilon}$ is less than zero and according to the definition of mean dose $\Delta\bar{\varepsilon}=\Delta\bar{\varepsilon}_+ + \Delta\bar{\varepsilon}_- = 0$. In the first approximation according to Eq.(30) ΔP is therefore equal to zero and the tumor control should be quite close to that expected for a uniform dose distribution. When the density of the tumor, $\rho(\boldsymbol{r})$, is constant Eq.(31) may also be expressed as a volume average:

$$\bar{D}=\frac{\int D(\boldsymbol{r})\mathrm{d}V}{\int \mathrm{d}V} \tag{32}$$

The simple "two volume case" treated in Eqs.(28) to (30) may serve as a clear illustration as D in this case is given by:

$$\bar{D}=\frac{(m-\Delta m)D+\Delta m(D+\Delta D)}{m}=D\left(1+\frac{\Delta m}{m}\frac{\Delta D}{D}\right) \tag{33}$$

By comparing this expression with Eq.(30) it is seen that the change in control probability is in the first approximation proportional to the change in mean dose to the tumor volume. It can thus be concluded that when the dose variations in the beam are small ($\Delta D<<D$ or

more exactly when $|\Delta\bar{\varepsilon}_-| = |\Delta\bar{\varepsilon}_+| << \bar{\varepsilon}$) the best possible correlation between dose delivery and tumor response is obtained when the mean dose according to Eq.(31) is used. This is particularly important if different dose distributions are being used for the same patient or group of patients. Owing to the requirement of small dose variations this conclusion is generally valid for external beam therapy [11]. In the above two sections it was assumed that the tissue was a tumor of strict "parallel" organization. However, all equations derived also pertain to normal tissues of essentially parallel organization as discussed in more detail by Källman *et al.* [7].

2.3. Non uniform dose distributions in general

Under assumption that the mean absorbed dose according to Eq.(31) is used as reference we will now try to find a more general expression for the control probability. This is possible by generalizing Eq.(29) in such a way that it holds for arbitrary dose distributions by allowing Δm and the number of mass fractions to decrease and increase, respectively, without bound and taking the product of all the individual probabilities. After taking the logarithm and conversion of the sum to an integral the control probability for this general case takes the form:

$$P(D(\boldsymbol{r})) = P(\bar{D})e^{1/m\int \ln[P(D(\boldsymbol{r}))/P(\bar{D})]dm} \tag{34}$$

If it is assumed that $\bar{D}$ is on the linear part of the dose response curve and $D(\boldsymbol{r})$ does not deviate much from $\bar{D}$ Eq.(31) may be used to expand Eq.(34). The integral in Eq.(34) may thus be approximated by:

$$\frac{1}{m}\int \ln\left(1+\gamma\frac{D(\boldsymbol{r})-\bar{D}}{P(\bar{D})\bar{D}}\right)dm \approx \frac{1}{m}\int\left[\gamma\frac{D(\boldsymbol{r})-\bar{D}}{P(\bar{D})\bar{D}} - \frac{\gamma^2}{2}\left(\frac{D(\boldsymbol{r})-\bar{D}}{P(\bar{D})\bar{D}}\right)^2 \ldots\right]dm \tag{35}$$

where the first term inside the square bracket gives no contribution to the integral owing to the definition of $\bar{D}$. The complete Eq.(34) may thus to second order be approximated by:

$$P(D(\boldsymbol{r})) = P(\bar{D}) - \frac{\gamma^2}{2P(\bar{D})}\left(\frac{\sigma_D}{\bar{D}}\right)^2 \tag{36}$$

where σ_D^2 is the variance of the dose distribution in the tumor volume as defined by:

$$\sigma_D^2 = \frac{\int (D(\boldsymbol{r})-\bar{D})^2 dm}{\int dm} \tag{37}$$

The negative sign of the quadratic term in Eq.(36) shows that all variations in the dose distribution that introduce deviations from the mean dose level reduce the control probability for a given mean tumor dose or mean energy imparted. This should be expected as the increased survival in low dose areas can never be completely compensated by the decreased survival in high dose areas. By setting Eq.(34) equal to $P(D_{\text{eff}})$ the

effective total dose assuming uniform dose delivery can be calculated and it becomes simply:

$$D_{\text{eff}} = \overline{D}\left[1 - \frac{\gamma}{2P(\overline{D})}\left(\frac{\sigma_D}{\overline{D}}\right)^2\right] \tag{38}$$

This result again clearly shows that the effective dose is slightly lower than $\overline{D}$ by an amount determined by the relative variance of the dose distribution.

The above relations were all derived in terms of the probability of achieving tumor control. If we instead are interested in the effect on normal tissue injury the result is different particularly with regard to non uniform dose delivery. This is so since doses above the mean dose increase the complications more than doses below the mean dose decrease them. Thus, quite generally the probability for complications in normal tissue

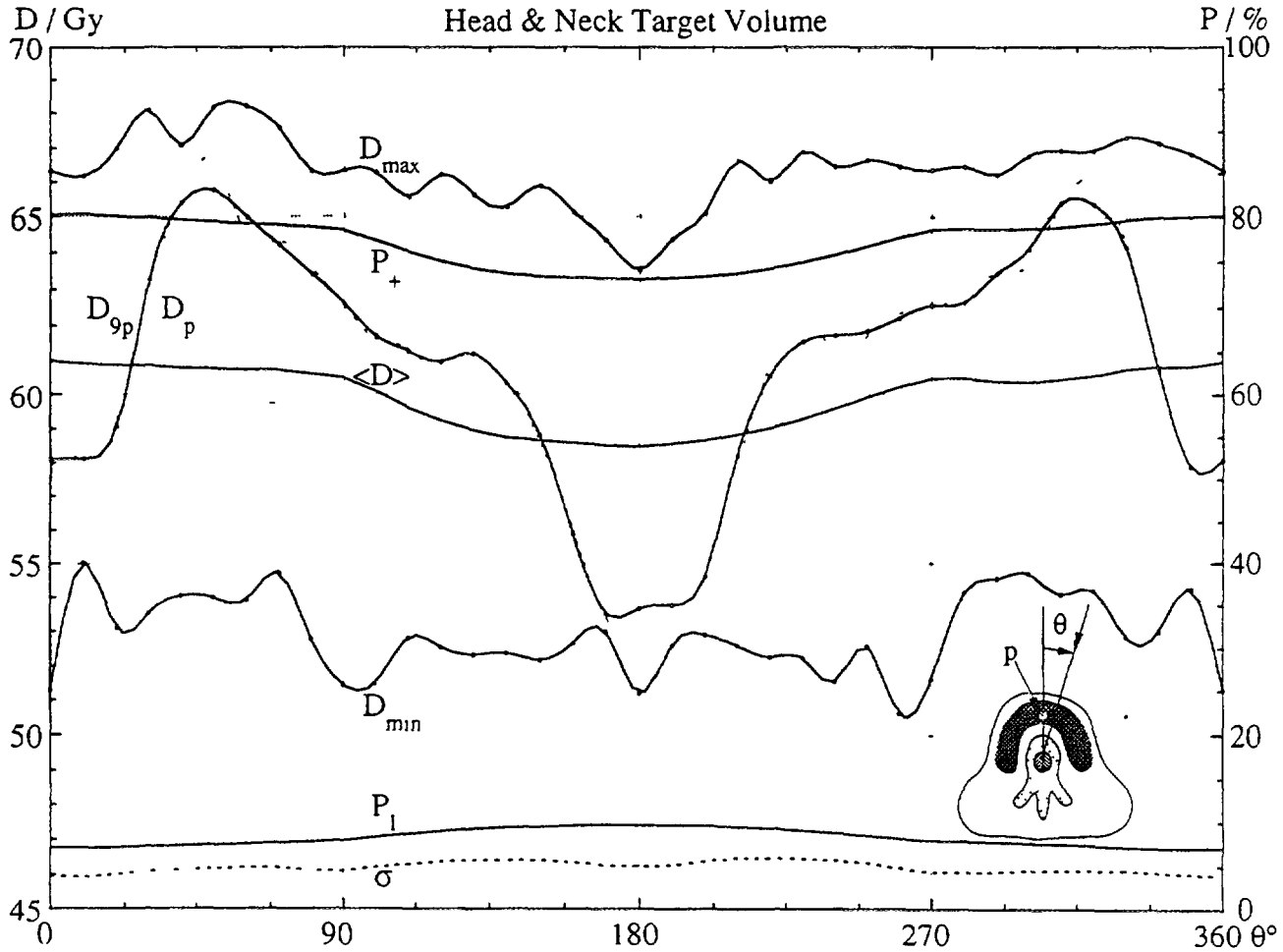

Fig. 5 The variation of P_+ and P_I with the angle of incidence, Ω, for a P_+ optimized treatment of the lymph nodes on the neck. D_p is the point dose at p, $D9_p$ is the mean value of 9 neighboring points over an area of 3 x 3 pixels, D_{max}, D_{min} and $\overline{D}$ are the maximum, minimum and the arithmetic mean value of the dose distribution in the target volume. The strong correlation between P_+ and $\overline{D}$ is evident!

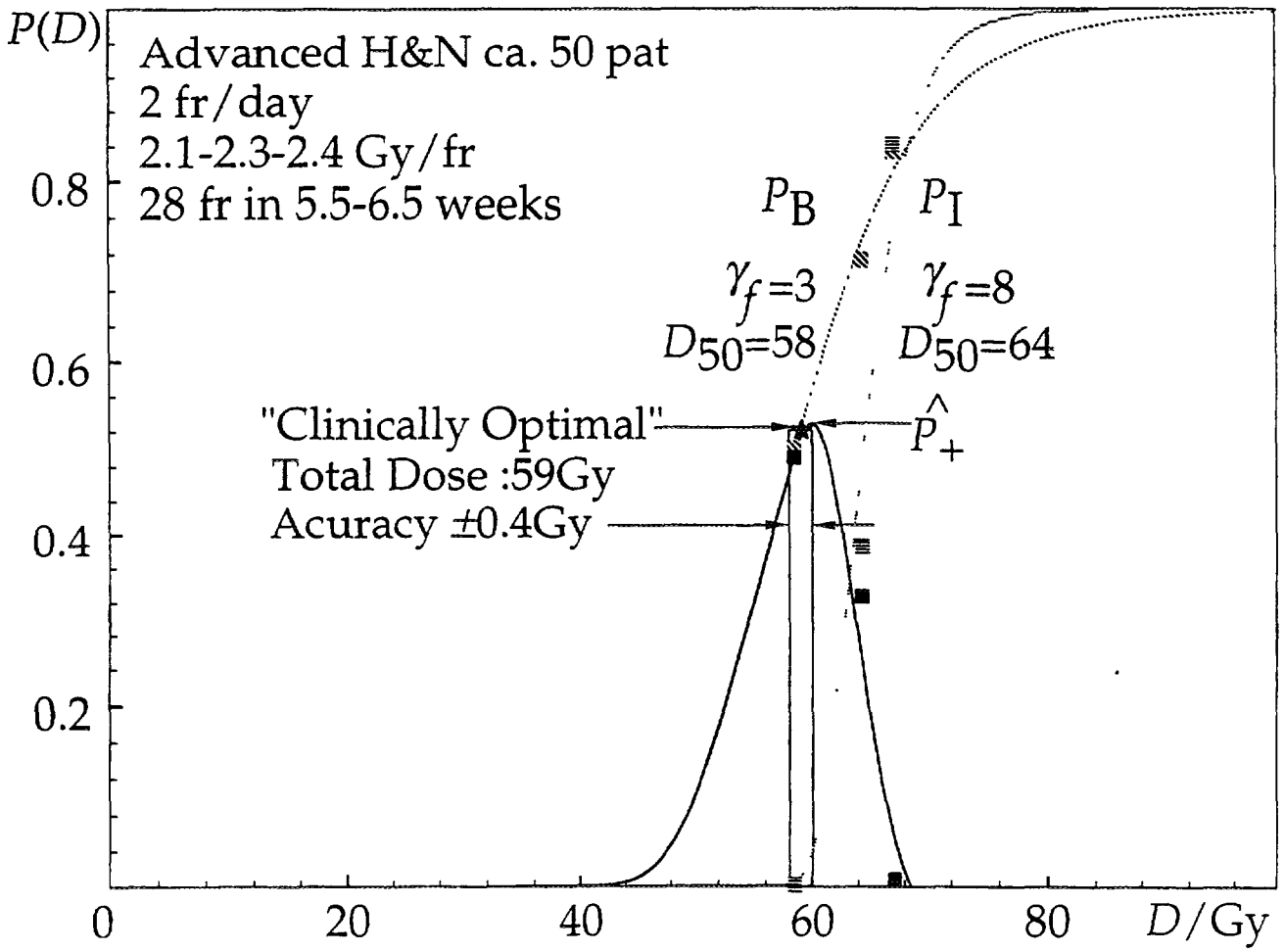

Fig. 6 Clinically established dose response relations for advanced head and neck tumors [12, 13]. The dashed curves labelled P_B (Benefit) and P_I (Injury) correspond to the tumor control and severe normal tissue damage probabilities, respectively. The solid curve represents the probability that patients are cured without inducing severe complications in healthy normal tissues. It is seen that the absorbed dose should be within about 0.5 Gy from the clinically optimal dose in order not to lose more than 5 percent of the patients that potentially can be cured.

increases with increasing fluctuations around the mean dose level! If the same analysis as above is made for the normal tissues we obtain:

$$P_I(D(r)) = P_I(\bar{D}) + \frac{\gamma^2/2}{1-P_I(\bar{D})}\left(\frac{\sigma_D}{\bar{D}}\right)^2 \qquad (39)$$

Thus the first order effect is similar for normal and malignant tissue and is related to the mean dose to the tissue in question. However, dose fluctuations in normal and malignant tissue will increase complications and decrease tumor control, respectively. Thus, in general the probability to achieve complication free tumor control, P_+, will decrease due to both these effects. This is so since in the first approximation [12]:

$$P_+ \approx P_B(1 - P_I) \approx P_B - P_I \qquad (40)$$

To illustrate the importance of the various dose distributional parameters the dose at a central point in the tumor, the arithmetic mean dose, the minimum and maximum dose are shown in Fig. 5 as a function of the angle of incident beams on a head and neck tumor. It is seen that all dose concepts varies substantially with the angle of incidence. However, the mean dose always seems very well correlated to the probability of achieving complication free tumor control (P_+). This figure clearly shows the importance of using the right dosimetric quantities when prescribing the dose delivery in radiation therapy.

2.4. Precision requirements in radiation therapy

In the above discussion several expressions were given both for the probability of achieving tumor control (Eqs. 6, 8, 20, 36) and complications in normal tissues (Eqs. 20, 39). What is most important in radiation therapy is that the probability, P_+, that patients achieve tumor control without severe complications in normal tissues, is as large as possible (*cf.* Eq. 40). As seen in Fig. 6 these patients follow a bell shaped curve as a function of dose with a maximum approximately half way between the doses causing 50% probability of tumor control and that causing 50% normal tissue injury. It has recently been shown [13] that this curve is well described by a Gaussian distribution according to

$$P_+ \approx P_+(\hat{D}) e^{-\pi\gamma^2\left(\frac{D-\hat{D}}{\hat{D}}\right)^2} \tag{41}$$

where $\hat{D}$ is the optimal mean dose to the patient. Basically, this is due to the fact that the difference between the two closely spaced sigmoidal tumor and normal tissue injury curves can be approximated by the derivative of the sigmoid provided their γ values are fairly similar. From this relation it can be shown that if one wants to have at least 95% of the maximum possible complication free tumor control $P_+(\hat{D})$ the dose delivered should be within ΔD from $\hat{D}$ as given by

$$\Delta D \approx \frac{\hat{D}}{10\gamma} \tag{42}$$

This means that for steep dose response gradients γ the clinically acceptable dose interval is quite narrow. For typical clinical data assuming $\hat{D} \approx 64$ Gy and $\gamma = 4$ the required accuracy in dose delivery is about 1.6 Gy or less than one dose fraction. This corresponds to an accuracy in dose delivery of about 2.5% which is a quite demanding value compared with what is generally achieved in clinical practice [4, 11].

3. CONCLUSIONS

The accuracy of the dose distribution and in particular of the mean absorbed dose delivered to the target volume need for many steeply responding tumors be as high as 1 Gy (1σ) as seen in Fig. 6. The main reason being that one want to deliver a dose which is as high as possible but preferably not higher than the peak of the bell shaped complication free tumor control curve. At higher doses the probability of inducing fatal injury increases very steeply so the optimum clinical dose is just below the peak of the curve. In order not to lose more than one out of 20 curable patients (95% of highest possible treatment outcome) the required accuracy in the dose distribution delivered to the target volume

should be 2.5% (1σ) for a mean dose response gradient γ in the range 2 - 3. For more steeply responding tumors and normal tissues even stricter requirements may be desirable.

When only one single dose quantity is required to analyze dose response data and to prescribe and report dose delivery the mean dose to the target volume ($\overline{D}$) is the most relevant concept at least in external beam radiation therapy where the dose variations over the tumor in general are quite small (*cf.* Eqs. 36 and 38). The mean dose is also most relevant for normal tissue damage under the same assumptions (Eq. 39). To get a physical and radiobiological feeling for the influence of the dose heterogeneity in the tumor and organs at risk the relative standard deviation of the dose around the mean dose is the most useful concept (Eqs. 36, 38, 39). In addition to the mean dose and its standard deviation the ICRU point dose should be stated to allow comparison with other recommendations [14, 15]. In particular when the dose variations are large (≥5% 1σ) the maximum and, in particular for tumors, the minimum dose are highly relevant too.

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II. EQUIPMENT REQUIREMENTS

LESSONS LEARNED FROM ACCIDENTS IN RADIOTHERAPY

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Abstract

Radiotherapy is the only application of radiation which intentionally delivers very high doses to humans. A gross deviation from the prescribed dose or dose distribution can have severe, or even fatal consequences. Since the patient is placed directly in the beam or sources are inserted in the body, any mistake made with the beam or the sources leads almost certainly to an accidental exposure. Lessons learned from previous incidents can be used to test the vulnerability of a given facility, provided that these are adequately disseminated. The purpose of this paper is to present a summary of the lessons learned from a relatively large sample of events. The analysis has been presented as a short description followed by an identification of the triggering event and the contributing factors. These have been grouped as follows: errors in commissioning or calibration machines and sources affecting many patients; mistakes affecting individual patients such as irradiating the wrong patient, the wrong, field or site, and mistakes when entering data into or reading from the patient's chart; errors due to unusual treatments or situations; equipment failure and human machine problems, including maintenance.

INTRODUCTION

There are situations that are unique to the medical use of radiation sources: patients are exposed to direct radiation beams and radiation sources are incorporated to their bodies as part of the diagnosis and treatment.

In therapeutical applications, doses are very high and a departure from the prescribed doses may have severe or even fatal consequences. Not only overexposure but also doses below the intended ones are accidental exposures in the case of the patients. Accidental exposures also include any treatment delivered to the wrong patient or the wrong tissue, or using the wrong source or the wrong radiation beam, or with a dose or dose fractionation differing significantly from the values prescribed or which may lead to undue secondary effects.

Lessons learned from previous accidents can avoid reoccurrence. For this purpose, the IAEA has collected information on accidents and made a review of 55 events. The result will be published as an IAEA special publication.

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The sources of information have been:

- papers published in scientific journals
- information provided by professional associations
- periodical reports available from national institutions, such as the USNRC

The lessons learned from this sample of events can be directly used as a checklist for testing the vulnerability of a given facility against the initiating events which triggered the accidents reviewed. An effective way of learning concrete lessons is the collection of contributing factors which made possible the initiating event culminating in an accident.

To illustrate the method of review, a few examples are given in the following section. It should be noted that, due to the length of this paper, the examples are extremely simplified. The review also includes two accidents involving the public and the environment.

EXAMPLES OF ACCIDENTS, INITIATING EVENTS AND CONTRIBUTING FACTORS

Event 5A

A new Co-60 source was installed. A mistake in the determination of output (dose rate) was made. The doses to patients were 25% higher than intended. A total of 207 patients were affected.

Initiating event:

Miscalculation of dose rate from measurements

Contributing factors:

1) There was no independent calibration of output
2) Subsequent reviews did not detect the mistake. Only an intercomparison exercise discovered it
3) There was no sufficient investigation of unusually slow healing of skin effects

Event 5B

A Co-60 decay curve was wrongly drawn. There was no calibration or verification of the output during 27 months. Overdoses increased with time, up to 40%, as a result of the departure of the curve from the real decay. A total of 450 patients were affected.

Initiating event:

Mistake in the determination of decay curve

Contributing factors:

1) Use of linear scales instead of simpler semi-log scales
2) No independent determination
3) No beam verification during 27 months
4) Priority was given to a new accelerator, so that Co-60 unit was assigned lower priority

Event 7

A computerized Treatment Planning System (TPS) was commissioned. A manual correction for distance is isocentric treatments was introduced. The correction factor was already included in the software of the TPS. The doses in all isocentric treatments were lower than intended by as much as 30%. The mistake remained for eight years. More than 1,000 patients were affected.

Initiating event:

A TPS was incorrectly commissioned. A distance correction was introduced twice.

Contributing factors:

1) There was no dosimetric validation of the TPS
2) Lack of written procedures made it likely that mistake remained undetected for years

Event 16

The wrong patient received a teletherapy fraction of 2,5 Gy to the lumbosacral spine.

Initiating event:

A patient responded when another patient was called

Contributing factors:

1) Procedures for patient identification (photograph) were not followed
2) Procedures for confirming the treatment site were not followed (tatoos)
3) Patient's objections were not given sufficient attention (by neither the technologist nor the oncologist)

Event 35

A patient was prescribed a teletherapy treatment, but received a treatment with Sr-90, 10Gy to the surface of the right eye.

Initiating event:

The wrong treatment was chosen

Contributing factors:

The information on the patient chart was not considered or was mistaken

Event 19

The interlock system of a linear accelerator stopped the beam. No beam was available. In order to obtain the beam, a decision to treat in "physical mode" was made. Twelve patients were treated normally. An equipment failure occurred when the interlock system ("non clinical mode") was disabled and one patient died.

Initiating event:

Equipment fault. No beam available from accelerator

Contributing factors:

1) Decision to treat patient in "no clinical" mode
2) A second failure (power supply) disabled both, the x ray positioning system (target, cone, ionization chamber) and the warning signal of their incorrect position

Event 20

The interlock system of an accelerator stopped the beam. No beam was available. An incorrect repair was made without notifying the radiation physicists. The control panel showed a fix energy (36 MeV) permanently regardless of the selected energy. Twenty-seven patients were treated with the wrong energy, dose and dose distribution. Several deaths have been admitted to be caused by the wrong treatment.

Initiating event:

Failure of the equipment and beam was stopped by the interlock

Contributing factor:

1) The repair did not correct the real fault. Energy fixed at 36 MeV. Energy selector at console disabled
2) The radiation physicists were not notified. Treatments resumed without verification of the beam
3) Staff assumed that the fix energy displayed at control panel was defective (the display was correct)
4) It was possible to operate the accelerator with the energy selection disabled (equivalent to non "clinical mode") from the normal "beam on" key

Event 21

An accelerator equipped with verification of the treatment parameters by software. The automatic verification process took about 20s. Selected parameters were modified during the automatic verification process, resulting in an operation with hybrid parameters. Six accidents involving overdoses occurred. Three patients died.

Initiating event:

An accelerator operated with hybrid parameters after the technologist changed previously selected parameters

Contributing factors:

1) Equipment not tested for operating conditions that occurred in practice in several facilities. Quality control and quality assurance of the software was not sufficient to avoid the occurrence of quick change of selected parameters
2) Commissioning did not include testing in operating conditions, that occurred in practice
3) The problem was identified by manufacturer only after occurrence of six accidents involving overdoses in four different hospitals. The first accident occurred in 1985 and the last one in 1987

Event 27

The wrong source was used in a brachytherapy treatment.

Initiating event:

The wrong source was selected

Contributing factors:

1) One drawer contained sources of two different activities
2) Labelling of source not adequate
3) No effective verification of sources prior to implant

Event 28

A treatment was performed with the wrong source.

Initiating event:

The wrong sources were selected for treatment

Contributing factors:

1) the colour coding was fading
2) no sufficient verification of the sources prior to implant

Event 25

The same event occurred in two different hospitals. Source ribbons of 0.79 mCi Ir-192 each were ordered. Ribbons with activities of 0.79 mg-Ra-equivalent were delivered. The wrong source was implanted into one patient in each hospital.

Initiating event:

The wrong activity was delivered

Contributing factors:

1) Different units of activity were used by the hospitals and the supplier
2) Insufficient verification was performed (only the number 0.79 but not the unit)
3) No source verification/calibration by the users

Event 36

A 53 mCi, Cs-137, radiation source was discharged after a brachytherapy treatment

Initiating event:

A radioactive source was discharged with inactive waste

Contributing factors:

1) Control of sources returned to the container after treatment was insufficient
2) Monitoring of the area after removal was not performed or was unsuccessful

Event 39

Brachytherapy treatment with high Dose Rate (HDR) Equipment. A source remained in the patient. The control panel indicated "safety" but the are monitor was sounding an alarm, indicating that the source was not returned into the container. The patient died. Members of the public were exposed to a HDR source.

Initiating event:

The source became dislodged from the equipment

Contributing factors:

1) Misinterpretation of two conflicting signals. Monitor alarm was ignored
2) Failure to ensure that the sources were not outside the equipment by monitoring of patient, clots and working area

Event 40

Remote control afterloading. Equipment failed to complete transferring the sources to the equipment. The sources remained for an unknown time somewhere near the patient's leg.

Initiating event:

The source became disconnected from drive mechanism

Event 50

A nursing mother was given 4.89 mCi of I-131 which resulted in an untended radiation dose of some 300 Gy to her infant's thyroid gland. The infant will require thyroid hormone medication for life. The mistake was detected when a whole body scan of the mother was done, which indicated an unusually high breast uptake.

Initiating event:

4.89 mCi was given to a nursing mother

Contributing factors:

1) No one (neither the referring physician, nor the staff in the nuclear medicine station) asked the mother if she was nursing. The usual measures for this kind of situation were not taken

Event 22 (involving public and environment)

A radiotherapy department moved to new premises and left a teletherapy unit in the old hospital. Members of the public had access to the unit, dismantled it and broke the Cs-137 source capsule. As a result of the spread of radioactive material, 112,000 persons had to be monitored for possible contamination, 249 individuals were found to have some contamination, 4 persons died, 159 houses had to be monitored for contamination, 42 houses required decontamination and 35,000 m^3 of wasted were generated.

Initiating event:

A teletherapy unit was dismantled and the source capsule was broken.

Contributing factors:

1) There was no proper decommissioning of the facility
2) The teletherapy unit was left in unsafe storage conditions
3) The national regulatory authority was not notified of the discontinued operation of the radiotherapy facility
4) The unit was not recognized by the members of the public as something dangerous
5) The chemical form of the source facilitated the spread of the contamination

Event 23 (involving public and environment)

A Co-60 teletherapy unit was illegally transported, imported into a country, stored during six years in unsafe conditions. A maintenance technician was able to get access to the unit, to dismantle the source driver, and to break the source capsule. Metal parts were sold to a scrap metal company and this in turn to a melting facility.

The result was that 30,000 metallic table bases made form contaminated material were distributed, as well as 6,000 tons of reinforced rods for buildings. 17,600 houses were checked

for contaminated rods. 814 houses were partially or totally demolished. 16,000 m^3 of contaminated earth were generated.

Initiating event:

A person had access to the teletherapy head, dismantled the unit and broke the source capsule.

Contributing factors:

1) Illegal import, transport and unsafe storage of a radiotherapy unit
2) The unit remained stored unsafely for six years
3) The unit was not recognized by the technician as something dangerous
4) The scrap metal company received contaminated material and did not detect it

LESSONS LEARNED FROM INITIATING EVENT AND CONTRIBUTING FACTORS

1. MISTAKES MADE DURING COMMISSIONING, BEAM AND SOURCE CALIBRATIONS, SUCH AS:

Determination of the dose rate for Co-60 teletherapy units and of the dose per monitor unit for accelerators

Decay tables or curves for the calculation of irradiation time per fraction

Determination of wedge factors

Validation of treatment planning systems

Verification of the activity for sealed sources (brachytherapy) and unsealed sources (metabolic therapy)

The lesson learned from these accidents is that the consequences may be severe or even fatal and affect a large number of patients. Examples are 207, 450, 1045 patients.

Recommendations on accident prevention and/or mitigation of consequences:

Human redundancy and independent determination and calculation for commissioning of equipment and facilities

Periodic internal and external independent audits

Dosimetric verification in phantom under real working conditions and with the normal operator using the equipment. This includes the validation of the computerized treatment planning system

When and where practicable, "in vivo" dosimetry (at least at the first fraction)

Implementation of quality assurance with periodic constancy checks

2) MISTAKES CONCERNING THE WRONG PATIENT, THE WRONG FIELD, THE WRONG BEAM OR SOURCE

These mistakes are relatively frequent and are usually related to procedural mistakes (either non written procedures, not well understood, not well verified or violated procedures)

Recommendation for prevention and/or mitigation. (In many teletherapy cases the mistake affects only one or two fractions. Therefore the severity can be kept low if the mistake is discovered by means of frequent verification)

Patient identification with photograph on the patient's chart

Communication with the patient

Clear and unambiguous procedure for tatoos

Human redundance, clear job descriptions, clearance with signatures by the operating staff for each treatment/fraction

3) MISTAKES RELATED TO ENTERING OR READING DATA FROM THE PATIENT'S CHART

These mistakes often deal with entering the wrong fraction dose, accumulation of the total dose, register of the wrong beam, wrong identification of wedges, the wrong wedge factors, or wrong activity or dose units

Recommendations for prevention and/or mitigation:

Redundant and independent, frequent review of patient's charts (two persons, twice a week)

Clear, concise and unambiguous written procedures

Clear job descriptions. Cross check of job descriptions to avoid gaps in responsibility

Clearance with signatures of the operating staff

4) MISTAKES SPECIFIC TO BRACHYTHERAPY

The wrong radionuclide, the wrong activity, the wrong units of activity, mistakes made in the identification of the sources, in the determination of the treatment time, the position of the sources in the applicator, the incomplete retrieval of the sources (sources left in the patient), mistakes in the accountancy of the sources at the storage, source damage or lost sources.

Recommendations for accident prevention or mitigation:

Clear identification of sources, double verification

Registration of all source movements without gaps

Accountancy and verification of sources before and after treatment

Monitoring the patient, cloths and area before discharging him

Clear definition of function of staff with regard to all steps: ordering, reception, labelling storage, retrieval for use, transfer between staff and return to storage

Clear job description and clearance by means of signatures

Quality control of remote control afterloading

5) COMMUNICATION MISTAKES

The mistakes are:

Lack of communication
Incorrect communication, or to the wrong person or by the wrong way
Oral communication of critical information wrongly understood
Mistakes when reading or transferring information
Unreadable or confusing handwritten communication, informal expressions or use of jargon which is not necessarily understood by everyone in exactly the same way
Equipment instruction in a foreign language
Telephone-only communications
Interpersonal problems
Noisy environment, prone to distraction or to loose concentration
Insufficient dedication (non availability) of key positions personnel in the process of communication

Recommendations for accident prevention and/or mitigation:

Identification and listing of all safety critical communication
Clear and concise written procedures for the safety-critical communications identified in the list
Clear assignment of responsibilities to the staff involved in the process of communication
Written and signed safety-critical communications
Communications check-lists

6) SPECIAL TREATMENTS, SPECIAL SITUATIONS, PERSONNEL CHANGES

Change of supplier of radioactive material

Non-typical dose for a specific treatment, unusual area or treatment with the patient in difficult and non-usual position

Change in units of activity

Recommendations for accident prevention and/or mitigation:

Anticipation of these situations by means of a safety assessment of all procedures

Clear and concise written formulation

Additional, specific training for special situations

Clear job description and clear assignment of responsibilities

7) EQUIPMENT FAILURE

The percentage of contributing factors to the overall causes due to equipment is small in the review of these histories. However, the consequences can be very severe and affect many patients. These contributing factors are:

No sufficient redundancy (single fault criterion) (interlock failure)

Software problems

Hardware incompatibilities

Recommendations for accident prevention and/or mitigation:

Implementation of basic design rules, such as:

Safety in depth (in terms of engineering this is call defence in depth)

Note: defence in depth consists of overlapping safety provisions, such as physical components, procedures or combination of both, so that a very low probability of failure can be achieved by combining protective layers such that the probabilities are multiplicative.

Single fault criterion. (e.g., one single faulty component should never cause an accident. An accident should be only possible if two systems or components fail simultaneously. By repairing the faulty component before the second fault appears, accidents can be prevented. The "single fault criterion" is a simple design approach to obtaining a minimal redundancy).

Redundancy, independence and diversity for safety critical components

Use of fault tree analysis for design

Systematic use of safety assessment methodologies

Equipment tests including possible operating mistakes (quality control of the manufacturing process including clinical conditions, and real machine operators) including special or extreme conditions and handling.

When the equipment is in "non clinical mode" (reduction of interlocks and elimination of layers of the safety in depth), it should be made impossible to meter a "beam on" order from the normal key in the keyboard. Rather, validation exercises of the whole, including clinical conditions should be made before starting treatments with real patients.

Manufacturers should investigate promptly and thoroughly any reported incident or unusual event and notify authorities and users of the findings and/or preventive measures. To achieve this, a system of formal reporting should be established in each country and disseminated internationally.

8) PROBLEMS OF HUMAN-MACHINE INTERFACE

Mistaken interpretation of signals and displays

Mistaken decisions of contradictory signals. Tendency to assume as a good signal, the one which fits expectations (tendency to accept unsafe conditions as those which allow resumption of operation).

Recommendations for accident prevention and/or mitigation:

Training of personnel to recognize abnormal situations (understanding of safety assessment).

Training which includes both normal and abnormal contradictory signals (in general and specific to the particular equipment)

Learning from previous accidents and unusual events (including maintenance personnel)

Training promoting a questioning and learning attitude as part of the safety culture. (See point 12)

9) BYPASSING OF INTERLOCKS AND OPERATION IN "NON CLINICAL MODE"

The bypass of interlocks reduces safety drastically

The decision of operating in a "non clinical mode" is often due to frequent or non-resolved maintenance problems, or intermittent faults non-definitely resolved, which causes the equipment to stop, thus disturbing the patient's treatment

The consequences of operation in "non clinical mode" are often fatal or quite severe

Recommendations for accident prevention and/or mitigation

Priority should be given to plan and implement efficient maintenance and avoid improvisation as far as practicable

Never treat patients in "non clinical mode"

Equipment design should not allow "beam on" from the normal key in "non clinical mode" (it should only be possible by using special tools or with special computer codes, for maintenance or physical work).

10) MAINTENANCE PROBLEMS

Mistakes made in maintenance can also be severe, even fatal, or affect many patients

Recommendations on accidents prevention and/or mitigation

The training of maintenance personnel should include the knowledge of the consequences of any manipulation, adjustment in all components. The training should be the result of an exhaustive safety assessment

There should be unambiguous communication procedures for transferring the machine, for initiating maintenance and returning the machine to the staff responsible for radiotherapy physics

Clear and unambiguous responsibility definition

11) PROBLEMS WITH DECOMMISSIONING OF SOURCES AND EQUIPMENT AND FACILITIES, AND UNSAFE STORAGE

Sources out of control can lead to catastrophic results

Recommendations for accident prevention and/or mitigation of results

Formal procedures for regulatory control of sources not yet in use or no longer in use, from import into the country to the proposer disposal.

Clear, unambiguous and authorized procedures making it mandatory to retrieve the source and to store it temporarily or definitely until disposal. Formal control during this period of time

Seal-off of equipment out of use to avoid connection to the electrical power supply

Storage of sealed sources after leakage or damage

Safe disposal of brachytherapy and nuclear medicine sources

Verification and clear labelling of empty containers

SAFETY CULTURE

In the preceding section, recommendations on accident prevention emerging from lessons learned from previous accidents have lead to the need for clear job descriptions, clear assignment of responsibilities, clear written procedures, compliance with procedures documented by signatures, written communication, safety assessment and training of personnel on anticipated problem situations. All this constitutes good practice.

However, even with strict procedural compliance there may be situations, events or combination of events, not exactly defined in the procedures where strict observance may not be sufficient. In these cases the effectiveness in dealing with initiating events and the culmination of accidents strongly depends on human attitude and judgement.

Therefore, good practice is essential but not sufficient. There is a need to go beyond the implementation of a good practice in order to ensure that both personal attitudes and habits of thought and organizational approaches and priorities are oriented toward the goal. This goal is that all duties

related to safety be carried out correctly, with alertness, due thought and full knowledge, sound judgement and a proper sense of accountability.

Furthermore, it implies a learning attitude in all organizations concerned, taking into account all relevant operating experience as well as new research results as a basis for safety improvements and reassessments.

The International Basic Safety Standards for Radiation Protection and the Safety of Radiation Sources [1] establish that the Regulatory Authority is to require all parties involved to develop a safety culture.

The safety culture involves therefore all levels: regulators, commitment of management and response of individuals. Measures to encourage a questioning and learning attitude and to discourage complacency should be put into place.

INTERNATIONAL DISSEMINATION OF INFORMATION ON ACCIDENTS AND UNUSUAL EVENTS

A questioning and learning attitude, combined with information on accidents and unusual events, their initiating events and contributing factors is an efficient way to drastically reduce the probability of further accidents.

The list of contributing factors can be use directly to test the vulnerability of any radiotherapy department.

The number of reported events in a single country is in most cases insufficient to provide a significant number of lessons in a reasonable time. Therefore, a compilation of accidents at the international level would allow all countries to benefit from the lessons learned by each of them. Moreover, unusual events which did not culminate in an accident can build up a body of operational knowledge to avoid real accidents. Therefore, the system should allow for anonymous reports of these cases that otherwise are not published in any scientific journal and would never reach the interested community.

For this reason, an International Reporting System of Unusual Events has been proposed by an advisory group. The reporting system will consist of a questionnaire and a narrative section. The database will include protected fields for confidential identity data which should not be included in the output of the system (reports).

REFERENCE

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REVIEW OF WHO/PAHO/IAEA RECOMMENDATIONS CONCERNING RADIOTHERAPY FACILITIES

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Abstract

Since the mid 1960s the World Health Organization, the Pan American Health Organization, and the International Atomic Energy Agency have provided recommendations concerning radiotherapy services, including organization, staff requirements and facilities. These are contained in various reports of WHO, PAHO and IAEA, which are reviewed and summarized.

INTRODUCTION

Beginning in the 1960s, international organizations with interest in the planning, organization, and provision of radiotherapy services have provided guidance for their member states. These are contained in various reports of the World Health Organization, the Pan American Health Organization and the International Atomic Energy Agency. For example WHO Technical Report Series Nos. 322 Cancer Treatment, 328 Planning of Radiotherapy Facilities (a joint WHO-IAEA meeting) and 644 Optimization of Radiotherapy. The earlier recommendations have been reconfirmed in more recent meetings convened by IAEA and WHO and today are still valid in many parts of the world. The need for a scientifically sound, robust, reliable treatment machine capable of high-quality performance has been universally recognized. Up to the present time the consensus of opinion is that a cobalt-60 machine is preferable and linear accelerators could not be generally recommended for use in developing countries.

Because of difficulties related to the timely replacement of cobalt-60 sources, including the proper disposal of used sources as well as the difficulties in providing the proper infrastructure to effectively utilize linear accelerators, IAEA, PAHO, UNIDO, WHO recently have organized (Washington D.C., December 1993) an Advisory Group Consultation to consider the design requirements for a megavoltage x-ray machine which could overcome the disadvantages of both the cobalt-60 unit and currently available linear accelerators.

REVIEW OF RECOMMENDATIONS

Cancer Treatment, Report of a WHO Expert Committee, WHO TRS No. 322 (1966)[1]:

EQUIPMENT;

Treatment Unites Recommended for a Radiotherapy Department or Centre

A. For conditions lying close to accessible body surfaces:

Low voltage x-rays, 60 - 100 kV

B. Megavoltage radiation for deep-seated lesions:

For x-ray therapy, 3 - 8 MV

C. Kilovoltage radiation:

250 kV or cesium-137 at 30 - 40 cm S.S.D.

D. Electron therapy, not essential for every radiotherapeutic department.

"The linear accelerator or the cobalt-60 teletherapy unit (whose gamma-rays are equivalent to 3-million-volt X-rays) must therefore be regarded as a standard piece of equipment for a modern department. As to which of these two types of machine is to be preferred there is much argument. Linear accelerators can produce rather more penetrating radiation and therefore make the treatment of the large patient somewhat simpler. They also provide sharper beams, which may be clinically advantageous. On the other hand, they require much more skilled technical attention to ensure their continued steady running, and the cobalt-60 machines do provide a non-fluctuating source of radiation. Offsetting this, of course, is the fact that the cobalt radiation can never be switched off, which introduces extra protection problems, and furthermore the natural decay of the radio-active source means that replacement has to be undertaken about once every three years, an expensive and somewhat troublesome business.

On balance, it is probably fair to say that if regular expert technical assistance is not readily available the telecobalt unit is to be preferred. Elsewhere the choice may depend on local opinion and can be allowed to do so since the differences are marginal. One point worth mentioning is that financially there is also little to choose between the machines, what advantage exists seeming to lie with the linear accelerator, provided it is used to capacity." ..... page 52

STAFF; full time dedication to radiotherapy of medical and auxiliary staff (physics, radiotherapy technicians, statistician).

Planning of Radiotherapy Facilities, Report of a Joint IAEA/WHO Meeting, WHO TRS No. 328 (1966)[2]:

EQUIPMENT;

Choice of Radiotherapy Equipment

I Deep-seated Lesions (following are available)

cobalt-60

2 MV resonant transformer

2 MV Van de Graff

4 - 8 MV linear accelerator

Betatrons, up to 42 MV x-ray and electrons above 20 MeV

II Lesions within few cm of Surface

200 - 400 kV x-rays

medium distance gamma-ray beam units

electron beams, 6 - 20 MeV

III Superficial Lesions (Skin or in Body Cavity)

short distance gamma-ray beam units

electron beams up to 6 MeV

RECOMMENDED: ONE UNIT FOR EACH CATEGORY AS BASIC EQUIPMENT FOR A RADIOTHERAPY DEPARTMENT

Considerations Affecting Choice or Specific Equipment within a Given Category

"Several physical aspects of competitive devices should be compared, including:

(a) the ease and uniformity with which a selected tumour dose can be delivered;

(b) the reliability and ease of maintenance:

(c) the versatility, i.e., ability to provide treatment in more than one of the above categories;

(d) the radiation safety;

(e) the capital cost;

(f) the economy of operation and maintenance."

... page 14

Treatment of Deep-Seated Lesions

"The most important single piece of apparatus to be selected for a new radiotherapy department is the supervoltage unit. Multiple-field or rotational treatment plans, using X-rays ranging from 2 MV to 30 MV or gamma rays from cobalt-60, do not differ remarkably. In all cases, adequate doses of sufficient uniformity are deliverable to the volume containing the lesion. In these circumstances, reliability, ease of maintenance and cost become the decisive factors." ..... page 14

Reliability of Supervoltage Equipment

"Probably the two most reliable types of supervoltage radiation sources are the 2-MV resonant transformer and the cobalt-60 teletherapy units. The former have been known to run for 10 years without being opened and without replacement of the glass accelerator tube. Tube lives of over 10 000 hours have been recorded."

"In the same way, modern cobalt-60 units frequently give continuous service for periods of several years between source replacement."

"It is generally recognized that Van de Graaff machines, linear accelerators and betatrons require the rapid availability of a skilled maintenance engineer or technician. Without such skilled personnel and without preventive maintenance routines, it would hardly be possible to operate these machines and not incur excessive loss of treatment time. Neither the resonant transformers nor the teletherapy units require the immediate availability of specially trained maintenance personnel. It is therefore concluded that reliability dictates the installation of one of these types of equipment in the developing countries."

..... pages 14/15

Economics of Supervoltage Equipment

Comparative estimates of operational cost per 200-rad (2 Gy) tumour dose at 10 cm depth.

ASSUMPTIONS

All units operated under conditions of full use during a treatment day.

Same or comparable source-surface distances.

Constant set-up time per patient.

Note: Comparisons do not include cost of personnel (personnel costs considered constant per treatment given).

"It is therefore recommended on grounds of reliability and economy that the supervoltage unit of choice for a developing country is a large cobalt-60 teletherapy unit." ..... page 15

STAFF; Medical radiotherapist, Radiological physicist, Radiotherapy technician, Physics technicians.

"Unless a fully-qualified radiotherapist and a radiological physicist will be available for staffing a new department, the wisdom of establishing it should be reviewed."

Optimization of Radiotherapy, Report of a WHO Meeting, WHO TRS No. 644 (1980)[3]

EQUIPMENT;

Choice of Radiation Energy for Radiotherapy with Photon Beams

"High-energy electron accelerators such as linacs and betatrons, when the electrons are directed at a metal target, give the radiotherapist a beam of X-rays of higher energy and sharper delineation than may be obtained with cobalt-60 gamma rays. Accelerators have other advantages also, such as higher radiation output and the ability to produce electron beams of specified energy (see section 3.1.2), but they are secondary to the central feature of higher radiation energy coupled with a sharper beam." ..... page 16

"From the available evidence the conclusion is inescapable that an electron accelerator (preferably with electron-beam capability) offers considerable advantages over a cobalt unit for the treatment of large patients and patients with lesions that are "difficult" owing to size, shape, or location. Altogether, these patients may constitute 10-15% of the total. For the remaining 85-90%, the accelerator may offer a slightly better treatment than the cobalt unit but this is in any case offset by problems of maintenance and dosimetry, which are particularly acute in developing countries."

..... page 17

The Role of Electron Accelerators in Developing Countries

"While the Meeting did not consider that high-energy electron accelerators could be generally recommended for use in developing countries, it did not wish to discourage the purchase of such machines where certain conditions are met. They can be justified for a centre of excellence in a country that has already achieved an acceptable standard of radiotherapy covering the majority of the population".

"An accelerator should be regarded as a machine of different capability from that of a cobalt unit - a machine able to treat patients who cannot be treated properly with a cobalt unit."

"It follows that, if an accelerator is to be purchased, it should be a machine capable of delivering 8-10 MV X-rays and an electron beam of variable energy up to at least 20 MeV and perhaps even 45 MeV."

"However, quite apart from the very large capital cost of such a machine and of the room required to house it, serious consideration should be given to the following prerequisites for the successful use of an accelerator."
..... pages 22/23

Prerequisites for the Successful Use of an Accelerator

(1) Expert personnel must be available, including radiotherapists specially trained in high-energy X-ray therapy, radiation physicists, a sufficient number of specially trained medical radiological technicians (more staff are needed than for a cobalt unit), and engineers or electronic technicians.

(2) The infrastructure of the radiotherapy department must be excellent. This includes a reliable electricity supply, not subject to interruptions or severe reductions in voltage, a good water supply at reasonable pressure, and access to a good machine shop and other back-up facilities.

(3) Good communication with, and ready access to, a service agency of the manufacturer is essential. Either the agency or the radiotherapy centre itself must keep a good supply of important spare parts and circuit

boards. Telephone communication with the service depot should be good and not liable to frequent interruption. arrangement should exist for the rapid passage through customs of spare parts that have to be obtained from abroad.

(4) If the hight cost of the accelerator is to be justified, it must be used intensively and efficiently for a large patient load. This implies that the radiotherapy centre must be efficiently organized so as to permit a high throughput of patients. If the treatment room remains unoccupied for even 5 minutes between each patient, the economic advantage of the high radiation output of the accelerator is lost. Furthermore, the "back-up" structure of the department (diagnosis, localization, treatment planning, dosimetry, follow-up, records) must be commensurate with a high throughput in the treatment rooms. These last conditions are not met in many radiotherapy centres, even in industrialized countries, and in such centres accelerators are employed for below their economically optimum potential.

(5) Excellent dosimetry and a daily check of the machine are further prerequisites for operating accelerators. If accelerators are not calibrated and operated with great care they can be extremely dangerous, since patient dosage can be in error by up to an order of magnitude.

Recommended Teletherapy Machines for Developing Countries

"The undoubted advantages of high-energy X-rays and perhaps electrons relative to cobalt-60 gamma rays are offset by the fact that an electron accelerator (linear accelerator, betatron, or microtron) entails considerably higher capital and annual expenditures. In addition, the accelerator is more liable to break down and more difficult to repair and to maintain in good running order. The Meeting therefore felt unable to recommend accelerators for general use in developing countries. It considered that, at least for the next 5-10 years, radiotherapy in these countries should rely principally on cobalt units and only in certain circumstances can accelerators be recommended for countries."

..... page 19

OTHER REQUIREMENTS

Report of Research Coordination Meeting on Testing of Dosimetry Equipment, IAEA, Vienna, 27-29 November 1989 (Internal IAEA Report)[4].

EQUIPMENT

Minimum Requirements to Perform Radiotherapy

Megavoltage units are strongly recommended for curative radiotherapy. When setting up a new facility it is strongly recommended that the first therapy unit be a cobalt unit, isocentric with an SAD of 80 cm. The unit should include a movable collimator, mechanical scales for all the motions, a mechanical distance indicator, and automatic timer and the necessary safety devices.

As additional equipment, the minimum requirement is:

- a set of wedge filters
- a tray with standard shielding blocks
- a convenient couch" ..... page 1

DISCUSSION

After completion of the present report, the Committee had the opportunity to consult the WHO technical report 644 and to compare its suggestions with the recommendations made in 1980.

The present recommendations are in excellent agreement with report 644 and can reinforce the conclusion of 1980. The only difference is that the Committee has, on purpose, considered the very minimum requirements and not the optimum ones."

..... page 3

Organization and Training in Radiotherapy for Developing Countries in Africa, IAEA TECDOC-614 (1991)[5]:

EQUIPMENT;

Recommendations

"The discussants recognized that African countries are in dire need of radiotherapy equipment. While most of the available equipment is sophisticated and expensive, they recommended that as much as possible the design of simple, sturdy, safe and reliable cobalt machines for radiotherapy should be encouraged. Such equipment should be cheap and affordable, while at the same time it will not compromise safety, reliability and efficiency. It is believed that this is feasible if such a machine is devoid of costly and sophisticated electronic and mechanical parts."

..... pages 219/220

"Adequate care must be taken in Africa to ensure optimal suitable power and air conditions wherever any major radiotherapy equipment is to be installed in order to reduce the risk of damage from the harsh atmospheric conditions, like temperature, humidity, dust, etc."

..... page 220

Radiation Dose in Radiotherapy from Prescription to Delivery, IAEA TECDOC-734 (1994)[6]:

Chapter VIII - Conclusions, Recommendations, and Future Work

Summary of Round Table Discussion

"The need for a scientifically sound, robust, reliable, treatment unit capable of high-quality performance with low initial and operating cost was universally recognized. The characteristics of such a unit were considered to be:

(a) Mechanically robust, incorporating only a minimum of electrical or electromechanical features, and with careful selection of components for resistance to deterioration due to high levels of heat and humidity. Mechanical scales must be provided for all motions.

(b) Isocentric mounting with 80 cm source-axis distance.

(c) Automatic timer (electronic).

(d) Adjustable collimator capable of rotation with all movements manually operated, and with a field size of 30 x 30 cm at 80 cm source-axis distance.

(e) Accurate light field and distance indicators (electrical), with mechanical back-up must be incorporated.

(f) The radiation sources must be cobalt-60 and must provide at least 1 Gy/minute at isocenter.

(g) Accessories must include a wedge filter holder and a beam-block holder.

(h) The treatment couch should be mounted on rails and not isocentric. It should be electrically driven with mechanical back-up features, and should be constructed so that it is essentially radiotransparent for treatments with the beam directed upward through it.

(i) Necessary safety features to comply with national and/or international requirements must be incorporated in the design."

... page 383

National Cancer Control Programme, Report of a Working Group on National Cancer Programmes, Geneva, 25-29 November 1991. WHO/CAN 92.1, Limited Distribution, 1992[7]

TREATMENT OF CANCER

"For as long as prevention of cancer cannot be fully achieved, treatment will remain important."

"While the basic principles of treatment are the same in all world regions, the emphasis accorded to treatment will depend upon the local pattern of cancer, i.e. the commonest types of cancer seen, and the relative proportion of early and late stage cancers."

"The primary objectives of cancer treatment are:

Ÿ cure;

Ÿ prolong useful life;

Ÿ improve quality of survival."

"The primary treatment modalities include surgery, radiation therapy, and chemotherapy. It is unlikely that newer approaches to treatment will be applicable to the less developed countries in the immediate future, since they are not only costly, but also, at the present time, of unproven or minor efficacy."

... page 55

"Radiotherapy can be curative for some cancers (eg head and neck, cervix) and provide substantial palliation for others. Treatment policies should be established."

"Relatively inexpensive cobalt therapy machines will be easier to maintain and will provide adequate therapy or palliation for the majority of patients without resorting to expensive and service-demanding linear accelerators or other high energy machines. For the great majority of treatable cancers in developing countries, linear accelerators offer no advantage over cobalt therapy."

... page 56

"Manpower needs should be reviewed. Where possible training should be sought in programmes with patients, training and equipment relevant to the needs of the country ..."

Advisory Group Consultation on the Design Requirements for Megavoltage X-ray Machines for Cancer Treatment in Developing Countries (Washington D.C., 6-10 December 1993), to be published by PAHO[8].

EQUIPMENT

The required machine dimensional and beam performance parameters to meet the needs of developing countries were defined and found to be very similar to those for developed countries. In fact, it was agreed that such a machine must be suitable for use in developed countries.

The major dimensional preferences were: (a) Low isocenter height, not more than 130 cm, 115 cm preferred; (b) 100 cm source-axis distance preferred; 80 cm acceptable; (d) Couch vertical travel 70 cm below isocenter. Rotation + or - 90 degrees; (e) Field size at isocenter at least 30 cm x 30 cm.

An x-ray energy of about 6 MV is preferred. There was a strong preference expressed that if the machine is to be an accelerator, it must provide significantly greater beam penetration than cobalt-60.

WHO Manual of Radiotherapy and Cancer Management (Manual being prepared, expected publication, 1995)[9].

MODALITIES

Medium energy (orthovoltage) x-ray machines with generating potentials in the range of 100-300 kV which originally were used to treat deep-seated tumours are no longer recommended for that purpose, having been replaced

by "megavoltage" machines operating at effective energies equal to or above the energy of cobalt-60 (1.25 MeV). The advantages of the higher-energy radiation which are now universally accepted are:

1. the skin-sparing effect due to the build up of electrons below the surface.

2. the greater power of penetration and hence the increased percentage depth-dose.

3. the decreased scatter sideways from the direction of the beam and consequently sharper delineation of the beam.

4. the smaller increase in specific absorption in bone compared to soft tissue.

ACCELERATORS

Advantages:

1. Sharper beam delineation (small physical penumbra) than cobalt-60).

2. Higher radiation output than cobalt-60.

3. Higher penetration in tissue of the photons than cobalt-60. Comment: 4-6 MV has only moderately higher penetration than cobalt-60. Higher photon energies (over 6 MV) often give some advantages in deep-seated tumours.

4. Greater ease in treating large patients, and lesions that are difficult to treat because of size, shape, or location. Comment: These may be 10% to 15% of all patients treated.

5. Units are available which can produce electron beams. Comment: If purchase is being considered, an electron beam of variable energy up to about 20 MeV covers most applications.

ACCELERATORS

Requirements:

1. Large amount of resources for the initial purchase and for construction of the treatment room.

2. Reliable electricity supply, not subject to voltage reductions or interruptions.

3. Availability of good service from the manufacturer or supplier, including spare parts locally available, and easy access to telephone communication.

4. Excellent radiation dosimetry and quality assurance capabilities in the department.

5. Well trained staff especially qualified to work with high-energy photons and electrons.
Comment: A very sharp beam delineation can only be utilized when the rest of the treatment procedure is very accurate, for example accurate diagnostic outlining of the tumour volume, and careful setting-up and positioning of the patient.

COBALT-60 TELETHERAPY MACHINES

Advantages:

1. Provides high-energy photons (1.25 MeV average).

2. Many years of experience in the use of these units has proven their dependability.

3. For most practical treatment situations adequate dose distributions can be obtained.

4. The initial capital investment is moderate, and routine operating costs are moderate.

5. Installation can be achieved in a relatively short time with moderately skilled workers.

6 Few total staff, and staff with moderate levels of training are sufficient for routine operation.

7. Maintenance and repair are not required frequently and when needed are moderate in cost

COBALT-60 TELETHERAPY MACHINES

Requirements:

1 Adequate premises and shielding are necessary and must be provided.

2 Provision must be made for replacement of the cobalt-60 source at periodic intervals

3. Trained staff in sufficient numbers are necessary.

4. For curative treatment of certain cancers (for example, very deep-seated lesions or lesions near critical tissues), more care in planning is required (as compared to accelerators) because of the less penetrating radiation, and the larger penumbra.

SUMMARY AND CONCLUSION

In view of the global range of situations now existing, most of the recommendations which WHO, PAHO, and IAEA have made concerning radiotherapy equipment and facilities since the 1960s, are still applicable in some place in the world. In 1994 in many places, reliable electrical power (voltage, current, frequency), water supplies (quantity and quality) and environmental modification systems (heat, humidity and dust control) are unavailable, or only available in a few large cities. Additionally the availability of adequate maintenance services, and spare parts and/or the funds to provide for them, is severely limited in many locations. Furthermore, the prerequisites remain unchanged for the successful use of any complex equipment, cobalt-60 or accelerator; expert personnel, excellent infrastructure, good communications, large patient load, excellent dosimetry, efficient organization and supporting structure.

Consequently, taking into account the existing situation and modifications which are feasible and sustainable in view of the economic and social situation, priority should be given to acquiring the kind of equipment which is most likely to be able to function in the local environment (climate, staff, supporting services, operating resources) where it is to be used.

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ALTERNATIVE DESIGNS FOR MEGAVOLTAGE MACHINES FOR CANCER TREATMENT IN DEVELOPING COUNTRIES

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Abstract

In developing countries radiation therapy is often performed with antiquated cobalt-60 units, the radioactive sources of which are long decayed and, thus, treatments are ineffective. Furthermore the cost involved in the disposal of spent radionuclide sources discourages owners from proper removal and storage, and accidents occur. Although present design of microwave electron linear accelerators provide excellent beam characteristics, developing countries in many locations do not have the infrastructure to maintain such machines. To explore the possibilities of designing, taking advantage of the latest advances in technology, a more elementary electrical teletherapy machine, inexpensive in first cost and maintenance, the Pan American Health Organization / Regional Office of the World Health Organization for the Americas organized in Washington an Advisory Group Consultation on the Design Requirements for Megavoltage X-Ray Machines for Cancer Treatment in Developing Countries, with the collaboration of the World Health Organization, the International Atomic Energy Agency and the United Nations Industrial Development Organization. It was attended by 40 radiation oncologists, physicists, technologists and engineers representatives from radiotherapy equipment manufacturers. After an analysis of the radiotherapy situation world-wide - especially from the viewpoint of maintenance - a consensus was reached on the radiotherapy equipment performance requirements. To meet these requirements, several accelerator designs were considered. Among the most promising new designs were the klystron/linac and the high frequency linear accelerator, the microtron in a radiation head, the high frequency betatron -also in a radiation head-, and DC accelerators. Possible treatment designs, including those of modular nature, were presented. Since it is estimated that by the year 2015 - barring a dramatic and unforseen cure for cancer - a total of 10,000 machines will be needed to provide treatment for an estimated 10 million new cases per year in developing countries, the impact of such high technology simple machine could be substantial in providing equity and quality for the management of cancer patients.

1. INTRODUCTION

According to WHO estimates, currently there are approximately nine million new cancer cases per year, worldwide. This number is expected to increase to about 15 million new cases by the year 2015, with about two-thirds of these cases occurring in developing countries.

Radiotherapy will, for years to come, be the most important therapy approach for most of these tumors, both for cure and palliation. Surgery is limited in its role due to the advanced stages of the diseases encountered in developing countries and chemotherapy is expensive.

In developing countries, the typical incidence is 75 to 150 new cancer patients per 100,000 population. To serve a current population of 4.4 billion, assuming 4.4 million new cancer cases per year - 50% requiring radiotherapy - and one machine per 500 new cancer cases treated, the current need is a total of 4,400 machines. By the year 2015, barring a dramatic and unforseen cure for cancer, a total of 10,000 machines will be needed to provide treatment for an estimated 10 million new cancer cases per year in developing countries.

Presently it is estimated that in developing countries approximately 2,300 megavoltage teletherapy units are installed, primarily cobalt-60. Unfortunately many of these units are antiquated, have received very little maintenance over the years and their radioactive sources are long decayed. Due to economical constraints, source replacement intervals may be up to 10 years, especially in private institutions [1]. Data obtained through the 1992 IAEA/WHO postal dosimetry intercomparison program for high energy radiotherapy units, show that more than 50% of the units tested in Latin American and Caribbean countries would require treatment times of over 4 min to deliver 2 Gy to the tumor. To compensate for the low absorbed dose rates at the treatment distance and still treat a very large number of patients, it has become a common practice to shorten the treatment distance -often without correcting the percentage depth dose tables in clinical use - and to give lower doses than necessary. In no case are doses increased to compensate for the low dose rates. The consequences of these practices are not only inaccurate doses being delivered (the 1992 intercomparison showed errors of more than 38%!), but the fact that treatments are ineffective, fostering the concept that cancer is incurable. Thus the health authorities do not assign proper budgets to radiotherapy services.

Furthermore, the cost involved in the disposal of spent radionuclide sources discourages owners from proper removal and storage and accidents like the ones in Ciudad Juarez, Mexico [2], and Goiania, Brazil [3] occur.

The industrialized countries have started the process of replacement of cobalt-60 units and most radiation oncology departments in the United States and in Europe have switched to electron accelerators. However, some of these units are very expensive and difficult to maintain and the infrastructure to properly use them is often lacking in developing countries. Thus, for the purpose of improving the availability of radiation therapy for cancer treatment, manufacturers and research laboratories are being encouraged, taking advantage of the latest advances in technology, to consider the design and development of a megavoltage x-ray machine much simpler than present microwave electron accelerators, a machine that could be used both in developed and developing countries.

To address this issue the Pan American Health Organization/Regional Office of the World Health Organization for the Americas organized in Washington an Advisory Group Consultation on the Design Requirements for Megavoltage X-Ray Machines for Cancer Treatment in Developing Countries, with the collaboration of the World Health Organization, the International Atomic Energy Agency and the United Nations Industrial Development Organization. It was attended by 40 participants: radiation oncologists, physicists, technologists and engineers representatives from radiotherapy equipment manufacturers.

After an analysis of the radiotherapy situation world-wide - especially from the viewpoint of maintenance - the Group reached a consensus on performance requirements and proposed various novel accelerator designs.

2. PERFORMANCE REQUIREMENTS

To reduce complexity and improve safety a single energy photon unit without electrons is recommended. (It is assumed that there is access to at least superficial x ray machines with energies between 100 and 300 kV for the treatment of tumors up to a depth of 3 cm.)

- Treatment time (average 2 fields/patient): 10-15 min/patient
- Patients treated/8 hours day: 32-48
- Dose rate at isocenter (depth of dose maximum): minimum 0.8 Gy/min
  recommended 2-3 Gy/min

2.1 Mechanical data

- Isocentric design recommended
- Isocentric height above floor level ≤130 cm, preferably 115 cm
- Isocentric clearance (with all devices) ≥35 cm
- Source-isocenter-distance ≥80 cm, preferably 100 cm

The floor surface should preferably be flat (no pit). (A small depression is acceptable).
Collimator jaw and distance indication: mechanical or electrical with mechanical backup.

2.2 Couch motions and radiation field size

- Isocentric, rotation is preferred.
- Angle of rotation $\pm 90^{o}$
- Lateral range ± 20 cm
- Vertical range 70 cm below isocenter preferred.
- Field sizes up to 42 X 42 cm^2 at the patient surface of 25 cm thick patient, should be available from above with lowered table (Maximum Field Size at isocenter ≥30x30cm^2)

2.3 Radiation Beam Quality

The beam quality is defined in a parallel opposed[1] beam configuration for a 10 x 10 cm^2 field size and a patient thickness of 25 cm using equal beam weights. In this configuration the following should hold:

- Depth of superficial 90% isodose[2] ≤5 mm
- Hot spot relative a central target[3] = 115%
- Penumbra width < 1 cm, and preferably <8 mm
- Uniformity over 80% of field (IEC) ±3%

It is highly desirable that the hot spot is not greater than 110% relative to a central target. (See Table I for $Dose_{maximum}/Dose_{axis}$ for various beam energies).

[1] About 65% of all radiation therapy is with two opposing fields and there is a strong preference for the higher energy (6 MV) instead of using more than two fields with lower energy (e.g., 2.5 MV equivalent to cobalt-60). (See Table II).

[2] To treat superficial lymph nodes.

[3] To avoid fibrosis.

Table I

D_M/D_A RATIOS FOR VARIOUS BEAM ENERGIES

Machine	Depth(mm)		SSD	Patient Thickness	
				20 cm	25 cm
	D_M	D_{90}	(cm)	D_M/D_A	D_M/D_A
Co-60	5	1.8	80	114%	127%
Co-60	5	2	100	111	123
4 MV	10	4	80	110	120
4 MV	10	4	100	108	117
6 MV	15	7	100	106	112
8 MV	20	9.3	100	104	110
10 MV	25	11	100	102	107

Table II

115% D_M/D_A REQUIREMENT
25 cm thick patient
1:1 Parallel Opposed Fields

Photon Energy

Co-60, 4 MV	No
5 MV	SSD treatments OK
6 MV	SAD treatments OK
8 MV	Better, D_M/D_A = 110%

Electron Beam Current *(3 Gy/min, 100 cm SAD)*

6 MV	100 μA
4 MV	200 μA

2.4 **Devices** - to be available for radiotherapy treatment

- light indication for field size with central axis indication
- distance indication with mechanical backup
- isocentric indication with mechanical backup
- wedges 15°, 30°, 45°, 60°
  light field preferably visible after insertion.
  orientation and wedge angle interlock.
- shadow tray(s) for standard and customized beam blocks.
- possibility to take megavoltage port films

2.5 **Safety**

- Compliance with FAO/IAEA/ILO/OECD-NEA/PAHO/WHO International Basic Safety Standards for Protection against Ionizing Radiation and for the Safety of Radiation Sources, as well as national and local safety regulations.

2.6 **Quality and Maintenance**

- Up time ≥95%
- Service interventions* < 1/month
- Preventative maintenance < 4/year
- Self diagnosis is recommended
- Component potential failure status read-out recommended

* Because clinical treatment cannot continue

2.7 **Serviceability/Reliability Specifications**

2.7.1 *Preventative Maintenance*

There shall be specified intervals for Preventative Maintenance. The integrity of the machine shall not be guaranteed if the manufacturer's schedule is not followed. These specifications consume other maintenance than the regular (daily/weekly) QA checks on the equipment.

2.7.2 *Failures requiring intervention*

These shall be classified according to the level of skill required to rectify the fault. With good education programs, some of these interventions may be handled "in house".

a) First Line - This is a local engineer trained by the manufacturer or the manufacturers local organization as appropriate. Diagnose problem to unit level using standard fault finding practices. In most cases isolate fault to Printed Circuit Board level. Solve 90% of problems.

b) Second Line - Manufacturer's engineers usually at a regional rather than hospital levels. System oriented - able to solve 80% of the remaining 10% of problems, i.e. 98%.

c) Third Line - Manufacturer's engineer of a senior level usually at the Head office.

2.7.3 *Meantime between failures*

Failures capable of first line repair > 3 months
Failures capable of second line repair > 1 year
Failures capable of third line repair > 10 years

2.7.4 *Target Planned Maintenance Schedules*

1 day required every 3 months
3 days required every year

2.7.5 *Spare parts*

A stocking strategy should be defined, e.g.:

First line repairs	on site
Second line repairs	regional
Third line repairs	manufacturer

3. ALTERNATIVE NEW APPROACHES

The various accelerator technologies considered were subjectively ranked A, B, or C depending upon their practicality and likely ability to meet the above requirements.

Ranking	Definition
A	Technology is most likely to meet the requirements and merits further exploration
B	Technology is probably relevant
C	Technology is not likely to meet the requirements, and is not recommended for exploration at this time but should be retained for future reference.

3.1 **Category A**

Microwave linac
Microwave power source - linac
Microtron in radiation head
2 Beam klystron/linac accelerator
Modular design linac

3.2 **Category B**

Betatron in radiation head
Rhodotron
Continuous wave rf linac or microtron
DC Accelerators

Cascade voltage multiplier
Laddertron or Pelletron charged electrostatic accelerator
Nested high voltage generator
Transformer coupled high voltage generator

3.3 Category C

Plasma wave accelerator
Induction linac
Interlaced accelerator structure
Superconductig linac using superconductor or beryllium coated cavities
Small synchrotron
Multiple low power magnetron
Accelerator activated short lived isotope

4. BRIEF DESCRIPTION OF PREFERRED UNITS

4.1 The Klystron/Linac and High Frequency Linear Accelerator

Each of these offered the possibility of a compact 6 MV gantry mounted accelerator. The klystron/linac and the integrated klystron/accelerator waveguide system do not require a particular resonance frequency. Higher frequency accelerators of either the linac or microtron type would be less massive.

4.2 "Mini-Microtron"

This design allows the possibility of a compact (35-40 cm diameter) 6 MV microton to be mounted in the gantry at the top of the radiation head. The small size of the accelerator can be achieved either by use of a high frequency or by a high energy gain per turn. It was considered that an integrated magnetron/cavity design might be advantageous. The microtron layout allows the field flattener to be in the fringing magnetic field thus reducing secondary electron emission. A photon beam spoiler may be employed for control of dose build-up. The accelerator should fit in a 80 cm SAD gantry, possibly even a 100 cm SAD gantry.

4.3 "Mini-Betatron"

In this design a small, 25 cm diameter donut betatron mounted in the gantry at the top of the head provides a compact 6 MV machine. A high frequency (possibly 10 Khz) is required to provide adequate output. A DC bias can be used to double the energy gain.

4.4 DC Accelerators

A relatively compact gantry mounted accelerator can be achieved using DC accelerator principles. This machine was considered to be potentially highly reliable. However, the 2-3 MV energy achievable was generally considered lower than required by most users. Heavy filtration of this beam would allow beam characteristics similar to a conventional 3 or 4 MV accelerator. The adequacy of the beam current obtainable was questioned.

4.5 Classical linacs (rf)

Commercial 4-6 MV linacs of compact design are widely available, though general concerns of cost and reliability were expressed.

5. CONCLUSIONS

The current manufacturers of electron linear accelerators and microtrons should be encouraged to design and prototype a super-reliable 6 MV x-ray system subject to the established performance specifications. This encouragement should come from accelerator designers who might cooperate with the manufacturers as well as representatives of the developing nations who can best make the case for their needs.

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SIMULATION AND RADIATION TREATMENT IN EXTERNAL RADIOTHERAPY

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Abstract

It is well known that in order to obtain a uniform dose in the treated volume as defined in ICRU 50, there should be a 10% maximum difference between maximum and minimum dose values in treatment planning. Clinical target volume (CTV) should be related to external areas of body sections where tumour is located. These areas are important because different radiation beams enter through them. Therefore, verification of the planning target volume (PTV) through the external areas is highly significant.

In this work we point out the importance of controlling that PTV is irradiated as planned considering some error sources usually found in radiotherapy practice with equipment that has been intensively used for a long time. Moreover, I think this experience will be helpful for those centers around the world where radiation treatment is carried out with reconditioned units.

1 PATIENT POSITION WITH RESPECT TO TREATMENT TABLE AND SIMULATOR EQUIPMENT TABLE

As several steps are involved on a simulation table before treatment is started, the differences between simulator table and different treatment tables must be considered in order to avoid incorrect transference to the daily set-up.

The treatment table and the simulator tables are built differently and what is more important with different pre-alignments. They are located in different bunkers, and many times the simulator room is smaller than the treatment one or vice-versa; so the simulator or the treatment table is limited in its movements, generating different use modes. Frequency of use is approximately 10 times higher in the treatment equipment than in the simulator causing deformations in the first one.

In the case of intensively used machines the treatment table is not flat, longitudinally and transversally, due to the fact that patients always use the same side to climb on it, and greater weight is exerted on the area where the table has no support.

An example of treatment table's deformation is observed in large fields such as the mantle field which encompasses almost the entire patient width. In this situation we found it difficult to encompass equally both mastoid epiphysis as well as axillar areas, without modifying patient's position by rotating his/her head one or two degrees, and slightly widening the treatment area, since these areas are included in the PTV.

A further problem is the fact that while many tables rotate 180 degrees and have sections with different materials this is not the case on a simulator or on other treatment tables. For instance, when irradiating a mammary volume, the patient can be set-up on the center of the couch if it has a mylar foil without metallic parts on the borders; otherwise the patient does not rest firmly on the couch and tends to move or rotate.

The weight of the patient's body on the mylar, racquet or on a rigid table as is the case with a simulator, changes the anterior-posterior diameter and the incidence of the lateral beams on the patient.

An example is given by the different contours of the same patient taken at the same section on the treatment equipment, the simulator, the scanner table or on a NMR used to determine the CTV.

If the contour is modified due to the factors just mentioned isodoses values around the PTV will become altered. If the contour does not coincide with the one that corresponds to the CT scanner, a 3 % error in location and/or size of lesion could be produced.

Further differences are due to the fact that treatment tables are usually narrower than simulation ones. Let us analyse the case of a mantle treatment in hyper extension position. Here the patient is able to move his arms. This causes a transversal expansion in the lungs seen in successive films. It sometimes originates an effect by which lung protections coincide with the film in which they were outlined, but not always with the verifying film in the equipment.

In order to treat the mantle (inverted "Y") in the same position as it was treated before with the normal mantle, the couch has to be rotated through an angle of 90° ; as it is usually not perfectly isocentric, the patient has to be repositioned. Besides, the change in arms position and the change in patient position when film cassette is placed under him/her, makes the gap different from that mathematically calculated.

A further problem easily solved is due to the quite common practice of using non-mountable mats to cover couches, and when removing one part of them where the beam impinges on, the flat support of the body is deformed.

In summary, the three main error sources when irradiating mantle fields are: 1) the difference in position of the patient's arms, 2)couch rotation around its isocenter, and 3)placement of the film cassette under the patient to obtain a better image.

2 RELATIVE POSITION BETWEEN THE PATIENT AND ACCESSORIES IN CONTACT WITH HIM

The same accessories must be used on the simulator and on the treatment unit. Pillows must fit the patient's head, neck, and back or shoulders fixing the patient so that uncontrolled movement can be avoided, allowing repeatability. The most common shapes should be available in small, medium or large sizes. For irradiation with electron beams, where surface flatness is very important, these supports are essential.

Immobilises

They are particularly useful for treating head and neck and essential for children or for patients with involuntary movements. Immobilisation can prevent lateral head rotation more easily than its anterior-posterior movement. Lateral head rotation may be controlled by lateral lights and by entrance and exit of radiation beams while these bonds do not register an anterior-posterior movement requiring tattoos or marks in the radiation field vertexes.

Given the fact that in certain cases each unit must handle as many patients as possible, time is saved and doses are accurately administered, if the patient is immobilised in the correct way. These immobilises should be easy to place. They do not only prevent movement but also make each treatment easy to reproduce, avoiding the uncertainty and thus the irradiation of volumes larger than necessary.

Lowering shoulders device

This is useful for lateral fields treatment where fields should not be discontinued taking into account lesion characteristics. Careful attention must be paid to possible deformations of head position and these devices should differ depending on whether they are fastened to the patient or on to the couch. When the lower part of a cervical field runs through a shoulder, the source surface distance (SSD) may vary within 10 cm. or more, and the dose rises up to 70% in the shoulders depending on the type of machine.

Belly flattening device

This should be used for those patients with overlapping adipose tissue due to which the maximum dose on the skin takes place on skin folds and produces skin lesions. The adipose tissue has to be raised in order to avoid its irradiation. It has to be done in such a way that it can be repeated daily, and keeping fixed references when these tissues move. Wide, stiff bandages fastened to the side of the couch with a Styrofoam wedge under patient's buttocks are used. References must be marked in non-movable areas of the patient, possibly not within treatment field, but which clearly define within an admissible margin the daily set-up.

Such a modification must be carried out locating the PTV on a contour taken with this new external configuration.

Support for the arm in breast treatment

The main objective is to achieve a position easy to reproduce daily in order to irradiate the movable mammary volume which depends on arm position, particularly when ganglion areas are to be irradiated. The breast support must be different for the different treatment units, depending on the type of table, the telemeter position and the SSD. At an SSD of 80 cm. there is collision risk, therefore it is convenient to place the lateral support behind the patient's arm. Support must be linked to an angular scale with respect to the longitudinal axis of the table. The main problem arising here is that, when using the same device for the different treatment units, in some cases it hides the distance scale. Notwithstanding the cumbersome procedure it is better to use it, than to place the patient's arm under the neck to irradiate the mammary volume and modify its position to irradiate the ganglion areas, as the PTV is defined for a specific arm position.

The common use of the different accessories makes it convenient to have CT scans taken at treatment position and with the accessories in place.

Bolus

Different bolus types are:

a) wet gauze bolus: skin attachment is good. However, tissue equivalence differs according to water absorbed, and they are difficult to adapt for widths in excess of 1 cm. Width errors plus water absorbed error, may reach to 2% in dose delivered.

b) wax bolus: tend to harden and must be heated to shape them adequately so they stick to skin; they are helpful in shaping sections to eliminate oblique incidence. Tissue equivalence must be calculated.

c) tissue equivalent bolus: they are the best but in some cases there are areas in which intermediate air cavities are difficult to avoid, due to lack of adherence.

Width, shape and type of bolus should be taken into consideration when carrying out planning.

3 RELATIVE POSITION BETWEEN PATIENT AND EXTERNAL ACCESSORIES

Some other widely utilised accessories will be analysed in terms of how their variations and/or deficiencies affect or alter treatment planning as well as daily dose.

Wedges

In some machines they may become loose with respect to their support and displaced even though they are well hooked up. Should wedge with its nominal maximum field size be used, in some cases there may appear a slit in which the dose value is much higher, thus altering dose uniformity. In addition, technicians must be required to place wedges last, once the gantry has been rotated to the incidence angle, and the collimator angle and depth have been controlled.

Protections

Routine protections must be fixed by means of threads and screws. The wearing out of the threads may generate displacement of the protections due to their weight when the equipment is rotated. They should have no rim which might deform shadows. The Pantograph should be checked mechanically at regular intervals. Patients' protections must be controlled on the basis of the X-Ray film on which they were built, and verified by x-raying. The simulator must have trays at the same distance of each unit and similar shields made of wood or Styrofoam.

X-ray film

When the cassette is placed under the patient, his/her position may become deformed. For example, in the mantle field it is convenient to place acrylic plates of the same depth as the cassette under the regions where the cassette does not reach; so the patient is at the same hyper extension position as before. In oblique field it is convenient to utilise a lectern with angles allowing the placement of the X-Ray film perpendicularly to the beam.

4 RELATIVE POSITION OF HEAD, ARMS AND LEGS

To fix a patient's body in space the number of degrees of freedom should be considered, and the bond condition should be fixed. Usually the different immobilisation modes for head and neck take into consideration a transversal axis to the couch. To control rotation around a longitudinal axis is more complex and can be done taking the SSD at two fixed points as references. It is important to position the patient in the simplest possible ways, trying to move couch, gantry and collimator instead of moving the patient especially when using electron applicators.

Treatment position with arms above head is frequently used in breast and oesophagus irradiation but must be avoided unless absolutely necessary. This position promotes patient movement, especially rotation and is less comfortable. When irradiating the ganglion areas and the mammary volume in the case of the breast, a new error is added when fields are overlapped or disjoined, if the arm is at waist level for the ganglion field and under the head for tangential fields. In such cases it is more difficult to transfer data from the simulator to the treatment unit and to reproduce daily treatment accurately. It is very important to register patient positioning with a photograph, controlling tattoos in different planes and the SSD in different points of the field, where skin surface is not perpendicular to the beam axis

5 POOR NUMBER OF TATTOOS AND THEIR MOBILITY

Parallel, opposite, and lateral or slanting fields rotating around an axis are erroneously defined by only one point at an angle of 0° If this is the initial position and the patient has no tattoos in field entrance and in case the SSD is not clearly registered in the patient's treatment report sheet, it is very difficult for the technician to know whether the treatment field is well located. This method saves time for each patient, and is usual when there is only one technician per team Coincidence of tattoos, depths, protection and wedges cannot be verified, This way of working is the simplest one but it leads to mistakes; the appropriate way is to train the staff in order to improve their skills for such a job Otherwise conflicts tend to arise between physicists and technical staff, usually closely involved in these issues. Tattoos should be registered in photos and in field films and be permanently available for each patient

6 THE MOBILITY OF THE ORGANS INVOLVED IN THE TREATMENT

The following are the most important causes which make PTV considerably higher than CTV.

a) the movement of the diaphragm muscles when diaphragmatic couple must be irradiated in ovary treatment

b) lung movement due to pleuro-pulmonary synchrony.

The following organs may fall into the irradiated volume due to different movements:

a) kidney position according to patient's positioning, for example when they must be protected in ovary treatment and are marked ecographically with patient standing.

b) bone marrow in head and neck treatment when there are neck movements.

These last two cases may modify target volume position with respect to treatment portal volume and therefore, should be taken into consideration in the original planning.

7 CONCLUSION

Given the type and amount of some of the possible causes of a difference between dose delivered and dose prescribed, an adequate and close relationship among operators, physicists, and radiotherapy physicians is crucial Regular staff meetings for patient control in treatment rooms, as well as regular workshops, are highly recommendable

ANALYSIS OF VARIATIONS IN THE DOSE DELIVERED IN RADIATION THERAPY

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Abstract

The outcome of radiotherapy in cancer care is heavily dependent on the quality of the treatment. This work presents a review of how daily practice and the current availability of equipment for treatment planning and simulation as well as a number of other factors affect the radiation therapy quality in Argentina. The establishment of refreshing courses for all type of staff involved in the treatments, modernisation of equipment and strict routines in patient set up and quality control would give a significant contribution to a higher quality in radiation therapy.

1 INTRODUCTION

In Argentina radiation therapy is mostly accomplished with Co-60 machines and linear accelerators. The number of linear accelerators is increasing very fast; the first one was a 6 Mv installed in 1979. Now there are 23 linacs, 8 of them have been installed between 1993 and 1994. Half of them have 6 Mv X-Ray beams; the others have X-Ray beams of 10 Mv and 15 Mv and electron beams with energies ranging from 4 Mev to 18 Mev. For the end of this year 3 new linacs will be operating. On the other hand, the number of simulators had not increased in the same way: the first one was installed in 1978 in a center with a Co-60 machine as the high energy unit; now there are 11 simulators all over the country.

Medical centers operating with linacs, have at least one medical radiotherapy physicist working in it. Exceptionally centers take on more than one physicist, even when they have several treatment machines. Sometimes physics technicians (or dosimetrists) are taken on in Radiotherapy centers which only have Co-60 units. But many centers having a Co-60 unit as the main treatment machine are operating with no physics staff at all till now. Many of the variations in the dose delivered, analyzed in this presentation, have nothing to do with type of machine but with the number of patients per hospital and with the staff. So, the first and essential task is to get a staff including radiotherapists, physicists, dosimetrists and technicians (machine operators) working close together for the accurate performance of any Radiotherapy Center.

2 ANALYSIS OF DIFFERENT FACTORS CAUSING DOSE DELIVERED DIFFERENT FROM DOSE PRESCRIBED

In this presentation the starting point is the assumption that routinary mechanical and dosimetric checks have already been accomplished and results are within tolerance values. Also that treatment planning for each patient have been already checked. So, the analysis refers to daily errors which can be named "daily accidents". These daily accidents are more difficcult to record and to solve than, for instance, calibration uncertainties where figures like 2 or 3% can be established

3 SIMULATION

Due to the poor number of simulators operating at present, very few patients are really simulated. If the center is operating without simulators, the treatment machine is used as a simulator for some selected patients. It has the advantage of having identical conditions for simulation and treatment (e.g. the treatment table), but the disadvantage of poor definition, particularly in the case of Co-60, and few number of patients selected for simulation. On the other hand simulators are not always properly used: a) fields and references localized with the simulator are not always verified on the treatment unit, giving rise to different errors in field size, patient position, etc. and causing changes in the volume encompassed by the isodose curve already selected by the radiotherapist (treated volume) [1]; b) usually it is the radiotherapist who defines field sizes. And he only takes into account what he "sees" from localization X-Ray films and from a radioscopic view on the simulator; it is to say the Clinical Target Volume (CTV) [1] without considering that the treated volume may be different for different machines and different treatment techniques. In the very frequent and simple case of parallel opposed fields, when varying irradiation conditions, the CTV may become underdosed near the border [2]. This dose gradient within the CTV varies between 5% and 10%, depending on type of machine, separation distance between fields, SSD and field size. When a linac stops working and has to be repaired, a great number of patients go on with their treatment in another machine, most of the times different from the first one. Field sizes are not changed and dose distribution will be different. It is important to emphasize that there are centers having only one 6 Mv linac and one old Co-60 unit which operates for instance at an SSD of 60 cm. Usually with an SSD of 60 cm,the isocenter technique is not used, and so field sizeis defined on surface. With the same field size for linac and Co-60, dose gradient may reach to 12% within the CTV

4 TATTOOS AND REFERENCE POINTS

Very often the radiation field is set-up on the patient having only one tattoo, usually in the center of the field. If the technician has not many indications for patient positioning, (as it often occurs) the patient may happen to be rotated with respect to the right position; the effect is the same as having the collimator rotated a few degrees; if it happens many times along treatment, the result is a widening of the penumbra and a different treated volume, giving additional dose gradient to the PTV.

5 REPEATABILITY

In order to assure the repeatability of a treatment schedule, technician needs to have clear reference points and enough number of tattoos as well as SSD or depths clearly indicated. When using isocenter treatment techniques, with lateral or oblique incidence, it is very dangerous (but it happens) to have only one tattoo which is the reference when the gantry is at 0°.Usually technician does not verify SSD or depth and positioning for the oblique fields. Again CTV may not be receiving the right dose and the irradiated volume may be greater than it is necessary. There are a number of different "daily accidents" which have to do with the great number of patients per treatment unit. Some of them are; **1)** Error in 0° gantry positioning due to parallax; an error of 1.5° in the gantry angle gives a lateral field displacement of 2mm. This displacement happens always in the same direction as the technician looks the 0° gantry angle always from the same side. **2)** to leave the block tray for patients who don't need it; if calculations for these patients have been done without tray attenuation factor, the error may be as much as 6% if the tray is an entire one. **3)** error in SSD when it is read after the entire block tray is put in place; light scale is altered when tray is placed between focus and patient, and a difference of 5 or 6 mm can happen.

6 VERIFICATIONS

Agreement between treated volume and PTV must be checked with X-Ray films during treatment, as well as very simple changes like variations in patient diameter; a difference of 1cm in tumor depth results in an error of 6% in the dose delivered for a 10cm x 10cm field and a Co-60 unit [3]. When there is only one medical physicist doing all the treatment planning and calibration, verifications are scarcely carried out and even important accidents can happen.

The following case is an example of this:
A treatment plan with a fixed field and a wedge of 30° used as a compemsator was decided for a patient; and a series of mistakes were committed:
1) The physicist chose a wedge shorter than needed.
2) He/she forgot to indicate the collimator angle and so the technician put the wedge in place with the angle in the opposite direction.
3) As the physicist had a lot of work to do he/she had no time to verify field set-up on patient.
4)Tthe technician realized that the wedge was very short and decided to change it; he looked for another wedge longer than the first one, but he didn't take into account the wedge angle and he put a longer wedge but of 45° instead of 30° without notifying the physicist about the change.
Half of the dose was delivered to the patient in the wrong way

7 FRACTIONATION

Usually treatments are planned for 5 fractions a week; but in some circumstances patients miss some sessions.
For instance, a) centers with only one treatment machine are not able to send all the patients for treatment to another hospital when machine comes out of service; b) people with economic difficculties are sometimes in trouble to attend hospital everyday, in particular at the end of the month.
Therefore, in these cases effective dose delivered to the patient is different from dose prescribed.

8 EDUCATION AND TRAINING

In Argentina there are regulations for operating a Radiotherapy Center. The more important are:
1) Physicists (with a degree in Physics or Engineering) and physicians need to get a license in order to be registered as radiotherapy medical physicists and radiotherapists respectively. Only those who have the license can be in charge of a Center. Both have to pass the Course "Dosimetry in Radiotherapy" carried out by the Atomic Energy Office (CNEA); physicists have to pass in addition the Course "Physics in Radiotherapy" also carried out by CNEA. after passing the courses, radiotherapists must fulfill a training period of 3 years and physicists of 1 year, carried out in Radiotherapy Hospitals.
The requirements for being a dosimetrist are: a) to have a secondary ordinary degree; b) to pass the Course "Dosimetry in Radiotherapy" and c) to get practice in dosimetry, in radiotherapy hospitals, under the supervision of a medical physicist during 1 year.
2) A radiotherapy hospital where a linac is operating has to include in the staff a medical physicist having the corresponding license.
3) This license is renewed each 5 years in order to be sure that they have continuity in radiotherapy performance.
4) Each hospital has to take part in postal TLD intercomparisons. There are 4 TLD intercomparisons a year so that each center takes part in, at least, one intercomparison per year.

Intercomparisons are carried out by the SSDL (CNEA) in an anonymous way; the results are reported to each center with a code number.
5) To install a new radiotherapy center, requirements for equipment and staff have to be fulfilled: A) minimum equipment include: a) treatment machines with high energy photon beams (Co-60 units with an isocenter distance of, at least, 80 cm and/or linacs of, at least, 4 Mv X-Ray beam); b) an X-ray treatment machine operating at 50-90 Kv or a linac with electron beams having energies between 6 and 8 Mev; c) afterloading intracavitary applicators and sources for brachytherapy; d) a simulator machine or a diagnostic X-ray machine adapted for localization.
B) a licensed medical physicist or, at least, a dosimetrist supervised by a physicist, must be part of the staff for new radiotherapy centers where a Co-60 unit is the main machine.
So, the minimum staff for new radiotherapy centers having Co-60 units or linacs as main machines include radiotherapists, medical physicists and/or dosimetrists and technicians with their corresponding licenses.
Unfortunately there are not, till now, regular radiotherapy courses for the education of the technicians who operate the treatment machines; there are courses intended for those who want to be diagnostic radiologist technicians; some radiotherapy elementary notions and practice are imparted along these courses, but they are not enough. Refreshment courses partially solve this situation but they are not regular and not all the technicians attend these courses.

9 CONCLUSION

It is clear that each radiotherapy hospital or center has to put in practice a quality assurance program. It is clear also that regular courses are necessary but not enough to assure quality in daily practice. Regular meetings among all the members of radiotherapy staff are very helpful to coordinate treatment schedules, to analyze different errors and mistakes; it is important to insist on the need of precision in radiotherapy and how it can change treatment results; verification films must be analyzed also with the technicians in order they can realize the different errors and changes which happen in daily practice. And it is important to emphasize that every indications has to be written and recorded on each patient treatment sheet
Partial replacement of old treatment units is coming on now in Argentina. But new equipment are in many cases not new but repaired machines and they look as new machines and are very often out of service.The other important point is that the great number of patients per machine and the few number of physicists and technicians, conspires against quality. When technicians are accostumed to work very fast, to be only one for each unit and to reduce time for the set-up, it is very difficcult to reverse the habits. So, it is a good thing to remind that we are working with patients to whom high doses are delivered; when a wrong dose is administered to a wrong volume, little can be done after.

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III (a). INTERCOMPARISON

THE ROLE OF SSDL-HELSINKI FOR DOSIMETRY AND QUALITY AUDIT IN RADIOTHERAPY

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Abstract

Quality and dosimetry audit in radiotherapy has in Finland been implemented through inspections carried out by the Finnish Centre for Radiation and Nuclear Safety (STUK). In connection with the Radiation Metrology Laboratory of the Centre, the SSDL-Helsinki, there is a section for radiotherapy supervision. The inspection by STUK is an independent review of the quality and dosimetry control system which can be called quality and dosimetry audit by site visits. STUK is the responsible authority for the supervision of all use of radiation in Finland and that is why it also can set up requirements on the basis of results of the review. The disagreement of the measuring results between STUK and the radiotherapy department, of more than a given action level, will always lead to a through investigation of the reason and to a discussion on the most reliable results to be used for the treatments. The inspections include dose calibration for conventional X-ray therapy equipment and dose comparison, including field size dependence, for high energy equipment. For afterloading equipment the reference air kerma rate is checked. Additionally, the inspections by STUK include checks of the performance characteristics of the equipment and the accomplishment and the results of quality control procedures. Further, methods are currently being developed to supplement the direct measurements by TL-measurements in special phantoms in order to include the whole treatment chain (e.g. the treatment planning system) in the audit.

1. INTRODUCTION

Quality audit is a review of the quality control system and dosimetry audit is a review of absorbed dose determination, performed by an independent person or body which is not responsible for the performance of the product or process under review. The audit should, in the final stage of its implementation, cover all steps of the radiotherapy procedure, i.e. the final aim should be to ensure that the prescribed dose is given to the patient.

The quality and dosimetry audit can be established by three different ways: (1) postal measurements (i.e. by mailed dosimeters) (2) site visits (3) combination of postal measurements and site visits. Audit by site visits is the most efficient method, since all interesting parameters can be easily checked and the necessary actions can immediately be started. During a site visit all aspects of the work can be discussed with local personnel and a comprehensive review of the overall accuracy and possible problems can be attainded. However, site visits to each clinic of the area serviced by the "audit centre" may become relatively expensive, especially in bigger countries, and a suitable combination of postal measurements and site visits can be more appropriate.

Quality and dosimetry audit in radiotherapy has in Finland been implemented through the inspections carried out by the Finnish Centre for Radiation and Nuclear Safety (STUK). For practical reasons, the quality audit operations have been connected with the standard dosimetry activities, the SSDL-Helsinki. In fact, the both activities form together a special section or "the laboratory for radiation metrology" at the Department of Radiation Safety of STUK. The combination ensures, among many other things, the traceability of measurements and the maintenance of high dosimetric competence.

While the inspection by STUK can be regarded as an independent review of the quality control system of the radiotherapy department (quality audit by site visit), it differs from the general principles of quality audit in two aspects: (1) STUK can set up requirements on the basis of results of the review, and (2) the disagreement of the measuring results between STUK and the department, of more than a given action level, will always lead to a through investigation of the reason and to the discussion on the most reliable results to be used for the treatments.

2. THE DEVELOPMENT OF AUDITING ACTIVITIES IN FINLAND

The Finnish Centre for Radiation and Nuclear Safety is responsible for the supervision of all use of radiation in Finland. Because of the Finnish legislation a licence issued by STUK is required for use of equipment used in radiation therapy. A through inspection by STUK is carried out before any radiotherapy machine is taken into use, and then regularly (every two or five years) or whenever a significant repair has been made or a significant change in the output has been observed.

The dosimetric competence and activities were concentrated at STUK in the 70's mainly because of the legislation, but also due to the pressure on the hospitals side. In the 70's STUK offered dose distribution measurement services on order of hospitals and a computerized radiation beam scanner was developed at STUK.

In 1977 STUK was nominated as a member in the IAEA/WHO Network for Secondary Standard Dosimetry laboratories.

At the end of 70's the old ^{60}Co-units and betatrons were replaced by linear accelerators and Quality Assurance in radiotherapy became actual. The overall accuracy on the target dose of $\pm$ 5 % was considered important in the international litterature [1,2]. Some international protocols contained detailed requirements on QA. In 1980 the recommendations by the Nordic Association of Clinical Physics (NACP) were published [3]. In 1981 a "radiation dose committee" gave its report, where it proposed a detailed program for acquiring new radiotherapy equipment within the next 10 years, and stated the need for QA according to the principles by the NACP.

STUK came to conclusion that the central dose measurement system must be supplemented by the increased responsibility of the therapy clinics for dose measurements and QA of treatment units. In the beginning of the 80's the hospitals started their own dosimetric measurements.

The hospitals suffered from a lack of adequate personnel resources to undertake the increased duties on QA. In 1985 a meeting of STUK and the representatives of each therapy clinic was organized to discuss both organizational and technical questions of QA. In the meeting it was agreed that the QA programs should cover all procedures which affect the accuracy of dose to the patient.

In 1987 the Radiotherapy section was founded in the Department of Inspection and Metrology (today: the Department of Radiation Safety) to be responsible for the supervision of radiotherapy in Finland.

3. SUPERVISION BY STUK

In 1991 Finland got a new radiation legislation. The new Radiation Act stipulates that a regular control of the performance of each piece of radiotherapy equipment has to be arranged. Under the Radiation Act STUK has issued instructions concerning quality assurance for radiotherapy equipment (ST-Guide 2.1) [4]. In the following the main principles of the established practice are presented.

3.1. Safety license and advance control

A special safety license is needed for radiotherapy, for the use of each therapy equipment, according to Radiation Act. The list of equipment which are covered by the license is given in an appendix to the license. A piece of equipment may not be taken into use before an inspection by STUK has been carried out, unless otherwise specified in the license. To take into use a new equipment or to remove an old one, a modification of the license has to be applied for.

A condition of the license requires that the clinic has to specify the organization for safety inside the clinic. In particular, the persons responsible for dose measurements, for the quality control of radiotherapy equipment, and for radiation safety arrangements have to be specified.

The safety license and its modifications are granted by STUK on written application by the clinic. On request, STUK will give an advance statement on the radiation shielding plan for radiotherapy rooms, and if needed, carries out an advance inspection for radiation shielding.

Information on radiotherapy clinics and equipment is entered into a computer register maintained by STUK. The important inspection data, machine faults, radiation accidents and abnormal incidents are also recorded in the register.

3.2. Inspections

A new piece of radiotherapy equipment, the rooms and the compliance of the operation with the license are inspected by STUK before the equipment is taken into use. The inspection is carried out also after significant repairs of the equipment and if its location in the clinic has been changed. Thereafter the inspections are repeated regularly as follows:

At the minimum every two years:

- high-energy and conventional X-ray therapy equipment
- radiotherapy simulators

At the minimum every five years:

- afterloading equipment

The information received from the clinic may give rise to an extra inspection.

Table 1. Contents of the Inspection for Different Radiotherapy Equipment

Object of inspection	High energy treatment equipment	Conventional X-ray therapy equipment*	Afterloading equipment*	Radiotherapy Simulator
Compliance with the license	A, R	A, R	A, R	A, R
Structural radiation shielding (Guides SS 2.8, 2.9 and 2.10, ST Guide 3.6 when applicable for simulators)	A	A	A	A
Radiation safety arrangements (Guides SS 2.8, 2.9 and 2.10, and applicaple parts of machine standards)	A, R	A, R	A, R	A, R
Dose calibration (see clause 3.5).	A, R	A, R	A, R	
Radiation beam characteristics - uniformity, symmetry and penumbra - depth dose characteristics	A, R	A, R	A, R	
Dose monitoring characteristics - repeatability and proportionality - dependence on gantry and collimator angles	A, R	A, R	A, R	
Mechnical characteristics - radiation field indicators - angle, field size and distance indicators - isocentricity - treatment or imaging table movements - laser beams	A, R	A, R		A,R
Imaging and fluoroscopic characteristics (ST Guides 3.3 and 3.4 when applicable)				A, R**
Results of quality control (ST Guides 3.3 and 3.4 when applicable concerning simulators)	R	R	R	R
Dose planning system	A, R	A	A, R	

* when applicable
** when needed

A = first inspection before taking into use
R = regular inspection

APPENDIX A

Suggested Details of the Inspection for High-Energy Treatment Equipment.
Gantry angle (GA) and collimator angle (CA) are 0° unless otherwise stated. Only values of parameters or machine settings which will be used are considered.

Object of inspection	A. First inspection before taking into use		B. Regular inspection	
	1. Total procedure	2. Measurements by STUK	1. Total procedure	2. Measurements by STUK
Dose calibration	All radiation qualities, dose rates, wedges and field sizes.	All radiation qualities, dose rates and wedges. Selected field sizes.	As A2.	As A2.
Radiation beam characteristics Uniformity Symmetry Penumbra	All radiation qualities. Field sizes 10 cm x 10 cm, ref. size, max and one field size between the two last ones. Minumum: GA 0° and 90°. Minimum: CA 0° and 90°, when GA 90°. Measuring depths: ref. depth (all GA and CA) and another depth (only GA 0°, CA 0°).	Minimum: one setting for each radiation quality (i.e. field size, GA, CA, measuring depth).	All radiation qualities. Minimum: field sizes ref. and max. Minimum: GA 0° and 90°. Minimum: CA 0° and 90°. Minimum: one measuring depth.	Minimum: one photon and one electron quality. For each quality, minimum: one setting (field size, GA, CA, measuring depth).
Depth dose characteristics	All radiation qualities, wedges, field sizes. For one electron energy, minimum: two values of GA.	All radiation qualities, ref. field size.	All radiation qualities, selected field sizes.	As A2.

APPENDIX A (cont.)

Object of inspection	A. First inspection before taking into use		B. Regular inspection	
	1. Total procedure	2. Measurements by STUK	1. Total procedure	2. Measurements by STUK
		As A1.		
Dose monitoring charasteristics				
Repeatability	All radiation qualities.	Minimum: one photon and one electron energy. Minimum: One setting (GA, CA).	Minimum: one photon and one electron energy. Minimum: four field sizes for photons and all field applicators for electrons.	
Proportionality	Minimum: one photon and one electron energy.			
Dependence on gantry and collimator angles	All radiation qualities. Minimum: GA 0°, 90°, 270°. Minimum: CA 0° and 90°, when GA 90° or 270°.	Minimum: one photon and one electron energy. Minimum: two field sizes for photons and one field size for electrons.	As A2.	
Mechanical characteristics			As A1.	
Accuracy of radiation field indicators	All radiation qualities. Minimum: four field sizes for photons and all field applicators for electrons.	As A2.	As A2.	
		As A1.		
	All photon energies. Minimum: two electron ener-	As A1.	Minimum: one photon energy. Minimum: two	

APPENDIX A (cont.)

Object of inspection	A. First inspection before taking into use		B. Regular inspection	
	1. Total procedure	2. Measurements by STUK	1. Total procedure	2. Measurements by STUK
Angle, field size and distance indicators	indicator further at field sizes 10 cm x 10 cm and max, when GA 90° and CA 0°, 90°, 180° and 270°. Distance indicator further, when GA 90°.	Angle indicators, minimum: one angle. Field size indicator, minimum: two field sizes. Distance indicator, minimum: one distance.		As A2.
			As A1.	As A2.
	Location, size and consistency of mechanical and radiation isocentres.			As A1.
Isocentricity	Isocentricity. All movements and their indicators.	Depend on the case.	As A1.	
	All lasers.			
Treatment or imaging table movements		Selected movements.	As A2.	
Laser beams		As A1.		
Minimum: angle indicators at 90° intervals, field size indicator at 5 cm intervals, and distance indicator at three distances. Field size		As A1.		

Table 1 and Appendix A specify the contents of the inspections in details. For the inspections STUK maintains high quality equipment for absolute and relative dose measurements. A protocol on each inspection with the results of measurements is given to the clinic.

3.3. Quality control

A quality control program has to be submitted to STUK for approval within one year from the date when a new piece of equipment was taken into use. International recommendations are applied for the criteria of approval [3,5,6,7]. However, the quality control measurements have to be started from the beginning of the use of the equipment, as specified in connection with the first inspection by STUK. The quality control program of a high-energy equipment shall always include dose calibration of the equipment. The accomplishment of quality control procedures and the results are inspected by STUK during the regular inspections of the equipment.

3.4 Dose calibration of treatment equipment

3.4.1 Conventional X-ray units

The dose calibration of a conventional X-ray therapy unit is defined as the determination of absorbed dose rate produced by the equipment. "Absorbed dose" for this case is defined as the absorbed dose to water at the surface of water phantom at the depth of 0.5 mm.

The dose calibrations of conventional X-ray units are carried out by STUK. The dose rate is measured for one tubus (deep therapy) or separately to each tubus (superficial therapy). The results of the dose calibration are given in a Dose Table, where the measured or calculated dose rate is given for each tubus or as a function of field size.

3.4.2 High-energy treatment equipment

For radiotherapy electron accelerators, the dose calibration is defined as the determination of the relation between the dose produced and the setting of monitor units. For gamma beam therapy equipment, the dose calibration is defined as the determination of the dose rate. "Dose" is here defined as the absorbed dose to water on the radiation beam axis, at the depth of dose maximum in a water phantom, when the surface of the phantom is at the normal treatment distance. For the dose calibration, the above relation is determined for open field as well as for each wedge field, using the reference field size. In addition, the dependence of this relation on the field size is determined or checked, for both open and wedge fields. The check of the uniformity of the field shall always preceed the dose calibration.

The measurements for the dose calibration are carried out by STUK in connection with each inspection of the equipment (see Table 1 and Appendix A). The dependence on the field size, which the clinic has to determine before taking into use the equipment, is then checked at selected values of field size. The results of measurements at the reference field size are compared with the corresponding results obtained by the clinic at the time of the inspection. In the comparison, the following action levels are applied:

- consistency of dose in open field at the reference field size
  - photons 1 %
  - electrons 2 %
- consistency of wedge factor 1 %
- consistency of field size dependence 1 %

If the difference of results does not exceed the action level, the results by the clinic are used to prepare a Dose Table. In a Dose Table, the monitor unit setting which produces the dose of 1 Gy at the given depth in water, is given as a function of field size. The Dose Table is the ultimate result of the dose calibration. If the action level is exceeded, the reason will be examined, and the most reliable result of measurement is taken as the basis of the Dose Table.

3.4.3. Afterloading equipment

The dose calibration of afterloading equipment is here defined as checking of the reference air kerma rate. This check is made by STUK regularly.

3.5. Requirements for giving information

The Radiation Degree states a number of matters on which the clinic is responsible for informing STUK. All changes of information included in the license, considerable changes in the intended use of the treatment equipment or in the approved quality control programs, important machine faults and repairs, and radiation accidents or incidents affecting the radiation safety shall be informed to STUK. STUK must also be informed, if the result of dose calibration for a radiotherapy electron accelerator (dose per monitor unit), taking into account all possible adjustments of the calibration, differs by more than 5 % from the value agreed on during the latest inspection by STUK.

3.6. Calibration of radiotherapy dosemeters

STUK maintains a calibration service for radiotherapy dosemeters. The dosemeters for dose calibration of radiotherapy equipment shall be recalibrated every three years. Calibrations are performed against the Finnish national standard, which is a secondary standard and calibrated by the International Bureau of Weights and Measures (BIPM). The results of calibration are given in a Certificate of calibration, where the calibration factor for absorbed dose to water is tabulated as a function of radiation energy.

3.7. Supporting activities

Supervisory activities of STUK are backed by training activities and continuous development of methods and procedures. STUK gives training and advice on Quality Assurance for radiotherapy equipment, both through direct contacts with individuals and through special occasions. To maintain the high expertism required for its work, STUK undertakes also research on dosimetry and on other aspects which affect the overall accuracy of the dose to the patient. STUK maintains contacts with national and international organization dealing with the same subject area, and participates in national and international cooperation. The knowledge and information gained STUK transmits further to the clinics.

4. RESOURCES AND FUTURE DEVELOPMENTS

The above quality and dosimetry audit system concerns about 25 accelerators, 10 afterloading units, 11 simulators and 11 conventional X-ray units in 9 radiotherapy centres in Finland. About 2 physicist manyears is required to operate the system. The cost of the system is approximately US $ 250 000 per year, most of which is due to the salaries. The operation is mainly financed by annual charge for the hospitals.

The dose planning systems have not been properly included in the current quality audit procedures by STUK. A project has been undertaken to develope a special phantom, which could be used to check the whole chain of radiotherapy: from CT-scanning to calculation of the dose distribution in target volume. The tests with the phantom would then supplement the other measurements for quality and dosimetry audit.

The benefit of the quality and dosimetry audit is quite evident based on the experience through the site visits from several years. The discrepancies observed include lot of cases, where the observation would not have been possible or would have been less probably if only postal procedures had been applied.

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SSDL ARGENTINA: DOSIMETRIC INTERCOMPARISON PROGRAMME FOR COBALT 60 THERAPY UNITS

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Abstract

Thermoluminescence dosimeters (TLD) are widely used to verify absorbed dose delivered from radiation therapy beams. The Secondary Standard Dosimetry Laboratory (SSDL) of Argentina uses TLD for its mailed dose intercomparison programme for cobalt 60 radiation therapy units. Results obtained since 1978 as well as causes of dose discrepancies greater than 5% are analyzed. Results of the external quality control performed by the IAEA for this programme indicate that the dose evaluated by the SSDL TLD service for the participating centers is about 1% lower than that evaluated by the IAEA TLD service. This deviation is accepted taking on account that a ± 2% dose uncertainty for TLD dosimetry is reasonable.

1. INTRODUCTION.

The Regional Reference Center (RRC) for Dosimetry of Argentina is a Secondary Standard Dosimetry Laboratory (SSDL) belonging to the IAEA-WHO SSDLs Network. As similar laboratories existing in other countries, the RRC was established in 1968 due to the necessity of improving the dosimetry and treatment plannings in the radiation therapy centers and to increase the participation of physicists, specially trained in this field, within the staff of these centers.

The most relevant activities performed by the RRC in the field of radiation therapy are: dosimeter calibration service, calibration programme for cobalt-therapy units in the country, advisory about clinical dosimetry for radiation therapy centers, organization of post-grade courses for physicians and for physicists specialized in radiation therapy, and a dosimetric intercomparison programme for cobalt 60 therapy units.

In 1977, through the IAEA Research Contract RC 1791/RB, the RRC started to develope a national postal dose intercomparison programme or cobalt-therapy units using TL-dosimeters. The first dosimetric intercomparison took place in 1978. Since then, about 4 dosimetric intercomparisons per year have been made, including 88 cobalt 60 and 8 caesium 137 therapy units belonging to public and private centers.

The RRC has participated in the IAEA dosimetric postal intercomparison programme for SSDLs for cobalt 60 beams, and more recently in a similar programme for high energy X-ray beams with deviations lower than 1%.

In 1992 the RRC participated in the Coherent and Accurate Reference Instrument (CARE) Programme for the IAEA/WHO Network of SSDLs. During 1992 and 1993 the RRC participated in the IAEA Quality Control Programme for the RRC TLD postal intercomparison service in order to test this intercomparison system. From these results it is possible to evaluate the present situation and trends of the SSDL of Argentina with regard to its postal dosimetric intercomparison programme.

2. METHOD

The method and technique employed by the RRC for the postal dose intercomparison programme were described in previous papers [1], [2], [3]. Briefly, the dosimeters consist of LiF powder contained in plastic capsules. A batch of capsules containing annealed powder TLD-700 is prepared by the RRC and sent by post to radiation therapy centers. Each center receives 3 dosimetric capsules for irradiation and one control capsule irradiated to 2 Gy at the RRC.

The participating center is requested to irradiate each of the three capsules separately to a dose of 2 Gy to water, in a water phantom, at the central axis of a vertical irradiation beam, at 5 cm depth. The field size to be used is 10 cm x 10 cm at either the source to surface distance (SSD) or the source to capsule distance, depending upon the usual technique employed at the center. The irradiation is coordinated so that all participants and the RRC irradiate during the same week in order to avoid any fading correction. The participants have to filll in a data sheet giving the method used for the absorbed dose determination. This helps to find the reasons of dose discrepancies between the dose quoted by the participant and the dose evaluated at RRC.

For the calibration of TL-dosimeters the RRC uses the IAEA International Code of Practice [4] for absorbed dose determination:

$$D_w = M_u \times N_D \times (S_{w,air})u \times p_u$$

where:

N_D is the absorbe dose chamber factor;

M_u is the meter reading, (corrected for ambient parameters).

$(S_{w,air})u$ is the mass stopping power water to air ratio

and

p_u is the perturbation correction factor.

The measurements at RRC are made in a water phantom at the central axis of a vertical cobalt 60 beam, with the ionization chamber centered at 5 cm depth and correcting for effective point of measurement. A 10 cm x 10 cm field size at surface is used.

Once the TL-dosimeters return to RRC ll measurements corresponding to a batch are made on the same day. From each capsule 3 TL-readings are obtained. The mean value is determined for each capsule being the standard deviation of these readings better than 1.5 % . The TL-readings are normalized to reference powder readings. The calibration line is obtained and the dose delivered by the participant is determined by interpolation in this straight line. The total uncertainty of the RRC TLD-system is ± 2%.

After the dose delivered by the participant is evaluated at RRC, the per cent deviation, Dev(%), between the dose quoted by the particiant, QD, and the dose evaluated by the RRC, ED, is calculated for each participating center:

$$Dev(\%) = (QD - ED) \times 100/ED$$

3. RESULTS

Results for the 88 cobalt 60 therapy units in operation in Argentina are summarized in Figure 1 to 3. The results obtained during the first participation of each unit (year 1978 to 1981) are shown in Figure 1. Only about 45 % of units delivered the dose within the interval ±5%. The mean dose was D = 1.908 Gy (σ = 11.7%), where centers with dose discrepancies greater than ±30% were not considered for the calculation of D. Figure 2 summarizes the results for the second participation of centers in this programme (year 1982 to 1983). About 70 % of participants obtained dose deviations lower than ± 5%. The mean dose was D=1.970 Gy (σ = 6.4%). To abreviate, Figure 3 summarizes the results obtained during year 1991 to 1992. About 80% of participants delivered the dose within the interval ± 5%. The mean dose was D= 1.990 Gy with σ = 3.4%.

With data sheet information it is possible for the RRC to calculate the real dose, (QD)*, given to capsules by the participants. The per cent deviation between (QD)* and the evaluated dose ED can be calculated:

$$Dev\ *(\%) = ((QD)^* - ED) \times 100/ED$$

Figures 4 and 5 show the Dev* (%) distribution for first and second participation of centers in the intercomparison programme. Results improve significatively if Dev*(%) is considered. This means that in many cases dose discrepancies are due to errors in dose calculations. In order to correct this problem the RRC sent to each center the corresponding information, the method for dose calculation and a list with the last recommended factors for dose evaluation. Nowadays no significant differences between Dev(%) and Dev*(%) are found.

According with the information given in the data sheet about 45% of centers in Argentina applies the IAEA Internatinal Code of Practice [4] for dose measurements. About 30% of centers uses a water phantom and 25% uses measurements in air for dose determinations according to recommendations given in ICRU 23 [5] .

4. DISCUSSION

In 1980 the authorities of National Health Ministry and the National Commission for Atomic Energy approved regulations for radiation therapy centers. These regulations include equipment and staff requirements for the authorization of operation of those centers. The participation of physicists within the staff of centers is required . The participation in the RRC dose intercomparison programme is considered within this normative too: the centers are obliged to participate at least once a year in the national dose intercomparison.

Nowadays about 50 physicists specially trained in radiation therapy are included within the staff of radiation therapy centers. Through the intercomparison programme it has been noted that 90% of centers with physicists delivered the dose within the accepted interval ± 5%. Dose discrepancies greater than ± 5% are easily corrected in those centers with a physicist in the staff.

The use of the IAEA International Code of Practice for dose measurements [4] is linked with the presence of physicists at radiation therapy centers. The above mentioned 45% of centers that uses this Code for dose measurements have a physicist within the staff.

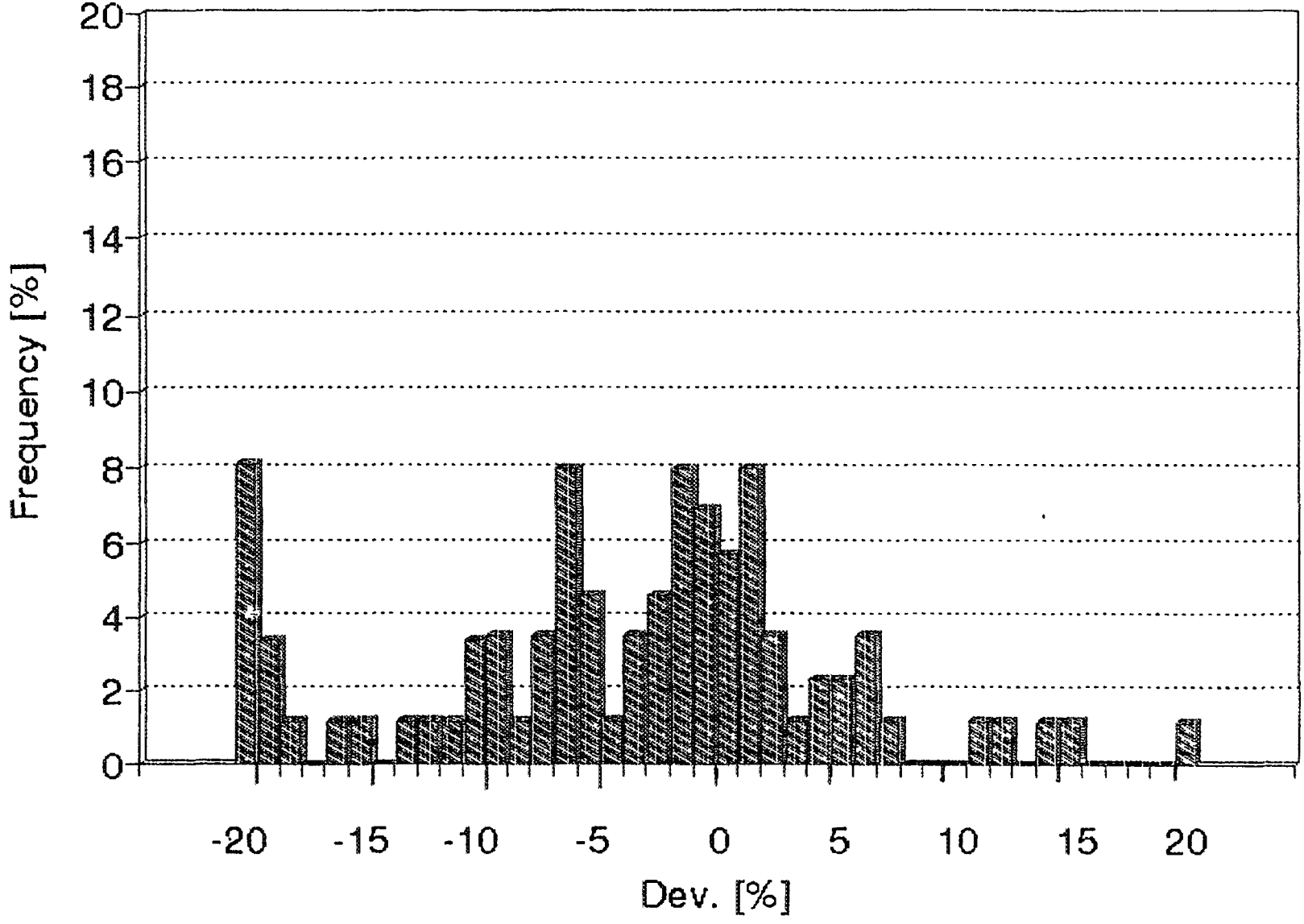

Figure 1. Results of RRC TLD service for cobalt 60 therapy units. First participation (year 1978 to 1981).

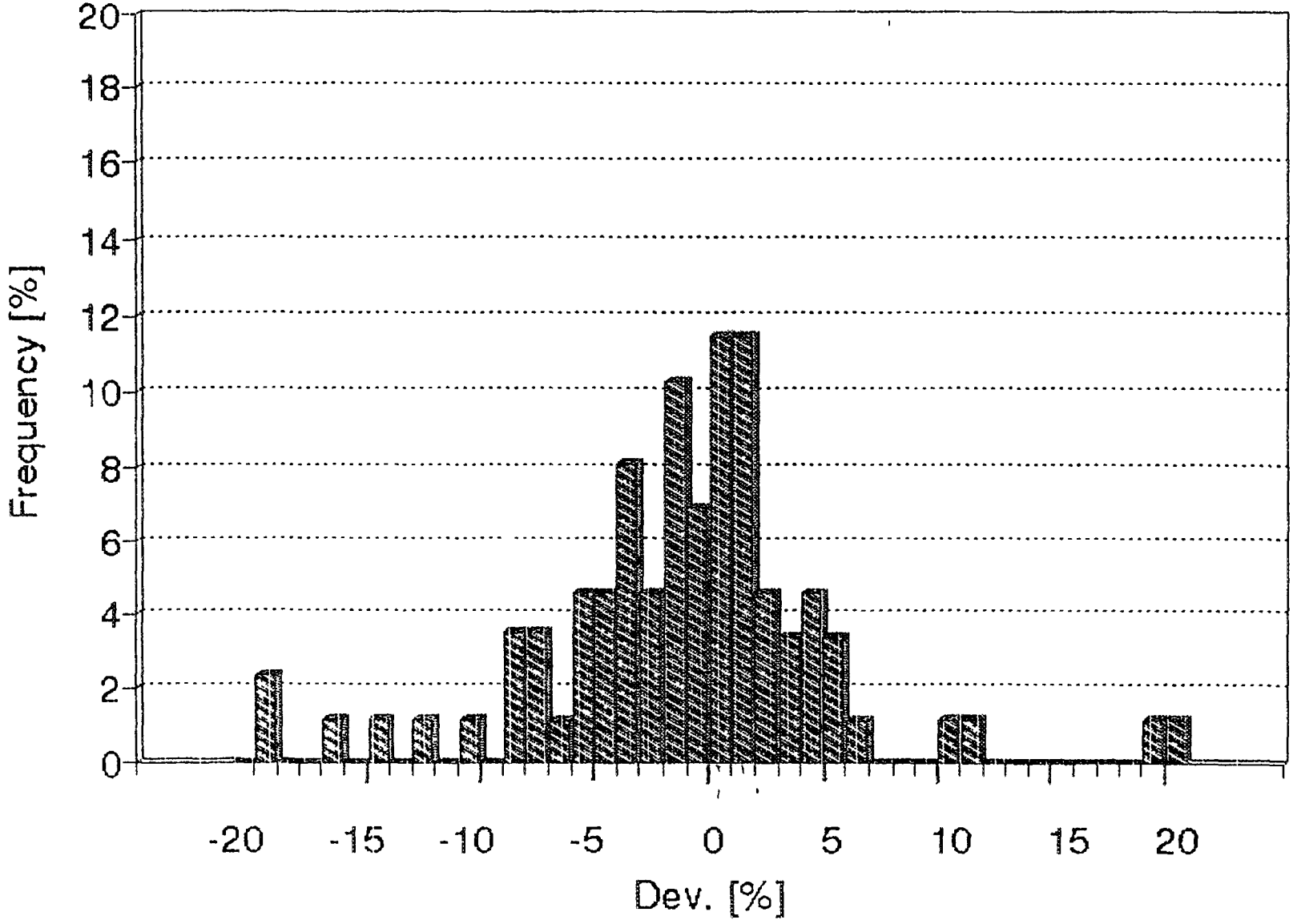

Figure 2. Results of RRC TLD service for cobalt 60 therapy units. Second participation (year 1982 to 1983).

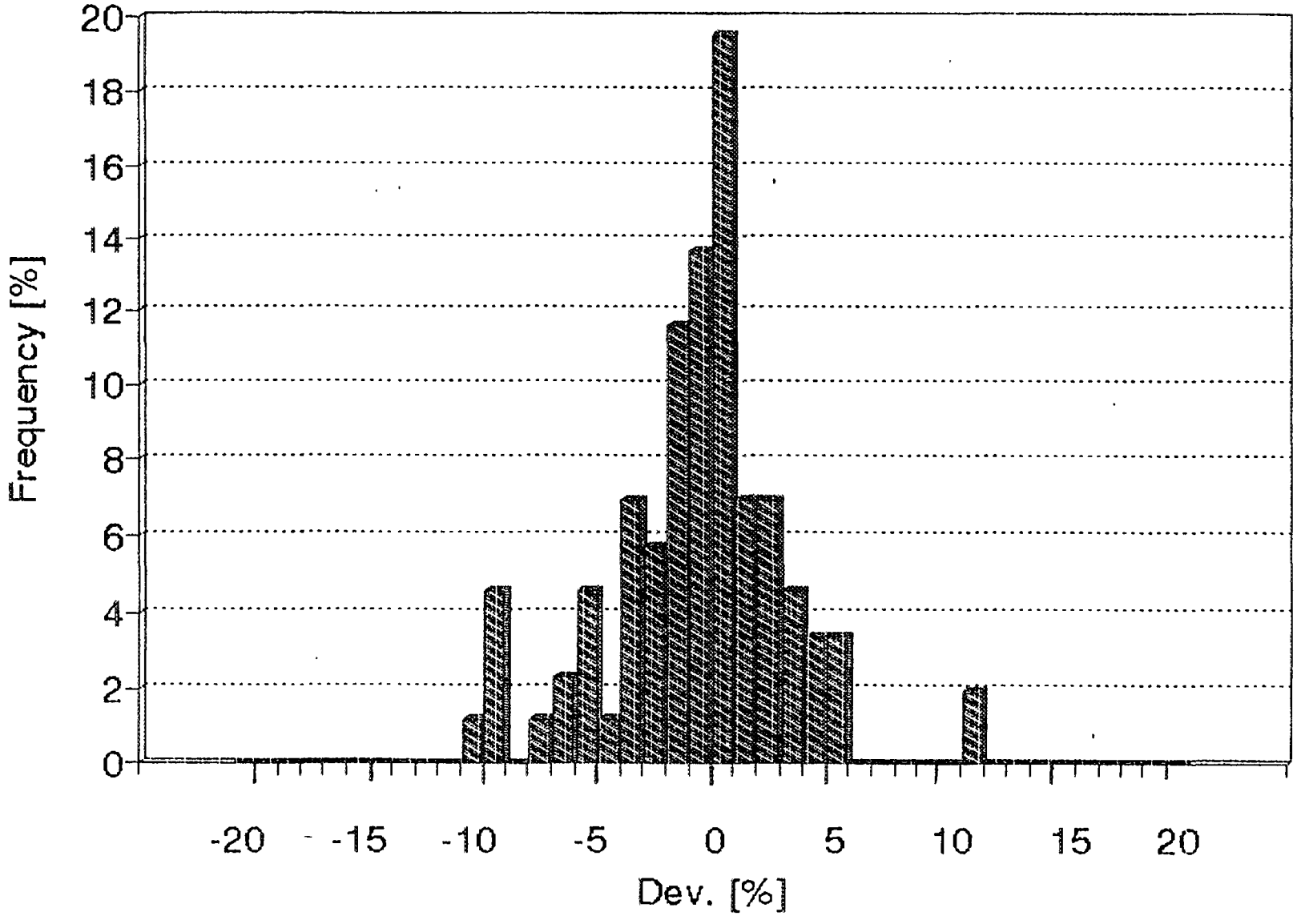

Figure 3. Results of RRC TLD service for cobalt 60 therapy units for year 1991 to 1992.

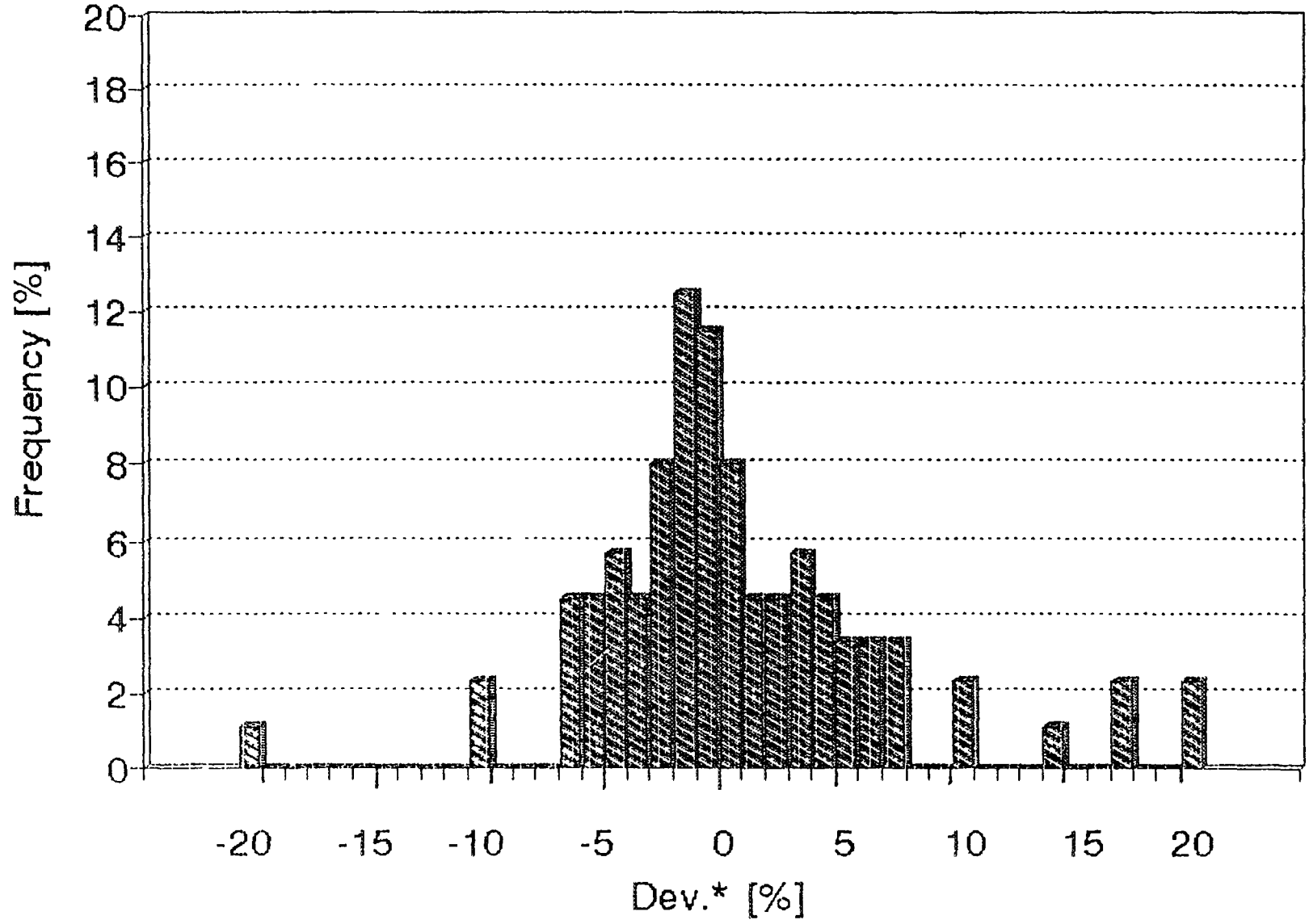

Figure 4. Results of RRC TLD service for cobalt 60 therapy units 1considering Dev(%): year 1978 to 1981.

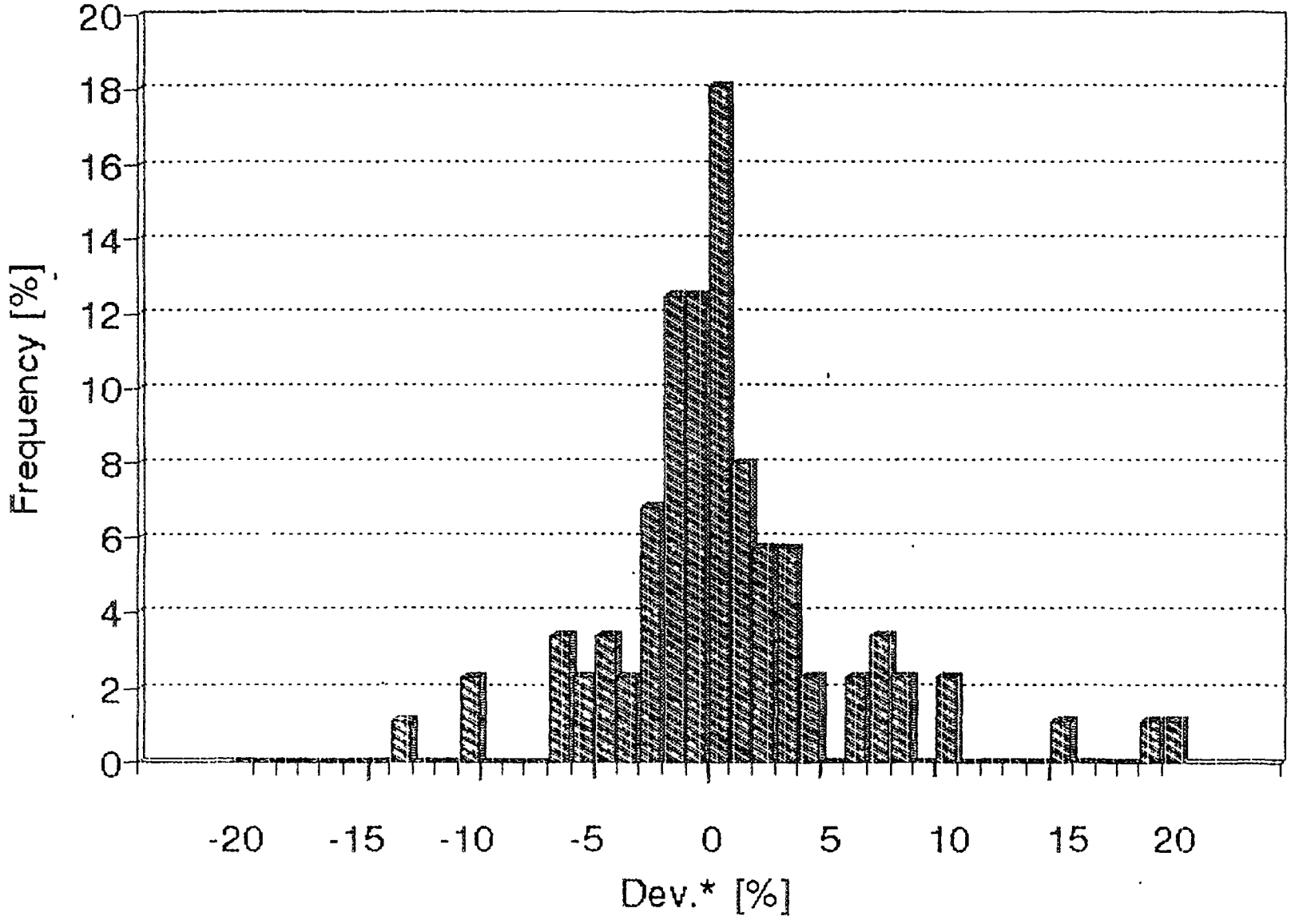

Figure 5. Results of RRC TLD service for cobalt 60 therapy units 1considering Dev(%); year 1982 to 1983.

5. TEST OF THE METHOD

During 1992 and 1993 the RRC participated in the IAEA Quality Control Programme for the RRC TLD postal dose intercomparison service. The QC programme included: a) reference irradiation at IAEA of TL-dosimeters from RRC; b) participation of IAEA as a radiation therapy center; c) run of IAEA TLD service in paralell with RRC TLD service. Distribution of capsules was coordinated in such a way that participating centers, IAEA and RRC irradiated during the same week.

In May 1992, previous to the irradiation window established for TL-capules, the RRC received the IAEA CARE system consisting in two electrometers and two ionization therapy level chambers to be calibrated at RRC. These dosimeters were calibrated in a horizontal cobalt 60 beam, in air, using the Secondary Standard NE 2560 Therapy Level dosimeter with graphite chamber NE 2561 belonging to the RRC. The calibration factor in terms of air kerma, N_K, was obtained for each CARE system. The CARE dosimeters were calibrated in a water phantom too, in a horizontal cobalt 60 beam at 5 cm depth on the central axis. The calibration factor in terms of absorbed dose to water $N_{D,w}$ for each CARE dosimeter was determined.

Results obtained in the IAEA CARE Programme participation are summarized in TableI. The maximum difference between the N_K factors obtained by the RRC and those reported by IAEA was -0.63 % . For the $N_{D,w}$ factors the maximum deviation of RRC values with regard to those reported by IAEA was + 0.31% .

TABLE I. RESULTS OF PARTICIPATION IN IAEA CARE PROGRAMME

	CAD 104 Ion chamber TK02 ser. No 104	CAD 105 Ion chamber TK02 ser. No 105
Mean air Kerma calibration det. by IAEA N_K (Gy/V)	1.599±0.8%	1.598±0.8%
Mean air Kerma calibration det. by the SSDL N_K (Gy/V)	1.589±0.9%	1.590±0.9%
Deviation	- 0.63	- 0.50
Absorbed dose to water cali factor det. by IAEA $N_{D,w}$ (Gy	1.599±1.0%	1.599±1.0%
Absorbed dose to water cali factor det. by the SSDL$N_{D,w}$ (	1.599±1.1%	1.599±1.1%
Deviation	+ 0.31	+ 0.17

Deviation = (RRC-IAEA)x100/IAEA

The results of participation in the IAEA QC Programme for TLD postal dose intercomparison service of the RRC are summarized in Table II where the ratio between the dose evaluated by the RRC, ED_{RRC1}, for the participating centers and the dose evaluated by IAEA, ED_{IAEA}, for the same centers are shown .According to these results the dose evaluated by RRC TLD-system is about 1% lower than that evaluated by IAEA TLD-system, beeing the standard deviation 1.1% .

TABLE II.

RATIO BETWEEN ED_{RRC} and ED_{IAEA}	
Participant TLD set No	ED_{RRC}/ED_{IAEA}
R002	0.9961
R003	0.9926
R004	1.0073
R005	0.9891
R007	0.9963
R008	0.9735
R009	0.9975
EL/1-93023	1.0055
EL/1-93024	0.9824
EL/1-93025	0.9931
EL/1-93026	1.0000
EL/1-93028	0.9664
EL/1-93029	0.9867
EL/1-93030	0.9888
EL/1-93031	0.9827
EL/1-93032	1.0054
Mean = 0.9915	
σ = 1.1%,	

6. CONCLUSIONS

The results obtained by RRC in the IAEA CARE Programme and in the IAEA QC Programme for the RRC TLD system show the coherence of dose measurements. The average deviation -1.0% obtained by the RRC in the IAEA QC programme for RRC TLD-system is acceptable taking on account that the total unceratinty for dose determinations with the RRC TLD system is ± 2%.

The RRC of Argentina has gained great experience in the TLD intercomparison programme for cobalt 60 therapy units. Increase of visits to radiation therapy centers should be implemented in order to improve the results of this programme. Increase of physicists trained in radiation therapy will help to improve the dosimetry in
those centers.

National regulations for operation of radiation therapy centers have given the necessary sustain for succesfull application of the RRC dose intercomparison programme. Improvements in the dose delivered by radiation therapy centers have been obtained through this programmme.

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A PROGRAM ON QUALITY ASSURANCE AND DOSE CALIBRATION FOR RADIATION THERAPY UNITS IN VENEZUELA

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Abstract

The results of a five year program (1988-90-91-92-93) on quality assurance and dose calibration for 12 Cobalt-60 units from public hospitals, which represents 30% of total radiation therapy units in Venezuela, are presented. The remarkable improvement in the general performance of these units can be seen in the IAEA/WHO Postal TLD Intercomparison results which gave 100% within ± 5% in 1990 and 1992, while 63% in 1990 and 44% in 1992, with errors up to 37% were obtained for the participants not included in the program.
The difference between the two groups lead the government to decrete through the Gaceta Oficial de la República de Venezuela, Resolution G-1397 on March 3, 1993, that quality assurance and dose calibration programs shall be established for all radiation therapy installations in Venezuela. The project for the standards was developed by the SSDL physicists and it was already approbated by the Health Ministry. It is expected that the Norms will enter into effect by the end of 1994.

1. INTRODUCTION

In 1988 a program on quality assurance and dose calibration was started by IVIC-SSDL on 12 Cobalt-60 units from public hospitals, which represents 30% of total RT units in Venezuela. All the job was performed by the SSDL task group and the program was sponsored by the Health Ministry.

This paper presents the methods and the results related to basic physics and dosimetric checks performed through postal dose intercomparisons and during on-site visits between 1988 and 1993, the SSDL projects for the next years as well as the main aspects covered by the Norms that will regulate radiation therapy installations.

1.1. Situation by 1988

1.1.1. No unit had preventive maintenance.
1.1.2. Some RT units worked only with the certificate provided by the source manufacturer. No beam calibration.
1.1.3. Most units were calibrated just once over several years.
1.1.4. None had a QA program.
1.1.5. Most institutions had empirical techniciens.
1.1.6. Many radiation therapy physiciens have never had a course on radiation physics.
1.1.7. Of 36 institutions (25 Co-60 and 16 LA), only 3 had a hospital physicist (part-time).
1.1.8. The role of a hospital physicist was not recognized. Often, salaries were lower than for techniciens.
1.1.9. As there were no legislation in this field, some private and public institutions never participated in Postal TLD Intercomparisons.

2. METHODS

2.1. On-site visits

On-site visits were performed at each of the 12 radiation therapy departments once a year. At each visit, a total QA inspeccion of unit parameters was carried out along with beam calibration for all field sizes. Two staff members of SSDL working 2-3 days were needed for each visit.

2.1.1. Mechanical verifications and beam alignments

The following main parameters with their indicated tolerances [1-7] were verified at each Co-60 unit (TABLE I).

TABLE I:
Main unit parameters with their tolerances

PARAMETER	TOLERANCES
Rotational system axis: isocenter and scales	± 2mm/0.5°
Field size congruence	± 2mm
Collimator symmetry and stability	± 2mm
Couch parameters	± 2mm
Radiation field alignment and homogeneity	± 2mm
Source transit time*	± 0.5%
Virtual source distance to isocenter**	± 0.5%

* influence for a dose of 2 Gy
** within consecutive calibrations

During 1993, the SSDL also conducted work sessions at each of the 12 radiation therapy departments to ensure that the process of QA and dose calibration and the IAEA/WHO Postal TLD Intercomparison were fully understood by the physiciens, techniciens and dosimetrists.

2.2. Absorbed dose determination

Determination of absorbed dose in water phantom.

The measuring procedure and calculations were performed according to the recommendations contained in the International Code of Practice published by IAEA [8].

A cylindrical ionization chamber, type NE 0.6 cm^3, connected to an electrometer type NE Farmer, calibrated at SSDL in a Co-60 beam, against a secondary standard dosemeter, in terms of absorbed dose to water and air kerma were used for all Co-60 beams. The constancy of the response of the chamber and electrometer was checked daily during the visits, with a Sr-90 reference source.

Absorbed dose determination was carried out in an IAEA water phanton at the reference depth of 5 cm, and at dmax, at the usual treatment distance, for all field sizes [4-6,8-13]. When available, absorbed dose to water was also measured with the breast tray.

Determination of absorbed dose to air.

Absorbed dose to air measurement was carried out for a 10 x 10 cm field at the treatment distance or at the isocenter with the chamber in air with the build-up cap.

Other parameters verified were:

- unit maintenance
- general mechanical performance
- head leakage
- radiological safety and area survey
- accesory factors
- personnel dosimetry
- requirements on source or unit replacement

2.3. Postal TLD Intercomparisons

WHO/IAEA Postal TLD Intercomparisons [14], coordinated by the SSDL, started in Venezuela in 1983. They were regularly performed in 1988, 1990 1992 and 1993 (TABLE III includes results up to 1992) with all the units under control and a small number of units not included in the program, even though TLDs were sent to all institutions nationwide.

3. RESULTS

3.1. Situation by 1993

3.1.1. The 12 units under control plus other 2 were already working under a contract between the government and the SSDL for annual QA and dose calibration, and with private companies for regular unit maintenance.

3.1.2. For Postal Intercomparison, any TLD with a discrepancy exceeding 5% was pursued by TLD reirradiation and, by dosimetry review visit if the discrepancy was unresolvable.

3.1.3. Five institutions calibrated their units and participated in Postal TLD Intercomparisons for the first time.

3.1.4. For the rest of the RT units, the situation remained without changes.

3.1.5. For medical physicists, the situation remained without changes.

3.1.6. Consequently to on-site visits, a better uniformity among the visited centers was reached.

3.1.7. The remarkable improvement in the general performance of the Cobalt-60 units can be seen in the IAEA/WHO Postal TLD Intercomparison results in 1990 and 1992 where 100% of the units under control were within $\pm$ 5 %, while 63% in 1990 and 44% in 1992, with errors up to 37%, were obtained for the participants not included in the program (TABLES II and III).

3.1.8. The difference between the two groups (Fig.1) lead the government to decrete on March 1993 that quality assurance and dose calibration programs shall be established for all radiation therapy installations in Venezuela. The project for the standards [5] was prepared by the SSDL physicists and it was already approbated by the Health Ministry. It is expected that the Norms will enter into effect by the end of 1994.

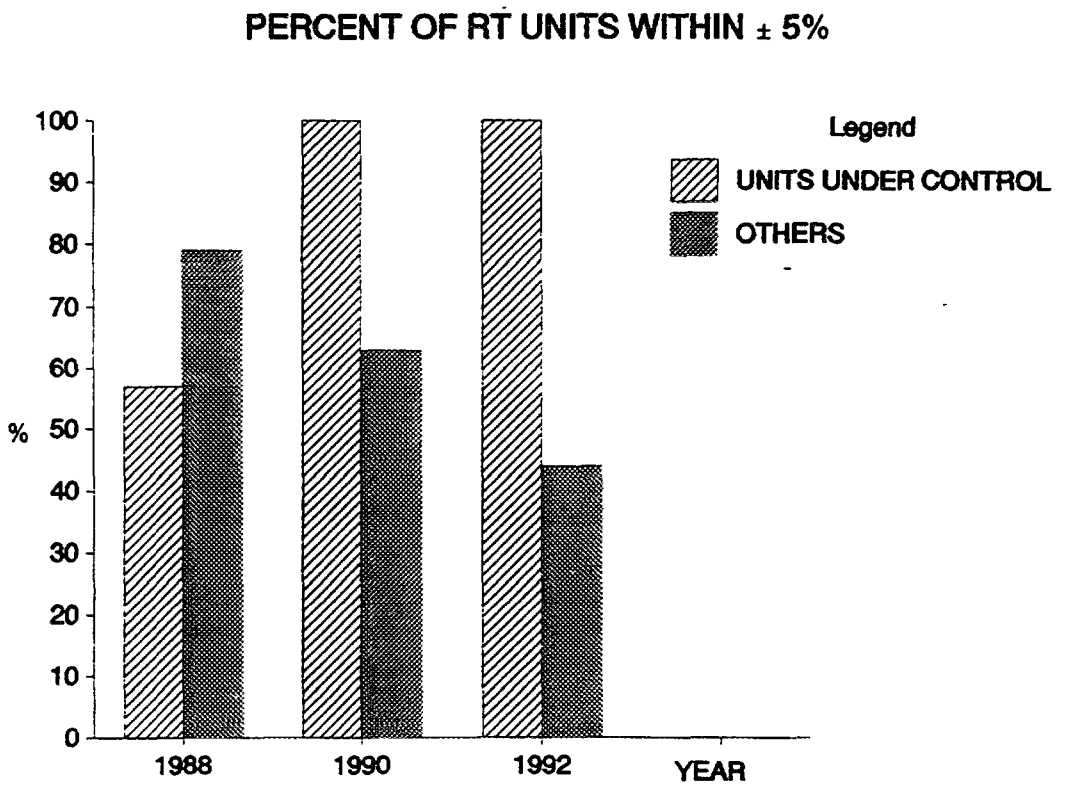

Fig. 1. Percent of beam calibration within criterion for the group under control and the rest of the units for 1988, 1990 and 1992 Postal TLD Intercomparisons.

TABLE II:
Percent of units under control with acceptable deviation

CHECKS	1988	1993
Unit preventive maintenance	0	100
General mechanical performance	42	80
Head leakage	100	100
Radiological safety and area survey	50	83
Rotational system axis: isocenter and scales	33	73
Field size congruence	43	73
Collimator symmetry and stability	50	73
Couch parameters	50	50
Radiation field alignment and homogeneity	75	82
Source transit time	90	90
Virtual source distance to isocenter	100	100
Personnel dosimetry	8	50

TABLE III:
Postal TLD Intercomparison results*

Year	1983	1984	1985	1988	1990	1992
Number of Participants	4	17	13	21	13	26
within ± 5%	75	47	41	67	70	54
within ± 5% with QA/CAL	---	---	---	57	100	100
within + 5% without QA/CAL	---	---	---	79	63	44

* In 1986, 19 institutions participated on the Intercomparison.
TLDs were not returned on time for processing.

4. DISCUSSION

4.1. SSDL Projects for the near years:

4.1.1. To establish a post-graduate program with a Master Degree on Medical Physics, starting by 1995, to be the first one in Venezuela. The goal is to supply a medical physicist to every institution with a RT unit.
4.1.2. To organize an national society of medical physics.
4.1.3 To work very closely with the Health Ministry to supervise the implementation of the Norms for RT installations in Venezuela.
4.1.4. To spread nationwide the use of SSDL protocols for QA and beam calibration for Co-60, linear accelerators (photon and electron beams) and ortvoltage units, in order to achieve a better uniformity among radiation therapy centers.

4.2. Main points to be covered by the Norms for QA and dose calibration for RT units: Cobalt-60, linear accelerators (photons and electron beams), and low to medium energy x-rays units [5].

4.2.1. Requirements on personnel and instruments for QA and dose calibration.
4.2.2. Requirements on personnel involved in RT installations.
4.2.3 Detailed description of QA tests and their tolerances.
Frecuency of constancy and reproducibility checks for all type of RT units: Co-60, linear accelerators (photons and electron beams) and ortovoltage.
4.2.4. Within the next 5 years, every institution with RT units must have at least one qualified physicist, minimum half journey per unit.
4.2.5. Mandatory participation on Postal TLD Intercomparisons.

5. CONCLUSIONS:

The Postal TLD Intercomparisons and QA surveys conducted in these units clearly demonstrated the need for upgrading the standards applied to radiation therapy installations.
The most important consequence of these surveys is the establishment of the Norms that will regulate radiaton therapy installations in Venezuela.

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III (b). DOSIMETRY PROCEDURES

Invited Paper

RADIATION DOSIMETRY WITH PLANE-PARALLEL IONIZATION CHAMBERS: AN INTERNATIONAL (IAEA) CODE OF PRACTICE

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Abstract

Research on plane-parallel ionization chambers since the IAEA Code of Practice (TRS-277) was published in 1987 has expanded our knowledge on perturbation and other correction factors in ionization chamber dosimetry, and also constructional details of these chambers have been shown to be important. Different national organizations have published, or are in the process of publishing, recommendations on detailed procedures for the calibration and use of plane-parallel ionization chambers. An international working group was formed under the auspices of the IAEA, first to assess the status and validity of IAEA TRS-277, and second to develop an international Code of Practice for the calibration and use of plane-parallel ionization chambers in high-energy electron and photon beams. The purpose of this work is to describe the forthcoming Code of Practice.

1. INTRODUCTION.

The advantages of using plane-parallel ionization chambers in the dosimetry of therapeutic electron beams have been recognised in all dosimetry protocols. The design characteristics, mainly regarding the shape and size of the collecting volume, make this instrument theoretically ideal for measurements in regions with large dose gradients in the beam direction.

A number of chambers are available today, a few of them having completely new designs, with practically negligible perturbation effects in electron beams. Large correction factors have

been found, however, for other chambers, mainly at low electron energies. There is still controversy on the use of plane-parallel chambers for photon beam dosimetry. Most chambers are far from homogeneous in their construction as, in general, materials with different scattering and absorption properties are used in the various walls. It is likely that these effects approximately balance other effects in electron beams, but measurements and calculations in photon beams have shown the need for correction factors to account for the different materials in the chamber. This suggests that plane-parallel chambers should mainly be used for absorbed dose determinations in electron beams but only for relative measurements in photons. The remaining problem is the calibration of the chamber.

The lack of details on dosimetry procedures using plane-parallel chambers, particularly regarding their calibration or a practical determination of the N_D (N_{gas}) chamber factor, has been one of the major criticisms made of the IAEA Code of Practice, TRS-277 [1] where only a reference to the procedures described by NACP [2] was made. It was considered that these procedures were well established and therefore still to be recommended. The influence of the central electrode correction for cylindrical chambers in TRS-277, however, added an unexpected complication to experimental determinations of N_D based on a comparison in electron beams [3].

Research in the field since IAEA TRS-277 was published has expanded our knowledge on perturbation and other correction factors in ion-chamber dosimetry, and also constructional details of the chambers have been shown to be important. Different national organisations have published [4, 5] or are in the process of publishing [6, 7] recommendations including detailed procedures for the use of plane-parallel chambers. An international working group was formed under the auspices of IAEA, first to assess the status and actual validity of IAEA TRS-277 [1] and second to develop an international Code of Practice for the use of plane-parallel ionization chambers in high-energy electron and photon beams. The purpose of this work is to describe the new Code of Practice. Further details on the present situation regarding correction factors and quantities briefly discussed here can be found in [3].

2. AN OVERVIEW OF THE NEW CODE OF PRACTICE.

The contents of the Code of Practice is shown in Table I. It can be observed that together with a rather conventional distribution of the different sections 1-9, Section 10 contains a summary

TABLE I. CONTENTS OF THE *IAEA CODE OF PRACTICE* FOR PLANE-PARALLEL IONIZATION CHAMBERS

1. Introduction
2. Update of the information in TRS-277
3. Equipment
4. Beam quality specification
5. N_K-based formalism and determination of $N_{D,air}$ for plane-parallel ionization chambers
6. N_{D,w,Q_o}-based formalism and determination of N_{D,w,Q_o} factors for plane-parallel ionization chambers
7. Use of plane-parallel chambers in electron beams
8. Use of plane-parallel chambers in photon beams
9. The uncertainty in absorbed dose determination at the reference depth using plane-parallel chambers in electron beams
10. A **Code of Practice** for the calibration and use of plane-parallel ionization chambers

Appendix A. Worksheets
Appendix B. Stopping-power ratios in clinical electron beams.
Appendix C. Chamber perturbation factors in electron and photon beams

of all the procedures and data required; this Section is effectively the Code of Practice. The report also contains Appendices where different topics are covered in detail; they also include Worksheets.

The new Code updates information in IAEA TRS-277 regarding recent developments in radiotherapy dosimetry. In most cases differences from existing values or the magnitude of new corrections, are within half a percent but developments (and clarifications) in the field are taken into account. Of special interest for the calibration and use of plane-parallel ionization chambers are

- the effect of metallic central electrodes in cylindrical ionization chambers (included in TRS-277 as a global factor) has been separated into two components, one at the Co-60 calibration (k_{cel}=1.006 for a Farmer-type chamber) and therefore entering into $N_{D,air}$ [1], and another at reference measurements in a phantom (for a Farmer-type chamber p_{cel}=0.994 in Co-60; p_{cel}=0.998 in electron beams). This yields a global correction equal to 1.004 in electrons. It should be noted that cylindrical ionization chambers are used as reference instruments for the calibration of plane-parallel ionization chambers in most calibration alternatives. New values for these corrections, based on Monte Carlo calculations, are adopted [8]. The new expression for $N_{D,air}$ for cylindrical ionization chambers becomes

$$N_{D,air} = N_K(1-g)\,k_{att}k_m k_{cel} \qquad (1)$$

- a procedure based on an absorbed-dose-to-water calibration factor, $N_{D,w}$, is also introduced. This symbol was given in TRS-277 but in practice its use was restricted to low-energy X-rays. It is now becoming available for high-energy photons. At present the most common approach is to provide users with $N_{D,w}$ at a reference quality Q_o, usually ^{60}Co, and apply *beam quality* correction factors for other beam qualities. Users should be warned of the possibility of confusion arising from the notation N_D used by AAPM TG-21 [9] for the $N_{D,w}$ factor.

- a new scaling procedure for conversion of depths and ranges measured in plastic to equivalent quantities in water is given; this is based on the concept of *detour factors* as an alternative to ratios of csda ranges [10].

- a correction for the non-medium equivalence of the chamber wall material, p_{wall}. This factor has implicitly been assumed to be unity in electron dosimetry protocols to date. There is however considerable experimental evidence that this factor may not be unity for certain plane-parallel chamber designs; the probable mechanism here is backscattering differences between the material behind the cavity and that of the wall material. However only values for an *overall* perturbation factor p_Q=p_{cav} p_{wall} are given; p_{cav} replaces p_u as the correction for the in-scattering effect in gas cavities.

- new calculations of stopping-power ratios water/air, $s_{w,air}$, using several independent Monte Carlo codes where different density effect corrections were taken into account. Compared with the stopping-power ratios in TRS-277, differences are small for the electron energies most commonly used in radiotherapy, being close to 0.5% at most depths. The recommendation for the small change is justified in terms of the lack of ambiguity in the corrections used and the higher accuracy of the present set of data.

- the determination of the recombination correction factor for plane-parallel ionization chambers using the "two-voltage" method has been shown to have limitations for most chambers due to the lack of linearity of saturation curves in the region of interest. In order to decrease the influence in the dosimetry procedure it is recommended to use the same voltage ratio for the determination of $N_{D,air}$ and for the absolute dose determination.

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[1] Note that the factor N_D in TRS-277 is now denoted by $N_{D,air}$ in order to distinguish it from $N_{D,w}$, the factor in terms of absorbed dose to water.

Section 3 provides detailed description on phantoms and equipment available, with emphasis on the properties of plane-parallel ionization chambers both for electron and photon radiation. Chambers of new design (Attix, Roos, etc) are included in the compilation. As in TRS-277, water is the recommended reference medium although plastics may be used for measurements at low electron energies. Emphasis is given, however, to the high accuracy achievable today with modern equipment in positioning ionization chambers in water phantoms which thus minimizes the need to use plastic phantoms.

The uncertainty in absorbed dose determination at the reference point using the recommended procedure for determining $N_{D,air}$ is treated in detail, separating the different steps of the dosimetric procedure in a similar way to TRS-277 but incorporating an updated evaluation of uncertainties in the different steps. Uncertainties are also evaluated for the alternative calibration methods based on measurements in photon beams.

Further details on certain sections follow.

2.1. Beam quality specification

The specification of the quality of the beams used for the calibration of plane-parallel ionization chambers follows the recommendations given in TRS-277. As mentioned in the introduction, absolute dosimetry is to be performed in electron beams only, as is the recommended calibration procedure (see below).

For dosimetry purposes it has become customary to specify the quality of electron beams in terms of the mean energy at the surface of the phantom, $\bar{E}_O$, determined from empirical relationships between electron energy and the 50% range in water, R_{50}. $\bar{E}_O$ is needed for the selection of different quantities and parameters in the formalism, and mainly affects the choice of stopping-power ratios water to air, $s_{w,air}$, at the reference depth, namely $s_{w,air}(\bar{E}_O, z_{ref})$. As in IAEA TRS-277 [1] and most dosimetry protocols, the recommendation is to determine $\bar{E}_O$ using the energy-range relationship

$$\bar{E}_O = C\, R_{50}\, MeV \tag{2}$$

where C=2.33 MeV cm^{-1} and R_{50} is obtained from a depth-dose distribution measured with constant source-chamber distance. As is well known, when the dose distribution has been obtained with a constant source-surface distance (SSD=100 cm) Eq. (2) is not strictly valid. As an alternative IAEA TRS-277 has provided tabulated data for determining $\bar{E}_O$ either from ionization curves measured at SSD=100 cm with an ionization chamber or from depth-dose distributions at SSD=100 cm, measured for instance with solid state detectors. These data can be fitted with the following second order polynomial:

$$\bar{E}_O = 0.818 + 1.935\, R_{50}^J + 0.040\, (R_{50}^J)^2 \tag{3}$$

for R_{50}^J determined from a depth-ionization curve and

$$\bar{E}_O = 0.656 + 2.059\, R_{50}^D + 0.022\, (R_{50}^D)^2 \tag{4}$$

for the case of a depth-dose curve, R_{50}^D. For energies above 3 MeV, Eqs. (3) and (4) yield stopping-power ratios, water-to-air, that on the average agree within 0.2% up to depths equal to 0.80 R_p with $s_{w,air}$ values obtained with $\bar{E}_O$ derived from TRS-277 Table IV, with a maximum deviation of 0.4% close to 12 MeV.

Improved energy-range relationships between $\bar{E}_O$ and R_{50}, based on Monte-Carlo calculations for mono-energetic electron beams, have been developed[3, 11] but all yield $\bar{E}_O$ values higher than the above expression. This would result in lower stopping-power ratios at the reference depth compared to those obtained with $s_{w,air}(\bar{E}_O, z_{ref})$ and $\bar{E}_O$ from Eq. (2).

2.2. Determination of $N_{D,air}$ for plane-parallel chambers

Several different methods have been proposed by Mattsson *et al* [12] for obtaining the absorbed-dose-to-air chamber factor $N_{D,air}$ for a plane-parallel chamber. These methods fall into two broad categories. In the first one, a Standards Laboratory calibrates the chamber in terms of N_K and then $N_{D,air}$ is obtained theoretically.

In the second one, the user determines $N_{D,air}$ directly by experimental intercomparison with a reference ion-chamber having a known $N_{D,air}$ factor. Both chambers are alternatively positioned at a reference depth in a phantom and the unknown $N_{D,air}$ is obtained from equating the absorbed doses with the two chambers. These procedures have been extensively discussed in Ref. [13] and in the recent TG-39 protocol of the AAPM [14]. Methods in the second category are generally performed in the user's beam, either ^{60}Co or high-energy electrons [12]. It can be noted that this method can in principle be applied to determining $N_{D,air}$ for any chamber that is to be used in electron or photon beams e.g. a second cylindrical chamber provided that $N_{D,air}$ is already known for a reference chamber [4, 15]. Consequently the chamber to be calibrated (not necessarily plane-parallel) and the reference chamber will be denoted by x and *ref* respectively.

The primary recommendation is the use of a high-energy electron beam. Following the formalism in TRS-277, and equating the absorbed dose at the reference depth with the two chambers, the expression for $N_{D,air}$ for the chamber x to be calibrated, becomes

$$N^x_{D,air} = N^{ref}_{D,air} \frac{M^{ref}}{M^x} \frac{p^{ref}_{wall} p^{ref}_{cav} p^{ref}_{cel}}{p^x_{wall} p^x_{cav} p^x_{cel}} \qquad (5)$$

where the numerator and denominator correspond to the D_w determination using the reference chamber (usually cylindrical) and chamber x respectively, and the stopping-power ratios cancel out. M^{ref} and M^x are ratios of the readings of the two chambers to those of an external monitor to take into account possible accelerator output fluctuations. They must be corrected for the polarity effect, for recombination, and for temperature and pressure. Note that p^{ref}_{wall} for the reference chamber is unity as recommended reference cylindrical chambers are assumed to have negligible wall effects in electron beams [16, 17]. For most plane-parallel ionization chambers and at the energies recommended for the calibration, the factors p^x_{cav} and p^x_{wall} are practically unity. The factor p^x_{cel} is not relevant for plane-parallel ionization chambers but as the procedure can also be extended to cylindrical ion chambers it has been retained in this Eq. For the case of x being a cylindrical chamber the value of p^x_{cav} should be interpolated from the data from Johansson *et al* [16] given in TRS-277 Table XI..

The phantom material should preferably be the same as that used for the absolute dose determination. This automatically ensures that the overall effects of any perturbation due to differences in backscattering between the material behind the cavity and that of the phantom (i.e. the component of p_Q due to p_{wall}) will be minimized. Water is the preferred material. The energy of the electron beam should be as high as possible in order to minimise the perturbation due to the air cavity of the reference chamber. As a guide p^{ref}_{cav} should be within 2% of unity. For a cylindrical reference chamber with an internal radius of 3 mm (approximately Farmer type) this means that $\bar{E}_o$ should be no lower than 15 MeV but should preferably be as high as possible; the lower limit on $\bar{E}_o$ may be lowered if the chamber radius is smaller. The depth should be the same as the reference depth z_{ref} used for absorbed dose determination in the chosen high-energy beam. The SSD should be 100 cm and the field size should be approximately 12 cm x 12 cm or larger - this is not critical. The chambers are to be placed with their respective effective points of measurement, P_{eff}, at the same depth. A Farmer-type chamber, i.e. approximately 6 mm internal diameter and 1 mm electrode diameter, for the reference cylindrical chamber is recommended here as a great deal of experience has been gained with such chambers and the correction factors can be said to be well known [18, 19]. The choice of a chamber with a radically different geometry, e.g. a very thick central electrode, can lead to larger uncertainties.

Alternative methods for obtaining the absorbed-dose-to-air chamber factor $N_{D,air}$ for a plane-parallel chamber based on measurements made in a ^{60}Co beam have been introduced. They are classified into two categories generally depending on the institution where the calibration is performed. The *in-phantom* method is generally performed in the user's beam at the Hospital, although it can also be performed at the Standards Laboratory. Measurements free in air are usually performed in a ^{60}Co beam at the Standards Laboratory.

The calibration in a ^{60}Co beam at depth in a phantom has been described by several authors. First by Mattsson *et al* [12] and then in more detail by Attix [20]. The approach is based upon the determination of $N_{D,air}$ from the knowledge of the absorbed dose in the phantom determined with a calibrated reference chamber, like that recommended in the electron-beam method, but in this case irradiated with a ^{60}Co beam. The formalism yields:

$$N_{D,air}^{pp} = N_{D,air}^{ref} \frac{M^{ref}}{M^{pp}} \frac{p_{wall}^{ref} p_{cel}^{ref}}{p_{wall}^{pp}} \tag{6}$$

where p_{wall}^{ref} and p_{cel}^{ref} are perturbation factors of the reference chamber at ^{60}Co; p_{cel}^{ref} is unity for a graphite electrode and 0.994 for a Farmer-type chamber. The standard SSD for a ^{60}Co unit and a field size of 10 cm x 10 cm at the surface should be used. In this method the effective point of measurement for both chambers should be placed at a reference depth of 5 g cm^{-2} in a phantom that matches the plane-parallel chamber material (to minimise p_{wall}^{pp}) or in water if the p_{wall}^{pp} factor is known. For cylindrical chambers in ^{60}Co beams P_{eff} is positioned at a distance equal to 0.6 r from the centre of the chamber.

It is important to note that when a non-water phantom is used, TRS-277 does not provide a direct determination of the perturbation p_{wall} at ^{60}Co as absorbed dose should only be determined in a water phantom. The perturbation factor of the reference chamber is determined according to the general equation that takes into account the thin waterproofing plastic or rubber sleeve normally used to protect the chamber in a water phantom (see also Refs. [21, 22]):

$$p_{wall}^{ref} = \frac{\alpha\, s_{wall,air}\, (\mu_{en}/\rho)_{med,wall} + \tau\, s_{sleeve,air}\, (\mu_{en}/\rho)_{med,sleeve} + (1-\alpha-\tau)\, s_{med,air}}{s_{med,air}} \tag{7}$$

where *med* is the phantom material and α and τ the fractions of ionization due to electrons arising from the wall and waterproofing sleeve respectively. A fit to the available data for α [23] is given. By applying this fit to the combined thickness of the wall and the sleeve, and substracting α from this, an expression for τ is obtained. This insures that $\alpha+\tau \leq 1$.

The perturbation factor p_{wall}^{pp} in ^{60}Co beams is the major source of uncertainty in this procedure and the reason why the electron-beam method for the calibration of plane-parallel ionization chambers is the preferred option in the new Code of Practice. Differences in p_{wall}^{pp} close to 2% have been reported, either between Monte-Carlo calculations and experimental data [24], or due to chamber-to-chamber variations for chambers of the same type (from the same or from different manufacturers) [25]. It has to be emphasised that p_{wall}^{pp} depends on the phantom material used for the calibration.

The calibration-in-air method is similar to the free in air approach used with cylindrical chambers in Standard Laboratories. The air-kerma rate, free in air, must be known at the position of the cavity centre and N_K of the plane-parallel ionization chamber is then determined. The plane-parallel ionization chamber with appropriate build-up material is placed free in air in a ^{60}Co beam, its center positioned at the point where K_{air} is known. The build-up material should have the same

outer dimensions as the chamber and preferably be of the same material as the predominant material of which the chamber is constructed. The procedure yields the N_K calibration factor of the plane-parallel chamber

$$N_K^{pp} = \frac{K_{air}}{M^{pp}} \tag{8}$$

and if the product $k_{att}\, k_m$ is known $N_{D,air}$ is determined according to the well-known expression

$$N_{D,air}^{pp} = N_K (1-g)\, k_{att}\, k_m \tag{9}$$

where for plane-parallel ionization chambers k_{cel} is not involved. In principle this procedure is used together with a *universal* value of $k_{att}\, k_m$ for a given type of plane-parallel ionization chamber. The limitations of this approach increase considerably the estimated uncertainty.

2.3. Determination of $N_{D,w}$ for plane-parallel chambers

The formalism for the determination of absorbed dose to water in photon and electron beams using a $N_{D,w}$-based calibration factor has been given in detail by Hohlfeld [26]. The absorbed dose to water at the reference point of the chamber (where the calibration factor applies) in a phantom irradiated by a beam of reference quality Q_o is given by the simple relationship

$$D_{w,Q_o} = M_{Q_o}\, N_{D,w,Q_o} \tag{10}$$

where $N_{D,w,Qo}$ is obtained at the Standard Laboratory from the knowledge of the standard quantity absorbed dose to water at the point of measurement in water for the calibration quality Q_o.

Efforts are at present being addressed to providing $N_{D,w,Q}$ calibrations for photon beams, mainly ^{60}Co gamma-rays and to a lesser extent high-energy photon and electron beams [27-31]. A practical approach in common use is to provide users with $N_{D,w,Co}$, i. e. calibration at the reference quality ^{60}Co, and apply *beam quality* correction factors k_Q for other beam qualities [26, 32]. For beams other than the reference quality the absorbed dose to water is then given by

$$D_{w,Q} = M_Q\, N_{D,w,Q_o}\, k_Q \tag{11}$$

where the factor k_Q corrects for the difference between the reference beam quality Q_o and the actual quality being used, Q. k_Q should ideally be determined experimentally at the same quality as the user's beam, although this is seldom achievable. When no experimental data are available an expression for k_Q can be derived comparing Eq. (11) with the formalism in TRS-277; this ensures consistency with the $N_{D,air}$ procedure when k_Q is calculated with the data in TRS-277 [26, 33, 34]. In therapeutic electron and photon beams the general assumption of $(W_{air})_Q=(W_{air})_{Qo}$ yields the equation for k_Q

$$k_Q = \frac{(s_{w,air})_Q}{(s_{w,air})_{Q_o}} \frac{p_Q}{p_{Q_o}} \tag{12}$$

which depends only on ratios of stopping-power ratios and perturbation factors. It should be noted that the chamber-dependent correction factors k_{att}, k_m and k_{cel} are not involved in the definition of k_Q. The only chamber specific factors involved are the perturbation correction factors p_Q and p_{Q_o}.

The connection between the $N_{D,air}$ and the $N_{D,w}$ based formalisms is established by the relationship,

$$N_{D,w,Q} = N_{D,air}(s_{w,air})_Q p_Q \tag{13}$$

In principle Eq. (13) could be used to determine $N_{D,air}$ independent of the factors k_{att}, k_m and k_{cel}.

The use of ^{60}Co as reference quality for determining N_{D,w,Q_o} for plane-parallel ionization chambers is an attractive possibility, especially for most SSDLs. Using the formalism at other qualities (both high-energy electron and photon beams) requires, however, the knowledge of p_{Q_o} at ^{60}Co in Eq. (12) which enters in k_Q; this is the main drawback of this procedure. This is also the case for the alternative option which enables users to determine N^{pp}_{D,w,Q_o} directly by experimental intercomparison in a ^{60}Co beam with a reference ion-chamber (cylindrical in this case, where p_Q is more precisely known) having a known N^{ref}_{D,w,Q_o} factor.

It is assumed that the water absorbed dose rate is known at 5 cm depth in a water phantom for ^{60}Co gamma rays. The plane-parallel chamber is placed with its reference point at a depth of 5 cm in a water tank where the absorbed dose to water D_w is known and $N^{pp}_{D,w,Co}$ obtained from

$$N^{pp}_{D,w,Co} = \frac{D_w}{M^{pp}} \tag{14}$$

D_w is obtained using a reference chamber having a calibration factor $N^{ref}_{D,w,Co}$. The calibration factor for the plane-parallel chamber becomes

$$N^{pp}_{D,w,Co} = N^{ref}_{D,w,Co} \frac{M^{ref}}{M^{pp}} \tag{15}$$

where it is assumed that the centre of the reference chamber is positioned at the depth of measurement. The alternative use of P_{eff} is also a valid option. All experimental conditions are identical to those for the determination of $N_{D,air}$ in ^{60}Co using in-phantom measurements.

2.4. Use of plane-parallel chambers

The use of plane-parallel ionization chambers both in electron and in photon beams is considered in line with the introduction above.

2.4.1. Reference conditions

In electron beams, reference conditions consider a reference depth z_{ref} (as in TRS-277) instead of the depth of maximum absorbed dose used in other dosimetry protocols. The absorbed dose to water is determined according to

$$D_w(z_{ref}) = M_u N_D s_{w,air} p_Q \tag{16}$$

where

$$p_Q = p_{cav} p_{wall} \tag{17}$$

Note that the perturbation factor p_u in TRS-277 is replaced here by p_Q which is the product of two factors (Eq. (17)). The first, p_{cav}, is the electron fluence perturbation factor, identical to the p_u factor in TRS-277 Table XI. The change in symbol attempts to emphasise that it is exclusively concerned with effects due to the air cavity, rather than the wall material, that is, a correction for the

effect known as *in-scattering* where electron tracks are scattered by the medium towards the air cavity. It is stressed that p_{cav} is strictly known at the reference depth only. The second factor p_{wall} takes into account the lack of backscatter of the back wall material compared to water, and has implicitly been assumed to be unity in electron dosimetry protocols to date. This factor is discussed in detail in Appendix C. It was not possible to make definitive recommendations regarding p_{wall} due to the present lack of consensus in the literature [35, 36]. However, all experimental determinations of perturbation in plane-parallel ionization chambers have effectively been of the overall factor p_Q. Figure 1 shows the values recommended for various plane-parallel chambers in the new Code of Practice.

Regarding water/air stopping-power ratios, new calculations have been performed [39] including the two sets of density-effect in water given in the ICRU-37 electron stopping power tables. They are density-effect corrections according to the Sternheimer's model and the more accurate calculations of Ashley based on semi-empirical dielectric-response functions (DRF). It was argued [39] that for electron energies used in radiotherapy, where the density effect in air is negligible, δ_{DRF}-based water/air stopping-power ratios provide a more accurate set of data. Differences in stopping-power ratios due to the different evaluations of the density effect correction are within 1%. The information on the density-effect correction in the set of values actually in use in TRS-277 and in other dosimetry protocols is, however, ambiguous. A new set of data is provided here based on Ashley density-effect corrections for water, Table II. Compared with the stopping-power ratios in TRS-277 differences are negligible for the most commonly used range of electron energies in radiotherapy, being close to 0.5% at most depths and high-energies (see figure 2). The small change is justified in terms of the lack of ambiguity in the corrections used and higher accuracy of the present set of data. It is interesting to note that if the comparison is made with Sternheimer-based electron stopping-powers, differences would be larger at shallow depths (up to -1.0 % for most energies) and slightly smaller at depths beyond 0.2 r_0.

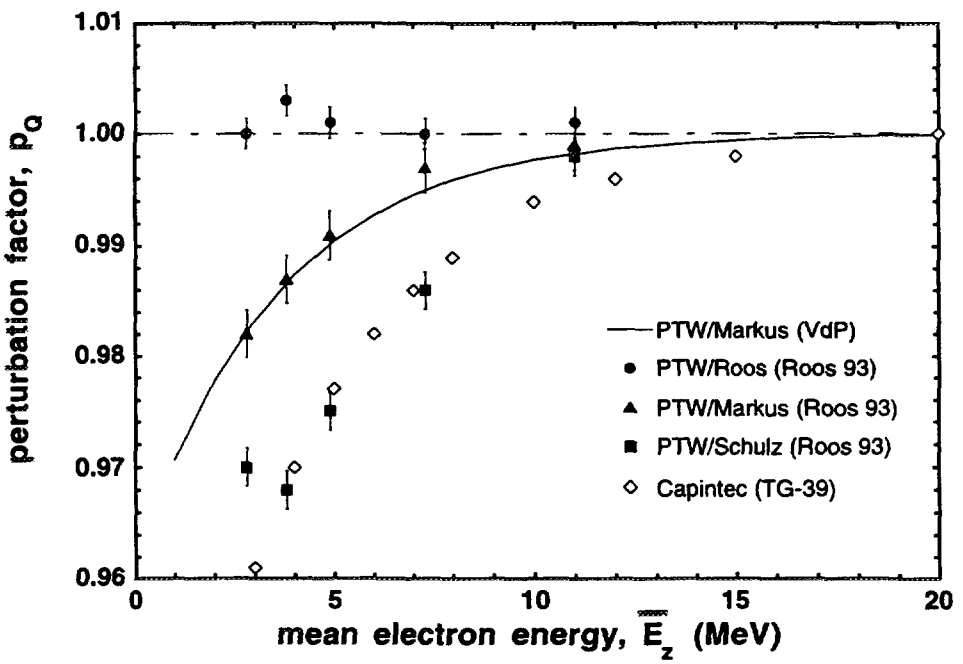

FIG. 1. The variation of the perturbation factor p_Q for several different plane-parallel chambers in common use, relative to the NACP chamber, indicated by the dashed line drawn at p_Q = 1.00. All the measurements were made at the depth of dose maximum and normalized to the quotient test chamber/NACP in a high-energy electron beam. The full line is a fit to 3 separate measurement series on different accelerators using the PTW/Markus chamber [37]. *The filled data points are measurements on three different PTW designs taken from* [38], *and re-normalized so that p_Q = 1 for the NACP chamber; the unfilled symbols are for the Capintec-PS-033 chamber as given in* [14].

As already mentioned, the specification of the "quality" of the electron beam in terms of the mean electron energy at the phantom surface is based on the "2.33 approximation", and stopping-power ratios selected with $s_{w,air}(E_o,z)$ using data from monoenergetic beams. The validity of these two approximations and their limitations is discussed in detail in Appendix B. In particular the influence of electron and photon contamination is demonstrated, showing maximum discrepancies up to 1% at z_{ref} between the $s_{w,air}(E_o,z)$ method and full Monte Carlo simulations. Differences are usually larger at shallow depths due to the difference in slope of the $s_{w,air}(z)$ distributions obtained with the two methods and increase further if analytic expressions yielding $\bar{E}_o$ values larger than the "2.33 approximation" are used [3].

It is emphasized that no accurate method exists today to predict the $s_{w,air}(z)$ dependence on the contamination of the beam unless a full Monte Carlo simulation of the complete accelerator treatment head is performed. On the other hand, the appendix on stopping-power ratios also describes two new methods recently proposed that, used in combination, could perhaps overcome the limitations described above.

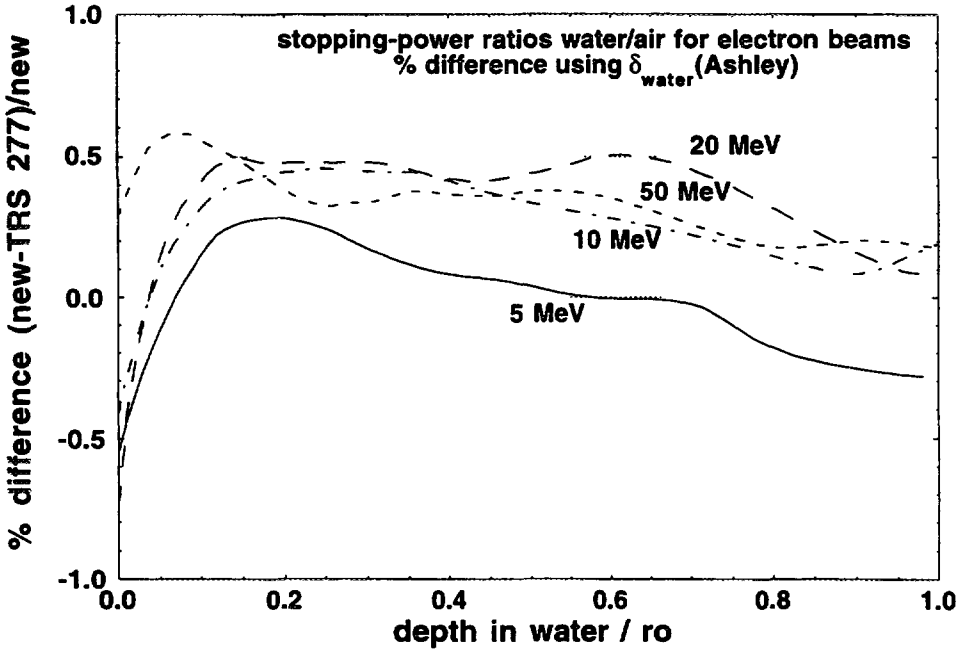

FIG. 2. Percent difference between the new stopping-power ratios for electron beams, $s_{w,air}$, given in Table III and those tabulated by TRS-277 [1] and other dosimetry protocols.

2.4.2. Non-reference conditions

Emphasis is given to the use of plane-parallel ionization chambers in non-reference conditions, especially to determine relative dose distributions.

For electron beams the need to take into account the depth variation of different quantities and correction factors for ion chamber measurements is stressed. This is a significant disadvantage compared with other detectors like TLD, diodes, plastic scintillators, synthetic diamonds, Fricke dosimeters or liquid ion chambers.

A common mistake in the application of TRS-277 for field sizes smaller than the reference field is to determine R_{50} for such fields and use equation (2) or alternative tables to determine $\bar{E}_o$, and then use $s_{w,air}(E_o,z)$ to select stopping-power ratios. As in TRS-277 it should be emphasized here that the validity of equations (2-4) or alternative tables is restricted to large field sizes. Users

TABLE II. SPENCER-ATTIX STOPPING-POWER RATIOS (Δ=10 KeV), WATER TO AIR ($s_{w,air}$) FOR ELECTRON BEAMS AS A FUNCTION OF $\tilde{E}_0$ AND DEPTH IN WATER. Density effect correction (δ_{Ashley}) and I-values from ICRU-37 and electron fluence Monte Carlo calculations from Andreo [39] using the EGS4 Monte Carlo system.

depth in water(mm)	Electron beam energy $\tilde{E}_0$																			
R_p (mm)*	1 MeV 3.6	2 MeV 8.8	3 MeV 14.0	4 MeV 19.1	5 MeV 24.3	6 MeV 29.4	7 MeV 34.5	8 MeV 39.6	9 MeV 44.7	10 MeV 49.8	12 MeV 59.9	14 MeV 69.9	16 MeV 79.9	18 MeV 89.8	20 MeV 99.6	22 MeV 109.3	25 MeV 123.8	30 MeV 147.7	40 MeV 194.1	50 MeV 238.8
0	1.117	1.088	1.066	1.049	1.034	1.026	1.014	1.006	0.998	0.993	0.981	0.969	0.961	0.955	0.948	0.943	0.936	0.924	0.912	0.907
1	1.125	1.096	1.072	1.055	1.040	1.030	1.018	1.010	1.002	0.996	0.985	0.973	0.965	0.959	0.951	0.946	0.938	0.927	0.914	0.908
2	1.131	1.104	1.079	1.060	1.045	1.033	1.022	1.014	1.005	0.999	0.988	0.976	0.968	0.962	0.954	0.948	0.941	0.929	0.915	0.909
3	1.134	1.111	1.085	1.065	1.049	1.037	1.026	1.018	1.009	1.002	0.990	0.979	0.971	0.964	0.957	0.951	0.943	0.932	0.917	0.911
4	1.136	1.117	1.091	1.070	1.053	1.041	1.029	1.021	1.011	1.005	0.993	0.982	0.973	0.966	0.959	0.953	0.945	0.934	0.918	0.912
5		1.123	1.097	1.075	1.057	1.044	1.032	1.023	1.014	1.007	0.995	0.984	0.975	0.968	0.961	0.955	0.946	0.935	0.920	0.913
6		1.127	1.102	1.079	1.061	1.048	1.035	1.026	1.016	1.009	0.997	0.986	0.977	0.970	0.963	0.957	0.948	0.937	0.921	0.914
8		1.132	1.112	1.089	1.069	1.055	1.041	1.031	1.021	1.013	1.001	0.989	0.980	0.973	0.966	0.960	0.951	0.940	0.924	0.916
10		1.135	1.120	1.098	1.077	1.062	1.047	1.036	1.025	1.018	1.004	0.992	0.983	0.975	0.969	0.962	0.953	0.943	0.926	0.918
12			1.127	1.107	1.086	1.070	1.054	1.042	1.030	1.022	1.008	0.995	0.985	0.978	0.971	0.964	0.956	0.945	0.928	0.920
14			1.132	1.116	1.095	1.079	1.061	1.048	1.035	1.027	1.011	0.998	0.988	0.981	0.973	0.966	0.958	0.947	0.930	0.922
16			1.135	1.123	1.104	1.087	1.069	1.054	1.041	1.031	1.015	1.001	0.991	0.983	0.975	0.969	0.960	0.948	0.932	0.923
18			1.137	1.129	1.112	1.095	1.076	1.061	1.047	1.037	1.018	1.004	0.994	0.986	0.977	0.971	0.962	0.950	0.933	0.924
20				1.133	1.118	1.103	1.084	1.068	1.053	1.042	1.023	1.008	0.997	0.988	0.980	0.973	0.964	0.952	0.935	0.925
25					1.128	1.120	1.102	1.086	1.069	1.056	1.034	1.016	1.004	0.994	0.986	0.978	0.969	0.956	0.938	0.928
30					1.133	1.131	1.118	1.103	1.086	1.072	1.047	1.027	1.012	1.002	0.992	0.984	0.974	0.960	0.941	0.931
35						1.132	1.129	1.118	1.102	1.087	1.060	1.038	1.021	1.008	0.998	0.989	0.978	0.964	0.944	0.933
40								1.128	1.116	1.103	1.074	1.050	1.031	1.016	1.005	0.996	0.984	0.969	0.948	0.935
45								1.130	1.127	1.115	1.088	1.062	1.041	1.026	1.012	1.002	0.990	0.973	0.951	0.938
50										1.125	1.102	1.075	1.053	1.035	1.021	1.009	0.995	0.978	0.955	0.940
55										1.127	1.114	1.088	1.065	1.045	1.029	1.016	1.001	0.983	0.959	0.943
60										1.124	1.123	1.100	1.077	1.056	1.038	1.024	1.007	0.987	0.962	0.946
70											1.122	1.120	1.099	1.078	1.058	1.041	1.021	0.998	0.969	0.952
80												1.118	1.118	1.099	1.078	1.060	1.037	1.009	0.977	0.957
90													1.114	1.116	1.099	1.079	1.053	1.022	0.984	0.963
100															1.114	1.098	1.071	1.036	0.993	0.970
120																1.109	1.104	1.065	1.012	0.984
140																		1.095	1.034	0.999
160																		1.099	1.058	1.015
180																			1.081	1.033
200																			1.091	1.053
220																				1.071
240																				1.084

* $R_p = -1.65 + 5.23\,E_p - 0.0084\,E_p^2$, average from Monte Carlo calculations for monoenergetic electrons using the EGS4 and ITS3 systems

should be aware that stopping-power ratios are almost independent of field size, see Figure 3, and using the incorrect approach just described to determine $\bar{E}_0$ will result in stopping-power ratios that correspond to a beam with a different energy.

In photon beams, plane-parallel ionization chambers are not recommended for absolute determinations, but for relative measurements on the central axis only and for output factors. Perturbation factors in photon beams are very sensitive to the details of the construction of a chamber and they cannot be predicted with an acceptable uncertainty. Furthermore, small changes from chamber to chamber in the manufacturing process render invalid the use of "general" factors for chambers of the same make. Plane-parallel ionization chambers should be avoided in very narrow beams such as those used in stereotactic procedures.

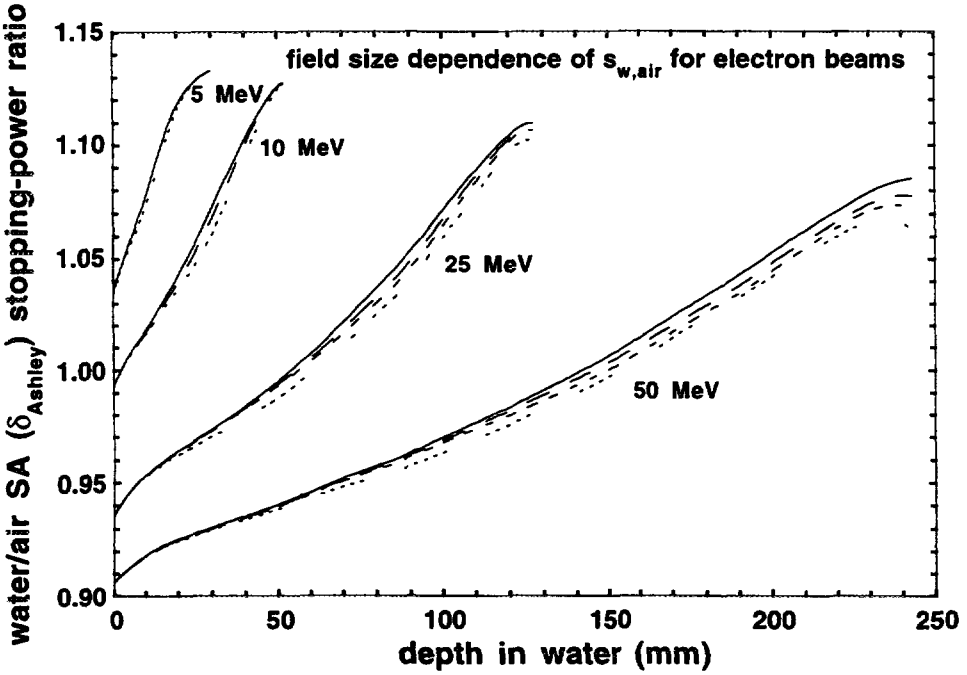

FIG. 3. Field-size dependence of water/air stopping-power ratios for electron beams determined with Monte-Carlo calculations. Radii shown in the figure are: for 5 MeV, 10 mm and broad beam; for 10 MeV, 10 mm, 20 mm and broad beam; for 25 MeV, 10 mm, 30 mm, 50 mm and broad beam; for 50 MeV, 10 mm, 40 mm, 60 mm and broad beam. The solid curves pertain to the broad beams.

3. TESTING OF THE NEW CODE

Tests at two different levels have been proposed to the IAEA by the working group:

- Category A - for checking that the Code is clearly written so that the procedure can be unambiguously carried out from a practical point of view. A comparison with absorbed dose determinations using TRS-277 will be included in this category. The group includes the (obvious) α-test by the authors followed by β–tests performed by independent persons. This category must be carried out before the new Code is published and it should not take more than two months. It should be undertaken by hospital physicists in several centres, some of which should not be in an English-speaking country.

- Category B - for testing that the correct absorbed dose to water is obtained by following the new Code of Practice. This category is a longer term project and represents a significant research project to be undertaken in a sophisticated centre or centres.

4. CONCLUSIONS

The forthcoming IAEA Code of Practice for plane-parallel ionization chambers should improve the accuracy of electron beam dosimetry and, to a lesser extent, of photon beam dosimetry too. Whereas efforts are being made to implement the latest developments in ionization chamber dosimetry, the verification of the Code will show if they are to be preferred to previous methods or to procedures recommended in other recent protocols in the same field. It is hoped that changes in structure, compared with TRS-277, will facilitate the use of the Code.

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AN ALGORITHM TO INCLUDE THE BREMSSTRAHLUNG COMPONENT IN THE DETERMINATION OF THE ABSORBED DOSE IN ELECTRON BEAMS

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Abstract

Currently used dosimetry protocols for absolute dose determination of electron beams from accelerators in radiation therapy do not account for the effect of the bremsstrahlung contamination of the beam. This results in slightly erroneous doses calculated from ionization chamber measurements. In this report the deviation is calculated and an improved algorithm, which accounts for the effect of the bremsstrahlung component of the beam, is suggested.

1. Introduction

None of the existing protocols or codes of practice for high-energy electron dosimetry (NACP 1980/81, AAPM 1983/94, HPA 1985, IAEA 1987, CFMRI 1987, NCS 1989, IPSM 1992, etc.) takes any amount of the bremsstrahlung component always present in electron beams. This results in a systematic error in the derivation of the absorbed dose.

The purpose of this study is to draw attention to this striking omission in dosimetric procedures. An algorithm is proposed for dealing with this deficiency in electron dosimetry.

2. Absorbed dose equation

Routinely, the recommended calibration depth for electrons is at the peak of the depth - dose curve (Klevenhagen 1994). In this position, the ionization chamber is exposed to both electrons and to bremsstrahlung (fig 1) but only the dose due to electrons is accounted for as seen in equation 1.

For the purpose of the beam calibration, an ion chamber is placed in a phantom so that its effective point of measurement is at the reference depth and the absorbed dose to the phantom medium is obtained from the familiar expression

$$D_W = M\, N_D s_{w,air} \pi p_i \qquad (1)$$

where D_W is the absorbed dose to water, M is the mean value of the dosemeter readings corrected for recombination and polarity effects as well as for any differences between ambient conditions at the time of measurement and the standard conditions for which the calibration factor applies, N_D (or its equivalent N_{gas}) is the absorbed dose to air (gas) chamber factor, $s_{w,air}$ is the stopping power ratio water to air, obtained for the appropriate electron beam mean energy and πp_i is the product of the various correction factors applicable to measurement in phantom at user's accelerator.

The calculation of the absorbed dose involves application of several correction factors represented in equ. (1) by the product, πp_i but these will be neglected further on in this analysis as being outside the interest in this case.

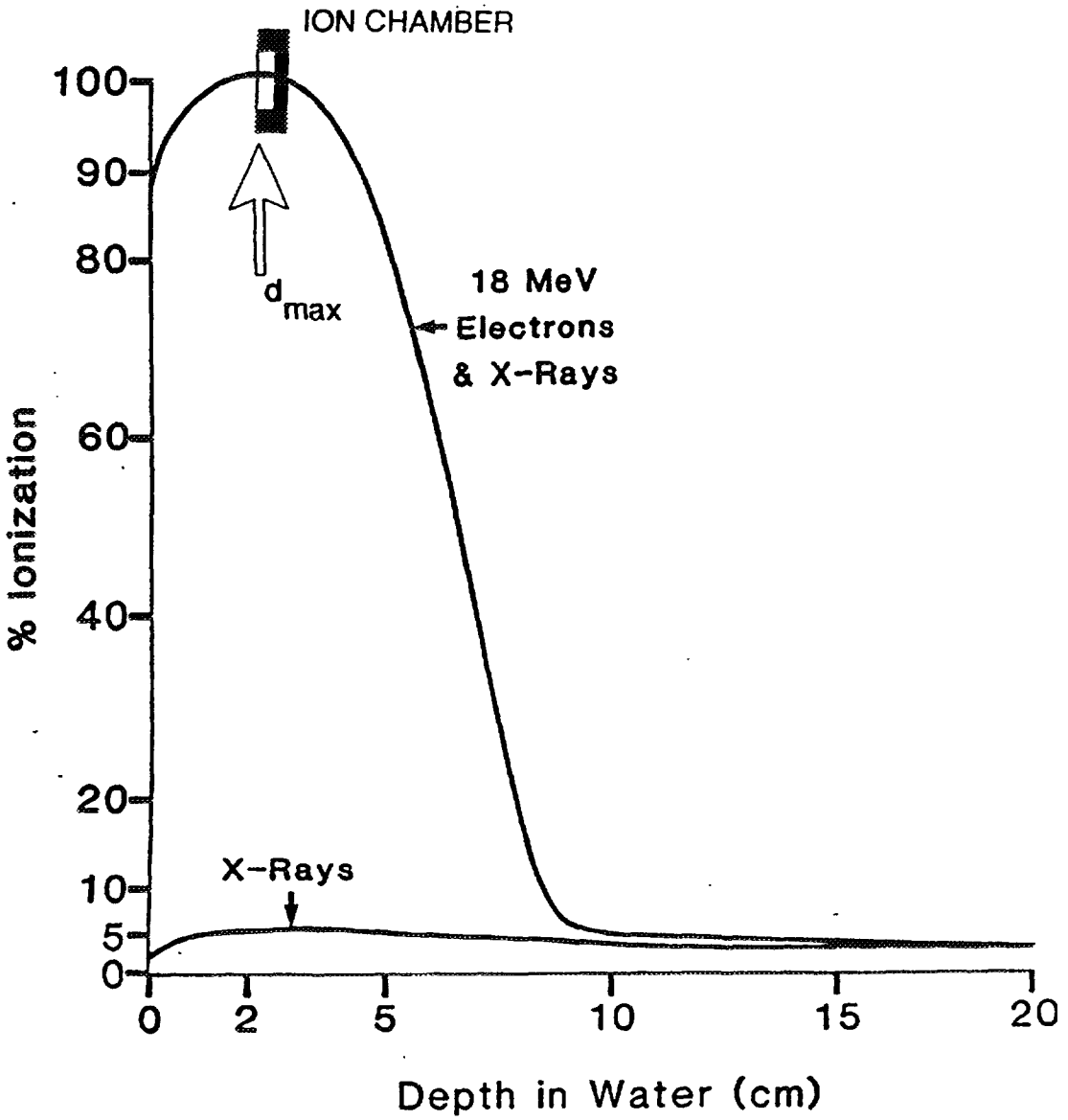

Fig 1 The response of an electron chamber placed at the reference depth in an electron beam is due to both radiation components; the main electron beam as well as to the bremsstrahlung contamination (adapted from Rustgi and Rogers 1987)

2.2. Modified absorbed dose algorithm

To allow for the bremsstrahlung component, equ. (1) may be rewritten as follows

$$D_W = M N_D [(s_{w,air})_{el}(1-\beta)+(s_{w,air})_{br}\beta] \qquad (2)$$

where $(s_{w,air})_{el}$ is the stopping power ratio for the electron fraction of the beam, $(s_{w,air})_{br}$ is the stopping power ratio for the bremsstrahlung fraction of the beam, and β is the bremsstrahlung fraction of the total radiation beam at the absorbed dose determination depth. The discrepancy, δ, in the absorbed dose determination with accounting for and without accounting for the bremsstrahlung component can be defined as

$$\delta = [(s_{w,air})_{el}(1-\beta)+(s_{w,air})_{br}\beta]-(s_{w,air})_{el} \qquad (3)$$

The first factor in the square brackets of equation 3 represents electrons alone, the second factor in the square brackets stands for the bremsstrahlung alone and the last component of the equation represents the total beam but for which only the stopping power ratio for electrons is employed.

3. Estimation of absorbed dose due to bremsstrahlung

The magnitude of the error due to the bremsstrahlung omission can be evaluated using equ (3). The methods of obtaining the required parameters for the calculation of δ are described below.

Stopping power ratios $s_{w,\,air}$ for the bremsstrahlung. Determination of the absorbed dose fraction which is due to the bremsstrahlung component requires the knowledge of the stopping power ratios water-to-air for the photon energy concerned. An approach similar to that used with the conventional high-energy photon beams was adopted, namely using the beam quality index concept and by measuring the tissue-phantom-ratios TPR_{10}^{20} which were performed in the tail of the appropriate electron beam. These TPR values were then used for finding the appropriate water to air stopping power ratios (fig 2) from the data provided by Andreo and Brahme (1986).

Stopping power ratios $s_{w,air}$ for the electron beam. The stopping power ratios employed in the AAPM (1983) dosimetry protocol and featured in many other protocols are considered the best and were used for these calculations. The water/air stopping power ratios derived for the depth of electron dose maximum (reference depth) are given in fig 2 as a function of the mean electron beam energy at surface.

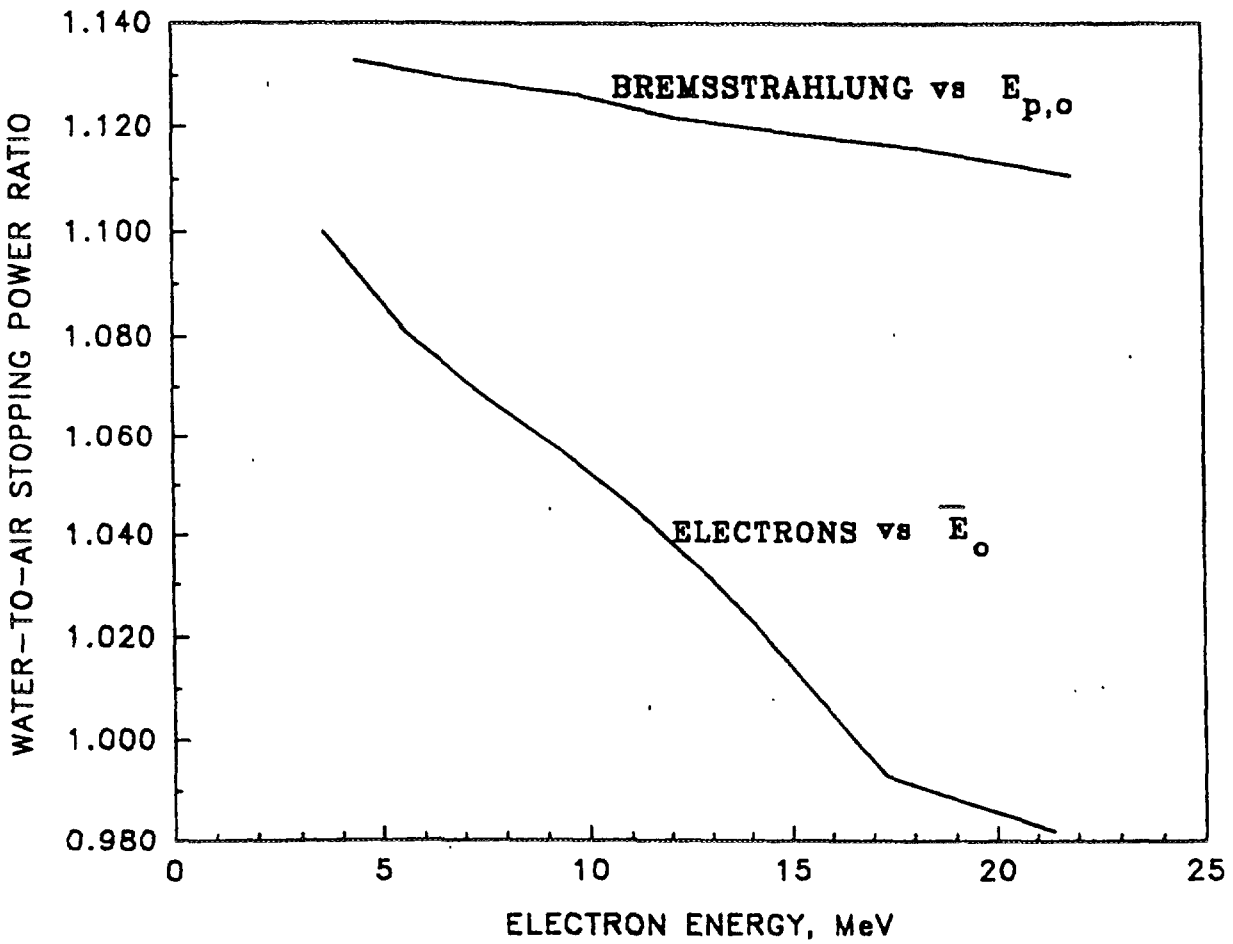

Fig 2 Stopping power variation with electron energy. For the electron beam the energy scale represents the mean electron energy at surface, for the bremsstrahlung component it represents the most probable energy at surface.

TABLE 1. UNDERESTIMATION OF THE ABSORBED DOSE DUE TO OMISSON OF BREMSSTRAHLUNG

BREMSSTRAHLUNG DATA FROM GUR et al 1979			BREMSSTRAHLUNG DATA FROM RUSTGI AND ROGERS 1987		
NOMINAL ELECTRON ENERGY,MEV	% OF X-RAYS AT D_x	UNDER-ESTIMATION OF ABSORBED DOSE	NOMINAL ELECTRON ENERGY, MEV	% OF X-RAYS AT D_x	UNDER-ESTIMATION OF ABSORBED DOSE
			4	1.7	0.05%
6	4.8	0.2%	6	2.2	0.1%
8	5.4	0.3%	8	3.5	0.2%
10	5.9	0.4%	10	3.8	0.3%
12	6.8	0.5%	12	4.2	0.3%
14	7.0	0.7%	15	4.8	0.5%
17	8.2	1.0%	18	5.7	0.7%
20	8.5	1.2%	22	7.5	1.0%

Bremsstrahlung fraction at the peak of electron curve. For the purpose of this analysis one needs to know the bremsstrahlung fraction of the total radiation beam at the depth of the absorbed dose determination i.e. at the dose (ionization) maximum.

For this analysis, the only useful data found on the bremsstrahlung content at the maximum electron depth dose curve are those obtained experimentally by Gur et al 1979 and by Rustgi and Rogers (1987). In both studies magnetic fields were used to deflect the electrons out of the beam allowing the dose from X-rays alone to be measured. The data on X-ray contamination obtained from this work, expressed in terms of the percentage of the maximum electron dose, are given in table 1.

4. Results and discussion

The discrepancy, δ, in the absorbed dose between the determination without considering the bremsstrahlung and determination with inclusion of the bremsstrahlung was calculated using equation (3). The results for the two accelerators expressed as a percentage of the total electron beam (electrons plus bremsstrahlung) are given in table 1.

It is seen that the underestimation of the absorbed dose is small at low electron energies increasing however with energy. In the clinically most relevant energy range between 8 and 15 MeV, δ, varies between 0.3 to 0.5% if both accelerator types are taken into consideration. Above 15 MeV, the discrepancy reaches 1.2%. This is consistent with the data in fig 2 where the stopping powers for the bremsstrahlung and electrons are seen to diverge with increasing energy.

5. Conclusions

This analysis has shown that omission of the bremsstrahlung component in the calibration of electron beams leads to a systematic underestimation of the absorbed dose. This error is comparable in magnitude with the existing correction factors and uncertainties pertaining to electron dosimetry and should therefore be taken account of in the dosimetric procedures.

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CLINICAL DOSIMETRY WITH PLASTIC SCINTILLATORS - ALMOST ENERGY INDEPENDENT, DIRECT ABSORBED DOSE READING WITH HIGH RESOLUTION

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Abstract

Clinical dosimetry is still far behind the goal to measure any spatial or temporal distribution of absorbed dose fast and precise without disturbing the physical situation by the dosimetry procedure. NE 102A plastic scintillators overcome this border. These tissue substituting dosemeter probes open a wide range of new clinical applications of dosimetry.

The scintillation light is transferred by a thin, multifibre plastic light guide to a very sensitive and stable miniature photomultiplier. The Cerenkov light signal generated in the light guide is compensated by differential measurement [Flühs, 1989]. Multichannel photomultipliers allow simultaneous measurement with detector arrays. The plastic scintillator NE 102A is tissue equivalent for all energies of electrons and ß-rays as well as for photons above 100 keV. The detector can be prepared in any size and shape, so a very high spatial resolution can be achieved. This dosemeter system has a large dynamic range. The detector does not have any dependence of its response to dose, dose rate, temperature, pressure, or to the incidence of the radiation, nor show a significant change of response or a significant radiation damage.

This versatile new dosimetry system enables fast measurement of the absorbed dose to water in water also in regions with a steep dose gradient, close to interfaces, or in partly shielded regions. It allows direct reading dosimetry in the energy range of all clinically used external photon and electron beams, or around all brachytherapy sources. Thin detector arrays permit fast and high resolution measurements in quality assurance, such as in-vivo dosimetry or even afterloading dose monitoring. A main field of application is the dosimetric treatment planning, the individual optimization of brachytherapy applicators.

Thus, plastic scintillator dosemeters cover optimally all difficult fields of clinical dosimetry. An overview about its characteristics and applications is given here.

1. Introduction

Radiooncology has done a big step forward by advanced methods of tumortherapy, such as intraoperative radiotherapy, afterloading brachytherapy, stereotactic treatment or conformal therapy, total body irradiation or rotational total skin electron radiotherapy. Improved

imaging technologies for localization, CT- and NMR-based three-dimensional treatment planning, Monte Carlo simulation based kernels for 3D-dose calculation, new real time verification methods, they all contribute to the success of radiotherapy.

However, there are still open questions. Treatment planning algorithms are approximations, only, covering most but not all fields of radiotherapy planning. There is still the need to determine doses in regions without secondary particle equilibrium, in steep dose gradients, in the build-up region, at interfaces, in tissue heterogeneities, in partly shielded organs at risk, or in regions with contaminated or mixed radiation beams.

Clinical dosimetry today is still far behind the goal to measure any spatial or temporal distribution of absorbed dose directly and precise. Clinical dosemeters do not fulfil the general requirements for measurement, the physical situation to be measured must not be disturbed by the measurement procedure itself. Dosemeter probes are not tissue substituting. The size of there sensitive volume is often too large.

2. Plastic scintillator dosimetry

The idea is, to use tissue substituting, direct reading detector probes for dosimetry with high response and high resolution [4]. Plastic scintillators are known since decades, but have been discovered for clinical dosimetry just recently [1-20]. There are many materials available, but the superior characteristics of the NE 102A plastic scintillator detectors indicate this material as an excellent dosimetry probe [4]. It has gained great importance for dosimetry.

2.1. NE 102A - Tissue equivalent dosemeter probe

NE 102A is a plastic scintillator with high light output. The material is almost fully tissue substituting (ρ=1.032 g cm^{-3}, N_{e^-}= 3.39 x 10^{23} cm^{-3}). The detector does not show a significant change of response nor a significant radiation damage, as checked in a long time investigation over one year of continuous irradiation with 1600 Gy by ^{137}Cs-γ-rays [4]. So, NE 102A is suitable as dosemeter probe.

The NE 102A scintillator is well known in nuclear and particle physics for decades. Due to its high response, detector probes can be prepared in nearly any size (e.g. 1 mm^3 for brachytherapy [5-11,13-20]) and shape (e.g. 0.1 mm thin for interface dosimetry [12,13]). Thus, a very high spatial resolution is achievable.

In our research group the dosimetry detector is used since seven years for different dosimetric applications [4]. Its characteristics open new possibilities of dosimetry with many physical and clinical applications [4-20].

2.2. The plastic scintillator dosemeter

The scintillation light is transferred to a photomultiplier by a thin, multifibre plastic light guide, e.g. 16 cladded fibres in a bundle of 2 mm Ø with an attenuation length of 16 m). The Cerenkov and luminescence light signal, generated in the light guide by electrons of energies above 175 keV, is compensated by differential measurement in a bundle of parallel, but blind ending fibres [Flühs, 1989].

As light detector, very sensitive (1 nA/1 mm^3 scintillator volume for 1 Gy h^{-1}), stable miniature photomultiplier tubes are used. Their spectral sensitivity covers optimally the emitted light of the scintillator in the 420 nm range [4,8]. This dosemeter system has a large dynamic range, more than 6 magnitudes of linear absorbed dose rate response in the current mode [4].

Multichannel photomultipliers with 16 channels (in use since two years) or even more separate channels (e.g. up to 256 channels, in preparation) allow simultaneous measurement

with many detectors. This enables linear or other detector arrays for 2D or 3D-dosimetry [5-10,13-20].

2.3. Physical application

The plastic scintillator NE 102A is tissue equivalent in considering the interactions of electrons and ß-rays of all energies as well as for photons above 100 keV (e.g. range of ^{192}Ir), see Fig. 1. Compared to water, the mass absorption coefficient decreases slowly above 10 MeV, due to pair production, while there is a steep decrease between 100 keV and 40 keV, with an almost constant mass absorption coefficient between 35 keV and 5 keV (range of ^{125}I [8]. For protons (e.g. eye tumor treatment) the non linearity of response can be taken into account during calibration.

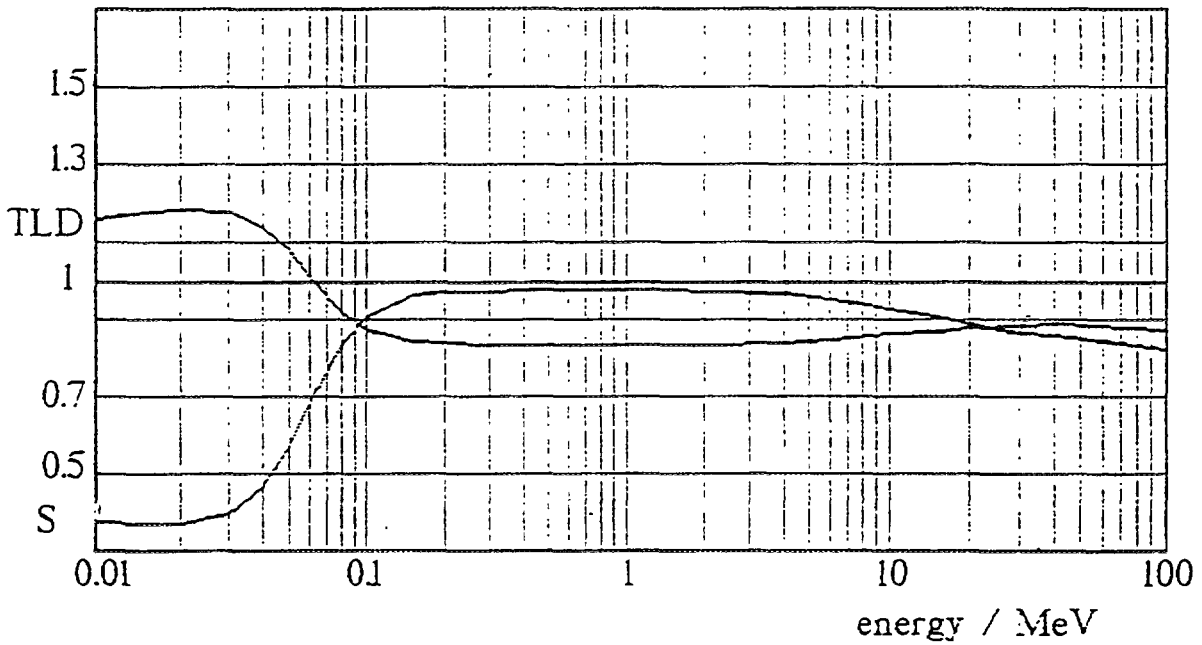

Figure 1. Mass absorption coefficient for NE 102A plastic scintillators (S) relative to water, compared to TLD

The detector does not show any dependence of its response to dose or dose rate (e.g. brachytherapy), to temperature (e.g. intraoperative radiotherapy) or pressure, nor to the incidence of the radiation (e.g. stepping source afterloading).

Thus, the detector is suited for most areas of physical application, dosimetry at steep dose gradients, at interfaces, at the surface, or in tissue heterogeneities like lung substitutes. They can be applied for electrons and ß-rays of all energies, for photons of external beam radiotherapy, as well as for those of brachytherapy.

3. Clinical application

This versatile new dosimetry system allows fast measurement of the absorbed dose to water in water also in regions with a steep dose gradient (e.g. stereotactic treatment, brachytherapy), close to interfaces (e.g. in the build-up region), in tissue heterogeneities (e.g. in lung substitutes) or in partly shielded regions (e.g. the lung as organ at risk in TBI).

3.1. Basic dosimetry

Fast and with high resolution all dose measurements which are needed in basic dosimetry can be performed in a water phantom [4]. Different to ionization chambers, in electron beam dosimetry the absorbed dose is directly indicated. No energy dependent correction is needed. Plastic scintillator dosimetry allows not only direct reading basic dosimetry in the energy range of all clinically used external photon and electron beams, but also around all common brachytherapy sources.

3.2. Dosimetric treatment planning

Dosimetric scanning with a single detector or detector arrays permit fast 2D-dosimetry [5-20], e.g. for ^{192}Ir afterloading applicator optimization [8,9,17-20], or for individual eye plaque preparation using ^{125}I seeds, emitting photons below 35 keV or even 3D-calibration of ^{106}Ru/^{106}Rh ophthalmic ß-ray applicators [5-8,10,13-17,20]. As the brachytherapy applicators can be optimized during a few minutes of measurement, this technique can be used as individual dosimetric treatment planning [5-10,13-20].

3.3. Quality assurance

The small size as well as the possibility to prepare detector arrays allow all measurements needed in quality assurance of external beam and brachytherapy [8,9,14,15-20].

One main interest of application is in-vivo dosimetry. The detector can be placed at the body surface (in the build-up region). The detector array can be positioned in body cavities (no temperature dependence), in tiny catheters, e.g. in stereotactic therapy treatment (no directional dependence of response), or in interstitial ^{192}Ir-afterloading brachytherapy needles (no energy or depth dependence).

A new era of dosimetry is opened by fast, high resolution plastic scintillator dosimetry. Afterloading dose monitoring is possible now [8,9,14-20]. A tiny detector array can be integrated into the afterloading applicator. Due to the high spatial resolution and the high time resolution the stepping source brachytherapy can be monitored independently and directly. To gain this technical information no in-vivo dosimetry is needed. All deviations in step position or step size in dwell time or stepping time can be indicated and measured directly.

4. Conclusion

Besides basic dosimetry the main field of clinical application is the fast 3D-dosimetric treatment planning of brachytherapy applicators (e.g. eye plaques) and the high precision quality assurance of afterloading applications. Thus, plastic scintillator dosemeters cover optimally all difficult fields of clinical dosimetry in radiotherapy.

ACKNOWLEDGMENT

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ABSORBED DOSE BEAM QUALITY FACTORS FOR CYLINDRICAL ION CHAMBERS: EXPERIMENTAL DETERMINATION AT 6 AND 15 MV PHOTON BEAMS

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Abstract

Ion chambers calibrated in terms of absorbed dose to water need an additional factor conventionally designed by k_Q in order to determine the absorbed dose. The quantity k_Q depends on beam quality and chamber characteristics. Rogers [1] and Andreo [2] provided calculations of the k_Q factors for most commercially available ionization chambers for clinical dosimetry. Experimental determinations of the k_Q factors for a number of cylindrical ion chambers have been made and are compared with the calculated values so far published. Measurements were made at 6 MV and 15 MV clinical photon beams at a point in water phantom where the ion chambers and a Fricke dosimeter were alternatively irradiated. The uncertainty on the experimental k_Q factors resulted about ±0.6%. The theoretical and experimental k_Q values are in fairly good agreement.

1. Introduction

The usual dosimetry procedure in which the ionization chambers are calibrated in terms of air-kerma is based on a rather complex formalism and requires the determination of several parameters the uncertainty of which is not low. Thus systematic uncertainties and mistakes in data handling may be not negligible. Moreover the primary standards upon which this procedure is based are, everywhere, graphite cavity chambers. This makes it difficult to detect possible systematic uncertainties as these standards are based on the same measurement procedures. At present time however the majority of national standards laboratories have developed primary standards of absorbed dose to water, D_w, for the Co-60 gamma radiation. These standards are based on different methods so that their intercomparisons [3] make it possible to detect possible systematic errors. The calibration in terms of D_w is based on a formalism that is conceptually more straightforward than that relevant to the air-kerma standards. Moreover the parameters entering into the expression of D_w can be determined, on principle, with a better accuracy. When expressed in terms of the absorbed-dose-to-water calibration factor, the dependence of D_w on the photon or electron beam quality is described by a factor originally denoted [4] as k_Q. The uncertainty on this factor may result not necessarily low if k_Q is calculated from the value of the single parameters upon which it depends.

Andreo [2] and Rogers [1] computed the k_Q values for many ionization chambers with an uncertainty of up to 1.5% about. The possibility of achieving better accuracy levels in absorbed dose determination is important in deciding whether or not to adopt calibrations in terms of D_w. The present work was then undertaken with the aim of developing an experimental procedure to determine the k_Q values with an accuracy appreciably higher than that resulting from the calculation of these factors. The expression of k_Q can be easily obtained just from the definition of absorbed-dose-to-water calibration factor, N_w, from which one has:

$$D_w = M N_w \tag{1}$$

where M is the chamber reading corrected for ambient conditions and ion recombination.
If the factor N_w is determined at the Co-60 calibration beam, equation (1) holds only at this quality. On the other hand the expression of D_w to which the dosimetry protocols refer for measurement at any beam quality, Q, is that based on the air-kerma calibration factor, N_K, (e.g. IAEA protocol [5]) and is given by:

$$D_w = M N_D \left(s_{w,a} \Pi p_i\right)_Q \tag{2}$$

where

$$N_D = N_K (1 - g) k_m k_{att} k_{cel} \tag{3}$$

the other symbols having the usual meaning [5].
Equation (2) holds at any quality, including the calibration quality, c. Then for the same ionization chamber having both the calibration factors N_w and N_D and irradiated at the calibration quality, one obtains from equations (1) and (2):

$$N_D = N_w \left(s_{w,a} \Pi p_i\right)_c^{-1} \tag{4}$$

Finally from equations (2) and (4) one finds:

$$D_w = M N_w \frac{\left(s_{w,a} \Pi p_i\right)_Q}{\left(s_{w,a} \Pi p_i\right)_c} = M N_w k_Q \tag{5}$$

where

$$k_Q = \frac{\left(s_{w,a} \Pi p_i\right)_Q}{\left(s_{w,a} \Pi p_i\right)_c} \tag{6}$$

The factor k_Q corrects the calibration factor N_w for beam quality dependence in radiation beams different from the Co-60 gamma-ray.

2. Experimental equipment

The experimental determination of k_Q for some ionization chambers was made at two photon beam qualities, 6 and 15 MV, obtained from a Siemens Mevatron MD Dual Photon linear accelerator. Type and characteristics of the ionization chambers are described in table I. Chamber calibrations were performed at the Co-60 gamma-ray quality using a gamma irradiator mod. AECL Eldorado 6 with a 60 TBq Co-60 source. The chambers were irradiated in water phantom with their individual waterproof sheath. Chamber sheaths were made of 0.5 mm thick PMMA accurately machined to minimize the air thickness between the chamber wall and the sheath. Each sheath was suitably designed to assure chamber cavity venting even during measurement in water phantom. Two different water phantoms were used. Measurements at accelerator vertical beam were performed in a phantom realized with a 30 cm side open

Table I. Characteristics of the cylindrical chambers used in this work, according to the data taken from the manufacturer datasheet.

chamber type	wall material	wall thickness ($g\ cm^{-2}$)	internal length (mm)	internal radius (mm)	nominal volume (cm^3)	central electrode material
NE 2571	graphite	0.065	24.1	3.15	0.69	aluminium
NE 2561	graphite	0.090	9.22	3.675	0.325	aluminium
Capintec PR-06C	C-552	0.050	22	3.2	0.65	C 552
PTW M233642	PMMA	0.090	6.5	2.75	0.125	----
ENEA	graphite	0.087	20	2	0.24	graphite

ended cubic tank with 1 cm thick PMMA wall. The phantom for accelerator vertical beam was also provided with two additional supports for two small cylindrical waterproofed chambers. This pair of chambers was positioned symmetrically with respect to beam axis. The aim of these two supplementary chambers was to accurately monitor beam output and flatness. The water phantom for irradiation at the horizontal Co-60 gamma beam was similar to the previous one but was provided with a thin (3 mm) PMMA window on the side wall. All measurement data were processed in line by a computer interfaced with the electrometers and with the temperature and pressure probes. Measurements at Co-60 gamma ray were done with a 10 x 10 cm^2 field size at 100 cm SDD. The Co-60 gamma-ray dose rate at the measurement point in phantom was about 300 mGy/min. Measurements at accelerator photon beams were made with higher dose rates of about 1700 mGy/min (6 MV) and 1600 mGy/min (15 MeV), respectively, at 100 cm SDD. The photon collimator setting at accelerator

beams was always 10 x 10 cm^2. The accelerator beam quality was specified by the $TPR_{20/10}$ parameter. The $TPR_{20/10}$ was determined by depth ionization curves in water phantom. The ion chamber was kept in fixed position at 100 cm SDD and the water level was varied according to the procedure recommended by IAEA [5]. The $TPR_{20/10}$ was 0.673 and 0.757 for 6 MV and 15 MV photons, respectively. At all beam qualities the ionization chambers (including the two monitor chambers) were positioned with their centre at the reference depth. At the Co-60 gamma beam the reference depth was 5 cm. For the accelerator beams the reference depths were 5 cm and 10 cm for the $TPR_{20/10}$ of 0.673 and 0.757, respectively. Absorbance measurements for ferrous sulphate dosimetry were made by a precision double beam spectrophotometer, mod. Cary 1, interfaced to a computer for data acquisition and processing.

3. Methods

The experimental determination of the k_Q factors for the ionization chambers listed in table I was based on measurements with ferrous sulphate (Fricke) solution. When used as an absolute measuring method, Fricke dosimetry does not have sufficient accuracy if one needs to keep the uncertainty below ±1%. The solution adopted in this study was to design an experimental procedure in which the knowledge of absorbed dose to water had not to be needed for determining the k_Q factors. Accordingly, the only important characteristic required for the Fricke solution was a response reproducibility adequate for the objective here considered. The Fricke solution prepared for this investigation was thoroughly tested for several months and was irradiated in PMMA vials. The choice of the vial material for Fricke dosimetry is not immediate. Glass vials can be more easily cleaned thus allowing a more simple control of the storage effects. However the PMMA was definitely chosen because for this material the wall correction factor for possible non water equivalence is very close to one [6]. Measurements of absorbance were made on both the irradiated samples and a set of not irradiated controls. PMMA cylindrical vials were used. They were realized with 0.5 mm wall thickness, 10 mm internal diameter and 30 mm length. In order to remove residuals of substances used for the preparation of the vial (glue, PMMA particles, etc.), all the vials underwent a treatment consisting of an immersion for 30 minutes in an ultrasound bath, followed by several rinsings with distilled water and by a pre-irradiation at about 5 kGy. With the procedure above described, a reproducibility of better than 0.4% (1SD) on 5 samples was obtained with irradiations of not less than 30 Gy. A ferrous sulphate solution of about 5 liters was prepared and stored in a dark place at stable temperature in a glass container. Homogeneity and stability of the solution were periodically checked for a period of about 1 month during which the Fricke irradiations for the present study were performed. The time of permanence of the solution in the vials (about 2 hours) was kept constant and as short as reasonably possible. If necessary, in order to limit the storage time, the solution was, at the end of the irradiation, transferred to accurately purified glass ampoules. This was done mainly for irradiations carried out at accelerator beams, located far from the authors' laboratory. The storage

time in each vial was the same for both irradiated samples and controls. The centre of the vial was positioned at the measurement point in water phantom. At the same depth in phantom the ionization chambers under investigation were positioned with their geometrical centre coincident with the measurement point. The procedure to determine experimentally the k_Q factors was based on the following rationale: the absorbed dose to water at any beam quality Q as measured with a calibrated Fricke solution is given by:

$$(D_{w/F})_Q = (\Delta A_F)_Q N_{w/F} \qquad (7)$$

where $(\Delta A_F)_Q$ is the difference in absorbance between the irradiated and unirradiated solution, and $N_{w/F}$ is the calibration factor of the Fricke solution in terms of absorbed dose to water. The factor $N_{w/F}$, determined at the Co-60 gamma radiation, is assumed to be constant since no evidence of its dependence on electron energy (at least in the energy range of interest in this study) appears from the data so far available (e.g. ICRU 35 [7]) . If an ion chamber calibrated in terms of D_w is used, the absorbed dose to water at any beam quality Q is given by equation (5). In the same irradiation conditions the calibrated Fricke solution and the calibrated ion chamber must measure the same D_w value when alternatively positioned at the same phantom depth. Accordingly, by the equations (5) and (7) one has:

$$D_w = M_Q N_w k_Q = (\Delta A_F)_Q N_{w/F} \qquad (8)$$

On the basis of their definition the calibration factors N_w and $N_{w/F}$, determined at the calibration quality c, are given, respectively, by:

$$N_w = \frac{(D_w)_c}{M_c} \quad \text{and} \quad N_{w/F} = \frac{(D_w)_c}{(\Delta A_F)_c} \qquad (9)$$

Finally from equations (8) and (9) one obtains:

$$k_Q = \frac{(\Delta A_F)_Q}{(\Delta A_F)_c} \frac{M_c}{M_Q} \qquad (10)$$

where M_C and M_Q are the corrected chamber readings due to the absorbed dose giving rise in the Fricke solution to the difference in absorbance $(\Delta A_F)_c$ and $(\Delta A_F)_Q$, respectively.
According to the above procedure and to equation (10), the k_Q factor can be obtained as function of quantities that can be determined experimentally with fairly good accuracy.
Five series of measurements were made for each ionization chamber and beam quality. After one month these measurements were then repeated by repositioning the experimental setup. It was thus possible to check the long term reproducibility regarding the stability of measurement and the

mechanical equipment. Measurements with Fricke solution were made by five irradiations on five distinct vials for each beam quality and then taking the mean value of the five readings. The absorbed dose for ferrous sulphate irradiation was about 30 Gy. Also for Fricke dosimetry the series of measurements were repeated after one month about to assure the reproducibility of the overall dosimetric method.

4. Results and discussion

The experimental values of k_Q determined at two photon beam qualities are reported in table II for the ionization chambers listed in the first column.

The results obtained in this work have been compared with the values computed (according to equation 6) by Andreo [2] and Rogers [1]. The differences among the k_Q computed values in table II are due to differences in the sets of parameters considered by those authors for their calculation. Andreo used stopping power ratios recalculated by himself and other parameters from the IAEA protocol [5]. Rogers used the basic parameters either from the AAPM [8] or the IAEA [5] protocols, respectively. The above authors adopted in their calculations the "p_{repl} approach". For data comparability the same approach was then used in this study. Measurements by ionization chambers in water phantom were made by positioning the chambers with their centre at the measurement point. Therefore the experimental k_Q factors of this work include the correction factor for water replacement, p_{repl}. To allow the comparison, the data by the other authors reported in table II were interpolated to refer to the same $TPR_{20/10}$ values used in this work. For the ENEA ionization chamber, the reference chamber of the italian AIFB dosimetry protocol [9], the theoretical k_Q factors were not computed by Andreo and Rogers but by the present authors, using the same basic data considered by those authors. The average deviation between the present results and the computed data of Andreo and Rogers, respectively, is about 0.6%. This shows that all these determinations are rather consistent. Although the AAPM parameters are not the most updated, the data based on the AAPM parameters (column 3) result to be slightly closer to the experimental k_Q values, than the other computed data. The experimental data either of this work or of other authors are always slightly higher than the computed ones. Actually, it should be noted that all the data in table II are within the stated uncertainties. The uncertainty on the k_Q factors is due to the experimental uncertainty of the individual quantities entering into equation (10). Accordingly the overall random uncertainty on the k_Q values determined in this study is about ±0.6%.

The few experimental determinations of k_Q so far available are those determined at NPL [10] and PTB [3]. The comparison with the present results is shown in table III. With respect to the results of the present work the data from NPL and the data from PTB at $TPR_{20/10} = 0.673$ are in fairly good agreement, whereas a deviation of about 0.8% is obtained for the PTB result at $TPR_{20/10} = 0.757$. To this regard it should be taken into account that the NPL and PTB data are based on measurement of the parameters, as D_w or N_w, directly entering into equation 5. This procedure is not the most accurate because of the intrinsic uncertainty on D_w. Moreover primary

Table II. Experimental and calculated k_Q factors for the various chambers used in this work.

$TPR_{20/10} = 0.673$

chamber type	experimental this work	calculated Andreo 1992	calculated Rogers 1992-a	calculated Rogers 1992-b
NE 2571	0.998	0.993	0.995	0.992
NE 2561	0.999	0.994	0.995	0.992
Capintec PR-06C	0.999	0.995	0.998	0.993
PTW M233642	0.987	0.989	0.994	0.989
ENEA	0.998	0.993	0.996	0.994

$TPR_{20/10} = 0.757$

chamber type	experimental this work	calculated Andreo 1992	calculated Rogers 1992-a	calculated Rogers 1992-b
NE 2571	0.983	0.980	0.983	0.978
NE 2561	0.982	0.982	0.983	0.979
Capintec PR-06C	0.985	0.980	0.986	0.977
PTW M233642	0.970	0.975	0.981	0.972
ENEA	0.983	0.979	0.985	0.979

standards of D_w in the various national laboratories have at present deviations of up to 1% [11-12]. Therefore the deviations among the data in table III are well within the experimental uncertainty. As pointed out by Andreo [2] also the computed k_Q values have a not low uncertainty of about ±1.5%. These computations, based on equation (6) cannot be on principle very accurate because of the intrinsic uncertainties on the individual parameters in this expression.

The results of this work are instead based on measurements that can be performed with the highest accuracy and this procedure seems rather promising for determining the k_Q factors at any beam quality.

Table III. Comparison of the experimental kQ factors for the NE 2561 ionization chamber.

$TPR_{20/10}$	this work	NPL Owen 1993[10]	PTB, in Boutillon 1993[3]
0.673	0.999	0.9973	0.998
0.757	0.982	0.9789	0.990

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DISPLACEMENT CORRECTION FACTOR VERSUS EFFECTIVE POINT OF MEASUREMENT IN DEPTH DOSE CURVE MEASUREMENTS AT ^{60}Co GAMMA RAYS

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Abstract

The discrepancies in data sets of values of the Displacement Factor p_d recommended by different codes of practices for calibration purpose still demand further investigation to clarify this point. In this paper, we propose an experimental method to determine the displacement factor for cylindrical ionization chambers (thimble chambers) in photon beams. Measurement of p_d for several depths were performed for ^{60}Co gamma rays. From these results we calculated the shift of the effective point of measurement (z-z_{eff}) for different depths. The results obtained in this work shown : (a) there is no significant change in p_d from 2 cm to 17 cm of depth in water; (b) the value of p_d for a ion-chamber Farmer type (inner radius r = 3.15 cm) is p_d = 0.988; (c) the shift of the effective point of measurement has a smooth variation with depth; (d) the value of (z-z_{eff}) at the recommended calibration depth for ^{60}Co beams (5 cm) is 0.6r (with r: inner radius of the chamber). The result (b) confirms the value of p_d suggested by the SEFM and NACP protocols and differs with that of the AAPM. The value obtained for (z - z_{eff})(d) is very closed to that recommended by the IAEA TRS-277. Finally, the results (a) and (c) suggest that it should be preferable to use the displacement factor instead of effective point of measurement to perform measurements of depth dose curves, since the use of z_{eff} should take into account its dependence on depth.

1. INTRODUCTION

The Bragg-Gray theory relates the ionization produced in a *small* gas-filled cavity placed in an homogeneous medium to the *absorbed dose* in the medium. However, the particle fluence in the cavity of a ionization chamber inserted in a medium will no longer be representative of the fluence at the point of interest in the medium, because of the complex differences in attenuation and scattering of radiation due to the *replacement* of the medium material by the cavity material. The displacement factor $\mathbf{p_d}$, is one method of correcting for such perturbation, and so that it could be determined as the following quotient, for a given radiation quality, **u** :

$$p_d = \frac{J_{air,\,u(0)}}{J_{air,\,u(r)}} \qquad (1)$$

where $J_{air,u(r)}$ is the mean ionization in the air of an air cavity of radius r, and $J_{air,u(0)}$ is the mean ionization in an air cavity small enough ($r \cong 0$) to neglect the effect of the perturbation.

An other method of correcting for this effect is to define an effective point of measurement, z_{eff}, radially shifted from the geometrical centre of cylindrical ion-chambers through the front.

In order to have consistency in these definitions, both correction factors that are essentially the same perturbation correction, are related by the following equation :

$$p_d = \frac{PDD(z)}{PDD(z - (z - z_{eff}))} \qquad (2)$$

where z is the depth of the chamber axis (geometrical centre); z_{eff} the depth of the effective point of measurement of the chamber; and **PDD(z)** is the percent depth dose.

There are significant discrepancies between the values of the displacement factor recommended by different authors and protocols. For example, the AAPM TG-21 [1] suggests the use of the values calculated by Cunningham and Sontag [2], which differ by about 0.5% from those measured by Johansson *et al.*[3] at ^{60}Co. These last values of p_d are recommended by SEFM [4] as well as NACP[5], and they are consistent with a shift of the effective point of measurement of $(z - z_{eff}) = 0.6r$. IAEA TRS-277 [6] recommend (for ^{60}Co) to use the value $(z - z_{eff}) = 0.5r$, assigned to Johansson *et al.* [3]; however, in that paper, the average of $(z - z_{eff})$ for different depths is 0.6r whiled there is no value of 0.5r for any depth. To clarify the discrepancies in the existing data sets for displacement factors, by means of experimental or theoretical techniques, is very important, mainly if calibrations in terms of absorbed dose to water are to be adopted in the near future.

2. MEASUREMENT OF THE DISPLACEMENT FACTOR p_d

2.1. Method

The displacement factor might be obtained from the equation (1) by measuring the mean ionization $J_{air,u(r)}$ in cavities with radius r, for different values of r, and extrapolating to r = 0. Then,

$$p_d = \frac{\lim_{r \to 0} J_{air,u(r)}}{J_{air,u(r)}} \qquad (3)$$

To perform the experience exactly as this formula suggests, it would be necessary to have several ion chambers with different internal radii but the same wall material and thickness. However, because of our experimental limitations, we propose a method for the determination of p_d based on the measurement of the ionization in a chamber located in the centre of several cavities with different diameters. The main assumption we made is that the ionization measured by the chamber will accuse the variations in the mean ionization of the air cavity, i.e. the variations in the ionization inside the cavities as varying their radii due to the lack of phantom material.

Our experiment was carried out in a solid phantom (acrylic) in order to achieve high reproducibility in our measurements. The ion chamber was a NE 2571 ($0.6cm^3$ and graphite wall). The phantom was constructed with 10 slabs of acrylic 200mm x 200mm x 20mm. One of these slabs has a cylindrical hole in order to allow the placement of different accessories. Figure 1 shows a scheme of these devices in the phantom. In this way, it is possible to ensure an accurate position of a thimble chamber in the centre of a cavity whose diameter could be varied by changing the acrylic accessory cylinder opposite to the ion-chamber. Such cylindrical accessories consist in acrylic rods with wells of different diameters in one of its extremes in order to provide several cavities with different diameters surrounding the thimble chamber, as it is also shown in figure 1. To obtain electronic equilibrium (necessary for dose measurements in air) a graphite cap 2mm thick (same wall material) was made which fits the thimble diameter.

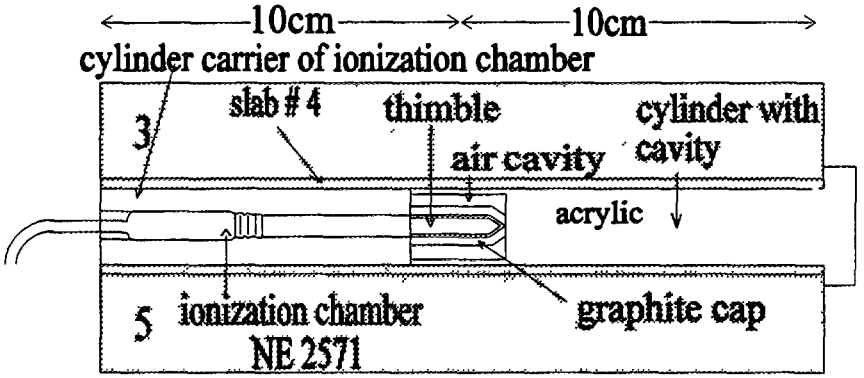

Fig. 1. Lucite phantom with accessories and ionization chamber.

Three cavities with radii r = 11.1, 12, and 16 mm were made to place in the phantom successively (mentioned wells in one end of the rods). The geometrical conditions of the irradiation were those recommended by the AAPM protocol [1] for calibrations in acrylic phantoms.

2.2. Results

The values of the displacement factor versus the radii of the air cavities obtained in this work (at ^{60}Co gamma rays) are shown in figure 2 together with the values given by Cunningham [2] and Johansson [3].

Graphs of p_d vs. r were obtained at several depths in the phantom: z = 1, 3, 5, 7, 9, 11, 13 and 15 cm. The curve shown in figure 2 corresponds to the average for all the mentioned depths, except that of z = 1cm because for this one the cavity breaks through the build-up region.

To illustrate the measurements of p_d vs. depth (z), figure 3 show these values for r = 3.5 mm.

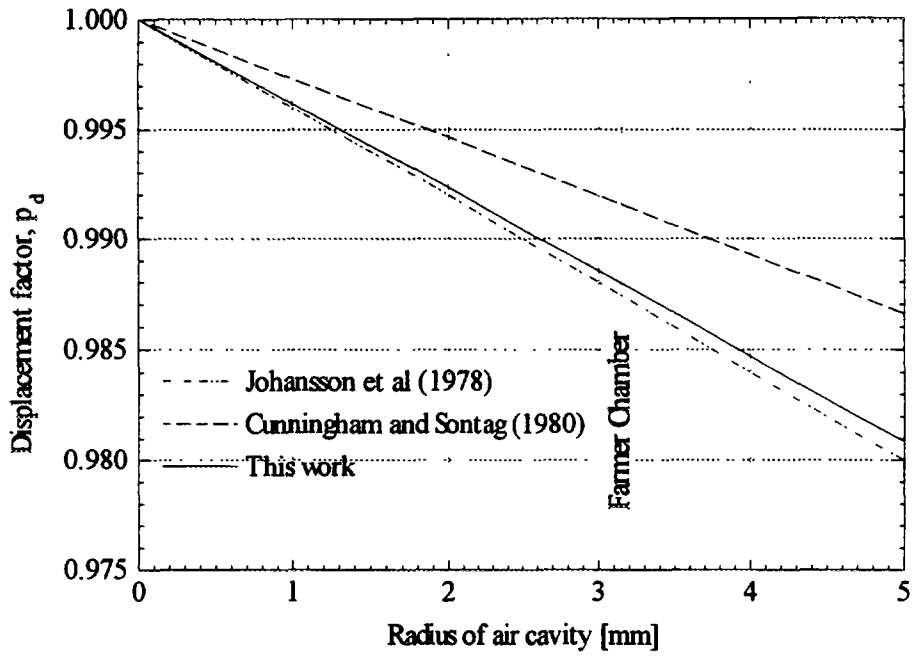

Fig. 2. Displacement factor vs. radius of air cavity for ^{60}Co.

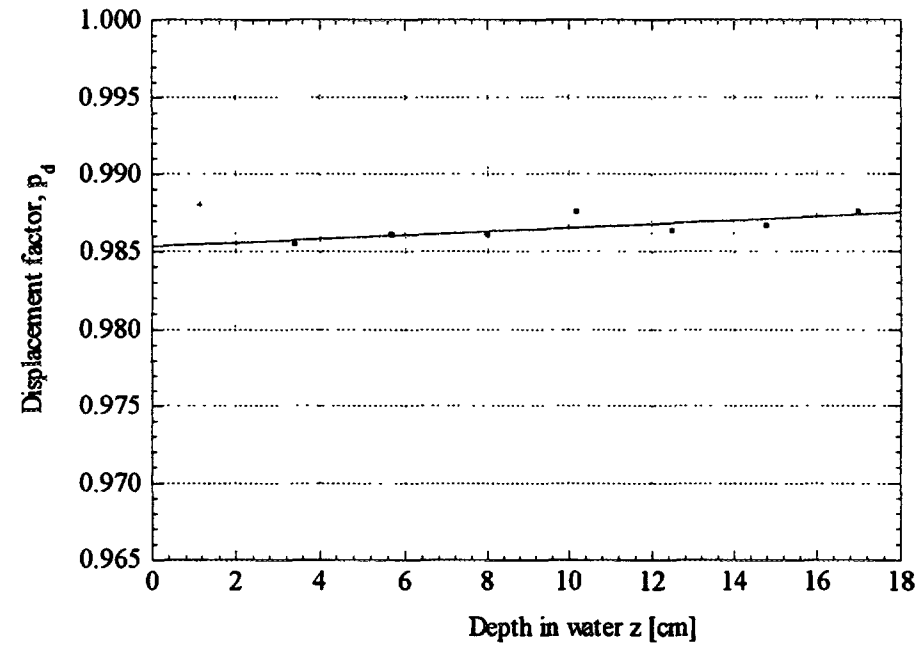

Fig. 3. The displacement factor vs. depth in water for a cavity of radius r = 3.5mm

3. STUDY OF THE SHIFT OF THE EFFECTIVE POINT OF MEASUREMENT

3.1. Method

The radial shift of the effective point of measurement was calculated from the values of the displacement factor pd obtained in this work, and using equation (2). The values of percentage depth doses were taken from the Br.J.Radiol. Supl.17 (1983) [7], for the corresponding field size and source-surface distance at ^{60}Co gamma rays. In our experience, DFS = 80 cm and reference field size 10cm x 10cm.

Since there is no significant change of p_d with z from 2cm to 17cm of depth in water, we used just one value of p_d for all depths.

3.2. Results

The values of the shift of the effective point of measurement versus depth is shown in figure 4.

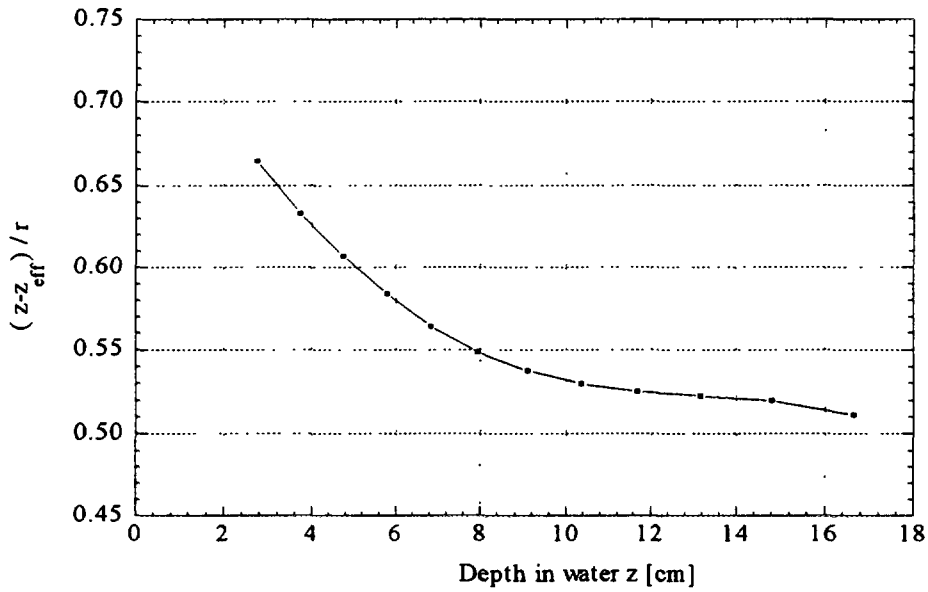

Fig. 4. The shift of the effective point of measurement (z-zeff) calculated from the displacement factor p_d (equation 2) and depth dose data in BJR suppl.17 [7].

4. DISCUSSION.

As it can be observed in fig.2, the values of p_d obtained in this work are in good agreement with those from Johansson *et al.* but there are differences higher than 0.5% with Cunningham & Sontag. Our results about no evidence of significant variation of p_d with phantom depth (see fig. 3) agree with those from Johansson.

From the conclusions mentioned above it is clear that the shift of the effective point of measurement has to vary, although smoothly, with phantom depth, as it is shown in Fig.4.

Consequently, to perform measurements of depth dose curves at depths greater than z_{max} (depth of maximum dose) it would be preferable to use the displacement factor p_d instead of using z_{eff}, since the last one should take into account its dependence on depth.

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AN ANALYSIS OF SOME ASPECTS OF THE ATTENUATION - SCATTER FUNCTIONS IN BRACHYTHERAPY DOSIMETRY

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Abstract

An analysis is presented of the attenuation-scatter functions radial dose functions) employed in brachytherapy dosimetry which accounts for the interplay between attenuation and scattering along the radial distance from the source. Some of the characteristics of these functions are still not established with certainty and are subject of misinterpretation. Such issues like whether they should be normalized or not, particularly in relation to the currently employed source strength specification in terms of air kerma, are not as yetagreed. In the literature, the functions are presented either as normalized or non-normalized but the differences between them are wrongly interpreted as being due to either computational or experimental uncertainties. Furthermore, there is uncertainty about the attenuation-scatter ratio very close to the brachytherapy sources and , in the case of some functions, at larger radial distances. Although the function's value at close distance may seem of lesser dosimetric relevance, it is important if one wants the underlying physics to be correct. These problems were studied in this analysis on the basis of the available data. An experiment was also carried out in order to determine the scatter component in the close vicinity to the source. The study is based on the data for Iridium-192 but the discussion and conclusions are relevant to all types of brachytherapy sources. It is concluded in this analysis that; i) it is incorrect to be comparing the normalised with non-normalised functions. ii) only non-normalised (the natural) functions such as that derived by Meisberger et al (1968) or Sakelliou et at (1992) are correct for dose calculation systems based on the recommended air kerma source specification iii) the function should not have a value of unity at $r = 0$ because of the scatter domination over attenuation in the space around the source and iv) the Van Kleffens-Star function is in error at larger radial distances.

1. INTRODUCTION

A number of workers have investigated the behaviour of radiation in water in the vicinity of brachytherapy point sources, applying the results to the dosimetry of nuclides. In particular, large interest has been shown in the relationship between attenuation and scatter along the radial distance from the source because of its significance in the dose calculation algorithm. In general, scattering compensates partially for attenuation, the combined factor, A-S, attenuation-scatter function varying in magnitude slowly with distance and energy. This function is also known as radial dose function.

In the days of radium therapy and until the early 1960s the attenuation-scatter contribution to dose was mostly ignored, only inverse square law was considered relevant. With the development of interest in the radium substitutes for brachytherapy this situation has changed. This analysis involves work done from this period onwards.

The published functions up to now have been derived either experimentally or theoretically. Although they all show similarity in overall pattern of behaviour, they differ in details. Several papers written on the subject deal with the comparison of the functions obtained by the various workers with the conclusion that the differences between them, particularly evident close to the source, are due to the experimental or computational uncertainties.

One striking feature to note is that the A-S functions published so far are presented either as normalised to unity, usually at 1 cm from the source, or non-normalised. Thus there are normalized and non-normalized A-S functions. This is an important characteristic and is a contributor to the differences in functions in the vicinity to a source both in values and shape of the curves (see fig.1). Because the numerical differences are not large, not exceeding 5%, they are misinterpreted (Glasgow 1981 Thomason and Higgins 1989, Thomason et al 1991) as being due to data uncertainties. This view is questioned in this analysis.

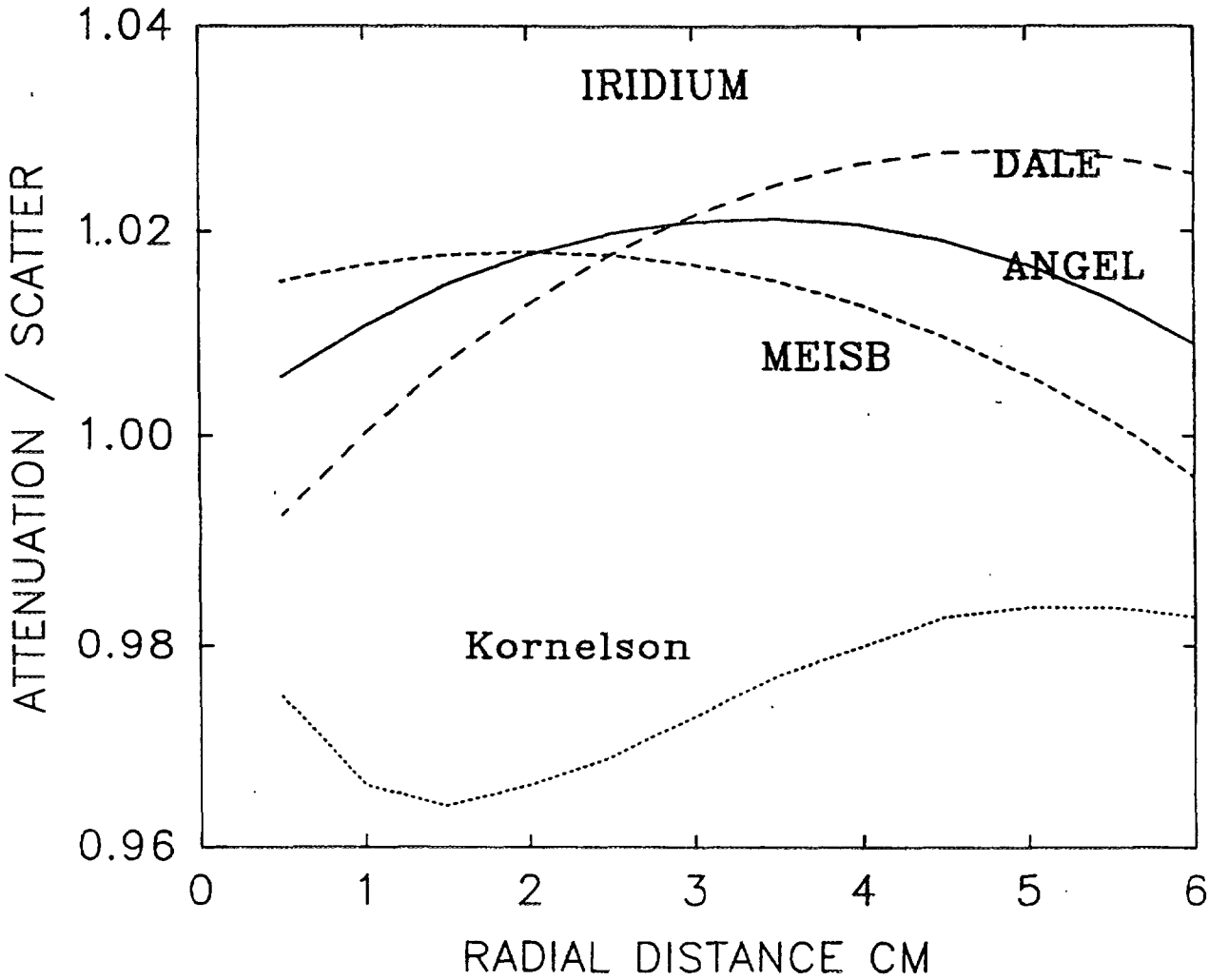


Fig. 1. Attenuation/scatter functions by various authors shown close to the brachytherapy source

The tendency to normalize arises from the "hang-ups" in the approach to dosimetry in the past when the radiation output constants (K-factor, specific gamma ray constant etc.) used to be referred to a distance of 1 cm. Dale (1983), in his Monte Carlo work, introduced the source dose constant concept (SDC for source strength specification which includes scatter component. This indeed involved radial function normalisation but this method of source specification found no use in dosimetry.

This subject has been discussed in several papers in which a comparison of the various available data have been analyzed (Glasgow 1981, Thomason and Higgins 1989, Thomason et al 1991, Sakelliou et al 1992, Parker and Almond 1992). However, the normalized and non-normalized functions were treated as if they belonged to the same set of data.

The purpose of this analysis is to consider the following aspects i) is normalisation, either at 1 cm or at zero distance, of the A-S function correct and consistent with the interpretation of physical phenomena involved, ii) is it correct to used any of the functions normalized at I cm in dosimetry based on the currently recommended air kerma calibration, and iii) is the Van Kleffens-Star function (1979), which differs in pattern from other functions, correct at larger distances.

2. PUBLISHED ATTENUATION - SCATTER FUNCTIONS

The division of the A-S functions into two groups: non-normalized and normalized will be used as a criterion in the review of the published work.

2.1 *Non-Normalized Functions*

One of the first investigation concerning radium substitutes was performed by Meredith et al (1966) who studied the attenuationscatter relationship for six brachytherapy isotopes for distances between 2 and 10 cm. The results were presented in numerical form and not normalized.

Meisberger at al (1968) used diffusion theory to calculate the ratio of exposure in water to exposure in air as a function of distance (1 to 10 cm) from isotropic point sources of several radionuclides. They then determined the mean values of the ratio from experiments by several authors and took the average of the theoretical and mean experimental values to derive curves which could be described by an empirical formula. This was the first mathematical function of this kind defined and found useful for computational purposes.

Thomason and Higgins (1989) derived a radial dose function defined as the ratio of the measured dose in water TLD technique) to the calculated dose in air (source activity, exposure constant, f-factor). A fit to a third order polynomial yielded a formula similar to the Melsberger's. For Iridium the function is unity at 5 cm having a value of 0.968 or 0.984 in platinum and steel encapsulation respectively at 1 cm distance. In making a comparison with Melsberger and Dale work the authors concluded that the differences are indistinguishable when considering uncertainties involved.

In 1992, Sakelliou et al carried out comprehensive Monte Carlo calculations of dose distribution around seven most popular brachytherapy sources for spherical phantoms of 15 and 20 cm radii. A polynomial expression has been derived for the A-S functions which show a very close agreement with the Meisberger polynomial. The important feature of their results is that the functions have not been normalised. That means that the function for a given isotope is unity at a distance at which the attenuation is compensated by scattering. That depth is a characteristic feature of a given isotope.

It is unfortunate, however, that in deriving the polynomial expression Sakelliou and colleagues set the first coefficient of the polynomial to unity without paying attention to the physical phenomena occurring in the close vicinity to source. This forces the polynomial to have a value of 1 at $r = 0$ for all nuclides considered. The Van Kleffens and Star function is discussed in detail in section 5.

2.2 *Normalized Functions*

One of the first Monte Carlo calculations of radial dose distribution was performed by Webb and Fox (1979) for several gamma-emitters. The data were normalized to unity at 1 cm Subsequently, they were used by Kornelson and Young (1981) as a basis for formulating an analytical attenuation-scatter function of a type suggested by Evans (1955) for an absorbed dose build-up. Thomason et al (1991) calculated radial dose factors (A-S functions) by Monte - Carlo for Ir-192 and Cs-137 normalizing at 1 cm and described them as undistinguishable from the Dale (normalized) and Meisberger (non-normalized) curves within the precision of the data.

Melgooni and Ravinder Nath 1992 produced radial dose functions for several isotopes by Monte Carlo code CYLTRAN. The functions were tabulated having been normalized at 1 cm. Thus for Ir 192 the function in solid water has the value of 1.00 at 1 cm rising to 1.15 at 2 cm.

3. WHICH APPROACH IS CORRECT, NORMALIZED OR NON-NORMALIZED?

One has to accept that the sensible way of describing the strength of a brachytherapy source is by the determination of its radiation output, an approach long established in external beam therapy. The practice of specifying a source by a quantity defining its radioactive content (i.e. activity or the milligramme radium equivalent etc) is being phased out from use.

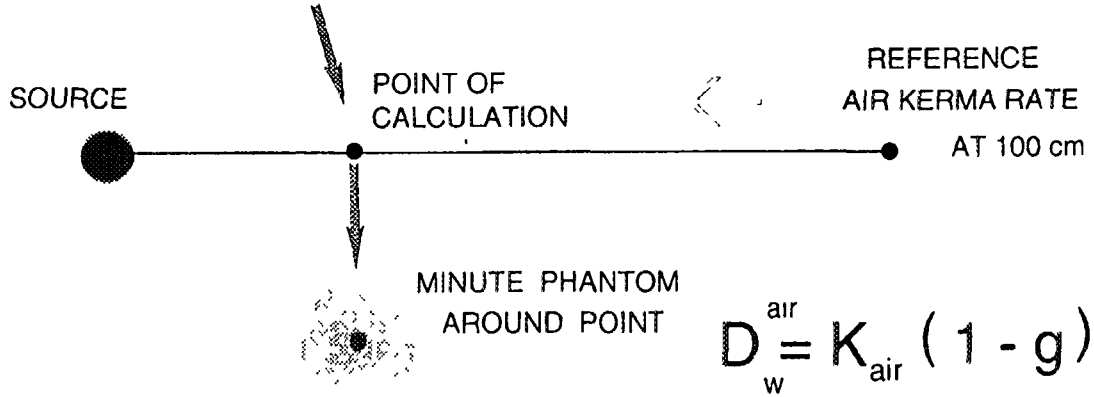
The quantity recommended currently for the specification of the source is the air kerma rate at reference distance but its exact definition differs slightly by involving the square of the reference distance (CFMRI 1983 ; AAPM 1987) or not involving distance (BCRU 1984; ICRU 1985; NCORD 1991; BIR/IPSM 1993 This particular aspect of the definition is, however, irrelevant to this discussion. What is relevant is that this type of source specification does not incorporate the scatter radiation.

Helpful in this discussion is to consider the dosimetric situation shown in fig 2. This model is designed to focus attention on the issue of attenuation/scatter and disregarding other factors such as source anisotropy, effect of encapsulation, oblique filtration, inverse square law etc. Let's consider a calculation point close to the source the strength of which is expressed in terms of air kerma rate in air (or in vacuo as in BIR/IPSM 1993).

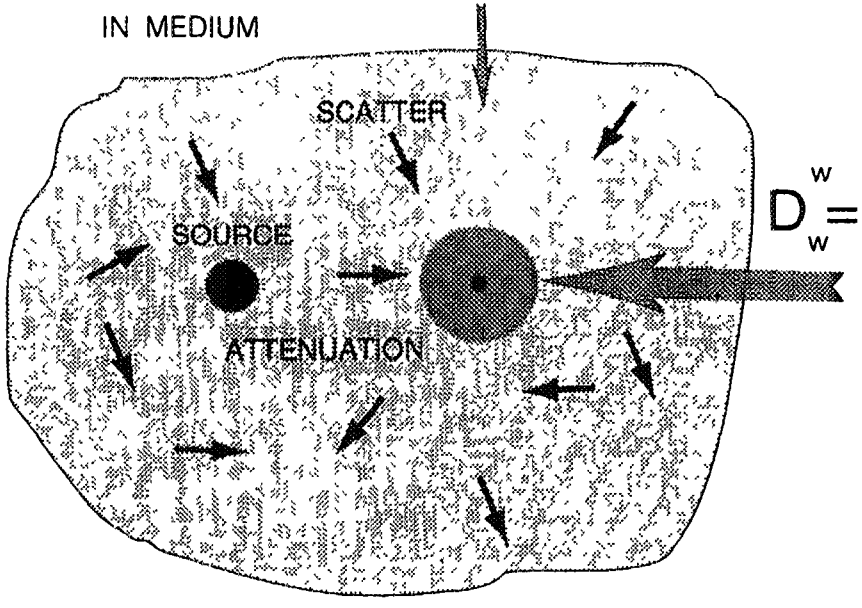
As the next step in the dosimetric procedure, the air kerma may be converted to absorbed dose to water assuming that the point of calculation is surrounded by a small water phantom of minimal mass of air. Theoretically, this does not involve scatter production or attenuation between the source and the point of calculation since the situation is considered "free in air".

Let's imagine now that the source is immersed in water or is placed in tissue. The dose at a point of interest will now be altered being affected by attenuation over the distance concerned. At the same time the scatter radiation will appears throughout the irradiated medium having a compensating effect on attenuation at short distances. Over larger distances, attenuation will become the dominant effect suppressing the compensating effect of scatter. Where they are in balance, the function will have a value of unity. At any other distance the attenuation/scatter function which is used as a multiplying factor in the dose calculation will have a value different from unity.

1. FREE IN AIR $K_{air} = X \frac{W}{e} \frac{1}{1-g}$



2. $D_w^{air} = K_{air}(1-g)$



3. $D_w^w = K_{air}^w (1-g) \frac{(\mu_{en}/\rho)_w}{(\mu_{en}/\rho)_{air}} (A\text{-}S)_r$

Fig. 2. Application of the attenuation/scatter function in a dosimetric system based on air kerma calibration

The conclusion from this discussion is that the attenuationscatter function whether it has been derived by theory or measurement must not be normalised to unity on purpose at one arbitrarily chosen distance, particularly not close to the source where absorption and scatter are not in balance.

4. THE FUNCTION VALUE AT ZERO DISTANCE

An interesting issue is also the value of the function at zero radial distance Although this information has more theoretical than practical value it is important to consider for the physical correctness of the attenuation-scatter function.

Unfortunately, there is no reliable information published on the A-S function value at a distance r = 0. This is understandable considering the experimental difficulties at distances close to the source. The closest measurements reported so far were made at 1 cm from the source. It is surprising, however, that the Monte Carlo calculations published up to now do not cover this region and do not provide solution to this problem. In the theoretical work by Sakelliou et al (1992) the polynomial A-S functions have a value of unity at r = 0 by the virtue of setting the first component of the polynomial to 1. The question is whether this is correct. No direct answer can be found to this among the published data either calculated or measured.

Some deductions can be, however, made considering a hypothetical dosimetric situation shown in fig 3. This assumes a minute point like source and a point of absorbed dose determination positioned at a vey close vicinity. Under such circumstances, attenuation between the source and the calculation point is negligible. The scatter component, however does exist

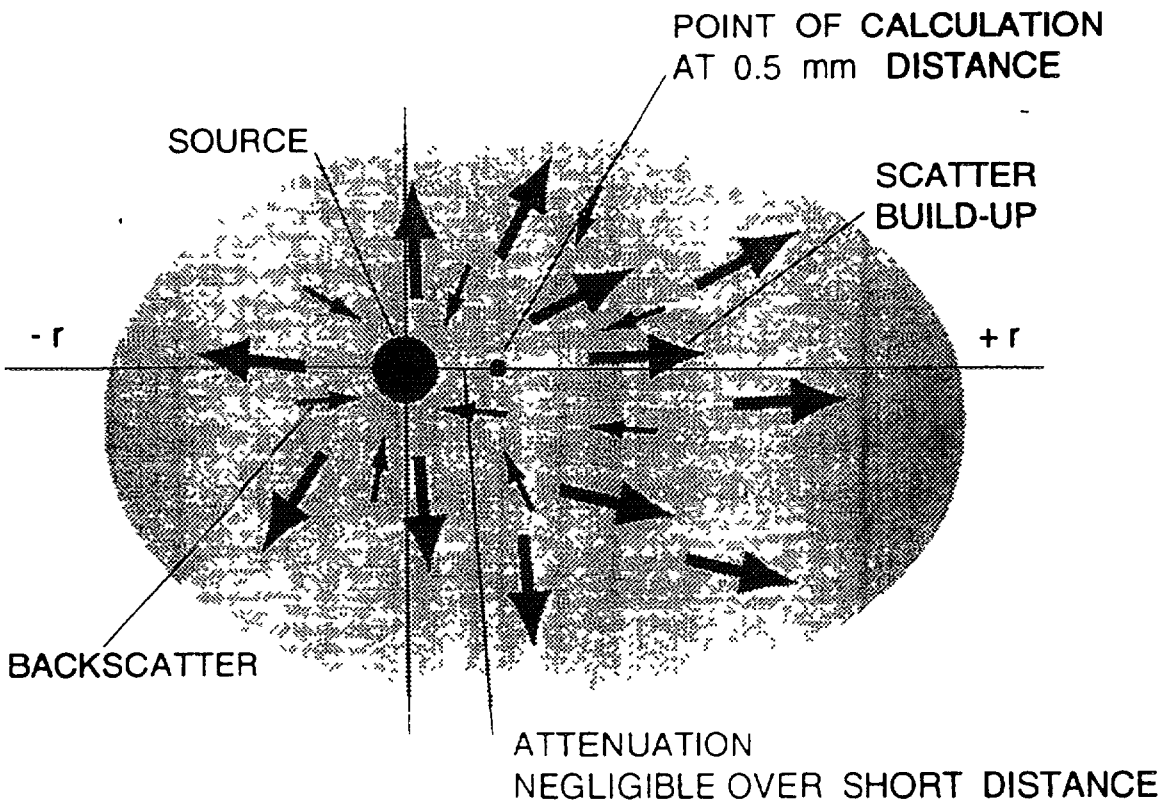


Fig. 3. Factors affecting the dose in the imediate vicinity of a brachytherapy source include forward scatter (large arrows) and backscatter (small arrows), the attenuation being negligible.

since the source is surrounded by a medium. There would be some backscattered radiation reaching this area. Thus at a point very close to the source, and indeed at r = 0, there would the unattenuated direct radiation plus the backscattered radiation arriving from -r direction. Thus on physics grounds, the attenuation-scatter function can be expected to have a value larger than unity. One can of course consider this issue also in terms of monte Carlo technique where the calculation model involves volume cylinders around the source.

4.1 *Theoretical and Experimental Evidence.*

An attempt was made to find support in published data for the above discussion and an experiment was designed to measure the scatter radiation in the immediate vicinity of brachytherapy sources.

4.1.1 *Monte Carlo.*

The computation by Thomason et al 1991) provided data on the scatter component calculated as the fractional scatter along the radial distance from the source starting from 1 cm onwards. By fitting a polynomial expressions to the Thomason's Iridium and caesium curves (fig 4) the data for closer distances can be derived. The fractional scatter (*FS*) variation with distance from source for Iridium was found to fit the following expression

$$FS_{Ir} = 0.0289 + 0.1034\ r - 4.732\ 1\ 0\text{-}3r^2 \quad (1)$$

and for caesium

$$FS_{Cs} = 0.0411 + 0.0596\ r - 1.875\ 1\ 0\text{-}3r^2 \quad (2)$$

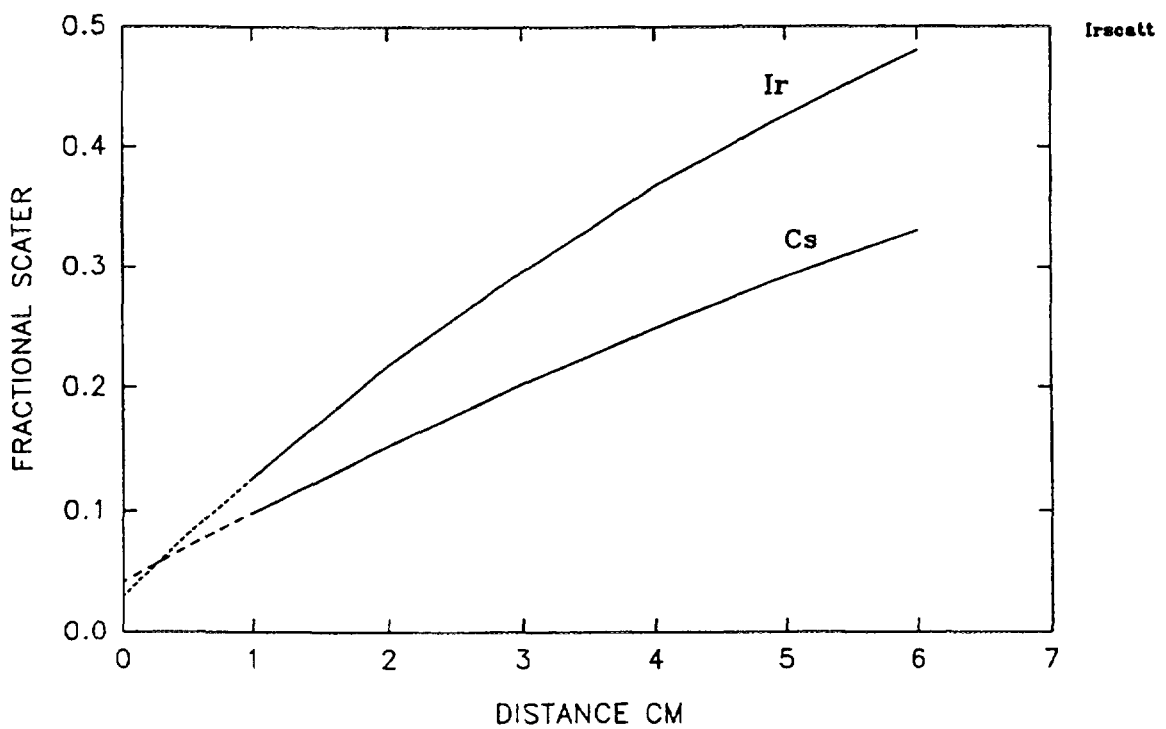


Fig. 4. Fractional scatter contribution to dose as a function of distance for Iridium and caesium sources (Thomason et al 1991 extrapolated by this author to r = 0 cm)

The first coefficients in these equations yield the fractional scatter value at r = 0 which, as it is seen, amounts to 2.9 and 4.1 percent for an Iridium and caesium sources respectively. This clearly indicates that the A-S function should have, at this point in phantom, a value larger than 1.0. This scatter component can be thought of as representing the build-up of scatter in forward direction (+r) due to the backscatter arriving from -r direction (fig 3 towards the origin of the calculation grid which coincides with the source geometric centre.

4.1.2 *Experimental Evidence*

Measurements in the immediate vicinity of the source are very difficult to perform and there is no method of sufficient resolution and accuracy. Nevertheless, an attempt in this work was made to measure the scatter-attenuation function very close to the source. For this purpose, a miniature ionisation "well" shape chamber was designed (fig 5). The chamber is a cylinder 7 mm in diameter and 15 mm in height with a sensitive volume formed by two concentric graphite

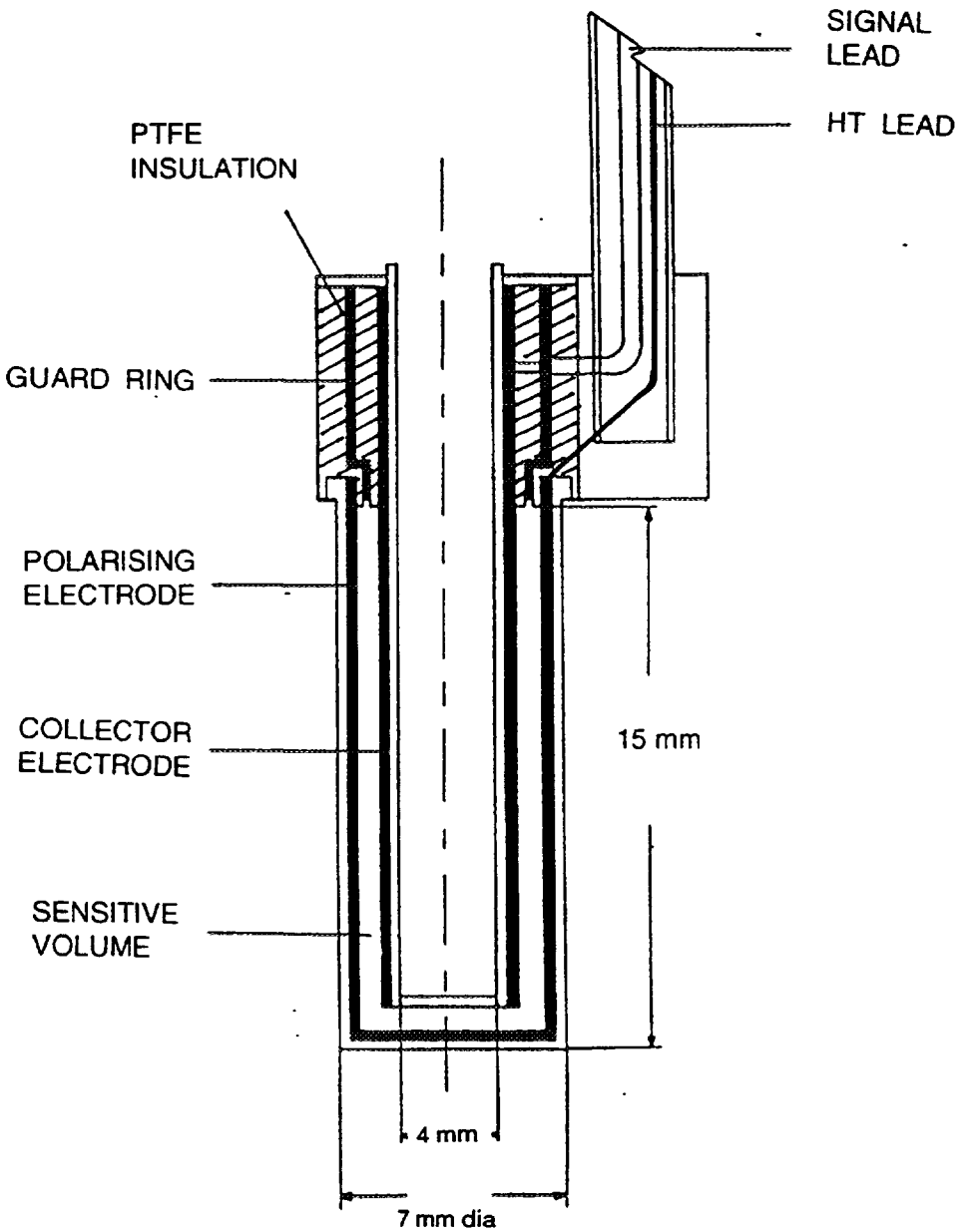


Fig. 5. Schematics of a miniature well ionisation chamber designed to measure the A/S function at close distance

cylinders (polarising and colecting electrodes) separated by 1.5 mm air space creating the sensitive volume. The brachytherapy source was dropped, during the measurements, into the chamber central channel. The geometric source centre was about 2.75 mm from the center of the chamber sensitive volume enveloping the source. This design facilitated close distance measurements of the attenuation/scatter function.

A number of conflicting experimental parameters had to be reconciled in such a difficult design. Because of its small dimensions, the brachytherapy sources of conventional (low) activity could not be measured with sufficient resolution i.e. high enough signal-to-noise ratio. The chamber could not be moved with distance, thus only one measurement point was possible obtain. The construction of the chamber, because of its miniature size is very difficult and is expensive.

Using this chamber, the measurements of the attenuation-scatter function for an - HDR Iridium-192 source were carried out by determination of the ratio of air kerma in water to air kerma in air using a 40 x 40 cm water phantom. The mean value of the A-S function obtained in 24 determinations was found to be 1.012 with a percentage standard deviation of the mean of 0.3% .

Thus, there is about 1.2 % scatter at the distance of 2.75 mm from the source centre. That is a low value compared to the expected 2.9 % on the basis of the lfhomason curve. The difference can be explained by the imperfect "in air" measuring conditions. The chamber body itself produces scatter as well as the environment in which the measurement were performed. Therefore the in air kerma value is not the perfect free-in-air value resulting in an underestimation of the measured A-S factor. Nevertheless, the measured value is a conclusive evidence of the scatter dominance over attenuation at this distance.

5. THE ATTENUATION-SCATTER FUNCTION AT DEPTH

A function of interest is the Van Kleffens and Star (1979 function used in the Selectron HDR treatment planning software Nucletron, Netherlands). The function was published without giving details about the method of its derivation.

Initially the function was presented without the parameter, δ, and it produced a curve not exceeding unity at distances- close to the source. Since this was subsequently found to be in variance with the Meissberger polynomial, the δ was added lifting the initial part of the curve to t-he value of 1.018.

The Van Kleffens - Star function used for Iridium has the following form;

$$(A\text{-}S)_r = \delta\,(1 + \alpha\, r^2)/(1 + \beta r^2) \qquad (3)$$

where for Iridium $\alpha = 0.0$, $\beta = 0.0006\ cm^{-2}$ and $\delta = 1.018$. The parameters for the original equation were limited to Cs, Co and Ra, for Iridium they were added at a later stage.

At depth, the slope of the Van Kleffens function differs significantly from all other functions. This is seen in fig 6 for Iridium in comparison with Meisberger and Sakelliou. The shape of the function is sensitive to the value of the arbitrary parameter P which is seen in fig 7. This suggests that this particular function may be an inappropriate algorithm to describe the attenuation/scatter relationship as required in a high doserate Iridium treatment planning system. This was also noted by Park and Almond (1992).

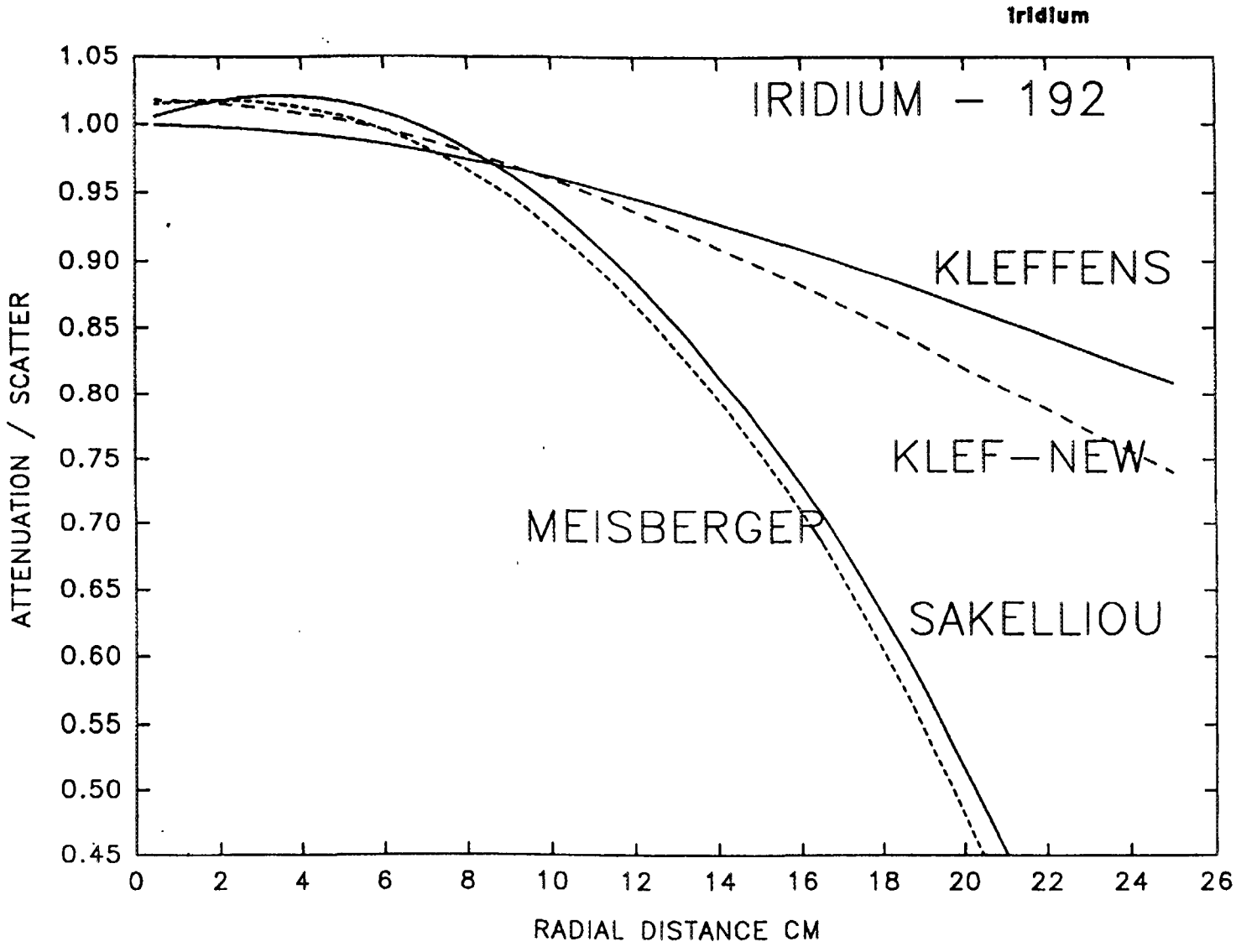


Fig. 6. Comparison of the Van kleffens-Star function with Meisberger and Sakelliou for Iridium

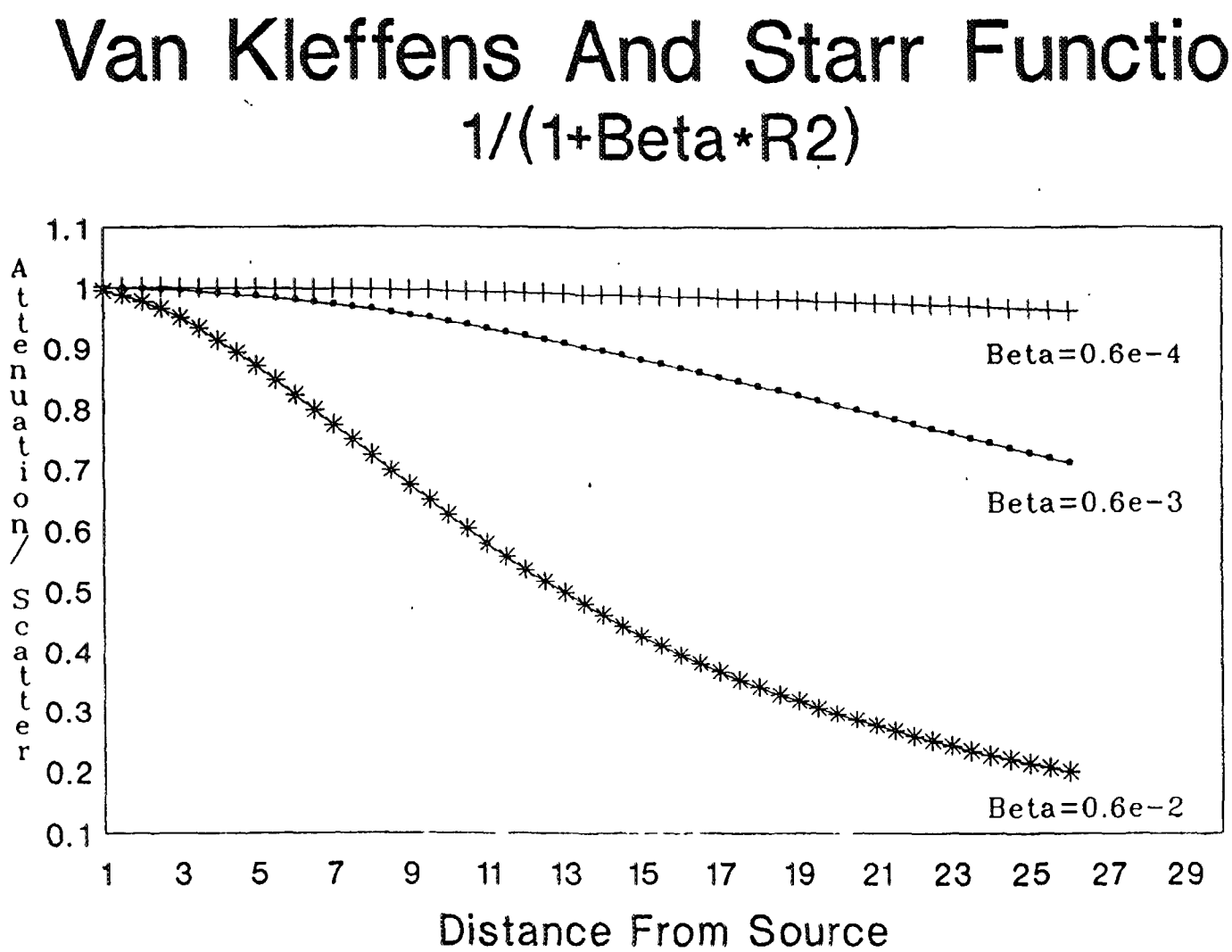


Fig. 7. Dependence of the shape of the Van Kleffens curve on the value of the parameter P.

6. CONCLUSIONS

It is concluded in this analysis that;

i) It is incorrect to be comparing the normalised with non-normalised functions asigning the difference to computational or experimental uncertainties
ii) Only non-normalised functions such as that derived by Meisberger et al (1968) or Sakelliou et at (1992) should be used in dosimetry systems based on air kerma source specification. However, functions from both studies are uncertain at zero distance.
iii) The A-S function, for isotopes such as Iridium and caesium, should not have a value of unity at r = 0 because of the dominance of scatter over attei-iuation close to the source
iv) The Van Kleffens-Star function is in error at larger radial distances.
v) The study is based on the data for Iridium-192 but the discussion and conclusions are relevant to all types of sources.

ACKNOWLEDGMENTS

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STANDARDIZATION OF IRIDIUM-192 COILED SOURCE IN TERMS OF AIR KERMA OUTPUT

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Presented by A. Brahme

Abstract

ICRU (1985) recommended that the output of gamma ray brachytherapy sources should be specified in terms of reference air kerma rate, defined as the kerma rate to air in air at a reference distance of 1 meter, perpendicular to the long axis of the source, corrected for air attenuation and scattering. As these measurements are difficult to carry out in the routine clinical use, it is the common practice to calibrate the re-entrant ionization chamber with respect to open air measurements and use the re-entrant chamber for routine measurements. This paper reports on the measurements carried out to correlate the nominal activity and air kerma rate of ^{192}Ir wire sources supplied by the Board of Radiation and Isotope Technology, Department of Atomic Energy.

Introduction :

One of the major factors which contributes towards dosimetric accuracy in brachytherapy is assessment of source strength used. According to ICRU Report 38 (1985), the specification of gamma ray brachytherapy sources should be in terms of reference air kerma rate , defined as kerma rate to air, in air at a reference distance of one meter from the center of source and perpendicular to the long axis of the source. The long distance measurement geometry minimizes the dependence of the calibration upon the construction of the source and detector, as both can be considered as points and effect of oblique transmission of gamma rays through source sheathing become negligible. But the long distance measurements under scatter free conditions are difficult to carry out in routine practice, especially with low activity sources. Hence it is a common practice to establish a calibration factor for the well chamber with respect to open air measurements and use the well chamber along with a reference standard for routine calibrations [1,2]. Under the EUROMET framework, a program of work was initiated at NPL to confirm the traceability to NPL secondary standard radionuclide calibrator of air kerma rate measurements made by Amersham International for wire sources of ^{192}Ir [3].

In India, radiation sources are supplied by the Board of Radiation and Isotope Technology (BRIT), Department of Atomic Energy. ^{192}Ir wire sources used for interstitial therapy is supplied to hospitals in the form of cylindrical coils, the nominal activity of which is measured in a re-entrant 4π gamma chamber. The coil is cut into required lengths by the users. This paper deals with the measurements carried out to correlate the nominal activity of coiled source with reference air kerma rate measured for coiled as well as linear form of sources.

Materials and Methods

^{192}Ir wire sources supplied by BRIT consists of iridium-platinum core (75 % platinum & 25 % iridium) of 0.1 mm diameter with 0.1 mm thick platinum coating, thus making an overall diameter of 0.3 mm. The wire in lengths of 50 or 100 cm is coiled and activated by thermal neutron irradiation in a reactor. The nominal activity of the coil is then measured in a calibrated well type ionization chamber and supplied to the users.

Measurements

An ^{192}Ir source of 3.7 GBq(100 mCi) produces an exposure rate of about 3.225 * 10^{-9} A/Kg (12.5 μR/sec) at one meter. The current per unit volume works out to be about 4.17 * 10^{-15} A/cm^3. To measure such low currents, a 400 c.c. ionization chamber coupled to a varactor diode amplifier with a calibrated capacitor in the feed back was used. The 400 c.c. bakelite chamber was calibrated against a spherical graphite chamber whose accuracy of air kerma rate determination is 2 % and the measurements are traceable to the ^{60}Co therapy level primary standard which is intercompared against international standards.

Iridium wire cut into small pieces of 1.5 cm length, irradiated to an activity of 222 MBq/cm (6 mCi/cm) was procured from BRIT for the calibration of 400 cc chamber. The sources were arranged in a matrix of size 1.5 cm x 2.0 cm and aligned in level with the center of spherical graphite chamber, 65 cm apart in a scatter free geometry. The source - detector alignment was verified using a laser beam. The chamber was connected to varactor amplifier set up with a calibrated capacitor in the feedback. The output voltage (V), over a time t seconds, was measured and corrected for background radiation, charge leakage, temperature and pressure. Corrections have also been applied for wall attenuation (K_{att} = 1.066) , stopping power ratio of graphite to air (S/ρ = 1.015). KCEP and (μ_{en}/ρ) were assumed to be 1.0 for energy corresponding to ^{192}Ir gamma rays. Current per unit volume was evaluated and correlated to air kerma rate. The graphite chamber was then replaced with 400 cc bakelite chamber and measurement was repeated as before. After applying necessary corrections for background radiation, charge leakage, temperature and pressure, current per unit volume was evaluated. The ratio of the two sets of readings was taken as the calibration factor of the 400 cc chamber.

The calibrated 400 cc chamber was then used to standardize the ^{192}Ir coils actually used in clinical practice. Two sets of measurements were carried out, one with the source in the coil form, as supplied to the users and the other after cutting into linear form, as used in clinical practice. ^{192}Ir coiled source and the chamber were aligned in a scatter free geometry. The source chamber distances were kept as 75 cm and/or 100 cm. Measurements were carried out as before and after applying the necessary corrections discussed earlier, current was calculated and using the calibration factor of the chamber, air kerma rate was evaluated. Measurements were carried out for five different coils, 1 of 100 cm and 4 of 50 cm each. Correction for decay of source activity during the course of measurement was applied assuming a half life of 73.83 days. Reference air kerma rate per unit length was evaluated as μGy.h^{-1}.m^2/cm.

Three of these coils, one of 100 cm and the other two of 50 cm each were then cut into linear sources, varying in length from 6 cm to 10 cm. Channels were drilled at 1 cm intervals in a perspex sheet and the nylon tubing used for implantation was fixed into these.

The perspex sheet and the 400 cm^3 ionization chamber were aligned in a geometry identical to coil source measurement. ^{192}Ir wires loaded in inner nylon tubings were then inserted into the outer nylon tubings fixed on the perspex sheet. Air kerma rate was measured as before. To account for the possible loss of small bits of wire while cutting, the actual length of wire used for measurement was determined from auto- and X-ray radiographs Autoradiograph also helped to ensure the uniformity of activity. Reference air kerma rate per unit length was evaluated and correlated to that measured for coiled source.

Discussion

Dose computation for linear sources require specification of source strength in terms of reference air kerma rate(RAKR) constant ($\mu Gy.h^{-1}.m^2.MBq^{-1}$). This was calculated from the measured reference air kerma rate using the quoted nominal activity of the source and assuming the coil as a series of rings of diameter 1.2 cm spaced at equal intervals for coil form and using Sievert's line source dose function, for linear form. The values thus obtained for sources in coil and linear forms are given in Table - 1. The correction factor to be used for linear source output is given as ratio of RAKR constants in the last column of the Table.

Table - 1

Measured Output Correction Factor for ^{192}Ir Coils

Source Chamber Distance (cm)	RAKR Constant ($\mu Gy.h^{-1}.m^2.MBq^{-1}$)		Ratio of RAKR Constant
	Coil	Linear	Linear/Coil
75.0	0.1025 ± 0.0013	0.1124 ± 0.0019	1.101
100.0	0 1035 ± 0.0009	0.1156 ± 0.0021	1.118

It may be seen that the RAKR constant for linear form is significantly higher than that of coiled form. This could be attributed to higher inherent self shielding for coil. In clinical practice , where long wires are used, the reference air kerma rate constant measured for linear form should be used for dose computations. As the published values of RAKR constant for ^{192}Ir sources show large variations, the measurements carried out at respective centers should be considered as appropriate.

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CALIBRATION OF ^{192}Ir HIGH DOSE RATE BRACHYTHERAPY SOURCES[1]

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Abstract

A method for calibration of high dose rate sources used in afterloading brachytherapy systems is described. The calibration factor for ^{192}Ir is determined by interpolating ^{60}Co gamma-rays and 250 kV x-rays calibration factors. All measurements were done using the same build up caps as described by Goetsch et al and recommended by AAPM. The attenuation correction factors were determined to be 0.9903, 0.9928 and 0.9993 for ^{192}Ir, ^{60}Co and 250 kV x-ray, respectively. A wall + cap thickness of 0.421 g.cm^{-2} is recommended for all measurements to ensure electronic equilibrium for ^{60}Co and ^{192}Ir gamma-ray beams. A mathematical formalism is described for determination of $(N_x)_{Ir}$.

1 INTRODUCTION

High Dose Rate (HDR) afterloading systems using ^{192}Ir sources are receiving considerable attention throughout the world as an economical and safe option for brachytherapy specially where the patient load is high. At the present time, there are 8 units in operation in Brazil and 4 other are expected in the near future.

The source manufacturer specifies its activity with an accuracy of ±10% which is unacceptable for clinical purposes and without traceability to a national standard. In the absence of a standard for ^{192}Ir calibration, the American Association of Physicists in Medicine (AAPM) proposes the use of an interpolation procedure [1,2] using the average calibration of a 0.6 cc ionization chamber obtained with ^{137}Cs gamma-rays and 250 kV x-rays (HVL=3.2 mm of Cu, effective energy of 146 keV x-rays). A graphite cap thickness of 0.31 g.cm^{-2} has been recommended for all measurements.

An alternative method using a well-type ionization chamber [3] has been carefully analyzed and compared with the AAPM method and an excellent agreement was found. However, its calibration relies on a thimble ionization chamber calibrated for ^{137}Cs gamma-rays. Since the majority of the Secondary Standard Dosimetry Laboratory's (SSDL) are not equipped with a ^{137}Cs therapy source, the users will have to send the ionization chambers abroad for calibration.

This paper proposes a calibration procedure to derive the N_x for a Farmer type ionization chamber for ^{192}Ir, by interpolation from a ^{60}Co gamma-rays and 250 kV x-rays (HVL= 2.5 mm of Cu, effective energy 131 keV x-rays) calibration factors [4].

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[1]**This paper is dedicated to the memory of Eugenio R. Cecatti.**

The calibration factors were determined using the same build-up cap for the measurements with ^{60}Co gamma-rays , 250 kV x-rays and ^{192}Ir with the appropriated wall attenuation factors taking into account.

2 METHODS

A. Wall attenuation measurements

The measurements were made using a NE Farmer thimble type chamber model 1975, with a 0.065 $g.cm^{-2}$ graphite wall. Six build up caps were made of PMMA with thickness varying from 0.194 to 0.496 $g.cm^{-2}$. The thickness was varied by adding additional buildup caps. Measurements for ^{60}Co gamma-rays and 250 kV x-rays (HVL = 2.5 mm Cu, effective energy = 131 keV) were done at the Laboratorio Nacional de Metrologia das Radiações Ionizantes (SSDL/CNEN). A 10x10 cm^2 field size and source to chamber distance of 100 cm for the ^{60}Co gamma-ray beam, and a 7 cm diameter circular field and 75 cm distance for the 250 kV x-ray beam were used. The ^{192}Ir measurements were made at the Hospital São Vicente de Paulo. For the ^{192}Ir source measurements, the Nucletron calibration jig was used with the center of the chamber positioned at 10 cm from the source [5]. Figure 1 shows the attenuation curves for all three energies. A combined (PMMA+graphite) buildup cap of 0.421 $g.cm^{-2}$ is sufficient for electronic equilibrium in ^{60}Co and ^{192}Ir gamma-ray beams.

B. ^{192}Ir Calibration Factor

The ^{192}Ir calibration factor, $(N_x)_{Ir}$, is obtained by interpolating the ^{60}Co gamma-ray calibration factor and the 250 kV x-rays as originally proposed by Ezzel [2] and following the more rigorous proposal by Goetsch et al [4]. A combined build cap of 0.421 $g.cm^{-2}$ was used for calibration in all three beams. As we interpolate the calibration factor for ^{192}Ir it is necessary to use the buildup cap and take into account its attenuation in the calibration process. The attenuation factors were determined from the slopes of the attenuation curves in Figure 1. The correction factor A_w was calculated by

$$A_w = 1 - (slope) \cdot (wall\ \ thickness)$$

The ^{192}Ir calibration factor is determined by interpolating the ^{60}Co gamma-rays and 250 kV x-rays calibration factors. Assuming that the calibration factor is linear with energy, it can be written that:

$$(A_w \cdot N_x)_{Ir} = (k \cdot A_w \cdot N_x)_{X-ray} + (k \cdot A_w \cdot N_x)_{Co}$$

where, $k_{X\text{-}ray}$ *and* k_{Co} are interpolation factors given by:

$$k_{X-ray} = \frac{E_{Ir} - E_{Co}}{E_{Co} - E_{X-ray}} = 0.7989$$

and

$$k_{Co} = \frac{E_{Ir} - E_{X-ray}}{E_{Co} - E_{X-ray}} = 0.2011$$

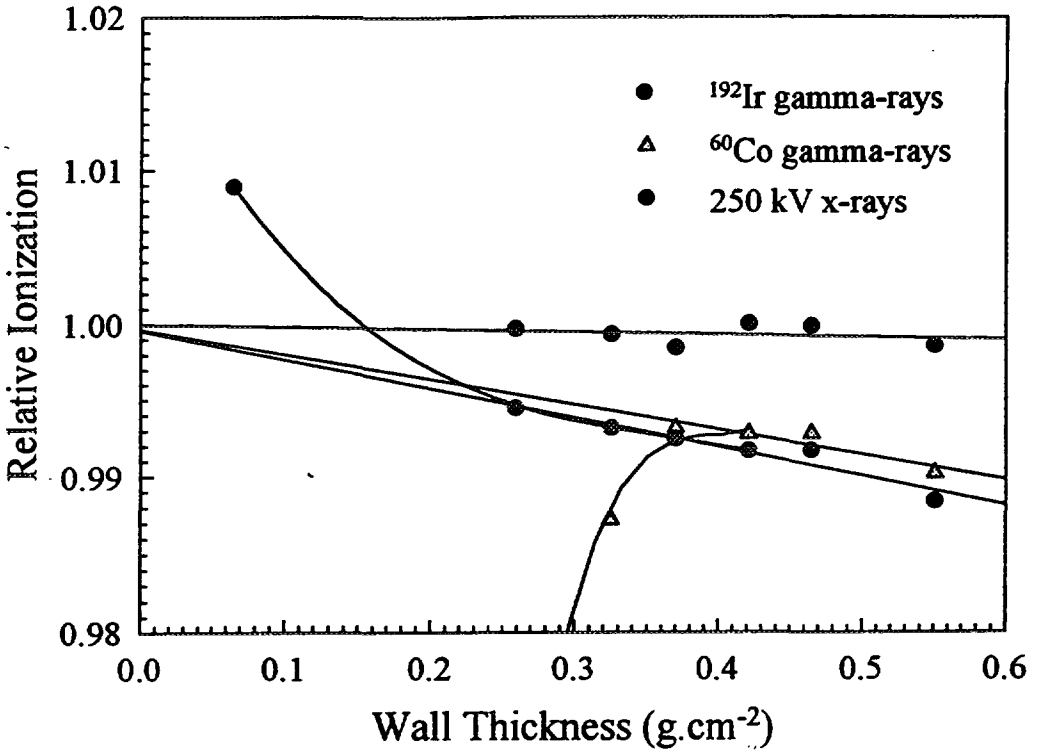

Figure 1. Attenuation curves for 250 kV (131 keV x-rays), ^{60}Co and ^{192}Ir gamma-rays in lucite caps covering a Farmer ionization chamber. The slopes are least-square fits to measured data with different cap thickness added.

E_{Ir}, $E_{X\text{-}ray}$ and E_{Co} are exposure-weighted average energy for ^{192}Ir gamma-ray , 250 kV x-ray and ^{60}Co gamma-ray beams, respectively.
$(N_x)_{Ir}$ is then given by

$$(N_x)_{Ir} = \frac{0.7989 \cdot (A_w \cdot N_x)_{X-ray} + 0.2011 \cdot (A_w \cdot N_x)_{Co}}{(A_w)_{Ir}}$$

3 RESULTS

The attenuation coefficients were found to be 0.023, 0.017 and 0.0016 cm^2.g^{-1} for ^{192}Ir, ^{60}Co and 250 kV x-rays, respectively. As one observes from Figure 1 a thickness of 0.28 g.cm^{-2} is enough for electronic equilibrium in a ^{192}Ir gamma-ray but not for ^{60}Co gamma-ray beam. A thickness of 0.421 g.cm^{-2} is necessary for ^{60}Co gamma-ray beams, the attenuation correction factors being in that case 0.9903, 0.9928 and 0.9993 for ^{192}Ir, ^{60}Co and 250 kV x-rays respectively.

The ^{192}Ir calibration factor is given by:

$$(N_x)_{Ir} = 0.8055 \cdot (N_x)_{X-ray} + 0.1997 \cdot (N_x)_{Co}$$

The total thickness for buildup cap is 0.421 g.cm^{-2}. If one subtracts the graphite wall thickness of the ionization chamber, a 0.356 g.cm^{-2} thick PMMA cap is needed. The combined uncertainty associated to the $(N_x)_{X\text{-}ray}$ and $(N_x)_{Co}$ and the experimental procedures results in 0.4% overall uncertainty for the ^{192}Ir calibration factor, assuming that the ionization chamber response is linear with energy in that region.

4 CONCLUSIONS

A method for determining the ^{192}Ir calibration factor for a thimble ionization chamber is described using interpolation from a ^{60}Co gamma-ray and 250 kV x-ray beams calibration factors. The method is valid for ionization chambers that responds linearly with energy.

To use this method at other SSDL only the attenuation factor for the build up cap in the 250 kV x-ray beam have to be determined because it is the only beam energy that might change. The attenuation factors for the buildup cap in the ^{60}Co and ^{192}Ir gamma-ray beams determined in this work can then be used for the derivation of the ^{192}Ir calibration factor. Since the attenuation factor for the 250 kV x-ray beam is small (less than 0.1%) even this factor could be used without compromising the accuracy of the method.

A 0.421 $g.cm^{-2}$ combined thickness build cap was used to calibrate the thimble ionization chamber. Since users normally have a ^{60}Co gamma-ray buildup cap we will in the future use the ^{60}Co buildup cap thickness (0.551 $g.cm^{-2}$) for this method. Future plans include direct comparison with a Farmer ionization chamber with PMMA wall to minimize wall artifacts. This ionization chamber will be calibrated for a ^{137}Cs gamma-ray beam to allow direct comparison between this method and the AAPM proposed procedure.

ACKNOWLEDGEMENTS

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QUALITY CONTROL OF Ir-192, Cs-137 AND Ra-226 SOURCES FOR USE IN BRACHYTHERAPY

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Abstract

In order to establish a certain degree of reliability in the use and managing of radioactive material in brachytherapy, a minimal quality control to each source was implemented. The purpose was to estimate the degree of radioactive leak, resistance to the mechanic traction and stress of use. Through a physical control of the radioactive material (a simple dip test and using a photon scanner or auto-radiography) the minimal conditions that guarantee safe use are established. This information is transmitted to a calibration laboratory for certification in exposure rate and/or activity. Systematic use of these tests, enables discovery of radioactive material leakage due to faults in the seal.

1 INTRODUCTION

During the last 10 years, sealed radioactive material has been used for radiotherapy especially in brachytherapy. In such practice, radioactive material with long half lives such as Cs-137, Ra-226 have been used for some decades, lately short half life nuclides, particularly Ir-192, came into clinical use. Such sources can be reactivated in nuclear reactors. In either case, the radioactive material must be sealed. For the purpose of radioprotection of patients and personnel involved in the treatment, these seals must be tested.

The radioactive material, normally Ra-226, Cs-137 or Ir-192 is in the form of spheres or cylinders and clustered inside a covering, sealed and mechanically labelled. This cylindrical covering is of Stainless Steel, Platinum and/or Platinum-Iridium.

For radiotherapy objectives a uniform dose distribution around a implant or therapy applicator is a primary goal. Thus, the radioactive material must be uniformly distributed in the isotope source. Simultaneously the uniformity of the seal is crucial to acquire a symmetrical dose distribution outside the capsule (axial symmetry). This enables less complicated calculations of dose distributions to the surrounding tissues. Also from the radiation protection point of view the control of the seal in terms of leakage and symmetry is important. Nevertheless radioactive leaks from such sources have been detected. The losses are due to mechanical loss of the covering material. A control applied on the covering would enable a determination of the degree of damage. The most meaningful tests turn out to be determination of;

1. Losses of radioactive material,
2. The degree of radioactive homogeneity,
3. The inherent resistance of the sealing material to physical-chemical stress of the covering.

As a whole, this series of tests determine if the source is suitable to use or if should be considered as radioactive waste.

2 METHOD

Loss of Radioactive Material

The determination of the speed at which radioactive material is lost and/or radioactive contamination occurs, is accomplished through either gamma and/or beta spectrometry. The determination of the radioactive contamination, it is realised through chemical pureness tests in a base solution, and Gamma spectrometry and/or Beta emission analysis of the same solution.

The base solution is a mixture of 50% of Ethanol and 50% Hydrogen Peroxide. The radioactive material is submerged in the base solution for a period not less than 76 hours at room temperature. After this time interval adjusted according to a radiological procedure previously established, the radioactive source is removed. The radioactive contaminants in the solution are then analysed.

In the case of Ra - 226, Cs-137 and Ir-192, the radioactive contamination analysis is based on Gamma spectrometry using a high efficiency detector. As detector, a NaI(T1) - crystal was used. It was connected through an amplifier and a pulse height analyser (PHA) to a multi channel analyser with 4096 channels.

Repeating the above sequence permits determination of external contamination due to:
1. Diffusion though the covering.
2. Interaction with external radioactive materials,
3. Micro-fissures or structural damages of the covering.

As a rule, the estimate of radioactive losses is based on the detection limits for each isotope, and the signal/noise ratio of the spectrometry system.

Degree of Radioactive Homogeneity

The degree of radioactive homogeneity, is determined through two complementary methods;

1. A longitudinal sweep with a semiconductor micro-detector, connected to a gamma spectrometry system.
2. Utilisation of radiographic plates to obtain autoradiographs.

The longitudinal sweep in based on the use of a Si(Li) mini-detector mounted on a table that permits precision movements. The detector is moved along the source and measures the event rate of each section of the radioactive material.

The autoradiography is based on the use of radiographic plates put in a mammography cassette of the type used in the clinic. If the autoradiography shows evidence of low filtration zones through the covering, a contact autoradiograph is made in order to detect any possible beta or Alfa emission through the seal (depending of the radionuclide).

The criteria used to determine the loss of covering material, is based on a study of the average thickness of the cover compared with the possible damage zone. It is obvious that the visual inspection of the material through an optical system is adequate.

Physical Chemical Resistance capacity

The physical-chemical resistance capacity is determined in order to ensure the source to overcome extreme and violent chemical reactions. Normally the same solution as that used for determination of radioactive leak is used.

The capacity of sustaining extreme mechanics actions such as beats and torsion was tested. It was shown that coverings of Platinum-Iridium do not sustain more than five re-activating with neutron flows in a 5 MW reactor, conditioned by mechanical effort accomplished through metallic tweezers without protection. In coverings of stainless steel, as is the case of the Ra-226, and having a normal manipulation history, surface damages are shown after 10 to 15 years of use.

3 RESULTS

The solid sources of Ra-226 with more than 10 years of use, show mechanical surface damages in about 15 and 30% of the total surface of the covering. This gives radioactive leaks in the range of 30 to 1000 times the detection limits. It is recommended not to use Ra-226 for intracavitary implants in Chile.

From a survey of 1000 Cs-137 needles surveyed, only 10 needles presented surface damages due to handling with metallic barbed objects.

In the Ir-192 wires activated by the first time, a 0,3% percent lack of radioactive material was shown. Wires reactivated three or four times present transverse fractures in the points of manipulation with metallic barbed objets. Ir-192 wires reactivated a third time are not used In Chile.

4 CONCLUSIONS

The application of a minimal quality control by testing the radioactive material used in intracavitary implant, permits:

a) To define and to guarantee the necessary technical specifications for the calculation of the Dosimetry of the implant.

b) To avoid radioactive contamination of the medical staff and the patient.

c) To maintain the conditions of the accounting and storage of the same.

An annual monitoring is recommendable. It can simultaneously accomplish a minimal quality control, based on a radioactive cleanliness of the covering and an autoradiography, both accomplished through a clearly detailed radiological procedure.

IV. QUALITY ASSURANCE NETWORK IN RADIOTHERAPY

QUALITY ASSURANCE NETWORK: THE EUROPEAN PILOT STUDY

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Abstract

Based on the IAEA/WHO experience in mailed dosimetry, a Quality Assurance (QA) Network, sponsored by the EC committee"Europe against Cancer", has been set up in 1991 for all European centres. Besides a survey of radiotherapy infrastructure, the project includes three measurement steps : primarily, a check of beam output and quality in reference conditions with a mailed TLD-procedure, in a second step, the mailed verification of other beam data and dose calculation procedures with a multipurpose phantom, and finally in vivo dosimetry at the individual patient levels with mailed dosimeters.

The results concerning 162 beams from 85 centres are analysed (58 ^{60}Co beams and 104 X-ray beams). 27 beams present minor deviations (3 to 6 %) and 15 beams (4/58 ^{60}Co beams and 11/104 X-ray beams) from 11 centres present major deviations (≥ 6 %). The analysis shows that 17/27 minor deviations and all major deviations have been detected in centres which have not benefited from an external check during the last five years; in 14 out of 15 large deviations, the measured dose is smaller than the stated dose. In most centres with major deviation, the physicists did not have the necessary experience and did not calibrate regularly the beams. In 6 centres out of 11 there was no dosimeter or the dosimeter available has not been calibrated recently. In 3 centres, the physicist did not give any explanation. The conclusions concerning the second step (multipurpose phantom), outline the larger magnitude of the deviations for off axis points, oblique surface and the use of wedge filters.

1. INTRODUCTION

The Interest of Quality Assurance in Radiotherapy (medical and physics aspects) has been stressed for decades with a number of dedicated Meetings, papers, reports and protocols.

Nevertheless, the 1989 world-wide investigation sponsored by the International Atomic Energy Agency (IAEA) and the World Health Organization (WHO) revealed that about 15 % of all cancer patients treated with radiation receive an inadequate dose due to systematic uncertainties in the dose delivery (1). This percentage is probably even much important in a number of Radiotherapy Departments not implementing Quality Assurance (QA) programmes.

In addition, it is well established that randomized clinical trials can improve the practice of Radiotherapy provided that suitable Quality Assurance Programmes are applied to ensure the compliance of the treatments with the treatment regimes (2, 3). Due to the need for a large number of patients in order to improve the significance of the conclusions, to allow the evaluation time to remain reasonable, more and more multi-institution trials are launched, the main requirement being that any patient of any institution be treated according to the same protocol and receive the same dose.

Consequently, in addition to the fact that both internal QA and external audits have proven to be very useful (4, 5, 6, 7), even when Radiotherapy Departments implement good internal Quality Assurance Programmes, the coherence must be ensured between the Radiotherapy Departments at larger scales, through **Q.A. networks**.

At the present time in Europe, a few large centres have developped QA Programmes and have shown the benefits of their implementation (8, 9, 10, 11, 12). A Few Countries have developped programmes of National Dosimetry Intercomparisons, often only in reference conditions (13, 14, 15, 16, 17), but no comprehensive system comparable to the US Centers for Radiological Physics is available at the European scale (18-1 and 18-2).

At the International level, a few international bodies have developed mailed QA Programmes. In particularly, since 1967, IAEA/WHO have developped a postal dosimetry service (1, 7, 19, 20) for developing countries, restricted to measurements at a reference point for ^{60}Co radiation. This service has recently been extended to high energy X-ray beams with the help of a few advanced centres. Since 1988, the European Organization for Research and Treatment of Cancer (EORTC) has also developed a mailed service as a part of larger QA programmes to check beam outputs in reference conditions for the European Centres participating in clinical trials (21, 22). A comparable on purpose service was also developed by the Institut Gustave-Roussy for a clinical trial on the use of the Etanidazole Radiosensitizer, including about 30 european centres, with a postal dosimetry service (reference conditions) associated with on purpose on site visits and patient treatment forms review (23).

So, in Europe, only a few national or international dosimetry intercomparisons and on purpose actions for clinical trials had been performed when, at the request of a number of centres not involved in clinical research, an attempt was made to extrapolate the expertise acquired in these first studies to set up a standardized QA procedure which could be the same in all European Community (EC) countries and which would provide guidelines and technical back-up to all radiotherapy Centres.

The principles of this attempt are that the practical responsabilities would be given to national bodies as soon as procedures are established. An European coordinating and advisory function would still be maintained in order to ensure EC guidelines to remain coherent.

At last, five radiotherapy centres from five European Countries (Belgian, France, Italy, The Netherlands and Sweden) succede when proposing the EC Committee "Europe against cancer" to support the project of a European Network for Quality Assurance in Radiotherapy.

2. DEVELOPMENT OF THE NETWORK

The Experimental European Network was implemented in 1991, with the following structure :

Coordinating Centre (CC)	University Hospital St Rafaël, Leuven, Belgium.
Measuring Centre (MC)	Institut Gustave-Roussy, Villejuif, France.

National Reference Centres (RC)	Belgium : University Hospital St Rafaël, Leuven France : Centre Georges François Leclerc, Dijon Italy : Universita degli Studi di Firenze, Florence The Netherlands : A. Van Leeuwenhoekhuis, Amsterdam. Sweden, University Hospital, Umea.

3. IMPLEMENTATION OF THE NETWORK ACTIONS (24)

3.1. Dosimetry checks

They include a preliminary investigation of the available infrastructure in the centers and three measurement steps :

3.1.1. Preliminary investigation

In the initial stage, before any measurement is performed, a questionnaire is sent by the National Reference Centre to the local centres, regarding staff, radiotherapy and dosimetry equipment, simulation and treatment planning systems, in order to be able to find out possible correlations between the radiotherapy department structures and the uncertainties. The questionnaire is identical for all countries in order to obtain a standardised data base. It is translated into the local language by the National Reference Centre.

3.1.2. First step : dosimetry check in reference conditions

At a first step, beginning in 1992, the output and beam quality of ^{60}Co-ray beams and high energy X-ray beams were checked in reference conditions with mailed TL Dosimeters placed in a small plastic holder, which were irradiated in a water tank ("simple phantom procedure"). following a procedure similar to the procedure used by the IAEA for the postal intercomparisons (fig.1).

3.1.3. Second step : additional checks in a "multipurpose phantom"

The uncertainty in the dose distribution due to beam dosimetry data other than in reference conditions is generally accepted to be much larger as additional errors originate in the estimation of beam flatness. wedge transmission and the calculation of dose distribution (25. 26. 6).

That is why, in a second step, a multipurpose phantom designed previously for the IAEA by a group of consultants (27, 28) is used to check various beam data (beam symmetry. wedge transmission) and dose calculation procedures (collimator output factor, surface obliquity correction, calculation of the dose at depth). The EC National Reference Centres have checked the reliability of the procedure in a pilot study conducted in cooperation with the IAEA. After the feasibility study, the multipurpose phantom is now being used in the first Local Centres of the EC project. The main features of the multipurpose phantom are displaid in fig.2.

3.1.4. Third step "postal in vivo dosimetry".

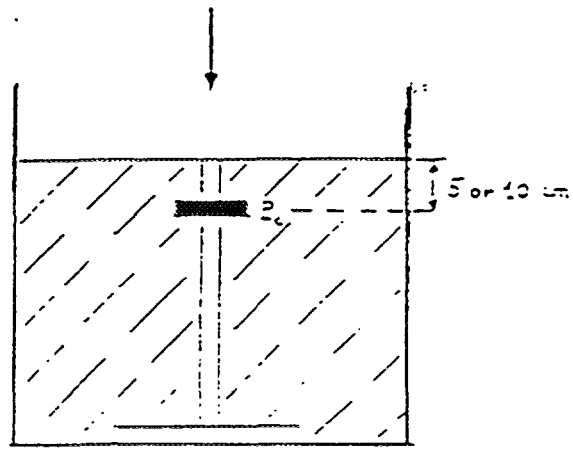
In the last step, yet to be implemented, it is proposed to measure the dose delivered at the individual patient level with mailed dosimeters.

A. CALIBRATION CHECK: Co.60 - RX

* *Vertical beam*

10 cm x 10 cm at usual SSD.

2 Gy to point Pc (5 cm or 10 cm depending on the beam quality)



B. BEAM QUALITY CHECK : RX

* *Vertical beam*

10 cm x 10 cm at SSD 100 cm

2 Gy to point P1

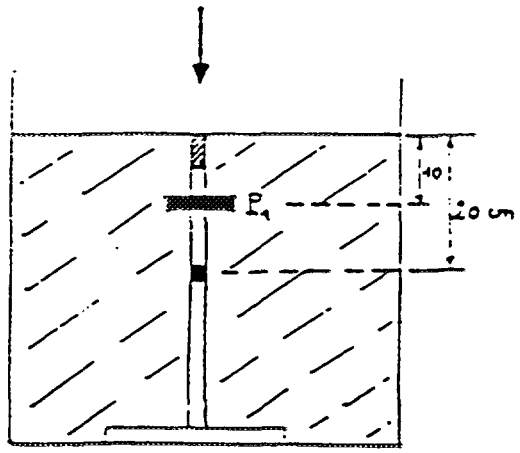
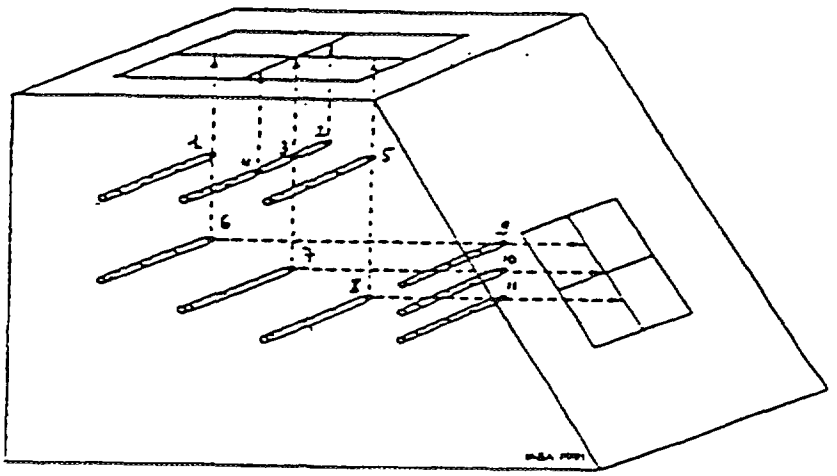


Fig.1 Simple phantom description and irradiation procedure

MULTIPURPOSE PHANTOM



11 points of measurements (LiF)

* *Basic dosimetry data to be checked ::*

- Calibration and field size dependence
- Central axis depth dose
- Beam symmetry
- Obliquity corrections
- Wedge transmission
- Summation of doses from multiple fields

→ ***Cheks of dose calculation procedures and algorithms.***

Fig.2 Multipurpose phantom description and check capabilities

3.2. Organization

The role of the different partners is defined as follows :

The Co-ordinating Centre prepares the questionnaires, information and data sheets. It supervises the organisation of the mailing and prepares the time schedule. It ensures the follow-up of the programme and analyses the results.

The **Reference National Centres** contact the **Local Centres** (LC) that wish to participate and plan the timing and practical organisation. The NC assures the LC about the strict confidentiality of the questionnaires and of the results, which will be known only by the CC, the MC and the NC. The data will always appear in an anonymous way in reports and/or publications and could never be transferred to administrative or governmental bodies, without full written agreement from the Local Centres.

The NC mails the dosimeters received from the MC, to the LC, together with the information and data sheets. It receives the irradiated dosimeters from the LC, together with the completed questionnaire and data sheets. It mails the dosimeters and a copy of the data sheets to the MC, receives the results from the MC and mails them with comments to the LC; It decides on further action if needed.

The Measuring Centre is in charge of the study and optimisation of the methodology and the equipment in relation with the coordinating and National Reference Centres, and of all the measurements related to the Network checks.

The present network is represented on the flow chart of fig.3. In fact, with respect to Eastern Europe, a dedicated experimental network has been set-up in 1993, sponsored by the Belgian authorities (Ministry of the Flemmish Community), called EROPAQ. In this experimental network, the University Hospital St Rafaël, Belgium is acting as both the CC and the MC. Three countries are presently connected to this network, the Czech Republic, Poland and Hungary. The results presented below include only the first measurements performed for the Czech Republic before the EROPAQ set-up.

4. METHODOLOGY

4.1. The Measuring Centre equipment and methodology (29)

The basic MC methodology has benefited from the IAEA/WHO cooperation and the EQAG experience. At the present time, 162 photon beams (58 ^{60}Co and 104 X-ray beams) from 85 centres have been checked, corresponding to more than 2000 TLD readings.

4.1.1. TL material

The postal dosimetric checks are performed with LiF dosimeters (PTL 717 from the Desmarquest-CEC Company-France). This LiF is enriched with 7 Li, presenting a low sensitivity to neutrons (high energy photon beams) and a low fading (less than 5 % per year at room temperature). Used as a powder, it allows us to get a good precision, in the order of 1.5 % (1 s), for the dose measurements with one dosimeter (polyethylene tubes allowing 5 readings).

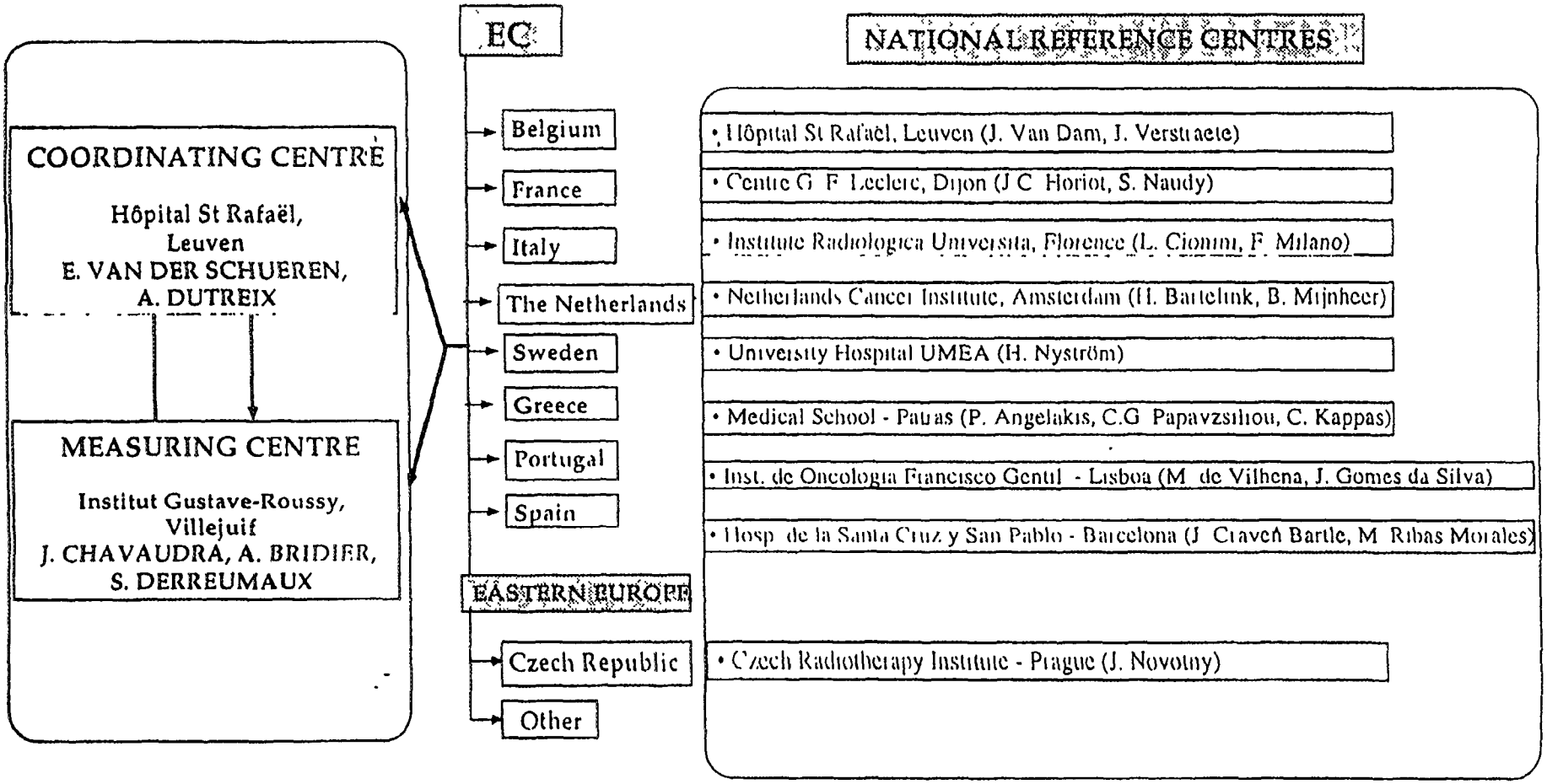


Fig.3 Flow chart of the European Pilot Study network

4.1.2. Reading equipment

The reading equipment includes two units :

• a manual reader (SAPHYMO LDT 21) used exclusively for two years (classical heating system with preheating).

• an automatic reader (PCL Fimel), allowing 90 samples loaders to be read in about 45 minutes and in routine use at the present time, using constant temperature ovens (30).

4.1.3. Calibration at the measuring centre

The reproducibility, dose response linearity, fading, accuracy and energy dependance of the TL dosimeters have been checked by the MC during the first year. The reliability of the mailed TLD measurements performed by the MC has been checked with the IAEA acting as an SSDL. The determination of the dose at the MC is made following the IAEA protocol (Report 277, 1987), with a reference dosimeter (calibration traceable to the French PSDL, the LPRI, Saclay, France, with an agreement better than 1 %).

4.2. Global postal checks procedure

4.2.1. First step ^{60}Co and X-ray beams output in reference conditions and beam quality of X-ray beams.

The dosimeters and their holders are identical to those used by the IAEA for the international network and have to be irradiated in a water phantom. The dosimeters are prepared at Institut Gustave-Roussy (Villejuif), acting as the European Measuring Centre (MC) and sent ot each participant centre with the holders.

The protocol sent to the participants for the irradiation of the dosimeters is similar to the protocol used by the IAEA. Two dosimeters are successively irradiated at 2 Gy to the required depth (usually 5 cm or 10 cm, depending on the beam quality), in a 10 cm x 10 cm vertical beam, at the source to surface distance (SSD) or at the source to dosimeter distance (SDD) normally used in the centre. The SEFM Spanish protocol recommends a reference depth of 7 cm for the photon beams of nominal energy between 10 and 25 MV. The irradiations have been performed at 10 cm but a set of holders will be adapted to allow the irradiation of dosimeters at a depth of 7 cm. It is recommended to the local centre to measure the output of the beam with a calibrated ionisation chamber just before the irradiation of the TLD capsules. For the check of the beam quality of the X-ray beams, two dosimeters are irradiated simultaneously, one of them at 10 cm depth (measuring a dose D_{10}) and the other one at 20 cm depth (D_{20}), in a 10 cm x 10 cm field, at an SSD of 100 cm. The two dosimeters are at right angle from each other, to minimise the shadowing of the lower dosimeter by the upper one. A dose of 2 Gy has to be delivered to the upper dosimeter. The quality index (QI) of the beam, defined as the ratio of the tissue-phantom ratios at 20 and 10 cm is in fact evaluated through the measured value of D_{20}/D_{10}.

The readings of the TL dosimeters are carried out at the MC. The repeatibility on the reading of one dosimeter is in the order of 0.7 %. A small correction for supralinearity is applied. When a dosimeter is irradiated in an X-ray beam, a correction, depending on the quality index of the beam and of about 2.5 % for 20 to 30 MV is applied to the reading, to take into account the energy dependence of the dosimeter response. Because in the actual conditions of international postal dosimetry, delays of one to two months can occur between the irradiation and the reading of the dosimeters, a correction is applied for the fading and for unexpected irradiations during the travel. For this purpose, any dosimeter mailing is made using "additional dosimeters" : one is irradiated (2 Gy) at the MC and remains in the laboratory, another is irradiated at 2 Gy at the MC and travels with the mailed dosimeters, as well as one unirradiated dosimeter, and the last one is irradiated at the MC (2 Gy) the day of the checking dosimeters return at the MC, before the readings. It is so possible to handle and overcome the fading, the consequences of unexpected irradiations and the possible reader response drift.

The accuracy of the dose calibration by the MC has been checked by intercomparisons with the EORTC (April 1991) and the IAEA (March and November 1992). TL dosimeters have been irradiated at 2 Gy at the MC, in the 60Co beam and in the 4, 18 and 25 MV X-ray beams, in reference conditions (water phantom, on axis, 5 or 10 cm depth, 10 cm x 10 cm field), and read by either the EORTC or the IAEA. The agreement was better than 1 % for the 60Co beam, and better than 1.6 % for each X-ray beam quality for both intercomparisons.

To check the MC TL postal procedure, the MC has sent dosimeters to the IAEA (November 1992) where they were irradiated at 2 Gy in reference conditions (water phantom, on axis, 5 cm depth, 10 cm x 10 cm field, 80 cm SSD) in a ^{60}Co beam. The dosimeters were read at the MC, and the agreement between the absorbed dose in water determined by the MC and the absorbed dose in water stated by the IAEA is better than 1 %.

The dosimeters are sent to the local centres, with instructions sheets explaining how to proceed for the irradiation of the TL dosimeters. Data sheets are also mailed to each centre which has to specify the date of the irradiation of the TL dosimeters (for the estimation of the fading), the dose delivered to the dosimeters (stated dose), and the quality index of the X-ray beams determined by the local centre (stated quality index). The dose and the quality index stated by the centre are compared to the values measured by the MC. The stated quality index is also used by the MC to determine the correction to be applied to the TL reading for the energy dependence of the dosimeters response. The participating centre is also asked to report the date of the last intercomparison or quality audit in which it has been involved, if any.

A quadratic summation of the uncertainties on all the quantities entering in the determination of the absorbed dose in water, with the MC TLD postal measurement procedure, gives an estimation of the final uncertainty on the order, for one standard deviation, of s = 1.5 % for 60Co beams and s = 2 % for X-ray beams.

The results are sent to the National Reference Centre (NC) which report to the local centres. The results are also sent to the Coordinating Centre in Leuven for analysis. The NC conducts an inquiry to trace the origin of large deviations and suggests a new TLD check. Because of the limited resources, the second checks have been performed mainly for the centres with major deviations. Depending on the local situation, the NC can perform an on-site visit or only discuss the method applied for dose measurements with the local centre. Deviations lower than twice the standard deviation do not justify any further investigation by the NC since there is a high probability for the observed deviation to be due to the procedure uncertainty. The NC reports to the Coordinating Centre the results of the inquiries.

A strict confidentiality is maintained all over the procedure and no details are given in any publication, either on the centres, on the countries or on the characteristics of the radiation units.

4.2.2 Second step : Evaluation of the multipurpose phantom (28)

The procedure has been developped at IGR, using a multipurpose phantom made at Houston by WF Hanson (27).

4.2.2.1. Preliminary evaluation of the multipurpose phantom at IGR

The geometry and the composition of the phantom have been determined with CT measurements. Attention has been paid to its reliability (suitability for mailing) and the water filling (bubbles require attention). The dosimeter positioning can be accurate (better than 2 mm). The inhomogeneous structure (PVC, water, perspex) can lead to corrections due to an increase of the values of the measured dose by up to about 1.5 % with ^{60}Co beams and 3 % with 25 MV X-ray with respect to the values expected in an homogeneous water phantom.

4.2.2.2. Evaluation of the TL dosimetry procedure

For each photon beam quality, 13 TL dosimeters are sent by IGR acting as the coordinating and measuring centre, to the participating centres : 11 to be irradiated in the phantom and 2 to detect respectively any unexpected irradiation and fading during transportation. In addition 2 TL dosimeters are kept at IGR to check the fading (see above).

Guidelines explaining how to proceed for the irradiation of the TL dosimeters in the phantom, and data sheets in which each centre has to report information on the irradiation conditions and stated doses have been established and sent, with the TL dosimeters, to the participating centres.

The conversion of TL dosimeter readings into absorbed dose to water is made as previously. The dose evaluated by IGR is then compared to the dose calculated by the centres according the method used in clinical practice.

• Step 2 A : Irradiation of the multipurpose phantom by European Reference Centres (Pilot Study)

In order to check the reliability of the procedure of the use of the multipurpose phantom for quality control by mail, a pilot study was performed during the period September 1991 - May 1992 by the IGR, in cooperation with 6 european centres (Centre GF Leclerc Dijon, France, UZ St Rafael Leuven, Belgium, AVLZ Amsterdam, the Netherlands, Sahlgren Hospital Göteborg, Sweden, Universita di Firenze, Italy, Western General Hospital, Edinburgh, UK).

In the study, the phantom was first sent by the IGR to the first centre and then directly from one participating center to the next. LiF capsules and documents (instruction and data sheets) were mailed from IGR to each centre and sent back after irradiation.

The TL dosimeters irradiation has been performed according to the following procedure (fig. 4) :

• one large vertical beam (15 cm x 15 cm) perpendicular to the phantom surface to check the beam flatness and symmetry at a depth of 4 cm, and dose calculation at 12 cm depth.

• one horizontal beam (8 cm x 8 cm) with a 30° wedge filter directed on the oblique surface of the phantom.

In the irradiation procedure in use, two sets of LiF capsules are considered : LiF capsules 1 to 8 are irradiated by the vertical beam with a 2 Gy dose delivered at 4 cm depth and, on another hand, capsules 6 to 11 are irradiated with the horizontal beam with a 2 Gy dose at 3 cm depth. LiF capsules 6 to 8 are therefore irradiated by both beams.

At each measuring point of interest inside the phantom, the dose measured by LiF capsules is compared to that calculated by the centre according to the dose calculation procedure used in daily practice.

• Step 2 B : Irradiation of the multipurpose phantom by local Centres

After having checked the reliability of the procedure in the pilot study, a run with the multipurpose phantom has been started in routine situations with the local centres of the EC project.

During the period 1992-1993, four local centres have particpated in the run with 6 photon beam qualities (2 ^{60}Co and 4 X-ray beams).

- *Step 2A* -

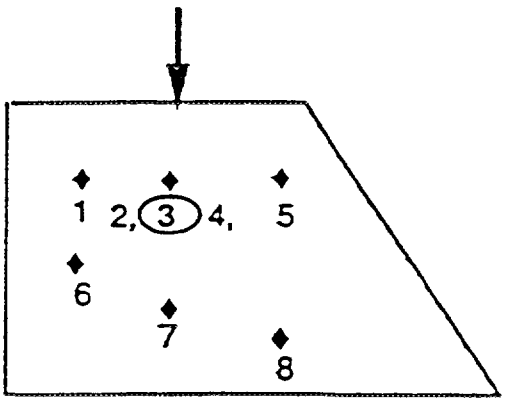
IRRADIATION OF THE MULTIPURPOSE PHANTOM BY THE PARTICIPATING CENTERS

IRRADIATION PROCEDURE

1) VERTICAL BEAM

15 cm x 15 cm at usual SSD
2 Gy to point 3 (4 cm)

- Lif capsules : 1 to 8



2) HORIZONTAL BEAM

8 cm x 8 cm at usual SSD
30° wedge if available
2 Gy to point 10 (5cm)

- Lif capsules : 6 to 11

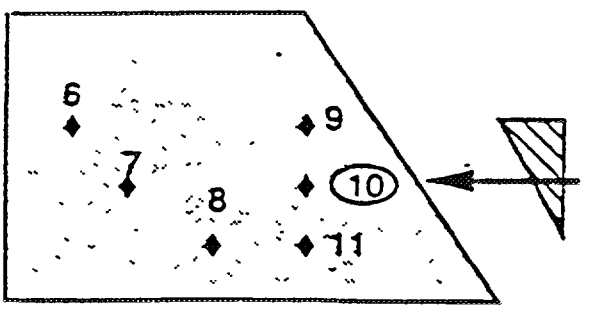


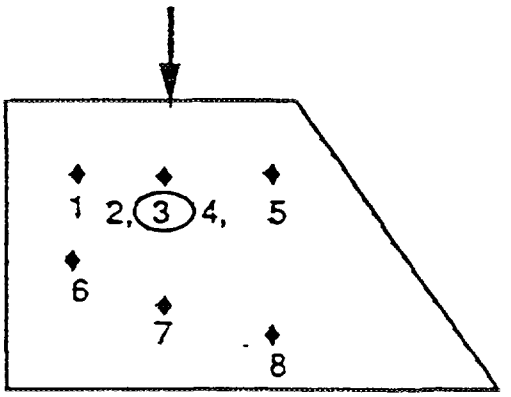
Fig.4 Multipurpose phantom Irradiation procedure according to the STEP 2 A protocol

- STEP 2B -
IRRADIATION PROCEDURE

1) VERTICAL BEAM

15 cm x 15 cm at usual SSD
2 Gy to point 3 (4 cm)

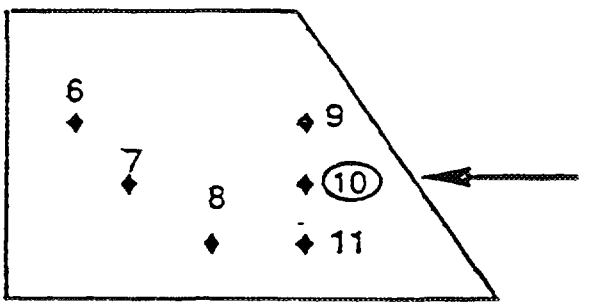
- Lif capsules : 1A to 8A



2) UNWEDGED HORIZONTAL BEAM

8 cm x 8 cm at usual SSD
2 Gy to point 10 (5cm)

- Lif capsules : 6B to 11B



3) WEDGED HORIZONTAL BEAM

8 cm x 8 cm at usual SSD
30° wedge if available
2 Gy to point 10 (5cm)

- Lif capsules : 6C to 11C

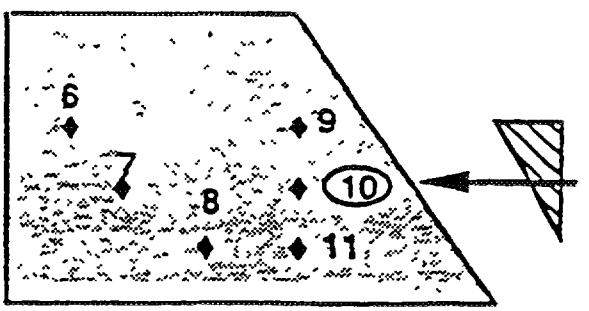


Fig.5 Mulupurpose phantom Irradiation procedure according to the STEP 2 B protocol

Taking into account the difficulties in dose interpretation encountered in the pilot study, on one hand with the combination of beams for TL dosimeters at depth and, on another hand, with the combined influence of the oblique surface and wedge filter, a new irradiation procedure has been set-up for the run.

It consists of three different beams (vertical, unwedged and wedged horizontal beams) for which three different sets of TL dosimeters have to be irradiated (fig. 5).

5. PRELIMINARY RESULTS

5.1. Structure of the Radiotherapy Departments.

The questionnaires on the structure of the Departments show large variations between different departments from the same country and are in good agreement with the data published by the EORTC for Research Centres (21); as an example, if we consider the physics staffing quoted by 27 answers, it can be expected that the level of physics support in a department will be directly determined by the patient load which has to be carried by the physicist. Overall, the mean number of patients per physicst is 440/year. However, while there are five centres where each physicist is responsible for less thant 200 patients, there are also five centres where the workload is above 600 patients/year, and even two with over 1000 patients. It is evident that in such extreme diverse conditions of workload, there will be differents levels of involvment of the physicist in a number of controlling and preventive measures.

There are systematic variations in some of the parameters from one country to another. but the total number of participating centres per country is still too small to draw valid general conclusions.

5.2. Dose measurements

5.2.1. Step one : Dosimetry checks in reference conditions (31)

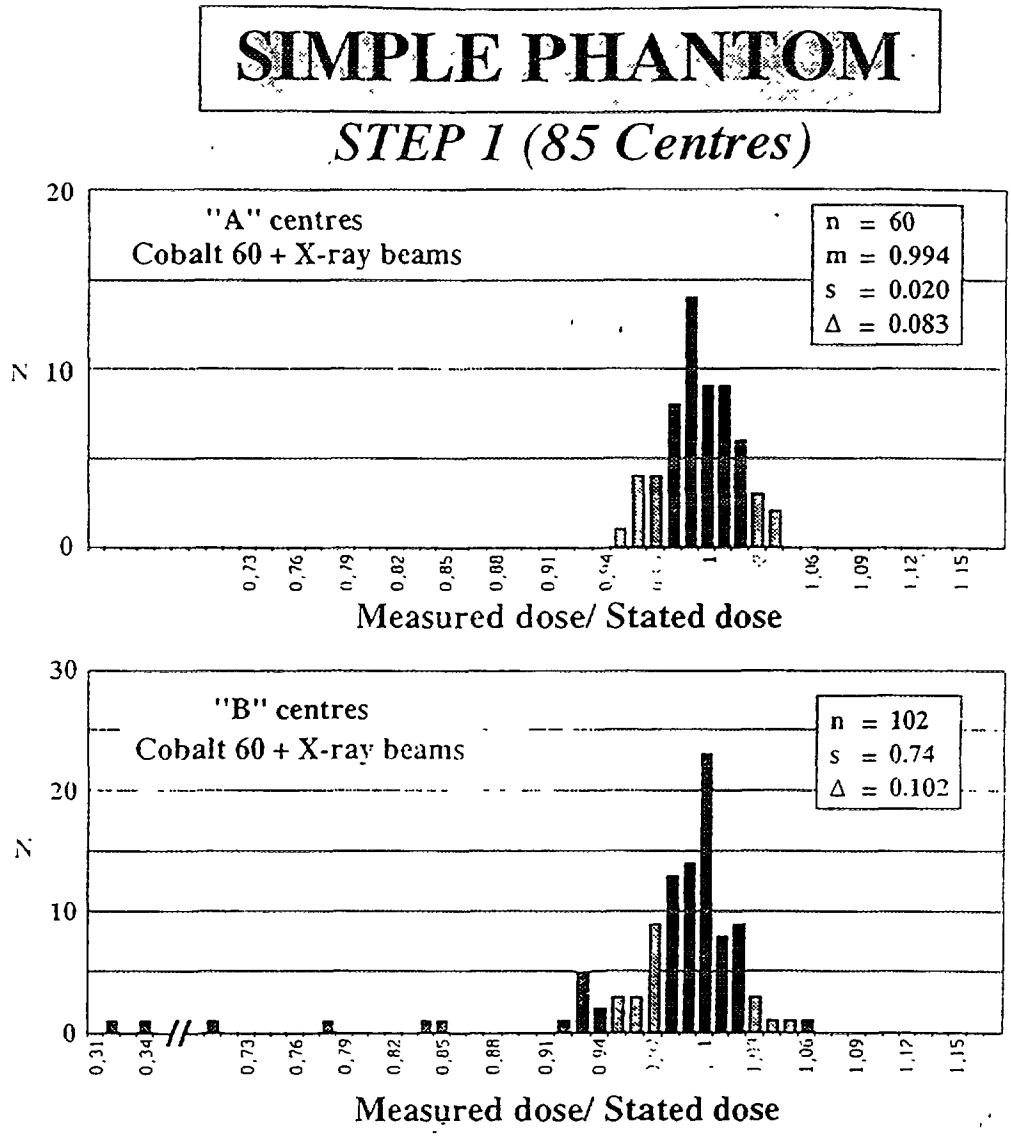
To date. cheks have been performed for 162 beams from 85 centres (58 ^{60}Co beams and 104 X-ray beams).

The postal measurements performed during the five first months of the project with research centres have presented, for the distribution of $D_{measured}/D_{stated}$, a mean of 1.0 and a standard deviation of 1.5 % with 26 ^{60}Co and X-ray beams. This uncertainty as good as 1.5 % for the measurement procedure is in agreement with the estimation by Kirby (32) of the uncertainty expected from TLD. Following this series of measurements, two critical levels of deviation have been defined :

The acceptable level corresponds to a deviation < 3 % (2 s) and the action level corresponds to a deviation $\geq$ 6 % (twice the accepted level). Below 6 %, the deviation is considered minor, whereas above 6 % the deviation is considered major.

The figure 6, representing the distribution of D_m/D_S for the check of the beam output in reference conditions exhibits, large deviations when all beams of all centres are considered. the ratios D_m/D_S ranging from 0.32 to 1.06.

Analysing the questionnaires, it appears that a large majority of the centres had not participated in any intercomparison nor had benefited from an external audit during the last five years. Sharing the distributions of the ratios D_m/D_S between such centres (called centres B) and the other centres (called centres A), it appears on figure 6 that the major deviations belong exclusively to the centres B distribution. On the contrary, the distribution related to centres A exhibits a mean value D_m/D_S very close to 1 (0.994), and a standard deviation of 2 %, for both the ^{60}Co and X-ray beams, whereas the centres B present a spectacular spread of the D_m/D_S



DEVIATIONS	"A" CENTRES		"B" CENTRES	
	3 TO 6 %		3 TO 6 %	
^{60}Co	2	0	10	4
RX	8	0	7	11
TOTAL	10/60	0/60	17/102	15/102
	(17 %)		(17 %)	

Fig.6 Distributions of the ratios D_m/D_S for the check of the beam output in reference conditions, considering "A" or "B" centres and all beams. The table presents the differences in the deviations observed between ^{60}Co beams and X-ray beams. n is the number of beams, m is the mean, s the standard deviation and Δ the total spread

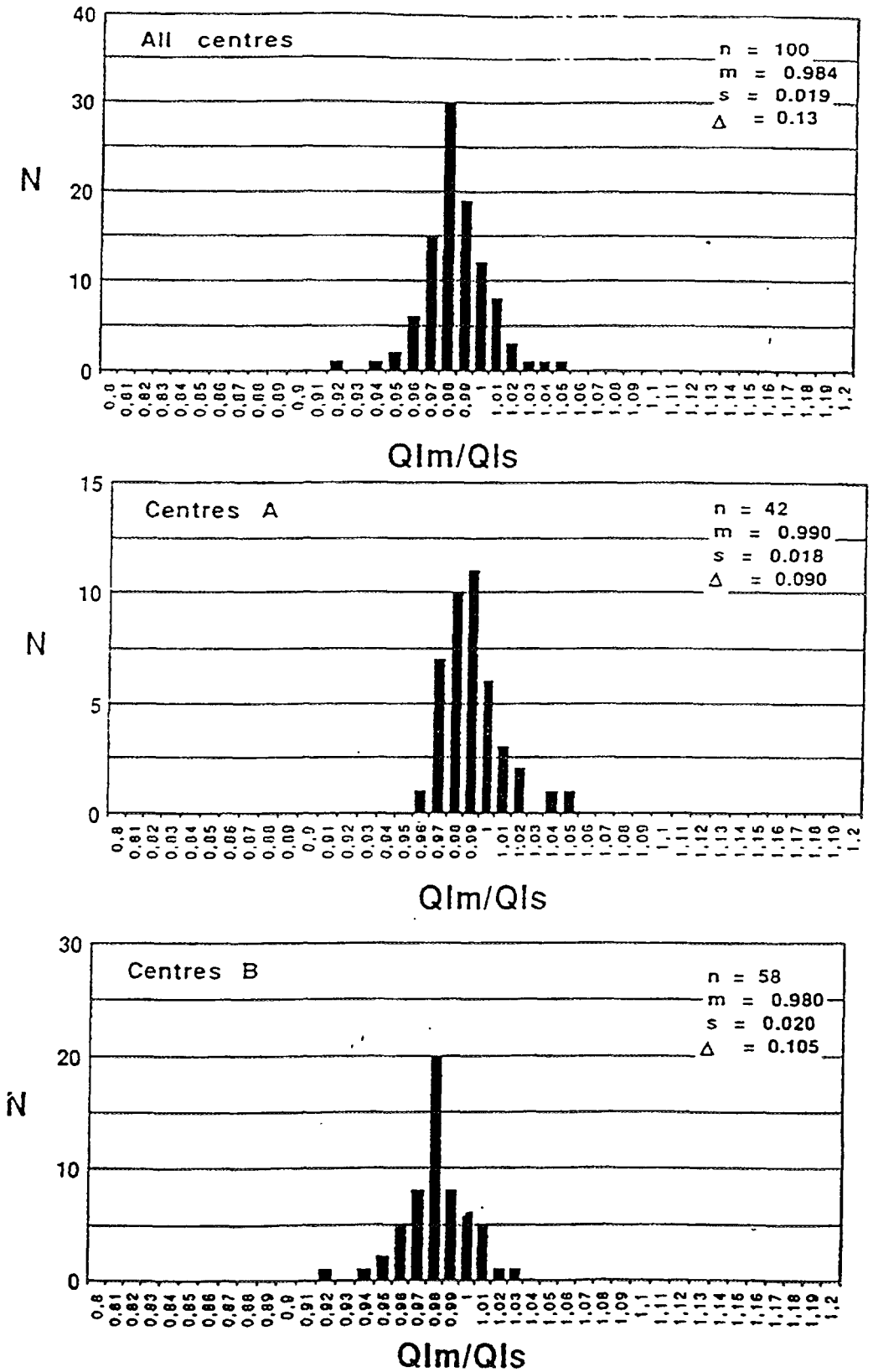


Fig.7 Distributions of the ratios QI_m/QI_S for the check of the beam quality, considering all centres and "A" or "B" centres.

n is the number of beams, m is the mean, s the standard deviation and Δ the total spread

values. The table of figure 6 presents additional data about this difference between centres A and centres B. The figure 7 is related to the determination of the Quality index (distribution of the ratios QI_m/QI_s). The analysis of the results does not show any significant difference between centres A and B. The only major deviation is nevertheless observed for a centre B. Among the 11 centres with deviations ≥ 6 %, 5 have accepted to participate in a second check, resulting in an improvement, with 3 large deviations instead of 9.

5.2.2. Step 2 : Evaluation of the multipurpose phantom

5.2.2.1. Step 2 A : Irradiation of the multipurpose phantom by European Centres (Pilot Study)

In this study, 22 photon beam qualities were checked, including 5 ^{60}Co beams, 9 X-ray beams < 10 MV and 8 X-ray beams > 10 MV.

The global results are presented in table 1.

From these results, we can report the following comments :

. for all centres and all beams, a ratio of the dose measured by TLD to the dose stated by the centre is very close to unity (1.004) with s = 2.6 %. About 21 % of minor deviations (between 3 and 6 %) and only 1.6 % of major deviations (> 6 %) are observed.

. better results have been obtained for ^{60}Co than for X-ray and for on-axis points compared to off-axis points.

Table 1 Multipurpose phantom experiments. Global results for the STEP 2 A.

Ratio of dose measured by TLD / dose stated by institution $\frac{D_m}{D_s}$:

Field	Points of meas. (Total number)	Mean of ratios $\frac{D_m}{D_s}$ (1 S) for all centres				Minor deviations (3% < - ≤ 6%)	Major deviations (> 6%)
		All energies	Cobalt-60	Low E (RX<10MV)	High E (RX>10MV)	All energies	All energies
Large anterior	Central Axis (21)	1.000 (0.024)	1.013 (0.017)	0.995 (0.025)	0.999 (0.026)	5 (24%)	0
	Off Axis:4pts (84)	1.005 (0.025)	1.017 (0.025)	1.006 (0.023)	0.995 (0.025)	18 (21%)	0
Wedged lateral	Central Axis (24)	1.000 (0.023)	0.997 (0.019)	1.001 (0.026)	1.001 (0.025)	4 (17%)	0
	Off Axis:2pts (45)	1.005 (0.031)	1.019 (0.025)	1.001 (0.024)	1.000 (0.041)	10 (22%)	3 (7%)
Combined field	Central Axis (22)	1.010 (0.030)	1.011 (0.042)	1.002 (0.021)	1.017 (0.033)	5 (23%)	1 (5%)
	Off Axis:2pts (48)	1.004 (0.025)	1.010 (0.024)	1.005 (0.025)	1.000 (0.028)	9 (19%)	0

22 BEAMS at 7 Institutions : Cobalt-60 to 25 MV X rays
Cobalt-60 : 5 beams
Low E : 9 beams : 4,5,6 and 9 MV
High E : 8 beams : 11,18 and 25 MV

. the best results have been obtained for on-axis ^{60}Co reference points for (s-value of 1.7 % to 1.9 %).

. the largest deviations have been observed for off-axis points in X-ray horizontal beams (s value ≡ 3.3 % with a Δ value of 15 %) and also for points irradiated by the two beams either for ^{60}Co and for X-ray beams (obliquity, wedge filter).

5.2.2.2. Step 2 B: Irradiation of the multipurpose phantom by local centres

The global results for this study are summarized in table II.

They lead to the following comments and conclusions :

As for the pilot study, the results are expressed, for each type of photon beam quality (^{60}Co and X-ray), as the distribution of the ratio of the dose D_m measured by TLD (at IGR) to the dose D_S stated by the centre. In each case, the distributions have been quantified by the following parameters. D_m/D_S mean value and estimation s of the standard deviation.

For each beam considered (vertical, unwedged and wedged horizontal beams), the data related to on-axis and off-axis points were examined separately.

The majority of the distributions of the ratio D_m/D_S are symmetrical and have a mean close to the unity value : however, for one X-ray horizontal unwedged beam, unexpected very low values of the ratio D_m/D_S were observed (≡ 0.33). The analysis of the results has shown that an instability of the linac has led to stop irradiation after only 77 of the 220 monitor units planned and the machine was not started again to end the irradiation.

Disregarding these very low values, the results show that, for on-axis points, deviations of the same order than for those observed in the pilot study have been obtained for the vertical beam. For that points, slightly larger deviations have nevertheless been observed.

However, for off-axis points, a large increase in the deviations has been observed for all beams. Compared to on-axis points, the estimation of the standard deviation is increased by a factor up to about 2, particularly for horizontal X-ray beams (s ≡ 4.4 to 5.5 %).

In short, the results of the run achieved in routine conditions with local centres show a large increase of the deviations for the fields which are not perpendicular to the phantom surface and, for all fields, a large increase of the deviations for off-axis points. Although a rate of minor deviations (21 % between 3 and 6 %) comparable to that obtained in the pilot study is observed for the local centres, the rate of major deviations (> 6 %) is much larger (about 15 %) than the pilot sutdy (≡ 1.5 %).

The large deviations observed in non reference conditions for two of these local centres which have obtained correct results in a previous mailed TL - intercomparison in reference conditions, point out the interest of performing such checks in order to verify on the one hand. the beam symmetry and the beam flatness and on the other hand, the accuracy of the algorithms used in routine for dose distribution calculations.

Table 2 Multipurpose phantom experiments. Global results for the STEP 2 B.

Ratio of dose measured by TLD / dose stated by Institution $\frac{D_m}{D_s}$:

Field		Points of meas. (Total number)	*Mean of ratios $\frac{D_m}{D_s}$: all centres (1 S)			Minor deviations (3% < - ≤ 6%)	Major deviations (> 6%)
			All energies	Cobalt-60	RX	All energies	All energies
Large anterior	Z = 4 cm	Central Axis (6)	1.000 (0.016)	0.997 (0.018)	1.001 (0.018)	0	0
		Off Axis (24)	1.008 (0.036)	0.997 (0.018)	1.013 (0.042)	4 (17%)	4 (17%)
	In depth	Central Axis (6)	0.999 (0.017)	0.995 (0.014)	1.001 (0.019)	0	0
		Off Axis (12)	0.991 (0.027)	0.980 (0.020)	0.996 (0.030)	3 (25%)	0
Unwedged lateral	Z = 5 cm	Central Axis (5)	0.861 ¹(0.997) (0.306) (0.017)	0.992 (0.011)	0.773 ¹(1.002) (0.397) (0.025)	0	1 (20%)
		Off Axis (10)	0.863 ¹(1.000) (0.289) (0.028)	0.994 (0.024)	0.777 ¹(1.006) (0.357) (0.033)	3 (30%)	2 (20%)
	In depth	Central Axis (5)	0.875 ¹(1.009) (0.301) (0.039)	1.012 (0.046)	0.783 ¹(1.005) (0.386) (0.049)	2 (40%)	1 (20%)
		Off Axis (10)	0.867 ¹(1.004) (0.292) (0.048)	1.007 (0.045)	0.774 ¹(1.001) (0.355) (0.058)	0	5 (50%)
Wedged lateral	Z = 5 cm	Central Axis (5)	0.974 (0.023)	0.980 (0.007)	0.970 (0.031)	2 (40%)	0
		Off Axis (10)	0.971 (0.048)	0.979 (0.020)	0.967 (0.062)	4 (40%)	2 (20%)
	In depth	Central Axis (5)	0.993 (0.023)	1.005 (0.035)	0.985 (0.013)	0	0
		Off Axis (10)	0.981 (0.046)	0.984 (0.047)	0.980 (0.050)	5 (50%)	1 (10%)

* : No correction applied for phantom composition and geometry.
1 : Without considering the very low values due to one beam (interruption of the irradiation).

6 BEAMS at 4 Institutions : Cobalt-60 to 15 MV X rays
Cobalt-60 : 2 beams
RX : 4 beams : 4,6,10 and 15 MV

6. DISCUSSION

6.1. Step 1 : Dosimetry checks in reference conditions

An important interest of this study was to show that 11 "B" centres could present deviations ≥ 6 % for the beam calibration, some being very large. We can notice first that these large deviations lead to underdosage of the patients (so avoiding alarming acute reactions) potentially inducing a lower cure rate, if the irradiation of the dosimeters is representative of the irradiation of the patients. Secondly, the physicists were supposed to check the beam output before irradiating the TLD's. From a detailed analysis, submitted to publication (31), it appears that, after on-site visits in most centres with major deviations, the physicists did not have the necessary experience and did not calibrate regularly the beams In 6 centres out of 11, there was no dosimeters or the dosimeters available had not been calibrated recently In 3 centres, the physicists did not give any explanation.

6.2. Step 2 : Multipurpose phantom

It is interesting to quote that, with centres B, the large deviations observed in non reference conditions appeared in centres having obtained correct results for the step 1. So, it is very important, in such a network, not to limit the investigations to the reference conditions It seems worthwhile to go further in the evaluation of such a phantom, an improved version being under developpement (WH Hanson). The procedure could also be improved (already, from step 2.A to step 2.B a better tracing of the reasons for deviations is obtained). The problem is the number of dosimeters involved, making the large scale use of this phantom expensive

7. CONCLUSION

This pilot study has shown the feasibility of the experimental european network, with its organisation relying on Coordinating, Measuring and Reference National Centres. This experience can allow each european country to plan the set-up of a national network connected to the european network.

From the scientific point of view, the accurancy of the dose checks is compatible with the action levels considered in most instances of the radiotherapy litterature.

The discovery of numerous major deviations outlines the need for any European radiotherapy department to be able to benefit from a QA network and support the initial application to the EC for such a project.

Further improvements are expected from the pilot study, regarding the accuracy, the optimisation of the procedures, the addition of electron beams checking and *in vivo* measurements.

The final outcome will rely on the capability of each country to develop its national network.

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V. QUALITY ASSURANCE PROGRAMME IN RADIOTHERAPY

MINIMUM REQUIREMENTS ON A QA PROGRAM IN RADIATION ONCOLOGY

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Abstract

In April, 1994, the American Association of Physicists in Medicine published a "Comprehensive QA for radiation oncology:" a report of the AAPM Radiation Therapy Committee. This is a comprehensive QA program which is likely to become the standard for such programs in the United States. The program stresses the interdisciplinary nature of QA in radiation oncology involving the radiation oncologists, the radiotherapy technologists (radiographers), dosimetrists, and accelerator engineers, as well as the medical physicists. This paper describes a comprehensive quality assurance program with the main emphasis on the quality assurance in radiation therapy using a linear accelerator. The paper deals with QA for a linear accelerator and simulator and QA for treatment planning computers. Next the treatment planning process and QA for individual patients is described. The main features of this report, which should apply to QA programs in any country, emphasizes the responsibilities of the medical physicist.

1. Introduction

The role of quality assurance in radiation oncology has received increasing attention in the last few years and its importance is now fully recognized in maintaining consistent accuracy of the absorbed dose delivered to patients undergoing radiation therapy. Sources of error can derive from deficiencies in tumor localization, patient immobilization, field placement, daily patient set-up, dose calibration and calculation as well as from equipment related problems.

Quality assurance in radiation oncology may be defined as those procedures that ensure a consistent and safe fulfillment of the dose prescription to the target volume, with minimal dose to normal tissues and minimal exposure to personnel and the public (AAPM 1984). [1]

The QA program in radiation therapy, therefore, covers a wide range of areas, often involving several medical disciplines and the medical institutions management. Coordination, therefore, is critical among radiation oncology physicists, dosimetrists, accelerator engineers, radiation oncologists, radiotherapy technologists (radiographers), other medical disciplines and management. In many institutions the medical physicist is best placed to oversee such a program.

The aim of a QA program for radiation therapy is an ongoing evaluation of the functional performance characteristics of the associated equipment and calculations, because these characteristics influence both the geometrical and dosimetric accuracy of the applied dose. There are two main parts of such a program: (i) periodic QA measurements and

evaluation and (ii) regular preventative maintenance. The medical physicist should be responsible for making sure that both parts of the program are carried out.

The three main areas for sources of inaccuracy in dose delivery can be identified as:

a) Physical dosimetry, i.e., the commissioning and calibration of treatment machines and sources.

b) Clinical dosimetry, i.e., the delineation of the target volume and critical structures, acquisition of patient specific factors and dose distribution calculations.

c) Patient treatment, i.e., the setup of the patient and the recording of the treatment and final verification of the accuracy of the delivered dose.

Any QA program will be based upon a complete determination of baseline values at the time of acceptance and commissioning of the equipment. Data for any machine must be measured and should not be assumed identical to data from similar equipment. Most manufacturers provide in written form, their acceptance test procedures which list the mechanical and radiation parameters which will provide the benchmark for the equipment. Commissioning provides the detailed information about the equipment, e.g., tables of beam data. This data obtained for each piece of equipment adds to the benchmark data. Once the acceptance tests, commissioning and calibrations have been completed a QA program must commence to insure that the accuracy of the treatments is maintained, i.e., the goal of such a program is to assure that the performance characteristics established during commissioning shows no serious deviations.

2. The QA Program in Radiation Therapy

QA in a radiation therapy department covers a wide range of activities and the treatment process can be viewed in many different ways. For the purposes of this discussion three main areas have been identified. They are:

a. External Beam Treatments
b. Brachytherapy Treatments
c. Measurement Equipment
d. Clinical Aspects of the Treatments

These areas are show in Table I. In discussing the topic of this seminar "Radiation Dose in Radiotherapy from Prescription to Delivery" all of the three aspects mentioned above are of importance to quality assurance.

Because it is not possible to cover this whole subject in a single paper the quality assurance program for a linear accelerator through the completion of a patient treatment will be presented in some detail and although the specifics will be different for other areas the general approach will be quite similar. Listed in Table II are some general references which describe quality assurance programs for all aspects of radiation oncology.

Table I
QA Program in a Radiation Oncology Department

A. External Beam Treatments
  1. External Beam Radiation Equipment
    a) Superficial and Orthovoltage Machines
    b) Cobalt 60 Units
    c) Medical Electron Accelerators
      (i) X-rays
      (ii) Electron Beams
  2. Simulators and CT Scanners
    a) Simulators
    b) CT Scanners
  3. Treatment Planning Computers
    a) Treatment Planning Program
    c) Test Procedures
  4. External Beam Treatment Planning
    a) Treatment Planning Process
    b) Treatment Planning for Individual Patients

B. Brachytherapy
  1. Sealed Sources
    a) Calibration
    b) Wipe Tests
  2. Unsealed Sources
    a) Calibration
    b) Contamination
  3. Source Inventories
    a) Long Lived Sources
    b) Short Lived Sources
  4. Brachytherapy Equipment
    a) Applicators
    b) Seed Inserters
  5. Remote Afterloaders
    a) Calibration of Source
    b) Source Position
  6. Treatment Planning and Dosimetry
    a) Source Localization
    b) Dose Calculation
    c) Delivery of Treatment

C. Measurement Equipment
  1. Standard and Field Instruments for Absorbed Dose
  2. Relative Dose Detectors
  3. Patient Monitoring Detectors
  4. Beam Scanning Equipment
  5. Accessories

D. QA of Clinical Aspects
  1. Chart Review
  2. Film Review
  3. Beam Modifying Devices

Table II
Physical Aspects of Quality Assurance in Radiation Therapy AAPM Report No. 13 Task Group 24 (New York, NY - American Association of Physicists in Medicine 1984) (AAPM 1984) [2]
Comprehensive QA for radiation oncology: Report of AAPM Radiation Therapy Committee Task Group 40 1994 (AAPM 1994) [1]
Report No 2 Radiation Control and Quality Assurance in Radiation Oncology: A Suggested Protocol. The American College of Medical Physics 1986 (ACMP 1986) [3]
John L. Horton: "Handbook of Radiation Therapy Physics" 1987, Prentice Hall, Inc., New Jersey (Horton 1987) [4]
Commissioning and Quality Assurance of Linear Accelerators, The Institute of Physical Sciences in Medicine, Report No. 54, 1988 (IPSM 1988) [5]
Quality Assurance in Radiotherapy Physics Proceedings of an American College of Medical Physics Symposium. May 1991, Medical Physics Publishing (ACMP 1991) [6]

3. A QA Program in Radiation Therapy Using a Linear Accelerator

The QA program for radiation therapy treatments with a linear accelerator will involve other equipment and areas and these are listed in Table III.

Table III
QA of Medical Linear Accelerators
QA of Simulators
QA of Measurement Equipment
Treatment Planning Computers
External Beam Treatment Planning: General Individual Patients
In-Vivo Dosimetry
Chart Review

In developing a QA program it is important to use measurement techniques which are simple, rapid, and reproducible and which can determine parameter changes smaller than the tolerance or action level and which will minimize the test time.

4. The QA of Medical Linear Accelerators

The most significant factors affecting the ability to deliver the correct tumor dose include: exact dose calibration, accurately determined depth dose, off-axis dose characteristics, wedge and block factors, etc., that will be obtained during commissioning of a linear accelerator. In addition, certain other data will be obtained during acceptance test for such machines, including measurements of the mechanical, radiation-beam and radiation protection specifications. A good QA program will, therefore, monitor each of these

parameters in order to ensure that the machine is performing within specifications and that the dose is being accurately delivered. The protocol for these QA tests should recommend the equipment to be used, the frequency of measurement, the techniques to be followed, and suggested performance criteria. Obviously the equipment and techniques should be a simple as possible in order to reduce the time, effort, and cost of such tests. The frequency is chosen depending upon the likelihood of a change and the effect upon treatment if a change should unexpectedly occur. Finally, action and notification levels must be established.

Table IV
QA of Medical Accelerators

Frequency	Procedure	Tolerance[a]
Daily	Dosimetry	
	X-ray output consistency	3%
	Electron output constancy[b]	
	Mechanical	
	Localizing lasers	2 mm
	Distance indicator (ODI)	2 mm
	Safety	
	Door interlock	Functional
	Audiovisual monitor	Functional
Monthly	Dosimetry	
	X-ray output constancy[c]	2%
	Electron output constancy[c]	2%
	Backup monitor constancy	2%
	X-ray central axis dosimetry parameter (PDD, TAR) constancy[c]	2%
	Electron central axis dosimetry parameter constancy (PDD)	2 mm @ therapeutic depth
	X-ray beam flatness constancy	2%
	Electron beam flatness constancy	3%
	X-ray and electron symmetry	3%
	Safety Interlocks	
	Emergency off switches	Functional
	Wedge, electron cone interlocks	Functional
	Mechanical Checks	
	Light/radiation field coincidence	2 mm or 1% on a side[d]
	Gantry/collimator angle indicators	1 deg
	Wedge position	2 mm (or 2% change in transmission factor)
	Tray position	2 mm
	Applicator position	2 mm
	Field size indicators	2 mm
	Cross-hair centering	2 mm diameter
	Treatment couch position indicators	2 mm/1 deg
	Latching of wedges, blocking tray	Functional
	Jaw symmetry[e]	2 mm
	Field light intensity	Functional
Annual	Dosimetry	
	X-ray/electron output calibration constancy	2%
	Field size dependence of x-ray output constancy	2%
	Output factor constancy for electron applicators	2%
	Central axis parameter constancy (PDD, TAR)	2%
	Off-axis factor constancy	2%
	Transmission factor constancy for all treatment accessories	2%
	Wedge transmission factor constancy[f]	2%
	Monitor chamber linearity	1%
	X-ray output constancy vs gantry angle	2%
	Electron output constancy vs gantry angle	2%
	Off-axis factor constancy vs gantry angle	2%
	Arc mode	Mfgrs. specs.
	Safety Interlocks	
	Follow manufacturers test procedures	Functional
	Mechanical Checks	
	Collimator rotation isocenter	2 mm diameter
	Gantry rotation isocenter	2 mm diameter
	Couch rotation isocenter	2 mm diameter
	Coincidence of collimetry, gantry, couch axes with isocenter	2 mm diameter
	Coincidence of radiation and mechanical isocenter	2 mm diameter
	Table top sag	2 mm
	Vertical travel of table	2 mm

[a] The tolerances listed in the tables should be interpreted to mean that if a parameter either: (1) exceeds the tabulated value (e.g., the measured isocenter under gantry rotation exceeds 2 mm diameter); or (2) that the change in the parameter exceeds the nominal value (e.g., the output changes by more than 2%), then an action is required. The distinction is emphasized by the use of the term constancy for the latter case. Moreover, for constancy, percent values are ± the deviation of the parameter with respect its nominal value; distances are referenced to the isocenter or nominal SSD.
[b] All electron energies need not be checked daily, but all electron energies are to be checked at least twice weekly.
[c] A constancy check with a field instrument using temperature/pressure corrections.
[d] Whichever is greater. Should also be checked after change in light field source.
[e] Jaw symmetry is defined as difference in distance of each jaw from the isocenter.
[f] Most wedges' transmission factors are field size and depth dependent.

The QA measurements recommended for medical accelerators is given in Table IV grouped by the frequency of such measurements and is taken from the latest AAPM report on "Comprehensive QA for radiation oncology" (AAPM 1994). [1] The beam parameters and the tolerances are given and the baseline for each measurement should be those established during acceptance and commissioning. It should be pointed out that some flexibility should be applied to this table, especially with regard to the frequency. Some monthly tests may be done weekly, for example a policy for the precise frequency at which each test is done should be developed at each institution.

5. The QA of Simulators

Since simulators are designed to reproduce the geometric conditions of the radiation therapy equipment (BJR 1989) [5] they are subject to the same mechanical checks as the accelerators. In addition, the simulator should be checked for image quality according to guidelines for diagnostic radiography units (AAPM 1984 [2], ACMP 1986 [3]). Table V summarizes the QA tests for simulators.

Table V
QA of Simulators

Frequency	Procedure	Tolerance[a]
Daily	Localizing lasers	2 mm
	Distance indicator (ODI)	2 mm
Monthly	Field size indicator	2 mm
	Gantry/collimator angle indicators	1 deg
	Cross-hair centering	2 mm diameter
	Focal spot-axis indicator	2 mm
	Fluoroscopic image quality	Baseline
	Emergency/collision avoidance	Functional
	Light/radiation field coincidence	2 mm or 1%
	Film processor sensitometry	Baseline
Annual	Mechanical Checks	
	Collimator rotation isocenter	2 mm diameter
	Gantry rotation isocenter	2 mm diameter
	Couch rotation isocenter	2 mm diameter
	Coincidence of collimator, gantry, couch axes and isocenter	2 mm diameter
	Table top sag	2 mm
	Vertical travel of couch	2 mm
	Radiographic Checks	
	Exposure rate	Baseline
	Table top exposure with fluoroscopy	Baseline
	KvP and mAs calibration	Baseline
	High and low contrast resolution	Baseline

[a]The tolerances mean that the parameter exceeds the tabulated value (e.g., the measured isocenter under gantry rotation exceeds 2 mm diameter).

6. QA of Measurement Equipment

The measurement equipment is equally important as the radiation treatment equipment and should be part of the QA program. The recommended QA tests, frequency, and tolerance limits are given in Table VI including tests for automatic beam scanners.

Table VI

QA of measurement equipment. I, initial use for each mode used or following malfunction and repairs; E, each use (measurement sequence) or ongoing evaluation; B, each batch or box at appropriate energy (dosimeter element position should also be considered); D, documented and correction applied or noted in report of measurement; M, monthly.

Instrument type	Test	Frequency	Tolerance[a]
Local Standard[b]	ADCL calibration	2y[c]	D
	Linearity	2y[c]	0.5%
	Venting	2y[c]	D
	Extra-cameral signal (stem effect)[d]	I	0.5%
	Leakage	E	0.1%
	Redundancy check[e]	E	2%
	Recombination	I	D
	Collecting potential	E	D
Field instruments	Local std. comparison	2y	1%
	Linearity	2y	D
	Venting	2y	D
	Extra-cameral signal	2y	D
	Leakage	E	0.1%
	Recombination	I	D
	Collecting potential	E	D
Output Check	Local std. comparison	M	1%
Relative dose			
TLD	Calibration	E	D
	Linearity	I	D
Film	Dose response	B	D
	Densitometer linearity	1y	D
	Processor uniformity/reproducibility	E	D
Ion Chamber	Linearity	1y	D
	Extra-cameral signal	I	1%
Diodes	Energy dependence	I	D
	Extra-cameral signal	I	D
	Linearity	I	D
Positioning	Accuracy	E	2 mm
	Hysteresis	E	2 mm
Automated Scanners	Mechanical	I	2 mm
	Positional accuracy	E	1 mm
	Collecting potential of detector	E	D
	Detector linearity	I	0.5%
	Extra-cameral signal	I	0.5%
	Detector leakage	E	0.5%
	Accuracy of data analysis	I	1%
	Accuracy of printouts	I	1 mm
Accessories	Thermometer Calibration	I	0.1 deg/C
	Barometer Calibration	3 mo	1mm/Hg
	Linear rule Calibration	I	0.3%

[a]Percent values are ± the deviation of the parameter with respect to the nominal, and distances are referred to the isocenter or nominal SSD.
[b]Local standard instrument has a calibration directly traceable to NIST and should be reserved for calibration of radiation beams, field instruments, and intercomparisons.
[c]Two years required by NRC. Without a redundancy program, this may be inadequate, with a redundancy program, dosimetry systems maintain calibration factors for significantly longer periods of time.
[d]With a radionuclide (e.g., SR-90) or chamber intercomparison.

Redundancy in dose calibration equipment is necessary to assure that instruments are holding their calibration. This can be established by comparing the response of the measurements equipment with an appropriate long-lived radioactive source (Strontium-90). If access to a Strontium-90 check source is unavailable, then at least two independent dosimetry systems should be maintained. A Cobalt 60 teletherapy machine can be used as part of a redundant measuring system but with care. If only one dosimetry system is available, a redundant system should be formed with a dosimetry system at another institution, with quarterly intercomparisons.

7. Treatment Planning Computers

The treatment planning computer is a critical component of the entire treatment process. Computers may be used to calculated patient dose distributions, monitor units for a

given prescribed dose and fixed point dose calculations for irregular fields, etc. All such systems should undergo rigorous acceptance testing and commissioning and a QA program implemented. Complete documentation by the manufacturer should include the methods for obtaining the beam data library and other data necessary to implement the system. A complete description of the physical models for dose calculations with expected accuracy and limitations should be provided along with complete input-output and operating instructions. Various test procedures need to be described to be carried out initially by the manufacturer and then by the user. Tests should also be done after program modifications and as part of an ongoing QA program. Table VII lists the recommended QA for treatment planning systems and monitor unit calculations.

Table VII
QA for Treatment Planning Systems and Monitor Unit Calculations

Frequency	Test	Tolerance[a]
Commissioning and following software update	Understand algorithm Single field or source isodose distributions MU calculations Test cases I/O system	Functional 2%[a] or 2 mm[b] 2% 2% or 2 mm 1 mm
Daily	I/O devices	1 mm
Monthly	Checksum Subset of reference QA test set (when checksums not available) I/O system	No change 2% or 2 mm[c] 1 mm
Annual	MU calculations Reference QA test set I/O system	2% 2% or 2 mm[d] 1 mm

[a]% difference between calculation of the computer treatment planning system and measurement (or independent calculation).
[b]In the region of high dose gradients the distance between isodose lines is more appropriate than % difference. In addition, less accuracy may be obtained near the end of single sources.
[c]These limits refer to the comparison of dose calculations at commissioning to the same calculations subsequently.
[d]These limits refer to comparison of calculations with measurement in a water tank.

8. External Beam Treatment Planning

In this section, QA for the treatment planning process is discussed, followed by a discussion of QA for individual patients. QA in treatment planning may refer to two distinct processes. (1) Nongraphical planning is often used for single or parallel opposed fields. In this approach, the monitor units (minutes) for the prescribed dose to a point on the central axis is usually calculated using central axis depth dose, tissue phantom ratios or tissue maximum ratios, and beam output calibration tables. Furthermore, the field apertures, which define the treatment volume, are usually designed on radiographs obtained during simulation; (2) Traditional graphical planning is used for many patients. In this method, a target volume is defined from CT or orthogonal simulation films, and the patient's contour is either obtained using a mechanical device (e.g., lead solder wire) or from CT. The field arrangements are designed and dose distributions calculated on one or a limited number of axial cross sections using a computerized treatment planning system. The radiation oncologist then prescribes the dose to a point or an isodose curve, and the field apertures are usually defined, as in procedure 1, from simulation films.

8.1. General Treatment Planning Process

Treatment planning is a process that begins with patient data acquisition and continues through graphical planning, plan implementation and treatment verification. It entails interactions between the radiation oncology physicists, dosimetrists, radiation oncologists, residents, and radiation therapists, and the use of a large number of software programs and hardware devices for graphical treatment planning. Each step of the complex treatment planning process involves a number of issues relevant to quality assurance. The process is represented schematically in Table VIII.

Table VIII
Treatment Planning Process

Process	Related QA Procedures
Positioning and immobilization	Port films. Laser alignment
Simulation	Simulator QA including image quality and mechanical integrity
Patient data acquisition (CT, MRI, manual contouring)	CT, MRI QA including image quality and mechanical integrity (Accuracy of mechanical contouring)
Data transfer to treatment planning system	QA of the entire data transfer process, including digitizers, digital data transfer, etc.
Definitions of target volumes	Peer review, e.g., new patient planning conference, chart rounds.
Aperture design	Independent check of delivery (e.g., port films, and peer review)
Computation of dose distributions	Machine data from commissioning and QA of treatment machines. Accuracy and QA of treatment planning system.
Plan evaluation	Peer review of plan, e.g., during chart rounds. Independent check by radiation oncology physicist.
Prescription	Written, signed, and dated.
Computation of monitor units	Treatment planning system QA. Independent check within 48 hours
Production of blocks, beam modifiers	QA for block cutting and compensator systems. Port film review.
Plan implementation	Review of set-up by treatment planning team. Chart review.
Patient QA	Treatment plan review. Chart review after new or modified field, weekly chart review, port film review. In-vivo dosimetry for unusual fields, critical organ doses (e.g., gonadal dose). Status check, followup.

8.2 Individual Treatment Planning Process

All parameters in the treatment plan should be verified during first set-up so that any ambiguities or problems can be corrected immediately. Special care should be taken to assure that all beam modifying devices (blocks, wedges, compensators) are correctly positioned. Although errors in block fabrication and mounting are often discovered during the review of port films, wedge or compensator misalignment is much more insidious, and may remain throughout the course of treatment if not discovered during initial patient setup. A check of the setup by the physicist will minimize errors that may be undetected due to misunderstanding of the physical concepts and details of QA recommendations for individual patients is given in Table IX.

Table IX
Summary of QA Recommendations for Individual Patients

Procedure	Recommendations
Monitor unit (minutes) calculation	Reviewed prior to treatment by an authorized individual who did not perform initial calculation, or when not possible (e.g., emergency treatment), then prior to 3rd fraction or before 10% of the dose has been delivered, whichever occurs first.
Graphical treatment plan review	1. Reviewed prior to treatment, or when not possible, then prior to 3rd fraction or before 10% of the dose has been delivered, whichever occurs first. 2. Reviewed by a radiation oncology physicist who did not formulate treatment plan. Where only one physicist and that person performed the plan, then reviewed by another authorized individual. 3. Review includes calculated monitor units, input-output and plan quality. 4. Independent calculation of dose at a point: Compare for each field - with an independent calculation of dose to a point using the calculated monitor units - the prescribed and calculated dose. 5. If these differ by more than 5%, then the discrepancy should be resolved before continuing treatment.
Plan set-up	Radiation oncologist present at first set-up for major changes in treatment.
Beam (portal) films---curative and high morbidity risk palliative patients	Initial films reviewed by radiation oncologist prior to first treatment. In addition, ongoing portal films (the standard is weekly) also reviewed by the radiation oncologist.
Beam (portal) films---palliative patients	Films reviewed prior to second fraction.
In-vivo dosimetry	1. All institutions should have access to TLD or other in-vivo dosimetry systems. 2. Should be used to measure dose to critical structures (e.g., lens, gonads). 3. May be used to record dose for unusual treatment conditions.

9. In-Vivo Dosimetry

In-vivo dosimetry can be used to identify major deviations in the delivery of treatment and to verify and document the dose to critical structures. Institutions should have access to TLD or other in-vivo systems. Thermoluminescent dosimetry (TLD) is often used because the device is small and relatively easy to calibrate, while diodes have the advantage of instantaneous readout. These in-vivo systems can have relatively large uncertainties, which should be assessed before using them. While in-vivo systems are useful for individual patient measurements, they should not substitute for an adequate QA program.

10. Chart Review

A procedure for checking the patient charts for the technical parameters of treatment should be developed. Below is given an outline of the parameters to be checked and verified:

10.1 Review of New or Modified Treatment Fields

The first task of the chart review is to identify any changes in the treatment or new treatment fields since the previous review. The following specific areas of the chart should be reviewed:

a) Treatment Prescription
b) Simulator Instructions
c) Isodose Distribution, Special Dose Calculation Measurement
d) MU (Minutes Calculated)
e) In-Vivo Measurements
f) Daily Record

10.2 Weekly Chart Review

In addition to the initial chart check a weekly review should take place:

a) Review of Previous Fields
b) Cumulative Dose

10.3 Review at Completion of Treatment

As a final review before the chart is placed in a file, the following items should be checked:

a) Prescribed dose delivered
b) Chart properly documented according to department policy
c) Treatment summary included

11. The Role of the Radiation Oncology Physicist

Radiation oncology physicists are primarily and professionally engaged in the evaluation, delivery, and optimization of radiation therapy. A major responsibility of the radiation oncology physicist is to provide a high standard of clinical physics and supervision. The roles and responsibilities of the radiation oncology physicist in QA are outlined below.

11.1 Specifications of Therapy Equipment

The radiation oncology physicist should help define the specifications for the purchase of treatment unit(s), therapy simulator(s), therapy imaging systems (e.g., CT Scanner, on-line portal imaging systems), and treatment planning system. The radiation oncology physicist is involved in the design of the facility and must assure that all radiation safety requirements are met.

11.2 Acceptance Testing, Commissioning and QA

The radiation physicist is responsible for acceptance testing, commissioning, calibration, and periodic QA of therapy equipment. In particular, the physicist must certify that the therapy units and planning systems are performing according to specifications, generate beam data, and outline written QA procedures which include tests to be performed, tolerances, and frequency of the tests.

11.3 Measurement and Analysis of Beam Data

Important components of the commissioning phase include not only the generation of beam data for each modality and energy, but also evaluation of the quality of the data and its appropriateness for treating different disease sites. Such evaluation may lead to the initiation of further measurements and refinements for different treatment techniques.

11.4 Tabulation of Beam Data for Clinical Use

It is the responsibility of the radiation oncology physicist to assure that the beam and source data are correctly entered into the treatment planning system and that a printed copy of the beam data is tabulated in a form that is usable by the radiation therapist, dosimetrists, and radiation oncologists.

11.5 Calibration of Radiation Oncology Equipment

One of the primary responsibilities of the radiation oncology physicist is to assure that all treatment machines and radiation sources are correctly calibrated according to accepted protocols.

11.6 Establishment of Dose Calculation Procedures

A major responsibility of the radiation oncology physicist is to establish the dose calculation procedures that are used throughout the department--and to assure their accuracy.

11.7 Establishment of Treatment Planning and Treatment Procedures

Along with the radiation oncologist and other members of the treatment planning team, the radiation oncology physicist is responsible for establishing treatment planning and treatment procedures. This includes both the technical aspects of the process (e.g., how block cutting is to be performed) and the flow of procedures entailed in the process (e.g., when different steps in the process of planning are to be performed).

11.8 Treatment Planning

The physicist should perform or oversee the determination of radiation dose distributions in patients undergoing treatment (i.e., computerized treatment planning or direct radiation measurements). This includes consultation with the radiation oncologist and the evaluation and optimization of radiation therapy for specific patients.

11.9 Establishment of QA Procedures

The physicist can ensure that the policies and procedures contain proper elements of good radiation oncology practice, delivery of treatment, radiation safety, quality control, and regulatory compliance (AAPM, 1987). Moreover, the radiation oncology physicist should perform a yearly review of the appropriate sections of the policies and procedures manual.

11.10 Supervision of Therapy Equipment Maintenance

Regular maintenance of the treatment machines is required and can be overseen by the radiation oncology physicist. While the supervising radiation oncology physicist may not perform the actual machine maintenance, he or she is responsible for releasing a treatment machine into clinical service after maintenance, and for documenting that any alteration caused by the maintenance and repair schedule does not affect the accelerator performance or calibration.

12. Discussion

The treatment of a patient with radiation is a complex undertaking with multiple steps. An error or miscalculation in any step effects the overall treatment aim. The quality assurance for such an undertaking is, therefore, complex, involving many steps and the medical radiation physicist is often the one person qualified to oversee the total program.

Although this paper describes in some detail the various steps for treatments done with medical linear accelerators, essentially the same steps must be carried out for treatments with Cobalt 60 teletherapy, High Dose Rate brachytherapy systems, brachytherapy with sealed sources or the therapy application of unsealed sources. More details can be found in the references and in particular the AAPM Task group 40 Report. [1]

Also, even though the QA program has been written as though various computers are available, e.g., for treatment planning or dose monitor calculations, the process applies also for hand calculations. In these cases the hand calculations should be checked by an independent calculation.

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ACCURACY IN RADIOSURGERY: THE INFLUENCE OF COLLIMATOR DIAMETERS AND ARC WEIGHTS ON THE DOSE DISTRIBUTION FOR SINGLE TARGET

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Abstract

The dosimetric characteristics of mini-beams and dose distributions in beams used for radio surgery defer substantially from beams used in common radiotherapy. The aim of radio surgery is to deliver a high dose to the lesion in one single fraction, while minimising the dose delivered to the surrounding normal brain tissue. This type of irradiation is performed with a number of continuous arcs located in various corneal (patient sitting) or sagital (patient in a supine position) inclined planes using a linear accelerator. A treatment planning system should take into account a large number of irradiation parameters such as the collimator diameter, number of arcs, their angular positions, length and weight of the arcs. We analysed the influence of collimator diameters in the range of 6 to 20 mm using 15 MV X-rays and stereo-tactic irradiation of ellipsoidal inclined arterio venous malformations (AVMs) with a single isocenter. Special arc weights were used to obtain an optimised dose distribution with 13 arcs distributed over an angular sector of 120° x130 °. In the two studies made we used 3 dimensional dosimetric calculations. The results were used for the treatment of patients and enabled the choice of the optimal irradiation configuration for each patient.

I Introduction

The dosimetric characteristics of minibeams and dose distributions used for Radiosurgery are very different from the radiation beams used in clinical Radiotherapy.

The aim of Radiosurgery is to deliver a high dose to the lesion in one single fraction, while minimizing the dose delivered to the surrounding normal brain tissue. This type of irradiation is performed with a number of continous arcs situated in different coronal (sitting position) or sagittal (supine position) inclined planes using a linear accelerator. A treatment planning system should take into account a large number of irradiation parameters such as the collimator diameter, number of arcs, their angular positions and angular lengths and the weight of the arcs, in order to obtain dose distributions adapted to each individual treatment plan. To stereotactically irradiate ellipsoidal inclined arteriovenous malformations (AVMs) with a single isocenter needs the special arc weights to obtain a optimise dose distribution.

We analysed the influence of diameter collimators in the range from 6 to 20 mm using 15 MV X-rays and arc weights (0 - 1.0) using 13 arcs distributed over an angular sector of 120^{o} x130^{o} (Fig. 1)

II Material and Methods

A. Material

The studies were realized with :

-Saturne 43 (GE-CGRMeV) linear accelerator (15 MV X-ray) with isocentric technique (ASD=100 cm),

-Talairach stereotactic fixation system,

-a special movable seat (O.Betti)[3] to hold the patient,

-3D dosimetric planning system (VaxStation 3200 and array processor Numerix-332 using ARTEMIS_3D planning program) with a dose matrix of 100mm x 100mm x 100mm,

-collimator diameter : 6,8,10,12,14,16, 18 and 20 mm.

B. The 3D treatment planning

In Radiosurgery, the treatment planning is a fundamentally three-dimensional task. The ARTEMIS_3D treatment planning system has an emphasis on three dimensional computations and graphics. It has been developed at Tenon Hospital and it is based on the general formulation proposed by Siddon [1]. The Tenon methodology the minimal therapeutic dose is delivered at the periphery of the lesion. It corresponds usually to the 60%-70% isodoses.

The dose contribution from each arc is simulated by stationary beams separated by 10^{o} increments, after which the computer system retrieves the necessary percentage depth dose data and the dose profiles and calculates the dose distribution on a 64 x 64 x 64 mm or 100 x 100 x 100 mm 3D dose matrix.

The system allows qualitative evaluation, such as isodose surface displays and quantitative evaluation such as dose-volume analysis.

C. Paramenters for the evaluation of dose distribution

The different means that can be used to demonstrate the quality of the treatment plans are one dimensional, two dimensional and three dimensional. Taking into account the importance of the functional structures surrounding the lesion, we propose in these studies the dose-volume analysis, because the planar dose distribution (coronal, sagittal, and transversal planes) do not give a complete evaluation. Dose-volume histograms are a useful feature of treatment plans and complement the information coming from planar dose distributions. To check the treatment planning quality by assessing the integral dose potential variations inside and outside the lesion, we employed:

(a) *The reference isodose surface volume* (VRI) which represents the lesion volume. The reference isodose volume represents the therapeutic part of the irradiation.

(b) *The volumetric dose fall-off* $(VFO)_{RI\text{-}IL\%}$ is the difference between the reference isodose (RI) and an arbitrary inferior limit isodose (IL) volumes. In these studies the RI is the 70% isodose and the inferior limit is the 10% .isodose. It represents the possible damage to the neurological healthy structures.

(c) The ratio of the volumetric dose fall-off to the reference isodose volume $(VFIR)_{RI\text{-}IL\%}$ expressed as the "multiplying factor" of the healthy irradiated tissues volume compared to the lesion volume. It represents the volume of healthy tissues which should be irradiated in order to achieve the planned lesion treatment. It is a measurement of the radiosurgical treatment quality outside the lesion. One should select the smaller VFIR irradiation configuration when a choice between the two different plans with the same reference isodose volume has to be made.

D. Method

We present 2 studies analysing the influence of the collimator diameter and the arc weight on the dose distribution.

1. Collimator diameter

The influence of eight different collimators (6 to 20 mm in diameter) over an angular irradiation sector of 130^{o} using 13 arcs spaced by steps of 10^{o} (total angulation of 120^{o})

2. Arc weight

This study was based on clinical cases. The AVM was considered to be in the center of the reference stereotactic frame (Ant=0 mm, Right=0 mm, Sup=0). We have studied the relationship between the AVM long axis and the superior-inferior direction angle β and the weight of the arcs. The irradiation space is represented geometrically in the same conditions to collimator diameters study (120^{o} x 130^{o}).

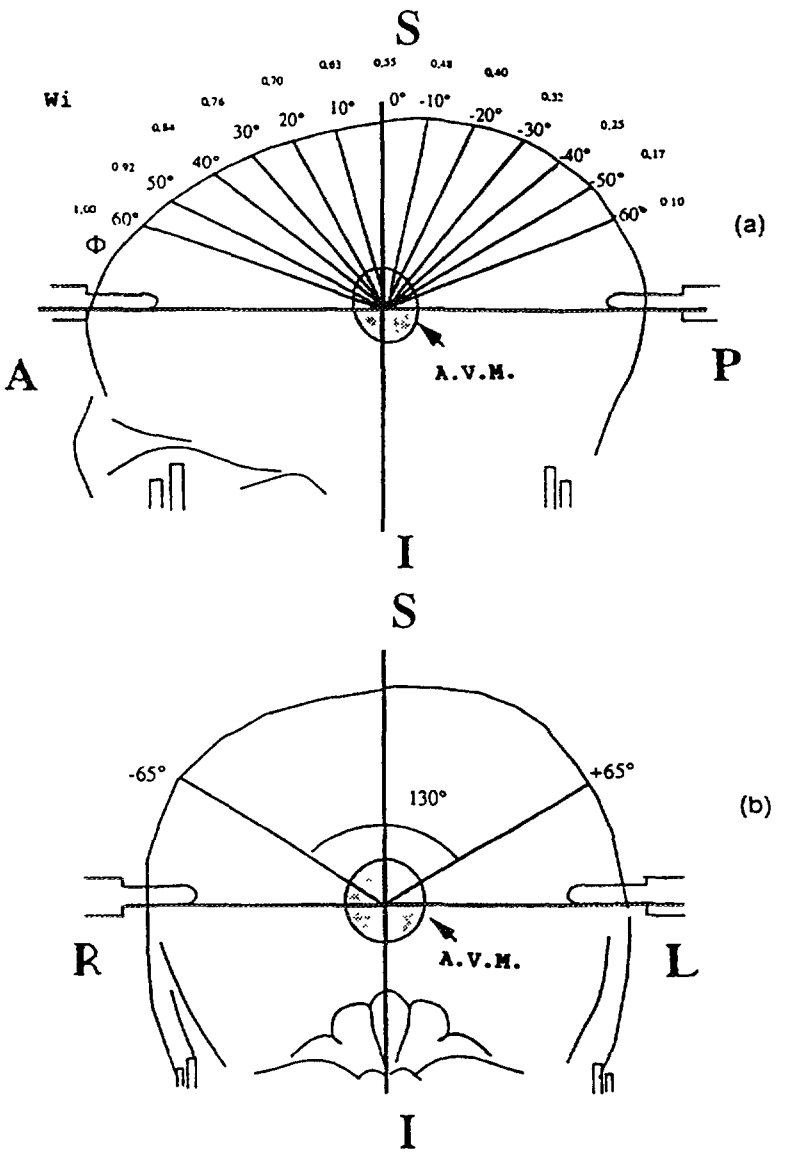

FIG 1.

Irradiation parameters for a centrally situated anterior-superior/ posterior-inferior ellipsoidal inclined AVM:
(a) 13 weighted arcs 10 degrees separated
(b) Angular legth of each arc , 130 degrees

To obtain inclined isodoses encompassing the elliptical shaped AVMs with different inclination angles on the sagittal angiogram, each arc should be differently weighted. We defined as "weighting vector" the vector having as elements the w_i arc weights (Fig. 1).

III Results

1. *Diamenter colimator*

-The 70% reference volume isodose (VRI) increases when increasing the collimator diameter (Fig. 2)

-The increase of $VFO_{70\text{-}10\%}$ is proportional to collimator diameter (Fig. 3) while the $VFIR_{70\text{-}10\%}$ is inversely proportional (Fig. 4).

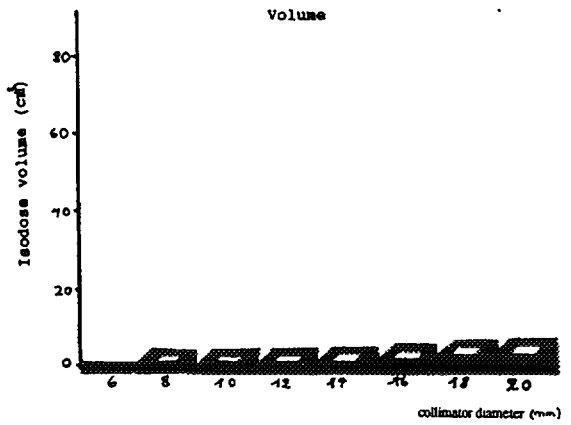

FIG 2.

70% Reference isodose volume (target volume)

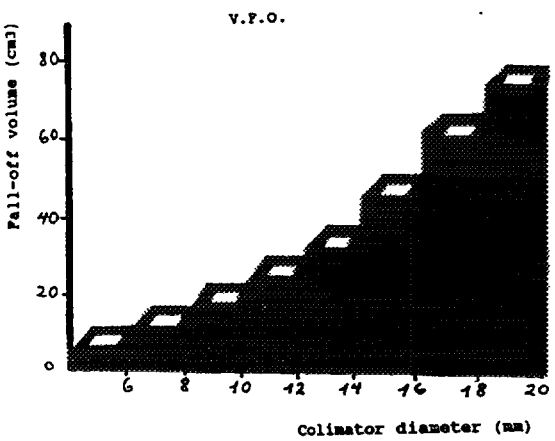

FIG 3.

Volumetric fall-off (70%-10%) outside the target.

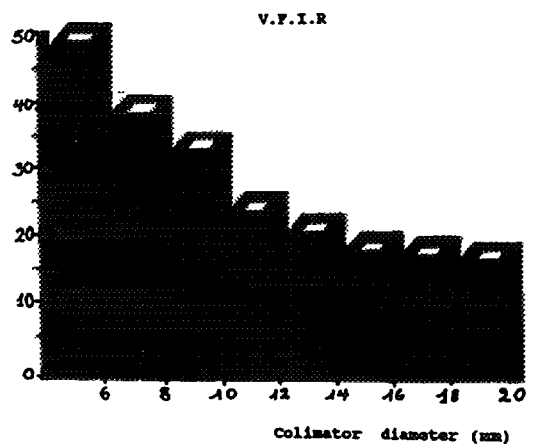

FIG. 4

The ratio of volumetric fall-off (VFO) by reference isodose volume (70%)

2. *Arc Weight*

We studied the arc weights, with seven linear weighting vectors (LWV1 to LWV7) having a simple shape (Fig. 5). We chose 1.0 to 0.1, as extreme values for these vectors. The parameter representing these LWV vectors is their angle α (Fig. 5).

-The minimal inclination isodoses $\alpha=0^{O}$ and the maximal inclination isodoses $\alpha=38^{O}$ in the sagittal planes after calculation of the dose distributions are presents in Fig.6

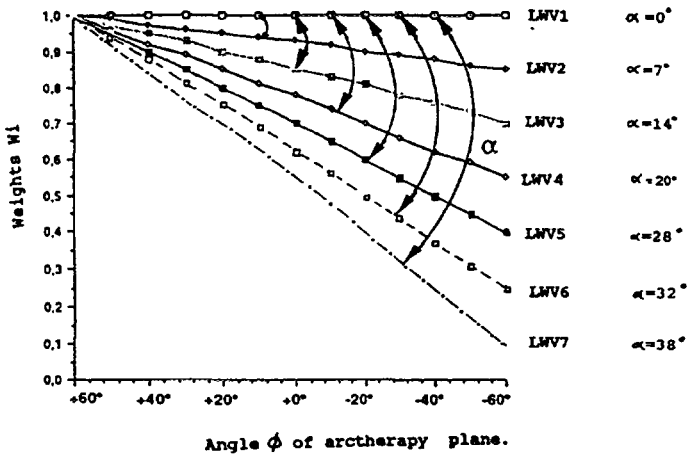

FIG 5.

Linear weighting vectors LWV as funcition of their α angle. (From LWV1: $\alpha=0°$ to LWV7: $\alpha=38°$)

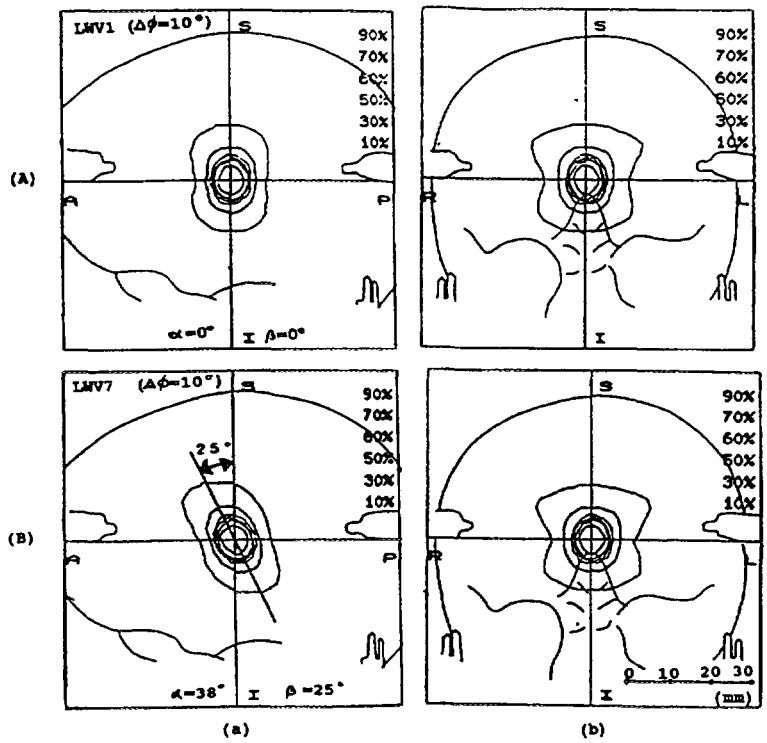

FIG 6.

Relative dose distributions in the : (a) coronal and (b) sagittal planes, corresponding to the linear weigting vectors (A) LWV1: $\alpha=0°$, $\beta=0°$ (B) LWV7 : $\alpha=38°$, $\beta=25°$.

These dose distributions permit the calculation of the realtionship between the isodose inclination angle β and the LWV angle α (Fig.7).

-Figure 8 shows the volumes enclosed by the 10%, 30%, 50%, 70% and 90% isodoses for the different LWV. The volumes encompassed by the isodoses from 30% to 90% remain relatively constant, the 10% isodose volume increases slightly as the angle increases.

-The $VFO_{70\text{-}10\%}$ remains unchanged and the $VFO_{30\text{-}10\%}$ presents a variation of 10.8%, passing from 39.7cm^3 (LWV1) to 44.0 cm^3 (LWV7) (Fig. 9).

-The relationships between the delivered doses inside and outside the lesion ($VFIR_{70\text{-}10\%}$) not change significatively 18,3 (LWV1) to 19,8 (LWV7).

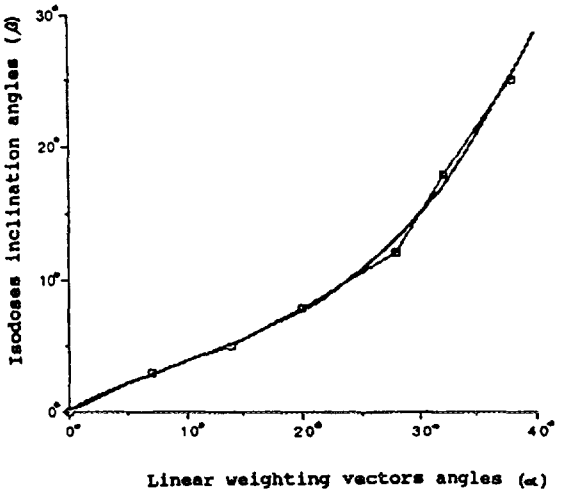

FIG 7.

Relationship between the isodoses inclination angle β and the linear weighting vectors, angle α.

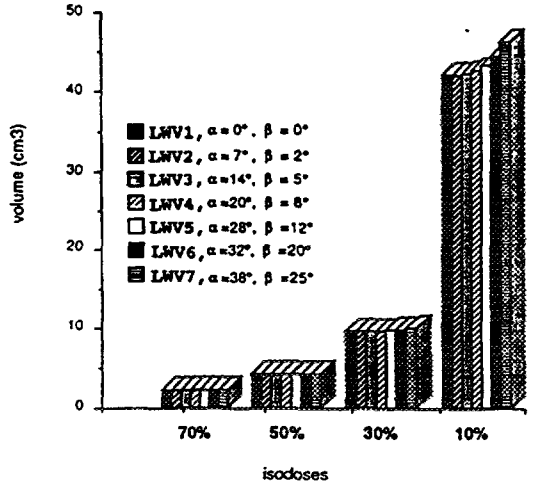

FIG 8.

The 90%, 50%, 30% and 10% isodose volumes for 13 arctherapies corresponding to the seven linear weighting vectors (LWV).

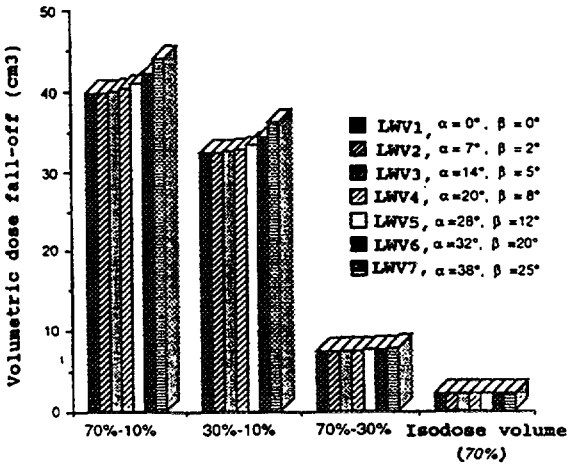

FIG 9.

Volumetric dose fall - offs 70%-10%, 30%-10% and 70%-30% and the 70% reference isodode volume for 13 arctherapies corresponding to the seven weighting vectors (LWV).

IV Conclusion

1. *Diameter Collimator*

-The reference volume (target volume $VRI_{70\%}$) increases when increasing the collimator diameter.

-The $VFIR_{70\text{-}10\%}$ shows that the doses outside the lesion decreases when increasing the diameter collimators.

2. *Arc Weight*

-The application of the linear weighting vectors produce inclined isodoses characterized by their inclination angle β. This last angle must coincide with the AVM long axis and the superior-inferior stereotactic direction angle. Studying a family of linear weighting vectors we have found the relationship existing between the α and β angles. This result (Fig. 7) constitutes an "a priori knowledge" which can be used in a dose distribution optimization procedure permitting the reduction of the treatment plan preparation.

-The dose distributions resulting from the application of the different LWV show that the isodoses volumes remain constant except for the 10% isodose volume which increases slightly as the angle α increases.

-The dose delivered to the healthy tissues (Fig. 9) $VFO_{70\text{-}30\%}$ remains unchanged whereas the $VFO_{30\text{-}10\%}$ increases for the angles superior to 28^{o}.

- Clinical experience shows that most AVMs can be classified in categories according to their shape and topopgraphy. To the inclined AVMs, we have found a relationship between the isodose inclination angle β and the weights of the arcs.

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DOSIMETRY OF BREAST CANCER

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Abstract

The systemic therapy of breast cancer has also changed profoundly during the last 60 years, and in this time the integration of treatment modalities involve a major area of investigation.(1). The dosimetry of breast cancer presents different complications which can range from the Physician's handling of the neoplasia up to the simple aspects of physical simulation, contour designs, radiation fields, irregular surfaces and computer programs containing mathematical equations which differ little or largely with the reality of the radiation distribution into the volume to be irradiated. We have studied the problem using two types of measurements to determine how the radiation distribution is in irregular surfaces, and designing an easier skill to be used with each patient, in order to optimize the treatment with respect to the simulation and verification process.

1 MATERIAL AND METHOD:

To measure the distribution of radiation in a volume of water we used a Wollhofer Phantom, to which was added an acrylic surface with a curvature similar to the Costal wall of a patient who has had a radical mastectomy performed. Moving the ionization chamber detector by means of a program that generates rectangle trajectories of different longitudes, like an integral step by step way, we got the isodose curves. These curves generated are the most exact distribution to a real measure within an irregular volume, just like the volume to be irradiated in the practice.

At the same time we initiated the film densitometer calibration of a Kodak X-OMAT V verification film in order to use a more precise dosimetry on each patient. We started constructing with wax (density = 0.94 gr./cc), a step wedge to generate a characteristic curve (H and D) of the film, which was irradiated locating the step wedge over the film and irradiated with a Cobalt source and 6 MeV, x-ray.

A Theratron-780 machine with a cobalt source of 2 cm of diameter and a Siemens Mevatron 6740 linear accelerator were utilized to irradiate the water phantom as well as the films. We constructed a wax phantom with the same form of a chest wall. The films were cut to form the figure of a chest wall in order to irradiate specifically the two tangential breast fields which are the ones that present the major problems of coincidence. Risk for tumor recurrence is a prerequisite to designing a radio-therapeutic, where the structures potentially at risk include the breast, the chest wall, the ipsilateral axillary, supraclavircular and the internal mammary lymph nodes (2), and additionally, that the target tumor areas stays within the proposed 80% isodose
line.

Equipment's used to measure the radiation were the Campintec 192X type WK92.WA10, with the ionization chamber PR-05 type IC-10 serial 477 (3), (4), (5).

Radiation beams of 6 cm x 12 cm for tangential fields with and without beam modifiers, like the breast cone manufactured by Theratronics, 45 and 60 degrees for the Mevatron 6740, to get the isodose over the irregular surface entrance were used.

These fields can take one of the three different forms. First, one can use standard tangential fields that enter and exit at the mid line of the patient. The second technique is to use deep tangential fields that enter and exit contra laterally across the mid line of the patient. The third technique matches shallow tangents to an en face internal mammary (IM) lymph nodes. In order to visualize whether the deep tangential fields encompass these points, one can obtain a CT scan in the treatment position with markers in the entrance and exit points and manually connect these points with a line to see if the fields are deep enough. Alternatively, on a simulator film the tangential fields should be seen to pass through the sternum.

The use of deep tangential fields has drawbacks. Foremost among them is the increased amount of lung that is included within the tangential fields as they are directed more deeply into the patient in order to treat the lymph nodes. Pneumonitis is a direct function of the amount of lung irradiated, (6), and since it is difficult to treat the IM lymph nodes in most patients without subtending at least 3.5 cm into the lung, the risk of radiation Pneumonitis from such treatment scales rapidly (7).

The films to be irradiated were placed between the two wax slices, in the form of a "sandwich", which are used to represent the patient's tissues, and irradiated with two tangential fields.

The films were developed under controlled conditions of darkroom and film processors, they were read with a film densitometer Macbeth TD-528, to connect the points with the same optic density, that represent the levels of radiation and generate the isodose of the different field combinations studied by us.

2 CONCLUSIONS:

The direct/indirect measures of the levels of distribution of radiation in irregular surfaces as is the breast and Costal walls, permits verification that the proposed dose or dosimetry, complies with what is proposed by the Medical Doctor or the treatment protocols.

With respect to our measures, we can visualize that according to the classification histological of the disease (C.A.), and the target areas (tumor bed and internal mammary lymph nodes), as well as the healthy tissues that should be preserved and diminish the morbidity of the patient (lungs, heart and spinal cord), the best options are presented in the distribution given when the treatment equipment is a 6 MeV Mevatron.

But the most important thing we found in this work was the confidentiality of the measures with the films since no differences larger than 2% were found in the whole range of measures of the film, close to the surface as well as deep.

Also important is the facility of how to construct the wax phantom of a patient's contours and verifying if the entrance angles of the fields, as well as the modifier implements (filter wedges, cone and bolus), presents the best real option of the treatment proposed in patients to which a radical or modified mastectomy has been effected.

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PLATON V2.0 & BRA V1.0 SYSTEM FOR TELETHERAPY AND BRACHYTHERAPY

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Abstract

A locally designed fully automatic Radiation Field Analyser (RFA) was constructed. The system is controlled by a PC and includes a graphic system ionisation chambers and an electrometer. The system is capable of reading doses instantly in any point inside a water phantom and provide graphs of the dose distributions (isodose curves) of any therapeutic unit. the information is automatically stored in the PC and can be transferred to Treatment Planning Systems (TPSs) such as the PLATON and BRA developed in Latin America.

1 INTRODUCTION.

During the last decade, there has been a vast projection of Radiotherapy on the general strategy of cancer treatment. This new dimension of the radiation treatment is due to the generalisation of computers, planning systems and to the modern advances in Brachytherapy (new sources and equipment).

As we proceed towards the 21st century, the goal is to optimise the quality of treatments improving their design, developing computerised systems for medical techniques. In this setting we worked hard for many years trying to develop the design of the product that we are introducing today in this meeting: Platon V2.0 & BRA V1.0. This product has been created in Latin America, more precisely in Argentina and in Uruguay.

Platon V2.0 replaces version 1.0, giving additional ways of treatment and a new interface with the user. This new system is completely graphic and operates by the use of a personal computer not demanding special hardware, the maintenance and availability of the equipment is easier than with other ones because they are standard, it can be configured for a single operator or may be part of a multiple workstation network design to optimise the effectiveness of the work group within the department. The integration of all the patient database is possible and readily accessible, it is designed in Spanish, Portuguese or English and it doesn't require previous experience for the user.

At the same time it provides a great variety of therapeutic modalities that are tools of calculation and analysis such as irregular tangential and rotational fields, beam's eye view, three dimensional calculations, irradiated volume calculations and representation of regions of interest.

2 MATERIAL AND METHODS

We can say that Platon V2.0 combines clinically oriented functionality with proven treatment planning methodologies, providing the professional with a very fast, friendly and simple to use decision support system. Through the use of system control files, the operating characteristics may be adapted to the department's work load and professional criteria to maximise the usefulness in documenting and evaluating alternative treatment planning approaches.

With this system we can see the representation of the treatment plan, with all the characteristics. The beam set up function reflects the every day clinical realities - all beams may be specified by their numeric parameters and/or positioned on the contour slice visually.

Otherwise, using a beam's eye view, collimator rotation, couch (isocenter) position and block placement over several contour slices may be interactively adjusted.

A treatment plan may have up to 5 separate patient contour slices defined and each contour slice may have up to 5 contours with different densities.

Also we can see isodose maps with the use of the different fields and weights.

The estimation of irradiated volumes and integral doses are also in this system. In a histogram way we can see the confined volume in each isodose curve, the median dose and the integral dose.

Distance and angle measurements, critical organ dose monitoring, isocenter localisation, dose normalisation, zoom, mantle fields and irregular fields, are able to be visualised on screen.

BRA 1.0 is the system that allows planning of brachytherapy treatments. Its design includes a variety of calculation and evaluation tools and it is extremely easy to use. This system calculates dose distribution in three dimensions for radioactive seeds, lines or ribbons in interstitial and intra-cavitary brachytherapy. Planes, cubes, cylinders, and cones may be automatically constructed by a few simple commands, and any combination of source types is possible. We are proud of this product and we can assure that our country is able to compete in the same line of the most developed countries.

Perhaps we can find similar or even better equipment, but our achievement is to have obtained a great design on less than a tenth of the cost of any other system on the market with the same quality standards of the most sophisticated and expensive ones.

Our ultimate purpose is not only to sell our product but to obtain the greatest credibility from our colleagues, so as to get the real technological exchanges which will allow us to achieve a greater growth. We would like to propose the international organisations the idea of buying this low priced system and to give it to the professional teams of the countries who might be in greatest need of them.

3 CONCLUSIONS

In Latin America there are more than 200 tele therapy machines using gamma sources. For example, if an international project distributes 100 systems of the kind we are mentioning, installs them, trains people for their correct use and if this investment is below the cost of one high energy linear accelerator, the impact in the change of radiotherapy treatment quality would be very important. And it could be easily financed by an international organisation.

In this way we will consciously increase the quality of radiotherapy around the world, working all together, integrally, in a universal project remembering that our relevant purpose is the cure. If we can understand that we are different but that we are all branches of the same tree, it will be the greatest achievement of this meeting and a indicative point of professional and personal growth.

THERMOLUMINESCENCE DOSIMETRY APPLIED TO QUALITY ASSURANCE IN RADIOTHERAPY, BRACHYTHERAPY AND RADIODIAGNOSTIC

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Abstract

Thermoluminescence (TL) dosimetry is very interesting for in vivo measurements because TL detectors have the advantages of being very sensitive under a very small volume and do not need to be connected to an electrometer with an unwieldy cable. The principle of the method being briefly recalled, criteria of choice of a TL material according to the applications to be performed are given. It is shown that to be used for in vivo measurements, TL material should have he same response at room and patient temperatures and be equivalent to soft tissue, lungs or bones for the energy ranges encountered in practice. Theoritical data are provided in order to facilitate the user's choice.

The different heating processes (linear or isothermal heating kinetics, hot gas, etc.) and light detection systems of TL readers are also presented. TL manual and automatic readers commercially available in 1994, and the emission temperature and wavelength the dosimetric peaks of usual TL materials are presented in two tables, respectively.

Then the principal properties of TL dosimeters to be used for in vivo measurements and their practical consequences are summarized: signal stability after irradiation, intrinsic precision, sensitivity, response with dose, dose-rate, mass and energy. At last some examples of applications as different as total body and skin irradiations, brachytherapy, diagnostic radiology and quality assurance purposes are given.

1. INTRODUCTION

Thermoluminescence dosimetry [1, 2] has been developed considerably over the past twenty years, the commercial availability of reliable detector materials and the commercialization of automatic readout systems being a decisive factor. A wide choice of TL materials in the form of powder, microrods, pellets, etc allow the dosimetry to be adapted to applications as different as radiation protection, radiotherapy, curietherapy, diagnostic radiology and quality assurance purposes such as calibration of treatment units and radioactive sources, verification of computer programs, validation of new protocols before clinical use, in vivo dosimetry either for particular techniques or to detect errors in individual patients.

For in vivo measurements (§ 6) TL dosimeters are competitive with other detectors [3, 4] and have the advantages of being very sensitive under a very small volume, to be tissue-equivalent and not to have to be connected to an electrometer with an unwieldy cable. As for time required for readout, it can be considerably decreased by a good choice of the equipment and a good methodology [4].

2. PRINCIPLE OF THE METHOD

Thermoluminescence dosimetry (TLD) is based upon the ability of imperfects crystals to absorb and store the energy of ionizing radiation, which upon heating is re-emitted in the form of electromagnetic radiation, mainly in the visible wavelength. The light emitted is then detected and correlated to the absorbed dose received by the TL material.

Many general theoritical models have been postulated to explain it, but still now difficulties arose when specific dosimetric materials are considered [1, 2]. One of the possible

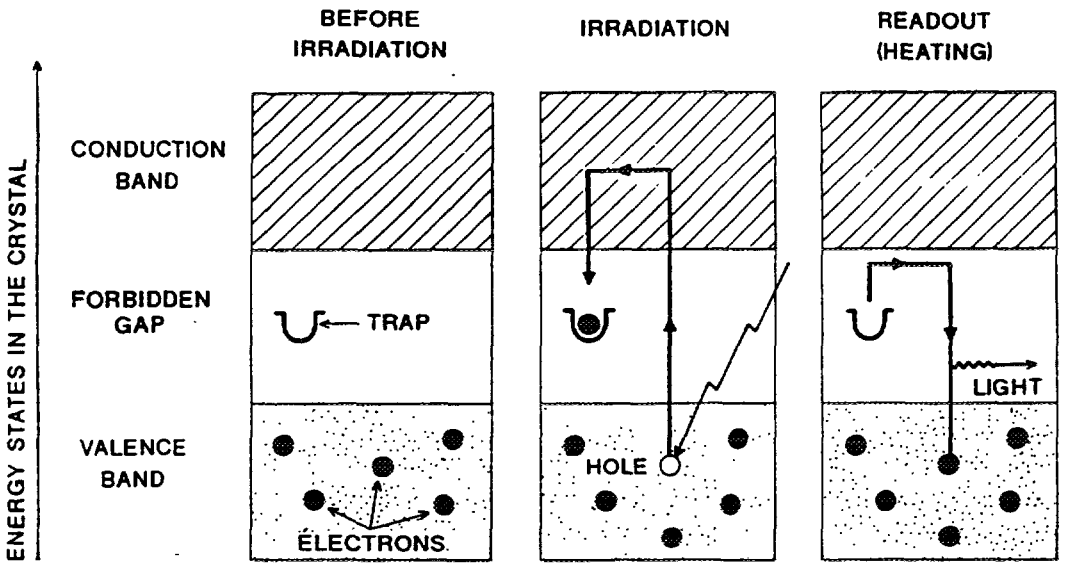

Fig. 1 : A possible mechanism for thermoluminescence. After [4].

mechanism for TL may be developed by refering to the band theory of the multi-atomic crystalline structure (Fig. 1). Energy states in a crystal can be represented with energy increasing upwards along the ordinate : irradiation produces free electrons and holes. The electrons are free to travel through the solid in the conduction band for a short time. They may be ultimally either trapped at defects (i.e at a metastable energy state) or fall back into the valence band and recombine either radiatively (fluorescence) or non radiatively with holes, or be captured at luminescent centers already activated by holes as a result of the irradiation, and deactivate the center with the emission of light.

Under the heating effect, the electron trapped at the metastable energy states are given sufficient thermal energy to escape from the trap into the conduction band again, where they are free to travel and have three possibles fates, as before :
- they are either be retrapped at defects
- or fall into the valence band and recombine radiatively or non radiatively with holes,
- or recombine radiatively at a hole-activated luminescent center.
The light emitted by the last process is *thermoluminescence (TL).*

Heating and light collection are performed in a readout system called *reader* (§ 4). The TL signal as a function of temperature is of complex nature and is called a *thermoluminescence spectra* or a *glow curve*. It consists of different TL peaks, each peak corresponding to a different energy state in the crystal, and depends on the TL material (nature and annealing procedures) and on the irradiation sources (Fig. 2).

After readout the TL material it is either entirely in its original state, and in this case it is just ready for re-use, or it requires a special heating treatments called *annealing* in order to restore it to its original state. When these treatments are not performed the sensitivity and background of TL dosimeters are considerably altered and their dosimetric properties do not remain constant. For instance $CaSO_4$, CaF_2 and LiF require annealing procedures depending upon the form in which they have been manufactured [5]. On the contrary $Li_2B_4O_7$ either doped with Mn or Cu can be used and re-used many times without thermal treatments between the successive irradiations and readouts.

3. CHOICE OF THE TL MATERIAL

Most commonly used TL detectors are obtained by doping phosphors such as lithium fluoride (LiF), lithium borate ($Li_2B_4O_7$), calcium sulphate ($CaSO_4$) and calcium fluoride (CaF_2) with impureties called activators: e.g. LiF:Mg-Ti is lithium fluoride doped with magnesium and titanium, $Li_2B_4O_7$:Cu is lithium borate doped with copper, etc. All TL materials are available either in the

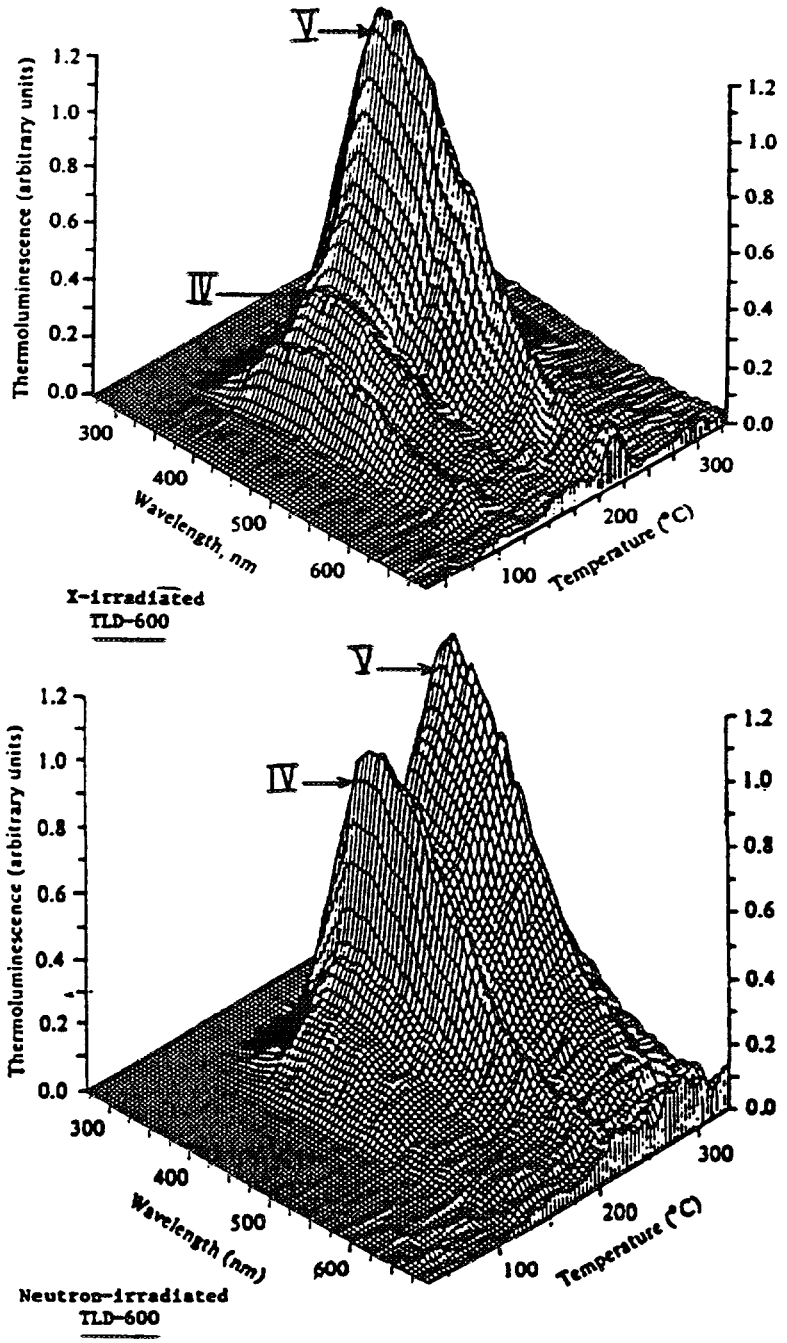

Fig. 2 : Thermoluminescence spectra of LiF 600. It depends on the previous annealing procedures and on the irradiation sources: X-rays [A] and neutron beams [B]. After [6].

form of powder or of solid dosimeters. The solid dosimeters may be made entirely of phosphors as single crystals or polycrystalline extrusions (extruded rods, sintered pellets or chips), or as homogeneous composite of the phosphor powder and some binding material. It should be noted that the characteristics of the pure phosphor dosimeters may be considerably different from those of the composite. To be used for in vivo measurements TL material should :

- have a high sensitivity under a very small volume
- have the same response at room and patient temperatures
- be equivalent to soft tissue, lungs or bones for the considered beam and in the energy range encountered in radiotherapy or radiodiagnostic.

The use of theoritical data such as Tables 1 or 2 for photon beams, or Fig. 3 for electron beams, allows a good choice of the TL material to be used according to the application. But in practice the influence of the surrounding material (build-up cap and patient), the size and the shape of the TL dosimeters have to be taken into account for energy correction (§ 5.7).

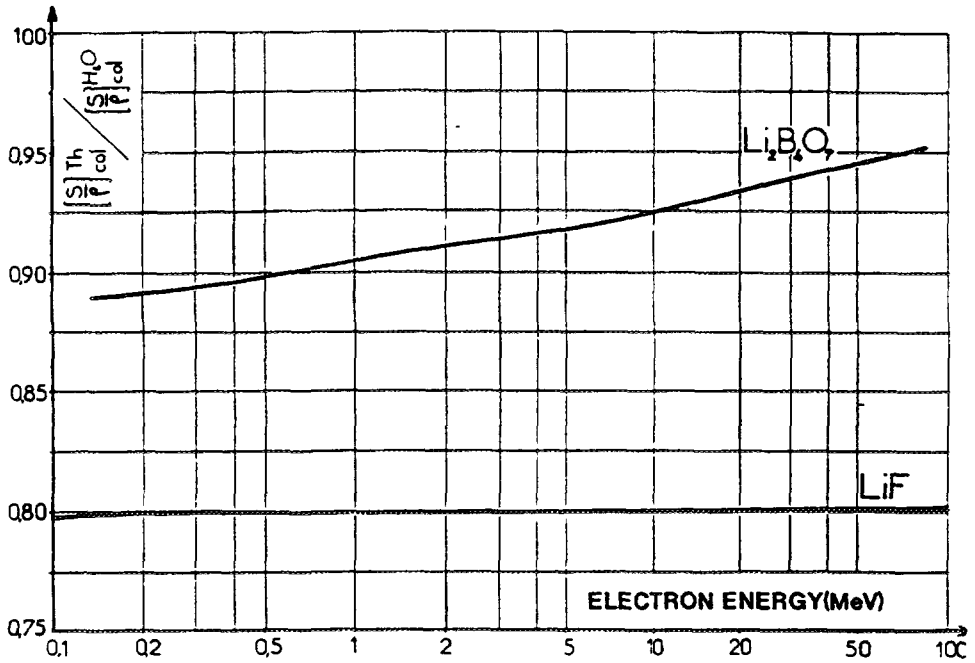

Fig. 3: Ratio of the mass radiative stopping power in the TL material to the mass radiative stopping power in water for TL dosimeters irradiated with electron beams. Curves have been obtained from I.C.R.U. N° 35 Report data [7].

Table 1 : TL materials equivalent to soft tissue for photon beams.

TL PHOSPHORS	Photo-electric effect Z eff	Compton effect e^-/g	pair production Z eff	ρ g/cm^3
LiF (Mg,Ti)	8.2	0.83	7.50	2.64
LiF (Mg,Ti,Na)	8.2	0.83	7.50	2.64
$Li_2B_4O_7$: Mn	7.4	0.87	6.90	2.30
$Li_2B_4O_7$: Cu	7.4	0.87	6.90	2.30
Soft tissue	7.42	1*	6.60	1.04
Air	7.64	0.90	7.36	0.0013

Table 2 : TL materials equivalent to bone for photon beams

TL PHOSPHORS	Photo-electric effect Z eff	Compton effect e^-/g	pair production Z eff	ρ g/cm^3
$CaSO_4$: Mn	15.3	0.90	-	2.61
$CaSO_4$: Dy	15.3	0.90	-	2.61
CaF_2 : Mn	16.3	0.88	-	3.18
CaF_2 : Dy	16.3	0.88	-	3.18
Bone	14	0.94	10	1.01 to 1.60

4. HOW TO READ OUT AN IRRADIATED TL MATERIAL ?

The principle of TL readers is shown in Fig. 4. They mainly consist of a heating and a light detection systems.

4.1. Heating system

Different heating systems are encountered in TL readers commercially available (Table 3). All of them offer the possibility to heat the TL dosimeter at two different temperatures: the *preheating temperature* used to clear unstable peaks and the *readout temperature* used to collect the information from dosimetric peaks. The readout chamber should be continuously flushed with nitrogen gas in order to reduce spurious phenomena [1] and then decrease the background.

The metallic support containing TL material may be heated by an electric current, or contact with ovens, or a hot finger moved by a lift mechanism. Generally the temperature of the support is measured by a thermocouple in close contact with it. Then heating kinetics are either linear or isothermal. When it is *linear* , the TL material is progressively heated until preheating and readout temperatures. When it is *isothermal* the TL material is quasi-instantaneously heated by isothermal ovens to both these temperatures and generally readouts take less than 10 seconds. In any case close contacts between TL dosimeter, support and heating system are necessary to obtain a good reproducibility (§ 5.2).

Some no-contact procedures, such as heating by hot nitrogen gas or air, optical infrared heating using an intense light pulse from an halogen lamp or heating by a laser beam [9, 10] can also be used. In these cases the heating kinetics are particular.

Readers which are designed for the readout of a great number of dosimeters in a short time have generally an isothermal heating kinetics [11, 12] or heat the TL dosimeters in hot gas.

4.2. Light detection system

Generally the luminous flux emitted by the TL dosimeter is collected and guided into the PM with a light guide leading to one or several filters placed in front of the PM window (Fig. 4). These filters have to be adapted to the spectral response of the PM and to the wavelength of the light emitted by the TL material to be readout (Table 4). The response of the P.M. depends on the composition of the photocathode and on the spectral transmission of the tube window. In most readers photocathodes are bialkalis with a peak sensitivity around 400 nm, in good agreement with the blue emission of LiF or $Li_2B_4O_7$: Cu (emission at 400 nm and 368 nm respectively) but not with $Li_2B_4O_7$

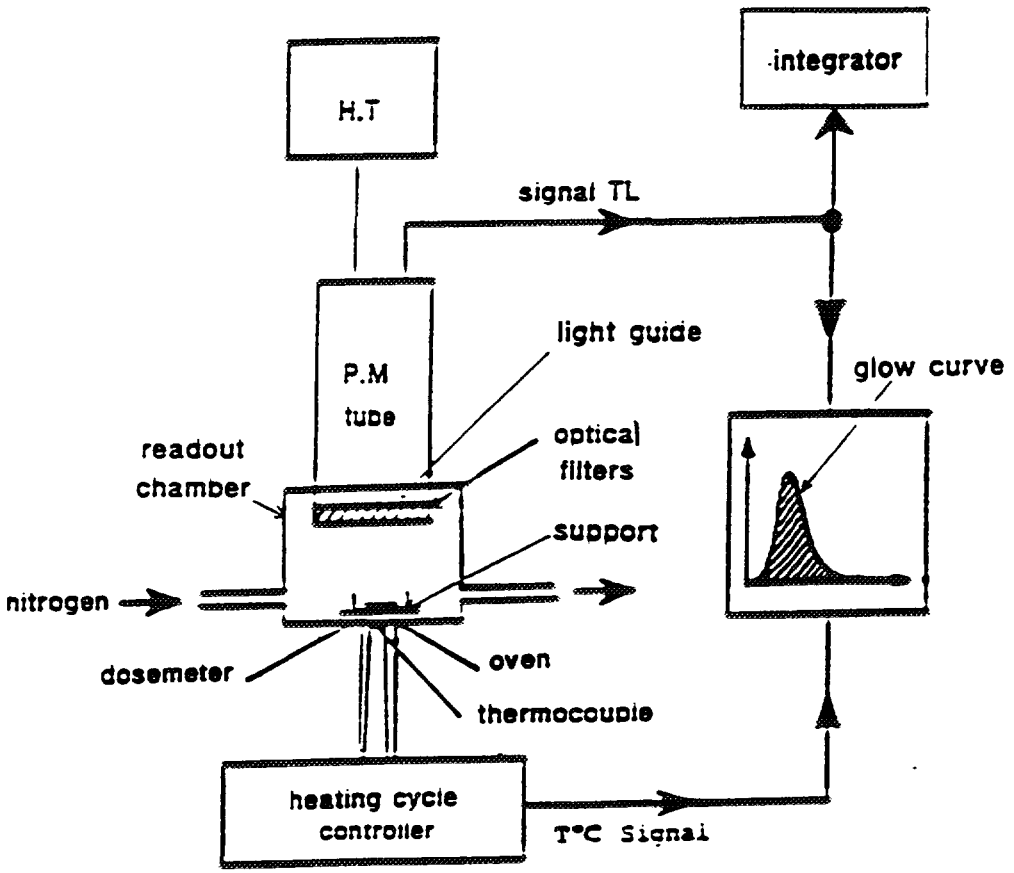

Fig. 4. Principle of a TL reader. After [8].

Table 3 : Some TL readers commercially available in 1994. After [4]

Manufacturer	country	model	heating system	detectors	manual / automatic
Panasonic	Japan	UD - 513A	hot gas	rods + P. G	manual
Harshaw	USA	3500	heated support	S.D. + P.	manual
Harshaw	USA	5500	hot gas	S.D.	automatic
Teledyne Isotopes	USA	310	heated support	S.D. + P.	manual
Alnor	Finland	DOSACUS	hot gas	S.D.	automatic or manual
Fimel	France	LTM	heated support	S.D. + P.	manual
Fimel	France	P.C.L	isothermal ovens	S.D. + P.	automatic

P. : powder S.D. : solid detector P.G. : powder within glass

doped with manganese (emission at 600 nm). Nethertheless adequate filters correctely chosen can improve considerably the response. A good reader should allow a quick interchange of the associated filters in order to be adaptable to different TL materials.

Table 4 : : Emission temperatures and wavelength of the most stable peaks used in dosimetry of different TL materials.

	WAVELENGTH (nm)							
EMISSION TEMPERATURE (°C)	368	380	400	450	478 and 571	480 and 577	500	600
80 - 90							$CaSO_4$:Mn	
200								$CaSO_4$:Sm
200 - 240						CaF_2:Dy		
210 - 220			LiF					$Li_2B_4O_7$:Mn
220 - 250				$CaSO_4$:Tm	$CaSO_4$:Dy			
240 - 270	$Li_2B_4O_7$:Cu							
260		CaF_2:Nat						
300							CaF_2:Mn	

Depending upon the type of the readout system used, the signal proportional to the light emission is either amplified and fed to an integrator (d.c operation regimen), or converted into pulses and fed to a scaler (pulse counting regimen) [8]. Irrespective of the regimen the voltage of the P.M. tube has to be correctly stabilised in order to get a good reproducibility of measurements (§ 5.2).

Results which have to be correlated to absorbed dose are either read out and stored by the operator, or safeguarded on the hard disk of a computer and most often printed. In many readers glow curves are also displayed during dose measurements so as to provide a maximum amount of information.

5. DOSIMETRIC PROPERTIES

The principal properties of TL dosimeters to be used for in vivo measurements and their practical consequences are summarized. For more detailed informations the reader could refer to general books on TL dosimetry [1, 2] or to *"Methods for in vivo dosimetry in external therapy"* [4].

5.1. Signal stability after irradiation

An important consideration in the choice of a TL dosimeter is the stability of the signal. In particular it is necessary to assess whether the charges trapped during the irradiation have not been lost before the readout by unwanted exposure to heat *(thermal fading),* light *(optical fading)* or any other factor *(anomalous fading).* This is expressed by a decrease of the TL dosimeter response depending on the delay separating irradiation and readout.

An appropriate preheating allows the elimination of that part of the signal (low temperature peaks) which presents an important thermal fading, and then reduces considerably thermal fading for most TL materials. In practice thermal fading should be evaluated on each individual reader with the TL material which is intended to be used. It should be approximately of 1% per month, or less, for the different preparations of LiF when correct readout and annealing conditions are reached [1, 2, 13]. For $Li_2B_4O_7$ it varies from 0,5 to 1% per week depending upon the doping [1, 14]. When a long delay separates irradiation from readout, a fading correction may be necessary.

Optical fading can be avoided by manipulating the dosimeters in a room illuminated with incandescent light and wrapping them in opaque containers or envelopes, when used for in vivo dosimetry in treatment rooms illuminated with fluorescent light.

Anomalous fading is much more difficult to detect than either thermal or optical fading because it generally occurs much more slowly. Possibly because of this, it has not yet been demonstrated to be a problem for in vivo dosimetry.

5.2. Intrinsic precision

Intrinsic precision is the reproducibility of a given TL material associated with a given readout system. It is very dependent on the quality of the TL material used, reader characteristics, the way in which the preheating and heating cycle have been defined, the purity of the nitrogen gas circulating in the readout chamber,etc. It can be evaluated by randomly taking 10 samples of TL powder or dosemeters out of the same batch and by irradiating them to the same dose. After readout, and annealing procedure when necessary, the operation is repeated several times.

When readout parameters have been optimised, a standard deviation of $\pm 2\%$ or less, can routinely be obtained with either manual or automatic readers of good quality associated with reliable TL materials [12, 15, 16].

5.3. Sensitivity

5.3.1. Solid dosimeters: identification

Some variations in sensitivity within a batch of TL dosimeters is unavoidable. Two methods can be used to limit the effect of these variations :

- one method consists of irradiating all the dosemeters in the same geometrical conditions, to read them out and to attribute to each of them a sensitivity factor S_i equal to $R_i / \overline{R}$ where R_i is the TL readout from dosimeter number i and $\overline{R}$ the mean of all values of R_i. Sensitivity factors should be checked periodically to take into account a possible loss of material occuring when TL dosimeters are not handled carefully;

- another method giving a similar accuracy consists of dividing the TL dosimeters into sensitivity groups without identifying them individually (e.g. groups of dosimeters with a response variation less than ± 1 or $\pm 2\%$ from the group mean) and to increase the number of dosimeters used for each point of measurement. When an automatic reader is available, such a method is very suitable because readouts take a very short time. As the distribution of sensitivities within a group can vary with time for the same reasons as above, it should be checked at a frequency which depends upon the accuracy required for measurements.

5.3.2. Powder: response variation with mass

If optimum accuracy is to be obtained when TL powders are used, the quantity of powder and the readout conditions must be accurately defined and corrections made when necessary. The response variations with mass of TL material should be established for the readout conditions used in practice because they depend upon the heating kinetics. For most TL materials the signal is

proportional to the mass when they are read out with linear heating kinetics: either a linear correction should be made with samples of different weight, or samples of equal weight should be used.

Some TL material such as $Li_2B_4O_7$:Cu have a response which can be considered as independent of mass, in a certain range of mass, when they are read out with automatic readers using isothermal kinetics [12]. In this case it is not necessary to weigh the powder, a simple volumetric measurement being enough (Fig.5).

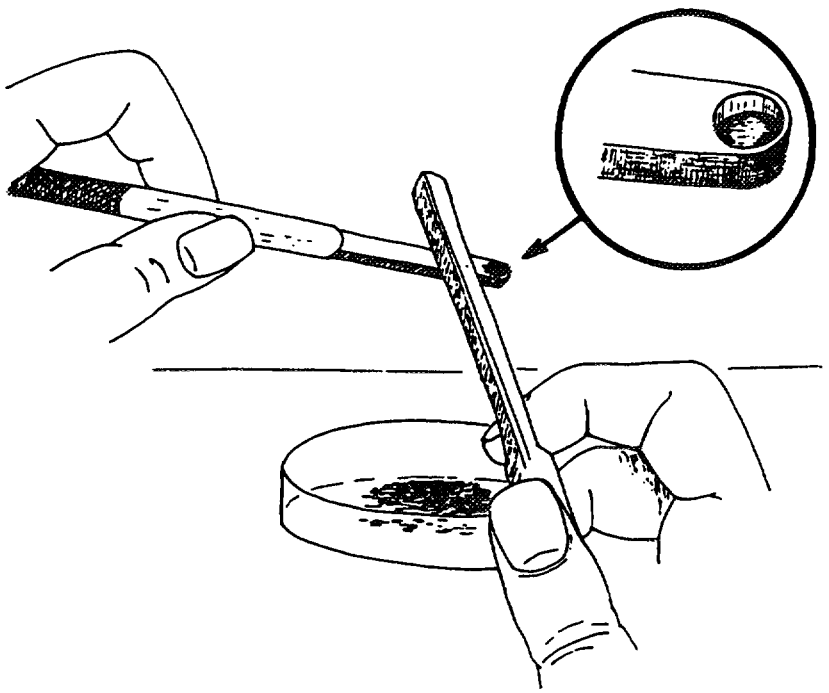

Fig. 5: Volumetric measurement consists of completely filling a calibrated hole with the TL powder.

5.4. Response variation with dose

In practice it is recommended to use TL dosimeters in the region where their response is proportional to the dose received (*linear region*). As an indication the dose ranges corresponding to the linear zone vary from 5×10^{-5} to 1 Gy for LiF [1], 10^{-4} to 3 Gy for $Li_2B_4O_7$:Mn [1] and 5×10^{-4} to 120 Gy for $Li_2B_4O_7$:Cu [14].

When TL dosimeters are not used in the linear region, a correction should be applied to the signal from a curve established with the TL material as well as the reader used, (and not from a published curve because the readout parameters may have an influence on its shape). This curve should be checked periodically.

TL dosimeters should not be used in the sublinear region approaching saturation. It should also be noted that supralinearity and saturation dose can both be affected by bad heating conditions, by previous exposures to irradiation and by thermal treatments (§ 2).

5.5. Influence of dose-rate

TL dosimeters are to a large extent dose-rate independent. As shown by Tochilin [17] and Goldstein [18] LiF and $Li_2B_4O_7$:Mn are independent of the dose-rate up to 45 Gy and 10^3 Gy per pulse of 0.1 ms, respectively. This property implies in practice that no correction for dose-rate is necessary for in vivo measurements. Even the extreme high dose-rates produced in scanned electron beams do not cause any special difficulty.

5.6. Influence of temperature

As the temperatures required to get the light signal out of the TL crystal is high compared to room or patient temperature, the response of TL dosimeters is indepedent of temperature variations in the range concerned by in vivo dosimetry. However care should be taken not to store the dosimeters close to a heat source.

5.7. Influence of energy

5.7.1. High energy photon beams

Except for superficial measurements, TL dosimeters should be surrounded by a suitable build-up cap corresponding to the energy and geometrical irradiation conditions considered to insure electronic equilibrium [4]. When the build-up cap is made of tissue-equivalent material, it is theoretically possible to evaluate the absorbed dose in TL dosimeters and associated build-up cap irradiated with photon beams, knowing the relative variation of mass energy absorption coefficient between the TL material considered and water as a function of photon energy. In practice, due to the influence of the surrounding material (build-up cap and patient), the size and the shape of the TL dosimeters may modify the expected results by a few per cent [19, 20, 21]. The heating conditions may also modify slightly the results [1]. So the most reliable method consists of comparing directly the TL dosimeters and associated build-up cap to be used to a calibrated ionisation chamber the energy response of which is well known, to irradiate both the detectors in the same beam as those used for patient treatments, using the experimental conditions shown in Fig. 6 and to compare their responses. Because of the slow variation of response versus energy of $Li_2B_4O_7$ and LiF in the energy range considered, the calibration factors obtained with this method can then be used for all patients treated in the same beam, or patients treated in photon beams of the same energy, irrespective of the geometrical conditions of the irradiation (field size, SSD, presence or not of compensating filters, etc).

It should also be noted that LiF type 6 and $Li_2B_4O_7$ respond to slow neutrons via reactions with ^6Li and ^{10}B. As X-rays of very high energy are sometimes contaminated by neutrons, a particular attention has to be taken when in vivo measurements are performed with X-ray beams of energy greater than 12 MV. The best solution consists of using LiF enriched in ^7Li which is not sensitive to neutrons.

5.7.2. Low energy photon beams

For photon energies below 300 keV, TL dosimeters should be very thin and applied without build-up cap. It is also preferable to use lithium borate instead of LiF, and a fortiori other TL material because the variation of response with energy is less important (Tables 1 & 2). In this case theoretical data of response versus energy can be used every time the TL dosimeter is of small dimensions [22].

For very low photon energies (below about 50 keV), theoretical curves or any other theoretical data, must not be used directly because the shape and dimensions of the detector can induce considerable response variations within the dosimeter volume [1, 23]. Moreover the nature of the activator may also yield too large differences in the response of TL materials in this energy range [24]. The only solution consists of comparing directly the response of the TL dosimeters to a calibrated ionization chamber using a method similar to the one shown in Fig. 6. Due to the low energy range, the effective point of measurement of the chamber, which should be adapted to these low X-ray energies, is then situated at the same level as the TL dosimeter.

5.7.3. Electron beams

Theoretically it is possible to evaluate the absorbed dose in TL dosimeters irradiated with electron beams knowing the variation of the ratio of the mass collision stopping power of the TL material to that of tissue or water as a function of energy (Fig.3). This variation is less than 2% and 5% for LiF and $Li_2B_4O_7$ respectively in the energy range from 200 keV to 50 MeV [7]. In practice and for the same reasons as for photon beams it is preferable to compare directly the TL dosimeters to be used to a calibrated ionisation chamber the energy response of which is well known for electron beams. The validity of the method has been verified by different authors [21, 25].

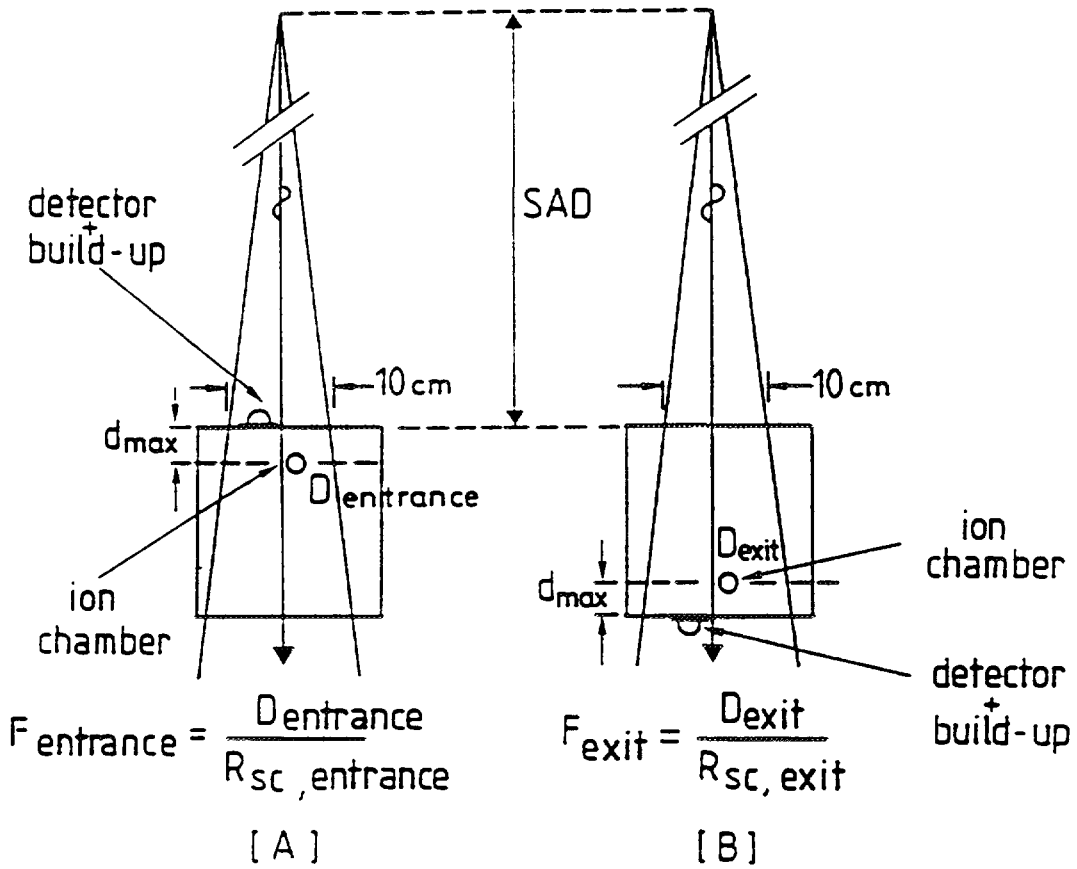

Fig. 6: Determination of the entrance and exit dose calibration factors of a TL dosimeter, $F_{cal,en}$ and $F_{cal,ex}$, respectively. The calibrated ionisation chamber is put at maximum depth d_{max} (for the entrance dose calibration factor [A]), or at a distance d_{max} from the exit side of the field for the exit dose calibration [B] and the detector is positioned at the entrance or exit surface, respectively.
After [4].

5.8. Directional effect

No correction for directional effect is necessary except if the dosimeter container and associated build-up cap have an asymmetric shape; even for the tangential irradiation of the breast or the thoracic wall no directional dependence of detector response is observed.

6. PRACTICAL APPLICATIONS

Many papers have been published showing the usefulness of TL dosimetry for in vivo dosimetry and quality assurance purposes. Its interest is still increasing with the apparition of modern automated readers, especially dedicated to medical applications, which can read out about 50 dosimeters in 15 minutes. Some examples of applications are presented below.

6.1. Calibration of treatment units and radioactives sources

Mailed TL dosimeters (generally LiF) are used by several national or international institutions (I.A.EA., OMS, EORTC, etc) to check beam calibration between different radiotherapy centers. Tl dosimeters are sent to the different centers, most often with a joined special holder (or phantom). After being irradiated by the user in water, in reference conditions, they are sent back to the official institution for readout. Dose delivered by each center is then compared to the dose actually received by the TL dosimeters. It has been proven that frequent quality audit by mailed dosimetry can improve treatment quality [26].

Tl dosimetry can also be used to check calibration of radioactive sources used in brachytherapy.

6.2. Verification of computer programs

It consists of comparing doses calculated by the Treatment Planning System (TPS) with measurements in phantoms of different shape and composition. The method can be used either for external beam therapy [27] or brachytherapy [28, 29]. Regions in which discrepancies are observed can be pointed out, and then the algorithms of dose calculation improved.

6.3. Dose mesurements in situations where the dose calculations may be inaccurate or impossible

An example of such situation is the evaluation of the dose delivered to the axilla during brachytherapy for carcinoma of the breast using iridium 192. The position of the radioactive material with respect to the axillary zone differing with the patient seated or lying, only in vivo measurements are able to yield to a correct dose evaluation [30].

In external beam therapy, due to the limited size of the detectors, and therefore their excellent spatial resolution, another typical application of TL dosimetry is the exploration of zones of high dose gradient such as junction zones, penumbra region,etc.

6.4. Check of correct delivery of external irradiation

A possible aim of in vivo dosimetry is to check the target dose in order to verify correct delivery of irradiation. Except when the target is the skin [31, 32] or when detectors can be introduced in natural cavities such as oesophagal tube, rectum, vagina, etc. this is impossible.

Nethertheless target dose can be deduced from entrance and exit dose measurements performed at the patient's skin provided certain precautions are taken [4]. Then it is essential to check each beam contributing to the target dose individually, at least at the first treatment session, in order to identify the possible causes of errors and to correct them. In the particular case of TL dosimetry, that implies the need to change the set of detectors after each irradiation beam.

Once the quality of the irradiation delivered individually by each beam has been checked at the first treatment session, some users may wish to check also the reproducibility of the treatment during the following sessions. In order to save time, they often prefer to leave the same *in vivo* detectors on the patient's skin for the full treatment session including all beams. In this case, it should be verified that entrance and exit doses of each beam are not influenced by contributions from other beams [4].

Such in vivo measurements can be used to detect errors in individual patients [33], to evaluate the quality of usual or special treatment techniques (total body irradiation before bone graft marrow) or to estimate the global accuracy of a department [34].

6.6. Diagnostic radiology

Different authors have used TL dosimetry for determining doses to patients during radiological procedures. Few difficulties are mentioned when lithium borate dosimeters are used for measuring gonad and maximum doses to the skin [35].

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SUMMARY AND CONCLUSIONS

ACCURACY REQUIREMENTS IN RADIATION THERAPY

Tumours can be thought of as "parallel" tissues (all the functional subunits i.e. clonogens act independently of each other). Therefore their viability remains if only one clonogenic cell survives. There are often as many as 10^7 to 10^9 clonogens in tumours at time of diagnosis. A possible explanation for the shallow dose-response curves observed clinically for tumours is that a relatively small number of radio-resistant clonogens (e.g. due to hypoxia) dominate the response; the much larger number of relatively radiosensitive clonogens are then practically irrelevant. An additional explanation for the shallower slopes is inter-patient variations in radiosensitivity. Both these explanations will apply to most tumour types but possibly in varying proportions.

Tumour cure is truly a statistical phenomenon unless the dose delivered results in tumour control probabilities (TCP) > 0.90 or < 0.10. The relative standard deviation at the 0.50 level is 25% for each patient. Normal tissues as well as organs at risk cannot generally be seen as parralell tissues. They are much more complex than tumours in their response to radiation with many more levels of increasing severity of damage. Broadly speaking there are two distinct classes of organs with regard to their response to partial irradiation: "parallel" e.g. lung and liver and "series" e.g. the gut and the spinal cord. The former exhibit a large "volume effect", the latter a small one.

In a group of patients the cure and damage probabilities may be correlated i.e. cells in the tumour and the organs at risk may be particularly radiosensitive or radioresistant; this has implications for therapeutic strategies as there is now evidence that such correlation may involve as many as 80% of patients for some types of tumours. Pilot studies of the use of assays of cellular response to radiation would be highly desirable in order to explore the potential for individualisation of dose prescription; normal-tissue assays will probably be more useful than tumour assays as the former determines the tolerance dose. Further, the most resistant tumour clonogen is very hard if not impossible, to assay.

Unless the tumour is very heterogeneous (in clonogen population, etc.) inhomogeneities in the target-volume dose distribution always result in a *decrease* in TCP but this decrease is small if the heterogeneities in the dose distribution are small (and the slope of the TCP curve, i.e. the gamma value, is not too large); consequently the mean dose, D, is generally the best single parameter to characterize the distribution (i.e. D is generally to be preferred over D_{min}.).

There is a large spread in clinically observed gamma values for tumours, but most of them are in the range 2-5; the few high values e.g. larynx are probably due to the absence of hypoxia. The data on normal-tissue gamma values is of very poor quality but in general it is likely that such gammas are larger than for tumours, especially if the number of subunits is large.

The much quoted "5%" figure from ICRU report 24 for accuracy requirements of the prescribed dose to the target volume (e.g. D_{ref}) is probably a 2-sigma value. Thus the one-sigma number is 2.5%. This is not realistic at the present time, however. A 2-3 % requirement is consistent with current radiobiological models when the mean dose is used as reference. Physicists should continue to strive to achieve as low uncertainties as possible in the dose delivered to the tumour. Only in this way will reliable clinical data ever be generated on which to base future improvements in clinical outcome, i.e. better estimates of gamma and D_{50} values.

Currently there is an unacceptable lack of uniformity between different clinics (even within one country) in the way that target doses are prescribed and reported, some preferring D_{ref}, others D, yet others D_{min}. For tumour response, D is generally the most relevant single measure of the target-volume dose distribution and physicists should endeavour to calculate this even in the absence of a 3D treatment planning system; this can often be done quite accurately by manual methods as heterogeneities in the dose distribution usually have a small influence on large-volume averages. It is good practice to also

- report the one-sigma of the dose distribution in the target volume (and organs at risk) as well as D_{min} and D_{max}.

The above suggestions regarding dose reporting strictly applies only to high-dose radiotherapy given with curative intent, i.e. not to palliative treatments.

EQUIPMENT REQUIREMENTS

The need to learn from accidents which have occurred in the past was recognised, and the analysis of a number of reported accidents indicated that there is often a root cause and contributing factors. In radiotherapy the analysis shows that human error is a much more important source of accidents than equipment failures and this underscores the continuing need for adequate training and qualifications of personnel in addition to adequate procedures and supervision. The identification of initiating events and contributing factors is a valuable exercise for testing the vulnerability of the radiotherapy system by all disciplines involved in the process: Radiotherapist, Physicist, technologist, maintenance engineer, administrator, manufacturer and regulatory authority. This exercise should include scenarios of unusual events which could cause an accident, as well as case histories of events which were successfully handled and did not result in an accident.

Regarding equipment, in addition to the traditional approach of specifications and standards, some concepts with origin in a safety culture philosophy, such as "defence in depth" should be considered during the design stage. The positive response of national professional and scientific organisations in preparing quality assurance protocols, was reported. Also the attention given to design considerations, procedures, calibration, quality assurance, investigating of accidents and adequate records in the revision of Basic Safety Standards for Protection Against Ionising Radiation and the Safety of Radiation Sources was mentioned. In an attempt to view the phenomenon in perspective, it was observed that in a single radiotherapy room thousands of patient set-ups are performed annually. In addition due to the nature of radiotherapy when considering misadministration of the intended dose, there may be a marginal or "grey" area between uncertainty and what might be considered an accident.

In reviewing the recommendations which IAEA, PAHO, and WHO have made concerning radiotherapy equipment and facilities since the 1960s, the participants recognised that in view of the global situations now existing, most of these recommendations are still applicable in some places in the world. It was observed that in 1994 in many places that reliable electrical power (voltage, current & frequency), water supplies (quantity, and quality) and environmental modification systems (temperature, humidity, and dust control) are unavailable, or only available in a few large cities. The availability of adequate maintenance services, and spare parts and/or the funds to provide them is severely limited in many locations. The perquisites remain unchanged for the successful use of any complex equipment, Co-60 or accelerator; Expert personnel, excellent infrastructure, good communications, large patient load, excellent dosimetry, efficient organisations and supporting structure. Taking into account the existing situation and modifications which are feasible and sustainable in view of the economic and social situation, priority should be given to acquiring the kind of equipment which is most likely to be able to function in the local environment (climate, staff, supporting services, operating resources) where it is to be used.

For external beam radiotherapy, the concept of an essential set of equipment to provide a satisfactory level of quality, for example a reliable source of megavoltage photons (Cobalt-60 or a low energy accelerator, depending local conditions), a simple and reliable treatment simulator and a simple treatment planning system was proposed.

Regarding the projections for the future, the World Health Organisation's estimate that currently there are approximately nine million cancer cases per year, world-wide, and that by year 2015 there will be 15 million new cancer cases annually, with about two thirds of these in developing countries was mentioned. Consequently, the need to provide simpler, more reliable and

less costly equipment (considering purchase, operating, source replacement in the case of Co-60, maintenance and decommissioning costs plus operating characteristics such as "up-time") was recognised. Furthermore, it was noted that the cost involved in disposal of spent radionuclide sources discourages owners from proper removal and storage, and accidents like that ones in Ciudada Juarez, Mexico and Gioania, Brazil occur.

Various components and factors that are involved in delivering the prescribed dose to the *planning target volume* (PTV, Re; ICRU report No 50), were identified and discussed: Patient marking, patient positioning (from simulation through treatment), treatment, mobility of organs, repeatability and verification. Even though considerable effort is made to carefully reproduce the patient positioning from simulator to treatment unit, and to maintain the correct positioning during treatment, considerable deviations between the prescribed and the delivered dose, estimated to be in the order of several percent, may occur due only to patient positioning variations. Thus, a large portion of the allowable uncertainty (perhaps even exceeding some estimates of the overall acceptable uncertainty) may be caused by patient positioning deviations.

The need for simulation was emphasised and it was observed that the number of simulators in use is often very insufficient for the number of treatment units. The need for X-ray films for treatment planning and verification was indicated and it was observed the fluoroscopy alone is not adequate. The need for multidisiplinary team work including radiotherapists, physicians and technicians, working closely together was recognised. The role of rules and regulations in creating a situation where quality and quality assurance are accorded the required priority was demonstrated, for example training criteria and standards, equipment standards, quantity and qualifications of staff and mandatory participation's in quality assurance activities. Details concerning requirements in one Member State (Argentina) are provided in Annex 1. These could serve as a good model for other interested Member States.

DOSIMETRY PROCEDURES

Absolute dosimetry procedures continue to be based on ionization chamber measurements using a N_K-N_D formalism. The status of the IAEA Code of Practice, TRS-277, has a solid ground due to the good agreement found in dosimetry intercomparisons with different Primary Standard Dosimetry Laboratories. These mainly refer to ^{60}Co gamma-ray beams, but deviations not larger than 2% have also been obtained for higher photon beam qualities and electrons.

IAEA has continued its program of disseminating ionometric standards and achieving high accuracy in radiotherapy dosimetry. At the same time efforts have been addressed towards a continuous critical analysis of new developments in the field to take into account their possible influence on TRS-277 future updates. Accordingly a new Code of Practice for the calibration and use of plane-parallel ionization chambers is under development which will incorporate most of the advances in radiotherapy dosimetry since the publication of TRS-277 in 1987. Some of these advances have been treated in detail in this session, such as proposals for corrections to take into account the bremsstrahlung contamination in clinical electron beams. The measurement of relative dose distributions will also be improved by the implementation of almost tissue equivalent plastic scintillator detectors which, hopefully, will be able to overcome some of the limitations of detectors in current use.

New trends in absolute dosimetry have also been included in this session. Efforts have been addressed to the use of clinical high-energy photon beams to verify calibration factors in terms of absorbed dose to water (N_w). This is an important difference compared to verifications done previously in non-clinical photon beams, such as those produced by accelerators in most Primary Standard Dosimetry Laboratories (PSDL's). The results presented in this seminar are encouraging and two aspects should be stressed. First is that the comparison of experimental and calculated photon beam quality factors (k_Q) gives support to the data in TRS-277 at high-energy photon beams. Second, that prior to implementing N_w at high-energy photon beams PSDL's should put efforts into the use of real clinical beams for possible direct N_w calibrations.

- BRACHYTHERAPY

Considering;

- The increasing number of Ir-192 HDR machines in the world and the problems associated with its dosimetric aspects and the need for uniformization of the source calibration methods.
- The increasing acceptance of Ir-192 wires for interstitial brachytherapy and problems associated with its production and dosimetry.
- The potential mismanagement of patient due to lack of training.

It is proposed:

I. To promote a working group reviewing the available information and propose guidelines on how to approach these questions;

II. To prepare guidelines for the SSDLs assisting the users with appropriate dosimetric methodology in order to assure the necessary metrological coherence in each country.

These documents should also contemplate: Radiation protection and emergency procedures, quality assurance, staffing, training, a review of available clinical data from LDR & HDR treatments and cost-benefit analysis of this new technology especially with regards to the use of it in developing countries.

QUALITY ASSURANCE NETWORK IN RADIOTHERAPY

Since 1966, the IAEA/WHO Postal Dosimetry Programme for hospitals has proven the usefulness and the reliability of external audits, not only for developing countries, but also for countries where quality assurance programmes are applied. Large deviations have been observed in most areas of the world and repeated checks resulted in substantial improvements.

The European Pilot Study on postal dosimetry service, which was set up in 1991 to explore the feasibility of enforcing the capabilities of such remote postal assistance through additional checks including not only the calibration and the beam quality at the beam axis, but also off-axis doses, the treatment planning system as well as patient set-up through in-vivo measurements and port films. In particular, the pilot study showed that special devices such as the "multipurpose" phantom can be useful and should be made available to other networks when completely evaluated.

The discussions following the presentations pointed out a number of important points:

I. Depending on the conditions existing in a given country, remote assistance may not be sufficient. Local conditions, regulations or staff might not even allow checks to be performed locally. The reasons can be lack of physicists or any qualified staff able to carry out the measurements, insufficient training of the physicists or the lack of equipment. This might require on-site visits to be the first action instead of a follow-up visit after low scores with the probe service (educational interest).

II. The need for more qualified physicists in many countries has also been stressed, as well as a reasonable ratio of dosimetrists, technologists and engineers.

III. Difficulties were also identified regarding the recognition by some radiation oncologists of the role and the need for physicists in clinical trials, especially for the physics part of the Q.A. and more generally in some radiotherapy departments. It has been suggested that improvements can be expected from the participation of the physicists for the training of the radiation oncologists in medical physics, to reciprocal contributions in scientific/medical meetings, and, more generally, to common efforts for cooperation.

IV. With respect to the need for more accurate radiotherapy, it was emphasised that in many countries a severe lack excists of devices for accurate tumour definition and localization as well as for immobilization and imaging.

V. Quality audits should include not only external beam therapy, but also brachytherapy, in particular some verification and calibration, software. The development of high dose rate brachytherapy machines stresses the need for such QA systems

Finally, in general there was a feeling that in many developing countries the lack of physicists in combination with the low recognition of this professional group from Radiation Oncologists is a main obstacle for introducing proper QA programmes and thus, improving clinical dosimetry. The lack of recognition might be understandable given the perspective that these physicians have worked with radiation therapy for many years without a single medical physicist in the country. Primary goals must therefore be to create resources for education of this type of specialists and to make clinicians aware of the value of such staff. The IAEA could play an important role in helping creating the resources for education and in combination with WHO the physicians could be informed on the necessity of medical physicists in the clinic

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