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Skin dose near compensating filters in radiotherapy

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Abstract. In radiotherapy treatments with MV beams, the use of tissue compensators affects the dose to the skin. Methods of calculating the relative skin dose (RSD) are described and a formula is derived to predict the contribution of a tissue compensator to the RSD. Measurements of RSD for various field sizes and distances from the compensator are presented.

1. Introduction

The absorbed dose at the surface of a patient or phantom irradiated with a beam of MV x-rays results from three components. There are contributions from electrons generated in the air above the phantom, from electron backscatter within the phantom and from electrons generated by any solid material in the beam. The latter component can arise from the collimators, blocking trays, etc. If a metal filter is being used to compensate for tissue variations in the patient then the compensating filter will remove electrons generated above it, but will itself be a source of electrons.

The term 'relative skin dose' as defined by Saylor and Quillin (1971) will be used. Relative skin dose is defined as

$$\mathrm{RSD} = \frac{\text{(absorbed dose near skin surface)}}{\text{(peak dose at equilibrium dose)}}.$$

This is made up of the three components, from air, backscatter and compensator, i.e.

$$\mathrm{RSD} = (\mathrm{RSD_a} + \mathrm{RSD_b} + \mathrm{RSD_c})/(1 + \mathrm{RSD_b}). \quad (1)$$

(The denominator in this equation is required since the methods used in this paper of calculating the air and compensator components are normalised without reference to the backscatter component.)

Methods of calculating the air-generated component are well established. Howarth (1951) calculated the relative skin dose by assuming this dose to be due to Compton recoil electrons generated in air, and by ignoring multiple scatter. By using the Klein–Nishina formula, numerically integrating over the volume of irradiated air and adding in a contribution for electron backscatter in the phantom, he obtained predictions of relative surface ionisation that showed good agreement with measurements for 2 MV x-rays at a distance of 70 cm from a perspex electron filter. Higgins *et al* (1983) found similar agreement for ^{60}Co.

The contribution from electron backscatter in the phantom was measured by Howarth (1951) for 2 MV x-rays and was found to give $\mathrm{RSD_b} = 0.13$ independent of field size. $\mathrm{RSD_b}$ should only vary with field size if the field radius is less than the maximum range of the electrons produced in the patient. As will be shown below, both RSD_a and RSD_c tend to zero as the field size tends to zero; RSD_b can therefore be determined by extrapolating, to zero field size, values of RSD measured for fields with a radius greater than this range.

2. Electrons generated in the compensator

Saylor and Quillin (1971) derived a completely empirical formula giving the RSD for ^{60}Co radiation as a function of field size and distance from a lead glass electron filter. Whilst usefully summarising their measurements, the value of the formula in predicting the dose for other types of fiiter and other energies of radiation is limited.

Howarth (1951) calculated the contribution from the Perspex electron filter in the same way as the air contribution, ignoring multiple scatter within the filter. This is a reasonable approximation in low atomic number materials, but it is a very poor approximation in high atomic number materials such as lead, where the product of the range and the angular stopping power is about ten times greater than for air or Perspex. Ignoring multiple scatter results in a large overestimate of the skin dose, as will be shown in the results section.

Multiple scatter within the lead filter is large enough at radiotherapy energies to enable one to ignore the Klein-Nishina coefficients and assume that radiation of electrons from the compensator is isotropic, with each element of irradiated compensator radiating electrons uniformly over a solid angle of 2π. This approximation enables one to calculate an analytic expression for the contribution of the compensator to the RSD. The calculation will assume a circular compensator and we will also assume that electrons travel in straight lines and that the energy spectrum is not altered by the air. We make the assumption that the irradiated part of the compensator acts as a uniform isotropic source of electrons, emitting n electrons per m per sr. We will further assume that RSD_c is proportional to the number of electrons per unit area of skin; the constant of proportionality will depend on energy and is discussed later. (Figure 1 illustrates some of the parameters used in the calculation of RSD_c.) Thus the total

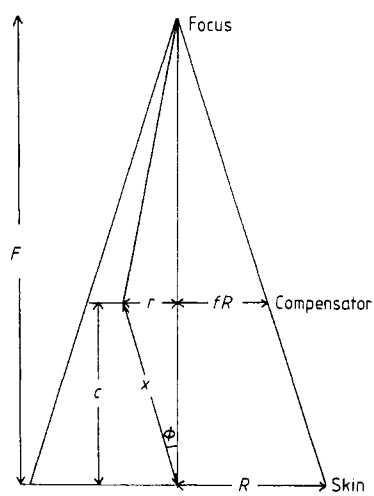

Figure 1. Diagram explaining the symbols used in the calculation of RSD_c.

number of electrons emitted in the forward direction will be

$$\pi f^2 R^2 n 2\pi. \tag{2}$$

For a point on the axis, the electron fluence will be

$$\int_{\theta=0}^{2\pi} \int_{r=0}^{fR} \frac{rn \, \mathrm{d}\theta \, \mathrm{d}r}{x^2}.$$

The number of electrons per unit area of skin will be

$$\int_{\theta=0}^{2\pi} \int_{r=0}^{fR} \frac{\cos\phi rn \, \mathrm{d}\theta \, \mathrm{d}r}{x^2}. \tag{3}$$

The need for the cos ϕ term is illustrated by figure 2. Fluence is defined as $\psi = \mathrm{d}N/\mathrm{d}a$, where $\mathrm{d}N$ is the number of particles incident on a sphere of cross sectional area $\mathrm{d}a$. However, a sphere of a given cross sectional area πA^2 will correspond to an ellipse on the skin of area πAB, where $A = B\cos\phi$.

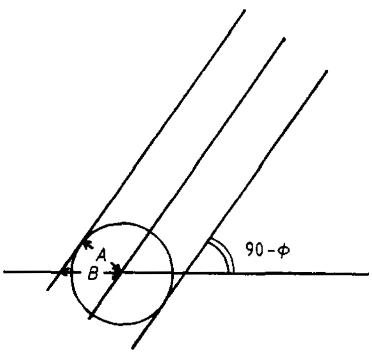

Figure 2. Diagram explaining the symbol ϕ in equation (3).

Integrating (3) gives the number of electrons per unit area of skin, which is equal to

$$\int_0^{fR} \frac{2\pi nr \cos\phi \, \mathrm{d}r}{x^2} = 2\pi n \int_0^{fR} \frac{rc \, \mathrm{d}r}{(c^2+r^2)^{3/2}}$$

$$= 2\pi n\left(1 - \frac{c}{\sqrt{c^2+f^2R^2}}\right). \tag{4}$$

As a fraction of the total number of electrons emitted in the forward direction, this is equal to

$$\frac{1}{\pi f^2 R^2}\left(1 - \frac{c}{\sqrt{c^2+f^2R^2}}\right). \tag{5}$$

To relate this to the absorbed dose, one should consider the situation when the compensator is touching the skin, when $c = 0$ and $f = 1$. This gives a fraction $1/\pi R^2$.

If a compensator made of tissue-equivalent material were used, this situation would correspond to full build-up, i.e. RSD = 1. Hence for a tissue-equivalent compensator we get

$$\mathrm{RSD_c} = \frac{1}{f^2}\left(1 - \frac{c}{\sqrt{c^2 + f^2 R^2}}\right)$$

irrespective of energy.

For a compensator made of metal, the RSD when the compensator is in contact with the skin will depend on the material used and on the energy of radiation. This will be called the factor $M(E)$, giving

$$\mathrm{RSD_c} = \frac{M(E)}{f^2}\left(1 - \frac{c}{\sqrt{c^2 + f^2 R^2}}\right). \tag{6}$$

$M(E)$ can be determined experimentally by measuring the skin dose with a compensator touching the skin and comparing it with the dose when a tissue-equivalent compensator is used. The readings should be corrected for attenuation in the compensator. $M(E)$ should equal the ratio of the mass energy absorption coefficients of the metal and the tissue. Both these coefficients will depend on the energy of radiation.

$$M(E) = \frac{(\mu_a/\rho)_{\mathrm{metal}}}{(\mu_a/\rho)_{\mathrm{tissue}}}. \tag{7}$$

Formula (6) enables predictions to be made. Tests of these predictions against experiment have been made, as detailed in the section below.

3. Experimental method

3.1. Radiation source

A Brown–Boveri CH5 accelerator was used to produce beams of nominal 5 MV x-rays. The depth of the 80% dose level for a 10 cm × 10 cm field at a 100 cm source–skin distance (SSD) is 6.2 cm. The narrow-beam attenuation coefficient in water, derived by extrapolation of the tissue maximum ratio data to zero field size, is 0.047 cm^{-1}, which is equivalent to a monoenergetic beam of 2.17 MeV (Johns and Cunningham 1983).

3.2. Other apparatus

A thin windowed ionisation chamber, designed for the measurement of low-energy electrons (Morris and Owen 1975), was used. The chamber has a cylindrical air volume 27 mm in diameter and 2 mm thick, with a front wall of aluminised Melinex 1 mg cm^{-2} thick, set in a polystyrene block. The chamber was in a block of polystyrene, 20 cm square and 5 cm deep. Since this depth of polystyrene was not sufficient to ensure that as much backscatter occurred as would occur in a patient, the chamber and block were placed on a 25 cm thick pressed-wood phantom. The chamber was connected to a Keithley 35617 electrometer.

A 'compensator plate' was made of 4 mm of lead stuck to a 1 cm thick Perspex plate. The lead-covered side of the plate faced towards the chamber in all experiments. Measurements of charge per 100 monitor units were taken for a variety of SSD and for a variety of compensator–skin distances (CSD). Each reading was related to the value for an identical irradiation with the compensator removed, with 1.5 cm of polystyrene placed on the chamber. The ratio of these readings, corrected for attenuation in the lead and in the plastics, gives the relative skin dose (more properly the relative surface ionisation).

4. Results

The value of $M(E)$ was measured for this radiation energy by measuring the surface ionisation (per 100 monitor units) with build-up of lead and polystyrene, correcting each value for attenuation. A value of 0.93 was obtained, which agrees with the value predicted by equation (7) for 2 MeV radiation of $(0.0235)/(0.0253) = 0.93$ (Johns and Cunningham 1983).

The backscatter component was measured by extrapolating, to zero field size, skin dose measurements measured for a number of field sizes from 6 cm × 6 cm to 30 cm × 30 cm. The graph of RSD against field size was not a perfect straight line. Several lines could be drawn through the data with intercepts from 6 to 9%. The value used was 7.5%, i.e. $\mathrm{RSD_b} = 0.075 \pm 0.015$.

Figure 3 shows the variation of RSD with CSD for a 10 cm × 10 cm field at 200 cm source–surface distance (SSD). The dotted curves show the contribution to the RSD from air, backscatter and compensator. The full curve shows the sum of these (divided by $(1 + \mathrm{RSD_b})$), which should equal the measured RSD. The broken curve shows the predicted RSD if multiple scatter is ignored in the compensator (see appendix). It will be seen that ignoring multiple scatter results in a large overestimation of dose, whilst the methods described above slightly underestimate it. Figure 4 shows the variation of RSD with field size at two different distances from the compensator.

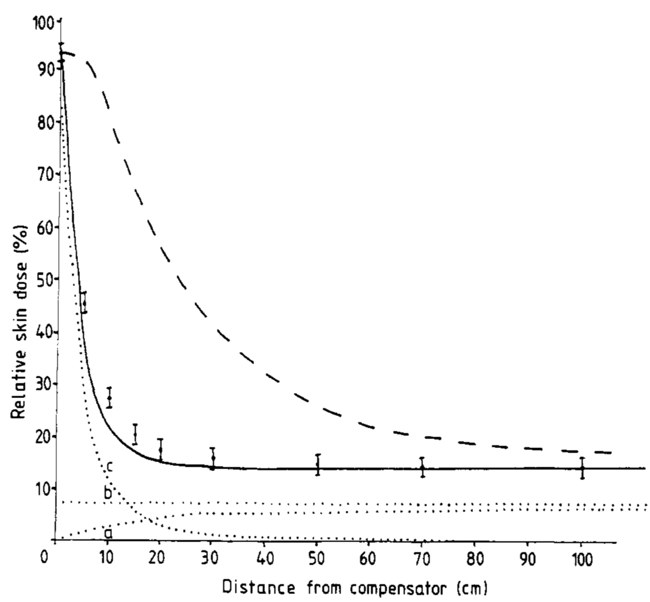

Figure 3. Variation of RSD with distance from the compensator for a 10 cm × 10 cm field size at 200 cm SSD. The dotted curves a, b and c show the predictions of the contributions from air, backscatter and compensator respectively. The full curve is the total predicted RSD. The broken curve shows the predicted RSD if multiple scatter in the compensator is ignored.

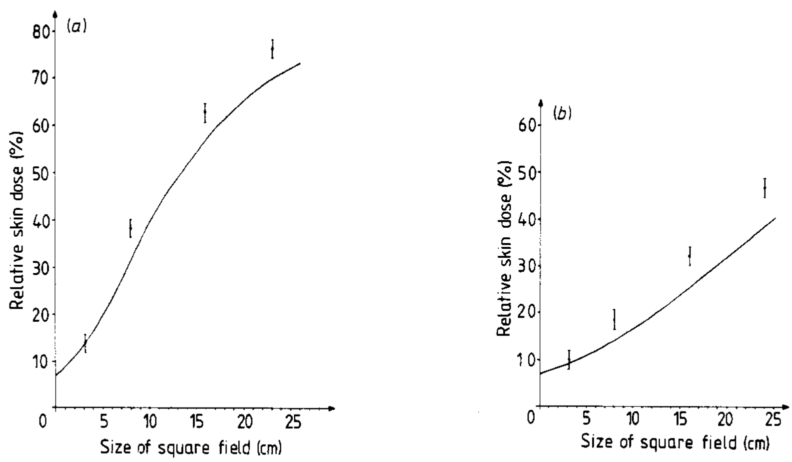

Figure 4. Variation of RSD with field size; (*a*) 5 cm from the compensator at 80 cm SSD; (*b*) 14 cm from the compensator at 80 cm SSD.

In these graphs, the data for square fields have been plotted against the predictions for circular fields of the same area. This has been done because of the difficulty of making calculations for square fields.

5. Discussion

ICRU report 24 (1976) states that 'the reduction of build-up in the skin is negligible for ^{60}Co gamma rays and other high-energy radiations if the compensating filter is separated from the skin by at least 15 cm.' As these results illustrate, this is only true for small field sizes. At 15 cm from the compensator a small field of radius 4 cm will only give a 3% contribution to the skin dose, but a 15 cm radius field would give a contribution of about 30%.

It is important, therefore, to consider the skin dose when using compensators in mantle therapy. When backscatter and air-generated electrons are included, the relative skin dose for a 15 cm radius field is about 40% at 15 cm from the compensator and 20% at 50 cm from the compensator. With most older treatment machines this was not a problem since the large field sizes required for mantles required treatments at extended SSD, and hence at large distances from any compensators used. Many newer machines have larger field sizes available at the isocentre. For example, the BBC CH5 has a maximum field size at the isocentre of 35 cm × 35 cm and a compensator tray system at 30 cm from the isocentre. If a patient of thickness 20 cm were to be treated isocentrically with a maximum field size, a RSD of over 40% would be received.

More work is needed to produce similarly simple equations to predict the air-generated component since the need to write a computer program to calculate RSD_a slightly reduces the usefulness of this formula in predicting total RSD. A formula to predict RSD_b at other energies would also be useful, although it is less essential in calculations for determining the effect of compensators since the presence of compensators does not alter RSD_b.

The formula given in this paper gives a simple means of estimating the approximate skin dose when near to a compensator. The model slightly underestimates the skin dose, suggesting that the assumptions are not completely correct. It gives considerably better predictions, however, than methods which use the Klein–Nishina coefficients and ignore multiple scatter in the filter; it also gives a simple formula rather than an algorithm needing numerical solution.

Electrons produced in Compton interactions are predominantly forward oriented. The more they are scattered out of the field, the lower the dose that will be received by the skin at the centre of the field. The assumption made in deriving the formula in this paper is that they are scattered so much that they lose their directionality. If slightly less scatter takes place, so that the electrons are still slightly forward oriented, a larger dose will be observed at the centre of the field than predicted by the formula. Figure 3 suggests that this is indeed the case, the measured values slightly exceeding the predictions of the formula. A more refined theory should include a small correction for this.

Appendix. Calculation of $\mathrm{RSD_a}$

Howarth (1951) showed that if multiple scatter is ignored, then

$$\mathrm{RSD} = \int_0^{\psi_0} \cot \psi C(\psi)\, \mathrm{d}\psi \left(\frac{1}{F^2} \int_0^{\pi/2} R(\phi) f(\phi) \sin \phi \, \mathrm{d}\phi \right)^{-1}$$

where

$$C(\psi) = \int_{x_0}^{F} B(x, \psi)\, \mathrm{d}x + B(x_0, \psi) R' \tag{A1}$$

$$B(x, \psi) = f(\phi) \sin^2(\phi - \psi)/x^2$$

and $f(\phi)$ is the number of recoil electrons per scattering electron per second per unit solid angle as calculated using the Klein–Nishina formula. F, ϕ, ψ, ψ_0, x are as defined in figure 5, $R(\phi)$ is the range in air of secondary electrons emitted at angle ϕ and $R' = R - (F - x_0) \sec(\phi - \psi)$.

Values have been calculated from these formulae by numerical integration using a Fortran program run on an Epson QX-10 computer. Data on electron ranges from Johns and Cunningham (1983) were used in the calculation.

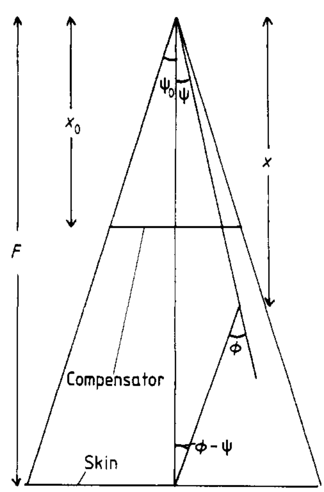

Figure 5. Diagram explaining the symbols used in the calculation of $\mathrm{RSD_a}$.

The first component in equation (A1) is for the electrons generated in the air, the second for electrons generated in a filter (a Perspex filter was used in measurements in Howarth's paper). In the calculations in this paper, only the first component of equation (A1) has been used to give an estimate of the air-generated component. The one exception to this is the broken curve of figure 3, where the second component has been used instead of using the formula in equation (6); the curve has been normalised to give $\mathrm{RSD} = 0.93$ at $c = 0$ and includes the additional contribution for RSD_b using equation (1). The purpose of this curve is to illustrate that equation (A1) gives a much poorer estimate of RSD_c than does equation (6) in cases where a lead compensator is used. This is to be expected from the theory, since a lead filter causes more multiple scatter of electrons than does a Perspex one.

Résumé

Dose à la peau á proximité des filtres compensateurs utilisés en radiothérapie.

Dans les traitements effectués avec des faisceaux de photons de haute énergie, l'emploi de filtres compensateurs en matériau équivalent-tissus modifie la dose à la peau. Les auteurs décrivent des méthodes de calcul de la dose peau relative (DPR) et proposent une formule permettant de prévoir la contribution du filtre compensateur équivalent-tissue à la DPR. Ils présentent des résultats expérimentaux relatifs à la DPR pour différentes dimensions de champ et différentes distances de la peau au compensateur.

Zusammenfassung

Die Hautdosis in der Nähe von Ausgleichsfiltern in der Strahlentherapie.

Bei der strahlentherapeutischen Behandlung mit hochenergetischen Photonen beeinflusst die Verwendung von Ausgleichsfiltern die Hautdosis. Es werden Methoden zur Berechnung der relativen Hautodis (RHD) beschrieben und eine Formel zur Vorhersage des Beitrages eines Ausgleichsfilters zur RHD wird hergeleitet. Messungen der RHD für verschiedene Feldgrößen und Abstände vom Ausgleichsfilter wurden durchgeführt und die Ergebnisse werden vorgestellt.

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