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Ionization gradient chamber in absolute photon and electron dosimetry

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A variable volume parallel-plate ionization gradient chamber was built to determine the absorbed dose in a polystyrene phantom. The sensitive volume of the gradient chamber is controlled by moving the chamber piston by means of a micrometer mounted to the phantom body. The displacement of the piston is monitored by a calibrated distance travel indicator which is accurate to within 0.01 mm. Irradiations were carried out with cobalt-60 gamma rays, photon beams ranging from 4 MV to 18 MV, and electron beams between 5 MeV to 18 MeV.

With the ionization gradient chamber the calculation of the absolute dose at a given depth in phantom is simple and based on first principles using the slope of the measured ionization as a function of the electrode separation, i.e., the sensitive air volume. The discrepancies between the doses determined with our uncalibrated gradient chamber and those obtained with a calibrated standard chamber are at most 1.08 % and 0.63 % for photon and electron beams, respectively, at all clinical energies, indicating that the gradient ionization chamber can be used as an absolute dosimeter.

Key words: radiation dosage; polystyrene; photons; electrons; absolute dosimetry

Introduction

An accurate determination of the absolute dose rate produced by photon or electron machines is one of the most important components of modern radiotherapy. Radiotherapy clinics most commonly determine the absolute absorbed dose with parallel-plate or cylindrical ionization chambers which are first calibrated at, or trace their calibration factors to, a national standards laboratory. The dose is calculated from the measured ionization in air using the chamber calibration factor and following one of several available protocols (*e.g.,* ICRU,[1] AAPM-TG21;[2] AAPM-TG25;[3] IAEA-WHO;[4] *etc.*) These protocols are based on the standard Bragg-Gray[5,6] or Spencer-Attix[7] cavity theories and incorporate various correction factors, which are used to account for effects of chamber dimensions and wall materials as well as disruptions in the photon and electron fluence caused by the chamber. These correction factors make the dose determination cumbersome and introduce uncertainties in the final result.

The basic Bragg-Gray and Spencer-Attix cavity relationships for the dose D_{med} in medium are:

$$D_{med} = \frac{Q}{m} \overline{W}_{air} \overline{S}_{air}^{med} \qquad (1)$$

and

$$D_{med} = \frac{Q}{m} \overline{W}_{air} \overline{L}_{air}^{med}, \qquad (2)$$

respectively, where Q is the charge collected under saturation conditions in the sensitive chamber air mass m, W_{air} = 33.97 eV[8] is the mean energy re-

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quired to produce an ion pair in air, and S_{air}^{med} and L_{air}^{med} are the ratios of unrestricted and restricted collisional stopping powers, respectively, for the medium and air for the electron spectrum at the position of the cavity. Both the Bragg-Gray and the Spencer-Attix formalisms assume that the air cavity within the medium is sufficiently small such that it does not alter the electron fluence in the medium. The Bragg-Gray formalism uses unrestricted stopping powers averaged over the slowing-down spectrum of only the primary electrons, while the Spencer-Attix formalism uses restricted stopping powers averaged over the slowing-down spectrum of all generations of electrons.

It is evident from *Equations (1)* and *(2)* that the dose in medium is proportional to the measured ratio Q/m which in principle should be straight forward to determine. In actuality, Q is easy to measure accurately in clinical beams, however, m is almost impossible to determine with an accuracy of better than 1 % required for clinical use, precluding the direct use of *Equations (1)* and *(2)* in absolute dosimetry. The standard method for obviating this problem is to calibrate the cavity chamber response in a known reference radiation field which has been calibrated previously with a standard free air ionization chamber. This determination of the chamber calibration factor is actually an indirect means of determining the mass of air in the chamber sensitive volume. The chamber calibration factor in conjunction with various troublesome correction factors is then used to determine the dose to the medium.

Investigation of *Equations (1)* and *(2)* has revealed that at least for small m the ratio of Q/m is a constant allowing its replacement with the derivate dQ/dm, resulting in the following modified Bragg-Gray and Spencer-Attix relationships for the dose in medium:

$$D_{med} = \frac{dQ}{dm} \overline{W}_{air} \overline{S}_{air}^{med} \qquad (3)$$

and

$$D_{med} = \frac{dQ}{dm} \overline{W}_{air} \overline{L}_{air}^{med} \qquad (4)$$

The advantage to this approach is that, in contrast to Q/m, dQ/dm is relatively easily measured accurately making the modified Bragg-Gray and Spencer-Attix relationships directly applicable in absolute dosimetry. Similarly to Klevenhagen,[9] we have developed an uncalibrated, variable volume, *ionization gradient chamber (IGC)* capable of measuring the absorbed dose directly in an absolute manner. The chamber developed by Klevenhagen was made of Lucite and required the use of a water tenk for dose measurement; therefore, corrections for the density and fluence differences between Lucite and water had to be considered. Our chamber material is the same as the phantom material (polystyrene); consequently, there is no need for such corrections to the measured signal when determining the absorbed dose in polystyrene. The determination of the absolute absorbed dose for clinical photon and electron beams at a given depth in phantom with the IGC is based on first principles, is simple to evaluate, and agrees well with results obtained with standard calibrated ionization chamber techniques.

Materials and methods

A 7 cm diameter polystyrene piston was fashioned to move inside a cylinder bored along the center of a 30 x 30 x 8 cm^3 polystyrene phantom. Graphite dag was painted on the top surface of the piston, and a 1.5 mm deep and 0.04 mm wide groove was cut through the graphite surface into the piston to form the 2.004 (1 ± 0.001) cm inner diameter measuring electrode and the guard ring of the chamber. The measuring electrode and the guard ring are both connected to ground (the measuring electrode through an electrometer) with electronically shield-

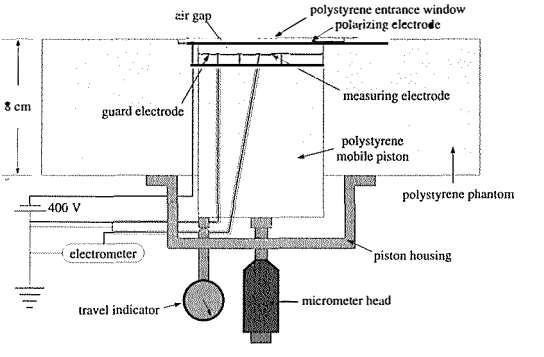

Figure 1. Schematic diagram of the ionization gradient chamber.

ed cables. The polarizing electrode consists of a 0.5 mm thick polystyrene disk painted with graphite dag and fastened to the top of the large phantom. The electronic potential of the polarizing electrode is maintained at ±400 V with respect to the collecting

electrode. The separation between the polarizing and measuring electrodes can vary between 0.5 mm and 10 mm, and is controlled by a micrometer mounted to the phantom body. The movement of the piston (*i.e.,* change in the air sensitive volume) is monitored by a calibrated distance travel indicator which is accurate to within 0.01 mm. In Figure 1 we show a schematic diagram of the IGC. Irradiations of the gradient chamber were performed with a cobalt-60 gamma source, photon beams in the energy range from 4 MV to 18 MV, and electron beams in the nominal energy range from 9 MeV to 18 MeV.

Results and discussion

The specific design of our IGC allows us to determine dQ/dm of *Eq. (4)* with relative ease and a high degree of accuracy. Since dm is directly proportional to the change dz in electrode separation, we can write *Eq. (4)* as follows:

$$D_{med} = \left(\frac{1}{\rho A}\right) \frac{dQ}{dz} W_{air} L_{air}^{poly}, \qquad (5)$$

with ρ the density of air at the ambient temperature and pressure, and A the area of the measuring electrode.

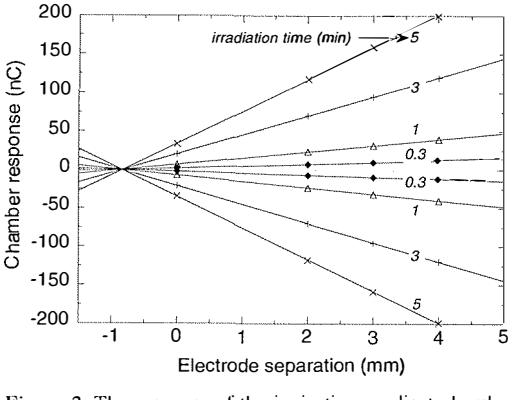

Figure 2. The response of the ionization gradient chamber as a function of electrode separation. The chamber was exposed to cobalt-60 radiation (field-size: 10 x 10 cm^2; source-surface distance: 80 cm; dose rate: 86.7 cGy/min). The buildup region consisted of 3.7 mm of polystyrene.

As shown in Figure 2, the response of our ionization gradient chamber to cobalt-60 radiation varies linearly with electrode separation (correlation coefficient ≥ 0.99995), with dose (irradiation time) a parameter. The chamber response is represented by the measured change Q corrected for the chamber collection efficiency[10, 11] at the polarizing voltage of 400 V and given electrode separation z. All ionization response curves for positive and negative chamber polarities intersect at the same location on the x-axis indicating the true zero electrode separation. We purposely did not calibrate our electrode separation to this intersection point in order to emphasize that there is no need to determine the separation in an absolute manner: only the relative variation in electrode separation is required in *Eq. (5)* The slopes dQ/dz obtained for the given doses in Figure 2 depend linearly on the dose as shown in Figure 3. Similar results were obtained in pulsed

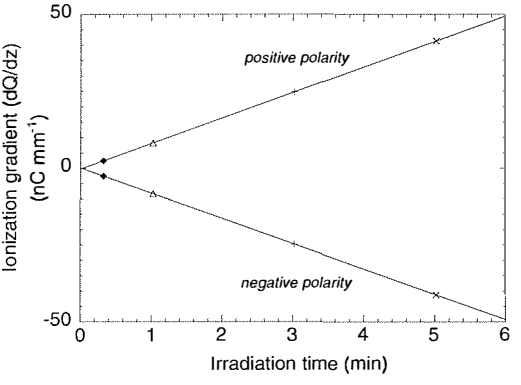

Figure 3. Ionization gradient dQ/dz as a function of the irradiation time. (dQ/dz was determined from data of *Figure 2.*).

photon and electron beams showing that (i) the chamber response is linear with dose and (ii) dQ/dz may be measured with a high degree of precision.

In Tables 1 and 2 we show how the ionization gradient chamber meets its main objective: the absolute dose determination in clinical photon and electron beams, respectively. Doses determined at a given depth in polystyrene with a calibrated Farmer chamber and the AAPM-TG21 protocol for photon beams and AAPM-TG25 protocol for electron beams are compared with doses determined at same depths in phantom with our polystyrene ionization gradient chamber. Tables 1 and 2 also give other relevant parameters used in the absolute dose measurements with the ionization gradient chamber. The discrepancies between doses determined with our uncalibrated gradient chamber and those obtained with the calibrated Farmer chamber are at most 1.08 % and 0.63 % for photon and electron beams, respectively, at all clinical energies indicating that the ionization gradient chamber can be used as an absolute dosimeter.

Table 1. Measurement of photon dose with the ionization gradient chamber.

1	2	3	4	5	6	7
Photon beam	Depth (mm)	$\bar{L}_{air}^{poly}$	dQ/dz ($nCmm^{-1}$)	Dose (IGC) (cGy)	Dose (TG 21) (cGy)	% difference
Co-60	3.7	1.113	8.274	83.41	83.18	+ 0.27
4 MV	10.1	1.108	9.640	97.26	96.79	+ 0.49
6 MV	50.1	1.103	8.313	83.81	84.23	– 0.49
10 MV	50.1	1.094	8.886	88.11	88.36	– 0.29
18 MV	50.1	1.078	9.397	92.56	93.57	– 1.08

(1) photon beam energy; (2) depth d in phantom; (3) ratio of restricted stopping powers[2] (Δ = 10 keV); (4) measured ionization gradient averaged over positive and negative polarities and corrected for charge recombination; (5) dose measured with ionization gradient chamber; (6) dose determined with the AAPM-TG21 protocol[2]; (7) percent difference between (5) and (6).

Table 2. Measurement of electron dose with the ionization gradient chamber.

1	2	3	4	5	6	7	8	9
Electron beam	$\bar{E}_0$ (MeV)	Depth (mm)	$\bar{E}(d)$ (MeV)	$\bar{L}_{air}^{poly}$	dQ/dz ($nCmm^{-1}$)	Dose (IGC) (cGy)	Dose (TG 25) (cGy)	% difference
9 MeV	8.1	15	5.24	1.017	10.469	95.56	96.01	– 0.47
12 MeV	10.8	15	7.96	0.988	10.825	95.83	96.44	– 0.63
15 MeV	13.5	10	11.60	0.964	11.233	97.74	98.21	– 0.49
18 MeV	16.1	10	14.42	0.952	11.484	98.94	98.51	+ 0.44

(1) electron beam nominal energy; (2) mean electron energy at phantom surface; (3) depth d in phantom; (4) mean electron energy at depth d; (5) ratio of restricted stopping powers[2] (Δ = 10 keV) at $\bar{E}$(d), (6) measured ionization gradient averaged over positive and negative polarities and corrected for charge recombination; (7) dose measured with ionization gradient chamber; (8) dose determined with AAPM-TG25 protocol[3]; (9) percent difference between (7) and (8).

Conclusions

Uncalibrated ionization gradient chambers built as part of the phantom in which the dose is measured behave as Bragg-Gray cavities and can be used reliably in the determination of absolute dose. In contrast to the dosimetry with calibrated chambers, the dosimetry with ionization gradient chambers appears simple and requires no troublesome correction factors to account for chamber properties and for the unavailability of high energy photon and electron calibrations at standards laboratories. With our gradient chamber design, no cumbersome apparatus is required to measure the plate separation absolutely in order to determine the absorbed dose in an absolute manner. The charge per unit air mass gradient can be measured accurately (to within 1%) with relative ease in a carefully designed and precisely built gradient chamber. This implies that absolute dose measurements with ionization gradient chambers could be added to the other three currently known absolute dosimetry techniques: calorimetry, chemical (Fricke) dosimetry, and standard free air ionization chamber.

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