Dosimetry issues related to the SSRT project at the ESRF

P. Berkvens[1], J. F. Adam[2], E. Brauer[1], A. Bravin[1], F. Esteve[2], C. Nemoz[1], Y. Prezado[1], M. Renier[1], H. Requardt[1] and M. Vautrin[2]

[1]*European Synchrotron Radiation Facility – BP220 Grenoble Cedex 09 France*

[2]*INSERM-U647/ESRF, Grenoble, France; MRI Unit, Grenoble University Hospital, Grenoble, France*

Abstract

The present paper describes the Stereotactic Synchrotron Radiotherapy (SSRT) project at the ESRF, emphasizing on the particular radiation safety issues related to this novel technique. The dosimetric characterization of the monochromatic X-ray beams used for therapy and for both 2D and tomography imaging is described. The dosimetry protocol is based on the IAEA 398 protocol, with the special provision to simulate uniform radiation fields by scanning an ionization chamber and a water phantom through the beam. The duality between the scanning ionization chamber geometry and the standard broad beam geometry is mathematically derived. It is shown that the use of a transmission ionization chamber scanned through the beam allows the on-line measurement of the integrated dose during patient treatment.

1. Introduction

The ID17 beamline at the European Synchrotron Radiation Facility in Grenoble, France is dedicated to medical applications. In the past a medical research protocol in the field of synchrotron radiation angiography has been conducted. Presently two radiation therapy programs, the Stereotactic Synchrotron Radiotherapy program (SSRT) and the Mircobeam Radiation Therapy program are being investigated and are progressing towards the clinical phase.
The SSRT technique consists in the stereotactic irradiation of a (brain) tumour using a monochromatic beam, with a dose enhancement effect due to the injection of a contrast agent in the tumour.
Synchrotron radiation beams are characterized by their reduced height (typically of the order of 1 mm). This particular aspect requires special attention concerning the dosimetric characterization of these beams.
International dosimetry protocols recommend the use of ionisation chambers to characterize and to calibrate X-ray beams used for radiotherapy purposes. The most widely used protocol for conventional radiotherapy in hospitals is the IAEA 398 protocol [1] and is based on absorbed dose in water. However, other protocols [2], using ionisation chambers calibrated in air in terms of air kerma, are used in reference dosimetry protocols for low- and medium-energy x-rays for radiotherapy and radiobiology.
These protocols assume broad beam irradiation geometries, which imply that the transverse dimensions of the X-ray beam are larger than the transverse dimensions of the ionisation chamber. In the case of radiotherapy techniques using synchrotron radiation X-ray beams the broad beam irradiation condition can not be fulfilled, since the X-ray beams will always have at least one transverse dimension which is smaller that the corresponding dimension of any commercially available ionisation chamber. The standard dosimetry protocols therefore need to be adopted to take into account this particular aspect. This is the purpose of the dosimetry protocol based on scanning the ionisation chamber through the X-ray beam, described hereafter.

2. The use of a thimble ionisation chamber in a flat beam

Ionisation chambers used for absolute dosimetry in radiation therapy must be calibrated in uniform broad radiation fields, whose transverse dimensions are much larger than the corresponding dimensions of the ionisation chamber. When using the ionisation chamber in a flat beam, the geometry is completely different from the standard calibration conditions. We must therefore proof that we can nevertheless use the broad beam calibration factors of the ionisation chamber.
When exposing an ionisation chamber to a uniform field, the different parts of the active volume of the chamber respond in a highly non-uniform way, depending on the chamber's geometry and due to the presence of the central electrode. Figure 1 shows the response of a PTW 31002 ionisation chamber to a flat beam, as a function of the transverse position of the beam relative to the ionisation chamber [3].

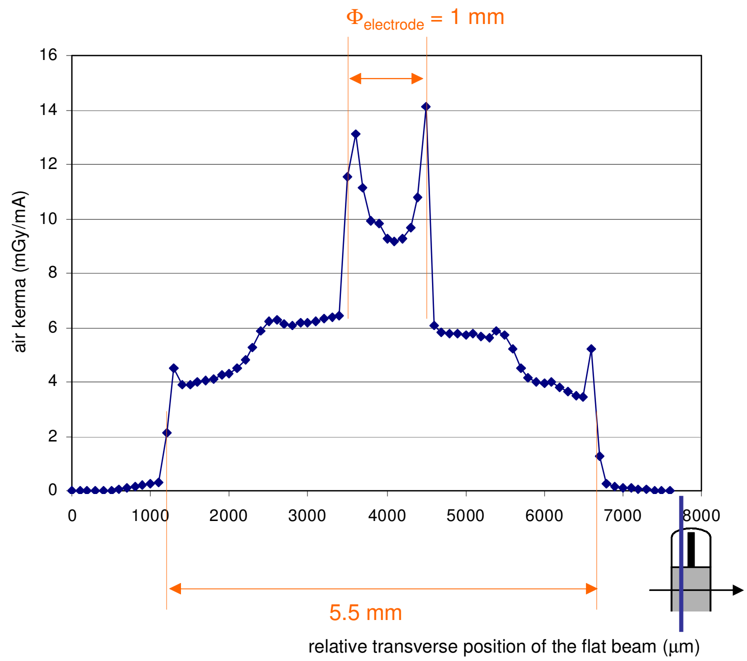

Fig.1 - Response of a PTW 31002 ionisation chamber to a flat beam.

If we denote by S(z) the relative sensitivity of the ionisation chamber as a function of one of its transverse dimensions, the measured dose rate $\dot{D}_{calibration}$ when exposing the ionisation chamber to a uniform calibration radiation field with an absolute dose rate $\dot{D}_{beam}$ can be expressed as:

$$\dot{D}_{calibration} = C \times \dot{D}_{beam} \times \int_{active\ volume} S(z) \cdot dz \ , \tag{1}$$

with C a calibration factor.

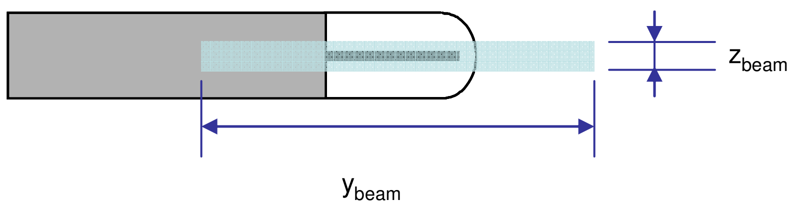

Fig.2 - Typical geometry encountered in synchrotron radiation X-ray dosimetry.

Consider the geometry of figure 2. An ionisation chamber is exposed to an X-ray beam that has one transverse dimension (y_{beam}) larger than the corresponding dimension of the ionisation chamber, while the second transverse beam dimension (z_{beam}) is smaller than the corresponding dimension of the ionisation chamber. Assuming that the beam is uniform over its entire surface, and is characterized by a dose rate $\dot{D}_{beam}$ (absorbed dose in water or air kerma), the dose rate $\dot{D}_{measured}$ (absorbed dose in water or air kerma) measured with the ionisation chamber will be given by

$$\dot{D}_{measured} = C \times \dot{D}_{beam} \times \int_{\Delta z = z_{beam}} S(z) \cdot dz \ . \tag{2}$$

Expression (2) shows that the accurate measurement of the ionisation chamber's transverse response S(z) allows the use of a standard ionisation chamber to carry out absolute dosimetry measurements in typical synchrotron radiation X-ray beams. The real difficulty of this method results from the fact that the transverse response function S(z) of the ionisation chamber depends on the energy spectrum of the X-ray beam at the point of measurement and from the fact that the measured dose rate will be very sensitive to the exact position of the ionisation chamber. These two complications are avoided by scanning the ionisation chamber through the beam along the z direction, at a constant speed v_z and measuring the integrated dose, instead of doing a static dose rate measurement (see figure 3).

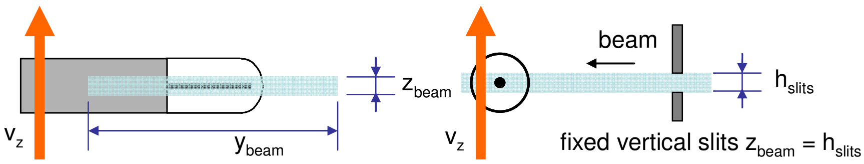

Fig.3 - Principle of scanning ionisation chamber.

Every plane cutting the ionisation chamber at a given position z will be exposed to a beam with a dose rate $\dot{D}_{beam}$ (absorbed dose in water or air kerma) during a constant time interval z_{beam}/v_z. The measured integrated dose $D_{measured}$ (absorbed dose in water or air kerma) will therefore given by.

$$D_{measured} = C \times \int_{active\ volume} \left(S(z) \times \int_{\Delta t = z_{beam}/v_z} \dot{D}_{beam} \cdot dt \right) \cdot dz = C \times \dot{D}_{beam} \times \frac{z_{beam}}{v_z} \times \int_{active\ volume} S(z) \cdot dz \tag{3}$$

or:

$$\dot{D}_{calibration} = D_{measured} \times \frac{v_z}{z_{beam}} \tag{4}$$

Expression (4) shows that from the measured integrated dose during the ionisation chamber scan a value for the dose rate can be obtained by multiplying the measured integrated dose, using the broad beam calibration factors, by the factor v_z/z_{beam}. Expression (4) is theoretically only valid if the intensity of the radiation field is such that the ionisation chamber is used in a regime where ion recombination is negligible. Indeed, the space charge distributions inside the ionisation chamber are obviously different when irradiating the chamber at once in a broad beam or when irradiating its different parts one after the other and the effects of ion recombination are expected to be different. Measurements using a PTW 31002 semiflex ionisation chamber seem to indicate that expression (4) would still be valid for ion recombination effect as large as 8% [3].

3. Relationship between a static uniform beam and a scanned synchrotron radiation beam

Consider a pencil-type beam, with transverse dimensions dy and dz and with a uniform distribution, e.g. characterised by its fluence F_0 (at x = 0). The dose rate in the point $(x_0,y_0,0)$ due to the pencil beam parallel to the x-axis, at the transverse position (y,z), for a given static geometry (infinite water phantom or irradiation on air) can be written as (see figure 4):

$$\dot{D}(x_0, y_0, 0) = F_0 \times T_{pencil}(x_0, y_0; y, z) \cdot dy \cdot dz \tag{5}$$

with $T_{pencil}(x_0,y_0;y,z)$ the geometry dependent conversion factor between fluence and dose rate (Gy·cm^2).

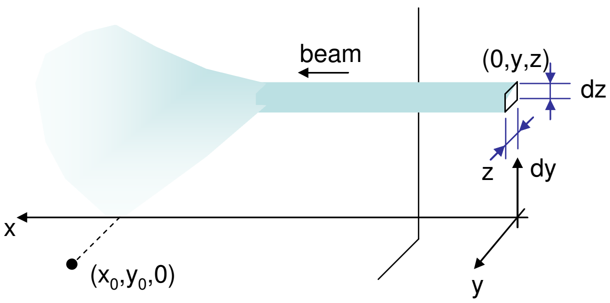

Fig.4 - Irradiation by a pencil-type beam.

The dose rate in the point $(x_0,y_0,0)$ due to a uniform broad beam, with field size Δy, Δz, centred around the x-axis is then given by:

$$\dot{D}_{broad}(x_0,y_0,0) = F_0 \times \iint_{\Delta y,\Delta z} T_{pencil}(x_0,y_0;y,z)\cdot dy\cdot dz = F_0 \times \int_{\Delta z} T_{flat\ \Delta y}(x_0,y_0,z)\cdot dy$$
$$\left(T_{flat\ \Delta y}(x_0,y_0,z)\cdot dz = dz \times \int_{\Delta y} T_{pencil}(x_0,y_0;y,z)\cdot dy \right) \tag{6}$$

$T_{flat\ \Delta y}(x_0,y_0,z)\cdot dz$ is the dose rate in the point $(x_0,y_0,0)$ due to a flat horizontal beam with width Δy, centred around the x-axis, and with a height dz, located at a vertical position z (see figure 5, left part).

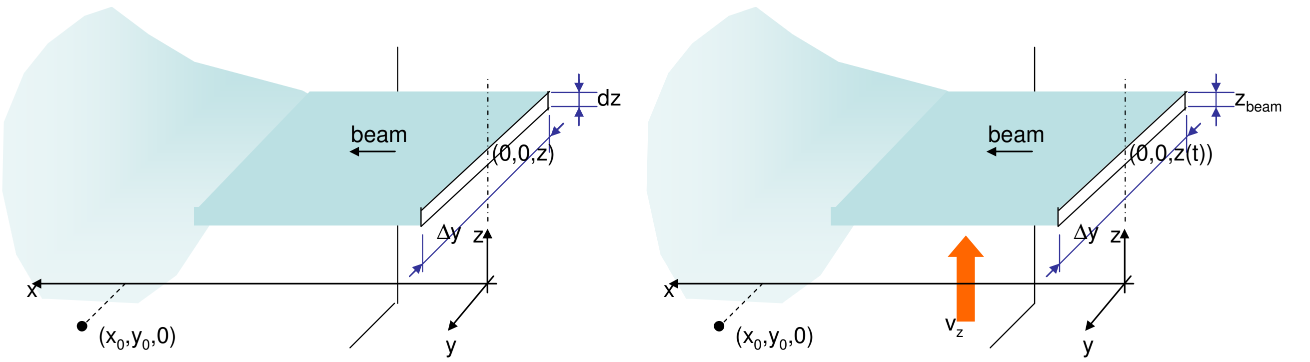

Fig.5 - Irradiation by a flat beam (left: static; right: scanning through flat beam).

We consider next the configuration used for SSRT, where we scan the object to be irradiated through the flat synchrotron radiation beam. If z_{beam} denotes the height of the beam, and Δy its width, the dose rate in the point $(x_0,y_0,0)$ is given by (see figure 5, right part):

$$\dot{D}(x_0,y_0,0,t) = F_0 \times T_{flat\ \Delta y}(x_0,y_0,z(t)) \times z_{beam}. \tag{7}$$

The object is scanned at a constant speed v_z through the beam. The dose in the given point, integrated during the scan over a height Δz, centred on the beam, is given by:

$$D(x_0,y_0,0) = \int_{\Delta t} \dot{D}(x_0,y_0,0,t)\cdot dt = \frac{z_{beam}}{v_z} \times F_0 \times \int_{\Delta z} T_{flat\ \Delta y}(x_0,y_0,z)\cdot dz = \frac{z_{beam}}{v_z} \times \dot{D}_{broad}(x_0,y_0,0). \tag{8}$$

Expression (8) is the theoretical basis for defining the dosimetric characterisation of the flat X-ray beam and to define the link with the treatment planning software.

4. Comparison between the scanning ionisation chamber protocol and the IAEA 398 protocol

The IAEA 398 protocol defines the determination, on the central axis, of the absorbed dose under reference conditions, the determination of the central axis depth dose distributions and the determination of dose profiles at different depths. Expression (8) showed that a broad beam irradiation is perfectly simulated by scanning the phantom vertically through the flat X-ray beam, at constant speed. The standard dosimetric quantities can be measured by scanning a thimble ionisation chamber vertically through the beam, at constant speed. Expression (4) showed that the standard, broad beam, calibration factors for the ionisation chamber can be used. The broad beam field size is obtained horizontally by setting the slits to the required width Δy and vertically by the height of the scan Δz (symmetrically around the beam axis). The measurement of the absorbed dose under reference conditions is obtained from expression (8) by using the appropriate value for x_0 and for $y_0 = 0$. The central axis depth dose is obtained by measuring the dose, using expression (8), for different values of x_0, always for $y_0 = 0$. An example of measured percentage depth dose profiles is given in figure 6. These measurements were carried out using a PTW 31002 semiflex ionisation chamber in a

30 cm$_x$ × 20 cm$_y$ × 20 cm$_z$ water phantom, for a quasi-monochromatic beam of 80 keV [4]. Finally by measuring the dose, using expression (8) for a given value of x_0 and for different values of y_0, horizontal dose profiles are obtained. Figure 6 shows an example of measured horizontal dose profiles.

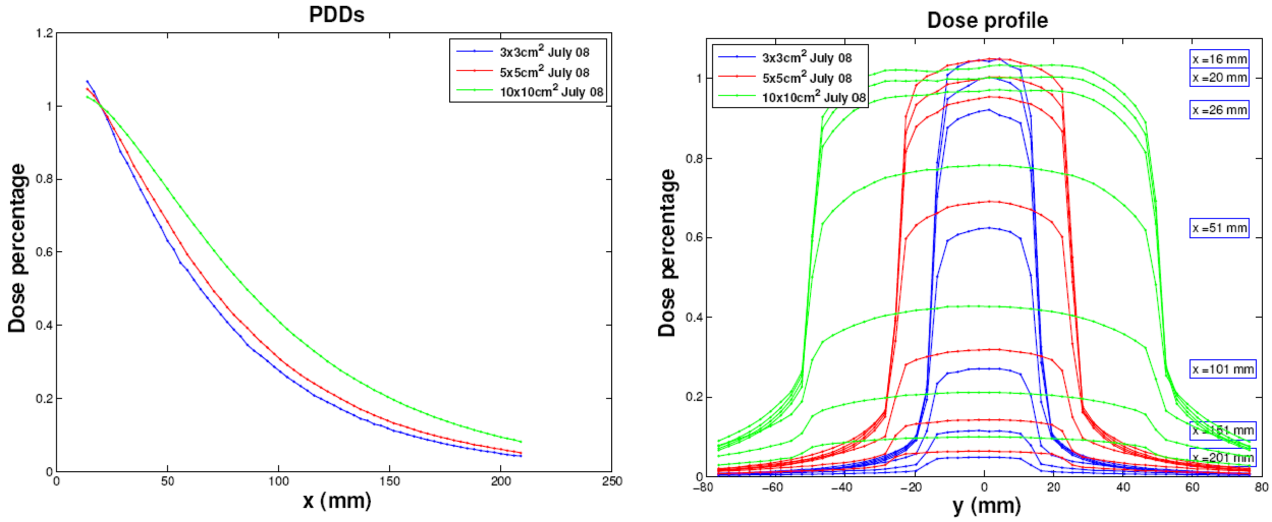

Fig.6 - Measured percentage depth dose profiles (left) and horizontal dose profiles (right).

5. Dose measurements during the patient treatment

Contrary to a conventional radiation treatment, where the integrated dose can be measured online using a calibrated transmission ionisation chamber, expression (8) shows that the use of a fixed, calibrated transmission ionisation chamber will not allow an online measurement of the integrated dose, because of the dependence of the integrated dose on the vertical scanning speed v_z. This would make the implementation of a reliable integrated dose interlock in the patient safety system too complicated. A different dose measurement must therefore be implemented. The solution will consist in the installation of a transmission ionisation chamber moving with the medical chair (see figure 7). As shown hereafter, the insertion of a transmission ionisation chamber, behind the 2D mask, moving together with the mask (and the medical chair) vertically through the beam, will provide a measurement for the integrated dose delivered to the patient for a given orientation.

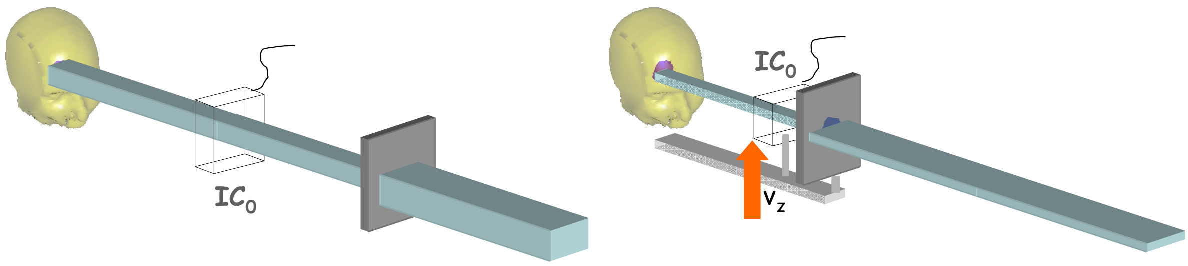

Fig.7 - Left: fixed ionisation chamber IC_0 in standard radiation therapy; right: scanned ionisation chamber IC_0 behind 2D mask in SSRT configuration.

The treatment planning software defines, for each individual irradiation orientation, the optimum two-dimensional shape of the uniform beam and the corresponding dose to the tumour. Using the same beam the treatment planning software gives a value for the corresponding on-axis dose at the reference depth in the standard water phantom (see figure 8). The initial (x = 0) transverse beam shape can be described by z_{min} and z_{max}, the minimum and maximum beam height respectively and by $(y_{max} - y_{min})(z)$ the z-dependent beam width. The dose rate in the reference point $(x_0,0,0)$ is then given by:

$$\dot{D}_{2D\,beam}(x_0,0,0) = F_0 \times \int_{z_{\min}}^{z_{\max}} \left(\int_{y_{\min}(z)}^{y_{\max}(z)} T_{pencil}(x_0,0;\,y,z) \cdot dy \right) \cdot dz\,. \tag{9}$$

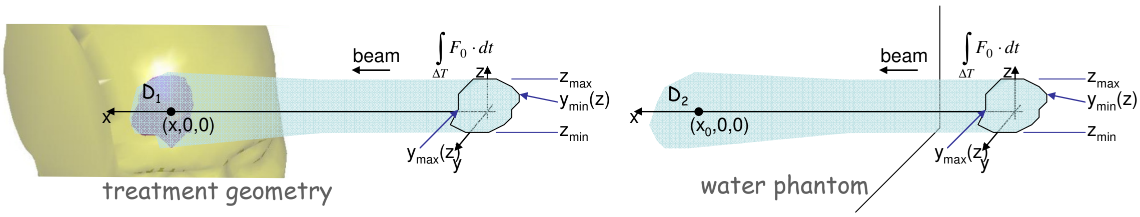

Fig.8 - Correspondence between dose D_1 delivered to the tumour and the absorbed dose to water D_2 at the reference depth in the water phantom.

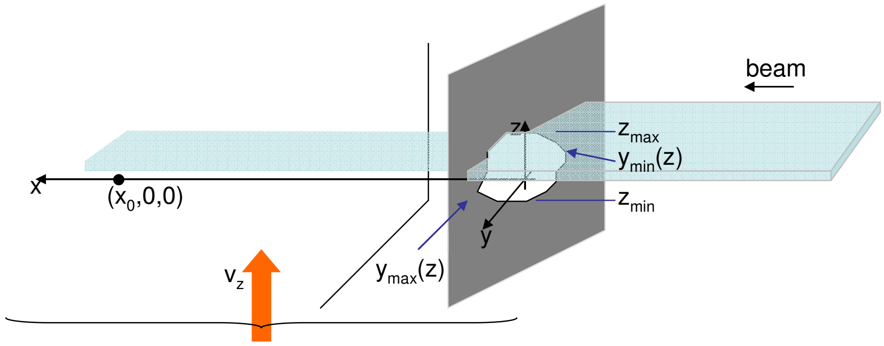

Fig.9 - Geometry for dosimetry measurement with moving 2D collimator.

The geometry of figure 8 is reproduced by inserting a 2D collimator, with a shape identical to the 2D beam defined by the treatment planning software, centred on the x-axis, fixed to the medical chair (therefore moving at the same speed during the vertical scan). If the 2D collimator is sufficiently thick to attenuate completely the beam, the dose rate in the reference point $(x_0,0,0)$ for the geometry in figure 9 is given by:

$$\dot{D}(x_0,0,0,t) = F_0 \times z_{beam} \times \int_{y_{\max}(z)}^{y_{\max}(z)} T_{pencil}(x_0,0;\, y, z(t)) \cdot dy \tag{10}$$

The integrated dose over the vertical scan (scan interval covers at least the interval $z_{min} - z_{max}$) is given by:

$$D(x_0,0,0) = F_0 \times \frac{z_{beam}}{v_z} \times \int_{z_{\min}}^{z_{\max}} \left(\int_{y_{\max}(z)}^{y_{\max}(z)} T_{pencil}(x_0,0;\, y, z(t)) \cdot dy \right) \cdot dz = \frac{z_{beam}}{v_z} \times \dot{D}_{2D\,beam}(x_0,0,0)\,. \tag{11}$$

Expression (11) shows that the correspondence between the integrated dose during a vertical scan and the on-axis absorbed dose rate in water obtained from the treatment planning software still holds. The integrated dose $D(x_0,0,0)$ can now again be measured using a thimble ionisation chamber, at the condition that the opening of the 2D collimator covers completely the active volume of the ionisation chamber (the minimum size of the tumours to be treated will be such that this condition will be fulfilled).

The measurement of the integrated dose with the ionisation chamber not only allows obtaining the dose rate in the 2D beam, using expression (11), but actually provides a direct measurement of the integrated absorbed dose to water in the reference point (the dose D_2 in figure 8), and which can therefore be related, via the treatment planning software, to the integrated dose delivered to the tumour, scanned with the same speed v_z through the beam with the same beam height z_{beam} (the dose D_1 in figure 8).

The integrated dose measured with a transmission chamber, inserted behind the 2D collimator, covering the entire collimator hole and moving together with the collimator during the scan, will be proportional to the integrated absorbed dose to water at the reference depth. It is therefore possible to calibrate this transmission ionisation chamber relative to the thimble ionisation chamber placed at the reference depth in the water phantom. In this way the transmission ionisation chamber will provide a direct measurement of the integrated absorbed dose in water in the reference point, the latter being directly related to the dose delivered to the tumour under the same scanning conditions.

It is important to note that the dose, measured with the thimble ionisation chamber, and the dose, measured with the transmission ionisation chamber, vary in the same way with the scanning speed v_z and the beam height z_{beam}. The calibration factor thus obtained for the transmission ionisation chamber will be independent

of v_z and z_{beam}. The online dose measurement with the transmission ionisation chamber, during the patient treatment, and the associated dose interlock will therefore be independent of the scanning speed v_z and the beam height z_{beam}. This is a fundamental advantage of the proposed dosimetry system.
Although the calibration factor of the transmission ionisation chamber is, as mentioned above, independent of the absolute value of the vertical speed of the chair v_z, it is important to keep this speed constant during the calibration of the transmission ionisation chamber, because of the z-dependence of the response function of the thimble ionisation chamber (or more precisely, because of the difference between the z-dependence of the response functions of the thimble ionisation chamber and of the transmission ionisation chamber).
In the same way, although the value for the dose delivered to the tumour, derived from the integrated dose measured with transmission ionisation chamber, is independent of the absolute value of the vertical chair speed v_z, , it is again important to keep this speed constant during the patient treatment, because a non constant speed will result in a vertically inhomogeneous dose delivery.
It should be noted that the duality between the dose D_2 obtained from the treatment planning software and the dose measured when scanning the ionisation chamber is mathematically only exact if the geometry used by the software treatment planning to obtain the dose D_2 is at all times identical to the geometry during the scan. This requires that the transmission chamber moves together with the collimator and the water phantom. Indeed, a transmission ionisation chamber, placed behind the 2D collimator, but remaining at the fixed beam height, would create a variable geometry during the scan, which cannot correspond to the static geometry used by the software.

6. Conclusions

We have shown that a commercially available thimble ionisation chamber can be used to characterise flat X-ray beams, by scanning it at constant speed through the beam, using the standard broad beam calibration factors of the ionisation chamber.
The Monte Carlo based treatment planning software relates the dose delivered to the tumour to the dose at the reference depth in a water phantom. The latter can be measured online by inserting a transmission ionisation chamber behind the 2D collimator, fixed to the medical chair.

References

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