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ASSESSMENT OF SCATTERED DOSE CONTRIBUTION TO HEALTHY TISSUE IN RADIATION THERAPY USING WATER PHANTOM

by

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In cancer therapy using gamma radiation one of the limiting factors in dose delivery is the safety of the healthy tissues and organs around the cancerous mass. Better collimation and dose fractionation are employed to achieve this. In the present paper results of scattered dose to healthy tissues around the incident beam cross-section or beam boundaries have been estimated using IAEA standard water phantom and Co-60 incident radiation. It has been observed that scattered dose to healthy tissues increases linearly from 4% to 7% of the incident dose of 185 cGy to 200 cGy at the centre of the beam, at 5 cm depth in water, as we increase the incident beam field size from 5 cm   5 cm to 10 cm   10 cm. Also the maximum unwanted scattered dose for any field size remains closer to the incident beam boundaries.

Key words: scattered dose, cancer therapy, healthy tissue, radiation therapy, water phantom

INTRODUCTION

The malignant growth of human tissue beyond normal limits, commonly known as cancer, has become a very common contemporary disease. Only during the calendar year 2007 more than 7 million people died of cancer in the world [1]. This number is 13% of the total human deaths occurred in the world in 2007. American cancer society estimated 20000 global cancer deaths per day during 2007 [2]. Among the curative efforts to fight cancer, the use of radiation is very common. During treatment of cancer patients, very high radiation doses are delivered to kill the cancerous cells as well as a fraction of healthy cells in the immediate periphery of the tumour mass. However, unwanted radiation dose is inevitably delivered to the healthy tissues of the patients' body during therapeutic process. Statistically driven radiation scattering going on in the cancerous mass gives rise to the scattered radiation dose to the surrounding healthy tissue.

Prevention of healthy tissue from receiving undue radiation dose is one of the fundamental responsibilities of medical physicist during treatment planning. If one has prior estimate of the extent of scattered dose to the healthy tissues, one will be in better position to devise safe and effective treatment planning.

In the present study we have used standard IAEA water filled poly methyl methacrylate (PMMA) phantom to estimate the scattered dose to water around the main beam area. Water is one of the best and easily manageable tissue equivalent material and water filled phantoms are widely used to calibrate therapy level dosimetry systems which are employed in cancer therapy [3, 4]. The scattered dose around the incident beam, from Co-60 therapy level irradiator GWGP-80, was measured at various depths and field sizes in the phantom.

MATERIAL AND METHODS

Collimated radiation beam from Co-60 teletherapy source was taken as the incident gamma radiation. The absorbed radiation dose was measured in water phantom placed in front of radiation beam at a distance of 80 cm from the radiation source – source to surface distance (SSD) as shown in fig. 1. This is the standard distance used for calibration of dosimetry

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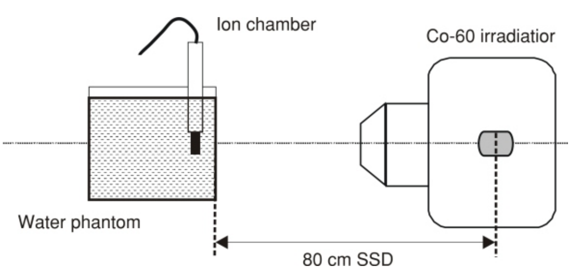

Figure 1. Irradiation geometry for scatter dose assessment with 80 cm SSD

systems in cancer therapy centres. The incident radiation dose of 185 cGy to 200 cGy was chosen since this the most probable dose fraction used in majority of the cases of cancer therapy [5]. The square field sizes of 5 cm × 5 cm, 7 cm × 7 cm, and 10 cm 10 cm were chosen at the entrance surface of the phantom. The absorbed radiation dose was measured using 0.62 cm^3 graphite thimble Farmer type ion chamber hooked up with SuperMax measuring assembly. During normal calibration of dosimetry systems the measurements are made at a standard depth of 5 cm in water below the incident surface of the phantom. Dose measurements in this study were made at depths of 5 cm, 10 cm, 15 cm, and 20 cm in water below the incident surface of the phantom.

The measurements taken along the central line of the incident beam at the above quoted depths represent dose mainly due to primary beam with some contribution from scattered radiation. For the sake of simplicity we assume it as the approximately unscattered dose (hereinafter termed as central dose) to be delivered to a cancerous mass at respective depth against which all other readings are normalized. Measurements along the either side of the incident beam inside the water phantom, as shown schematically in fig. 2, are the contribution due to scattering taking place in the main beam area. The measurement points along either side of the main beam cross-section are at a distance of 2.5 cm, 5 cm, 7.5 cm, and 10 cm from the outer field boundary of the incident beam. The dosimetry system used in these measurements was properly calibrated against secondary standard of SSDL PINSTECH, Pakistan [3, 4]. The system was calibrated in terms of absorbed dose to water by substitution method [6]. The positioning error in this method is assumed as zero since the chamber was fitted in the PMMA inserter sleeve which guided the chamber into its position and it was aligned using laser beam. The calibration factor in dose mode came out to be $N_{d,w}$ = 0.97789 mGy/mGy. The measurements were normalized at reference conditions *i. e.*, temperature of 20 ºC and atmospheric pressure of 1013.25 mbar, according to the following equation

$$D_w \quad X \; k_{pt} \; N_{d,w} \tag{1}$$

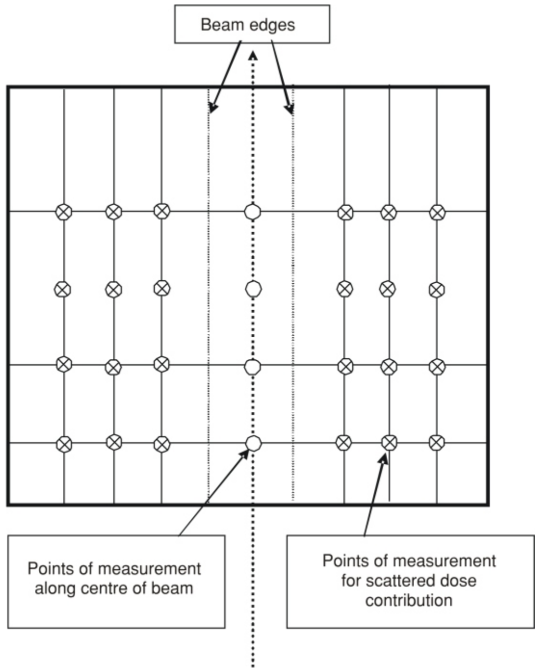

Figure 2. Top view of PMMA water phantom showing beam edges and points of measurement

where D_w is the corrected dose, X – the uncorrected dose, $N_{d,w}$ – the calibration factor, and k_{pt} – the temperature pressure correction factor, which is calculated from the equation

$$k_{pt} \quad \frac{(T_m \quad 273.15) \; P_0}{(T_0 \quad 273.15) \; P_m} \tag{2}$$

where P_0 = 1013.25 mbar and T_0 = 20 ºC are reference values of pressure and temperature, and P_m and T_m are the values of prevailing pressure and temperature measured during the experiment.

RESULT AND DISCUSSION

In our present study we have assumed that the cancerous mass is located at a depth of 5 cm below the surface, therefore, all the measurements of radiation doses along centre of beam line as well as scattered doses on either side of beam cross-section have been normalized with respect to absorbed dose at 5 cm depth. Measurements were made at depths of 5 cm, 10 cm, 15 cm, and 20 cm along central line of the incident gamma beam inside the phantom as well as scattered doses away from the beam edges at each depth. The field size at the entrance surface of the phantom was kept at 5 cm 5 cm. The data of scattered doses have been graphically presented in fig. 3. It can be seen that scattered radiation dose to healthy tissue at a distance of 2.5 cm on either side from the field edge is 4% of the central dose at 5 cm depth (tumour dose) and it reduces to 2% of the

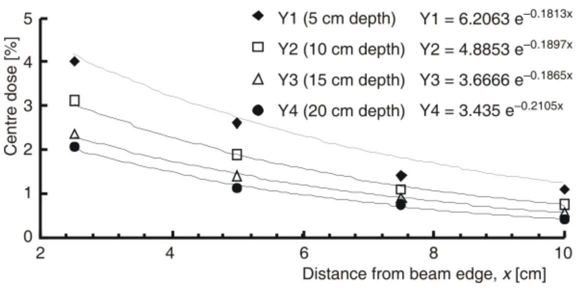

Figure 3. Exponential curve fitted to scattered dose as (percentage of tumour dose at 5 cm depth) and as a function of distance from field edge of incident beam at depths of 5 cm, 10 cm, 15 cm, and 20 cm for field size of 5 cm 5 cm

tumour dose at 20 cm depth. As we move away from the field edge the difference in the scattered dose reduces as we go deeper in the phantom. This is clearly seen when we compare the scattered dose along a line at a distance of 10 cm from beam edge. The scattered dose at 5 cm depth from front wall of phantom along this line is about 1% of the tumour dose and it slowly reduces to 0.4% of tumour dose at a depth of 20 cm in the phantom. This indicates that scattered dose to the healthy tissue in the vicinity of the tumour mass is relatively high for any treatment planning or irradiation geometry. Reducing this scattered dose around the tumour mass will be a defining parameter for safety of healthy tissue. When we compare the first point and last point on the horizontal axis in fig. 3, it is clear that scattered dose contribution is higher near the beam edge and at shallower depths as compared to farther from the beam edge and at deeper positions. It is also clear that appreciable effect of the field size to the scattered doses is only in the immediate vicinity of the tumour (column at 2.5 cm in the histogram in fig. 5).

The same data has been plotted in histogram form for each of the depths namely, 5 cm, 10 cm, 15 cm, and 20 cm in the phantom at various distances from the edge of the beam as shown in fig. 4. This shows that as we

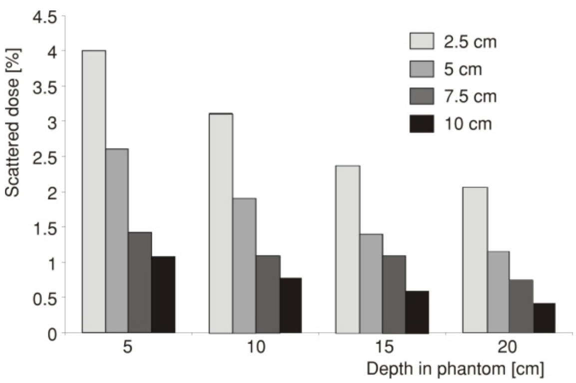

Figure 4. Histogram plot of percent scattered dose for each depth in the phantom at various distances from the edge of the beam for field size of 5 cm 5 cm

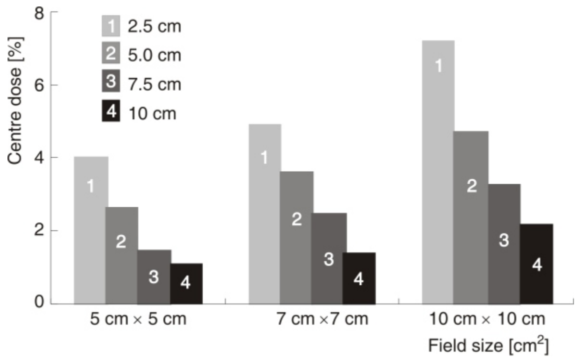

Figure 5. Scattered radiation dose for field sizes 5 cm 5 cm, 7 cm 7 cm, and 10 cm 10 cm at 5 cm depth from front wall of the phantom and for various distances from the beam edge

move away from the centre of the beam line the scattered dose falls almost exponentially. To see this, exponential curve has been fitted to the scattered dose data as shown in fig. 3. Equations of the fitted exponential curves indicate a characteristic attenuation coefficient of percent scattered dose in the phantom which ranges from 0.1813 cm^{-1} to 0.2105 cm^{-1} with average of 0.192 cm^{-1}. This linear attenuation coefficient in water (as well in soft tissue) corresponds to gamma energy in the range of 60 keV to 80 keV [7], which is thought due to Compton scattering inside the main beam.

Table 1 gives scattered radiation dose for field sizes 5 cm 5 cm, 7 cm 7 cm, 10 cm 10 cm. The behaviour of scattered dose at each field size at 5 cm depth as we move away from the beam edge has been plotted in fig. 5. It can be seen that at field size10 cm 10 cm the scattered radiation dose is quite high. This is because available volume of the scattering medium and thus the scattering centres have increased which gives rise to higher scattering doses to the healthy tissue. Therefore, in actual cancer therapy irradiation, the field size should be decreased as much as possible and optimum for the treatment.

When the gamma radiation beam emits from the Co-60 irradiator it undergoes scattering with the material of the beam collimator. This gives rise to well known penumbra region around the main radiation beam. To investigate whether this effect has any contribution to the scattered radiation doses in the phantom and reported in this study, a separate experiment was carried out for measurement of beam penumbra at 80 cm distance from the source in air for a field size of 10 cm 10 cm. This result has been plotted in fig. 6. It can be seen that the beam falls very fast to almost zero within 1 cm on either side of the beam edge. This shows that beam penumbra effect has got no contribution to the scattered radiation doses presented in this study.

As shown in tab. 1, the scattered radiation doses for a field size of 5 cm 5 cm and at a distance of 10 cm on either side of beam edge is 1.08% of tumour dose at

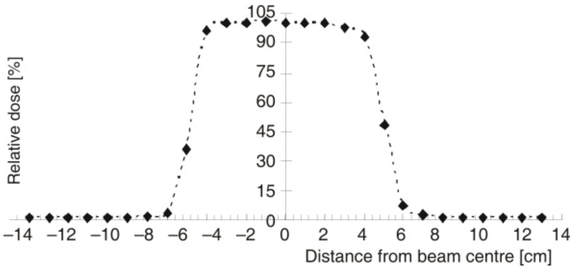

Figure 6. Beam profile for field size 10 cm   10 cm at 80 cm in air to see the extent of beam penumbra

Table 1. Scattered dose as percentage of dose at the centre of beam at 5 cm depth from wall of phantom for three field sizes and at various distances from the beam edges

Chamber position from beam edge	Field size at the entrance surface of phantom					
	5 cm 5 cm		7 cm 7 cm		10 cm 10 cm	
	Mean dose [cGy]	Percent of central dose	Mean dose [cGy]	Percent of central dose	Mean dose [cGy]	Percent of central dose
Centre of beam	186.36	100	185.71	100	201.13	100
2.5 cm	7.46	4	9.06	4.88	14.35	7.13
5.0 cm	4.84	2.6	6.58	3.54	9.43	4.69
7.5 cm	2.65	1.42	4.58	2.46	6.49	3.22
10.0 cm	2.01	1.08	2.52	1.36	4.32	2.15

the centre of beam. In a real situation these scattered radiation dose to left arm (almost 12 to 13 cm away from the beam edge in a breast cancer therapy case) has been 1.8% of the fractionated dose received [8]. This is slightly greater than scattered dose estimated in our case and the difference could be due to increased scattering from human skeleton which contains relatively higher Z material in the form of calcium. However, our value is not very much far away from the real case and gives a good approximation of the actual case. Scattered dose at a distance of 2.5 cm from the beam edge at a depth of 5 cm below the surface for field size of 10 cm   10 cm is 7.13% of the central beam dose. In a similar study, by Uddin *et al*., a scattered radiation dose has been determined for field size 10 cm   10 cm and at a distance of 3 cm from the beam boundary to be equal to about 6.7% of the central beam, which is close to our value [9].

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Инфтихар ХУСАИН, Зиа-ур-РЕХМАН, Первин АХТЕР, Мухамад АХТАР**

ПРОЦЕНА ДОПРИНОСА ДОЗЕ РАСЕЈАНОГ ЗРАЧЕЊА ЗДРАВОМ ТКИВУ У РАДИЈАЦИОНОЈ ТЕРАПИЈИ КОРИШЋЕЊЕМ ВОДЕНОГ ФАНТОМА

Један од ограничавајућих фактора при прописивању дозе у лечењу рака је сигурност здравих ткива и органа око канцерогене масе, те се ради тога врши боља колимација снопа и фракционализација дозе. У овом раду, коришћењем стандардног МААЕ воденог фантома и зрачења из Co-60, процењена је доза расејаног зрачења предата здравом ткиву око граница инцидентног снопа. Уколико се површина инцидентног снопа увећа од 5 cm 5 cm на 10 cm 10 cm, уочено је да се доза расејаног зрачења на 5 cm у води увећава линеарно од 4% до 7% вредности инцидентне дозе, која у центру снопа износи од 185 cGy до 200 cGy. Такође, максимум непожељне дозе расејаног зрачења за ма коју површину снопа остаје у близини граница инцидентног снопа.

Кључне речи: доза расејаног зрачења, терапија канцера, здраво ткиво, терапија зрачењем, водени фантом