Measurements of X-Ray Dose Delivered at Different Depths Applied on Water Phantom

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Abstract: ***Background**: Radiotherapy, is therapy using ionizing radiation in order to deliver an optimal dose of either particulate or electromagnetic radiation to a particular area of the body with minimal damage to normal tissues. The source of radiation may be outside the body of the patient (external beam irradiation) or it may be an isotope that has been implanted or instilled into abnormal tissue or a body cavity. Called also radiotherapy. The aim of work studies the relationship between the depth dose and the high photon x-ray energies (6MeV and 10MeV). **Patients and methods**: in our work, we studied the dose distribution in water phantom given at different depths (zero-18) cm deep at1cm intervals treated with different field size (5×5-,10×10,15×15,20×20) cm^2. **Results** show that the increasing the field size, the percentage of surface dose increases that this could be caused by an increase of the amount of scattering in the larger fields and on the other hand with increased energy, the percentage of surface dose will be reduced due to the reduced backscattered. **Conclusion**: Increases field size increasing surface dose and this is attributed to the scatter radiation. Higher energy has higher depth dose at higher depth.***

Keywords: high photon energy, water phantom, percentage depth dose

1. Introduction

Radiation therapy or radiotherapy, is therapy using ionizing radiation, generally as part of cancer treatment to control or kill malignant cells and normally delivered by a linear accelerator(1). Linear accelerator ("linacs") which produce megavoltage X-rays. The first use of a linac for medical radiotherapy was in 1953. Medical linacs produce X-rays and electrons with an energy range from 4 MeV up to around 25 MeV (2).

X-ray beams are used to deposit absorbed dose at depth within a patient at the site of the tumor. High energy X-rays produce a skin sparing effect whereby more dose is deposited at depth than in the skin tissue region (3).

The percentage depth-dose of diagnostic X-rays are important in evaluating patient dose from medical exposure (4). In radiotherapy, quality of a radiation beam is most usefully expressed in terms of its penetrating power, which is a function mainly of the mean photon energy, and may be fully described by its depth dose characteristics in water (5) but an increase in surface dose with field size is also noted due to electron scattering from intervening materials (6). Absorbed dose in the body is dependent on depth, field size, photon energy and Source to surface distance (SSD) (7).

2. Materials and Methods

This work was carried out in the Oncology Teaching Hospital, Medical City/ Baghdad, in the period between February and July (2017). Measurements were made Elekta linear accelerator, 6MeV and 10 MeV X-ray beams and a fixed SSD of 100 cm. Water phantom is usually used for measuring basic dose distribution, because it is very close to human soft tissue due to its similar density, atomic number and number of electrons per gram and universally available with reproducible radiation properties. The water tank (phantom) is large enough to allow full photon scatter. The applicator must then be positioned just at the surface of the water with the central axis precisely over the ionization chamber's sensitive volume. The data of different field size [(5×5)-(20×20) cm^2],for (6MeV and 10MeV) energies were analyzed the average decrease in dose with depth, same source-to-surface distance (SSD).

3. Results

Depth dose 6MeV (water phantom)
A selected square field sizes of; (5×5,10×10,15×15, 20×20) cm^2 were taken for this study. The depth dose was measures between (0-18)cm deep at1cm intervals. The maximum depth dose for 6 MeV was 1.5cm, table (1).

Table 1: Measured percentage depth dose (PDD) for 6 MeV (water phantom)

	Square field size (cm^2)			
Depth (cm)	(5×5)	(10×10)	(15×15)	(20×20)
0	42.43	47.62	50.36	54.75
1	95.35	96.03	96.56	97.41
1.5	100.0	100.0	100.0	100.0
2	99.63	99.12	99.18	99.21
3	95.33	95.30	95.67	95.70
4	90.65	91.16	91.79	92.10
5	86.04	87.11	88.06	88.41
6	81.43	83.19	84.23	84.77
7	77.06	79.16	80.44	81.24
8	72.81	75.42	76.96	77.72
9	68.80	71.71	73.47	74.40
10	65.18	68.26	70.06	71.20
11	61.38	64.90	66.80	68.07
12	57.96	61.55	63.60	64.91
13	54.77	58.44	60.60	62.01
14	51.83	55.53	57.75	59.24
15	48.95	52.72	54.96	56.60
16	46.25	49.98	52.32	53.88
17	43.75	47.46	49.83	51.47
18	41.31	45.07	47.44	49.01

The change of depth dose with different field size and for different depths was shown in Fig. (1). It shows a slight increase in dose with the increase in field size for the same depth and the deeper the more dose difference, this may be attributed to the increased scatter radiation reaching the dosimeter. The decrease in PDD with depth for all field sizes, is related to progressive attenuation with depth and the inverse square law. Table (1)

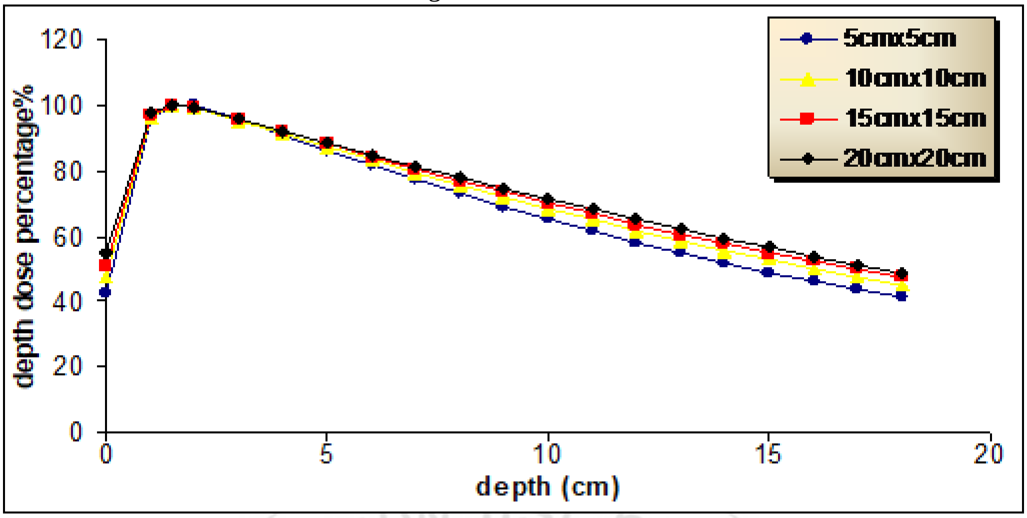

Figure 1: The change of 6MeV, depth dose with a different field size

Depth dose 10MeV (water phantom)
A similar results were observed for 10MeV in the relation to the decrease in the depth dose with increasing depth and a slight increase with the increase in the field size for the same depth, an important observation is that the maximum depth is 2.5cm deeper than 6MeV which is 1.5cm this is attributed to the increased dose build up depth because of increased mean free path for the higher energy, table (2).

Table 2: Measured percentage depth dose (PDD) for 10 MeV (water phantom)

	Square field size (cm^2)			
Depth (cm)	(5×5)	(10×10)	(15×15)	(20×20)
0	29.95	38.01	40.88	46.47
1	85.11	89.13	90.13	92.32
2	99.53	99.85	99.83	99.97
2.5	100.0	100.0	100.0	100.0
3	99.01	98.40	98.32	97.76
4	95.10	94.90	94.76	94.45
5	90.94	91.08	91.26	90.99
6	86.70	87.36	87.70	87.57
7	82.66	83.68	84.25	84.27
8	78.79	80.13	80.85	80.99
9	75.08	76.69	77.64	77.86
10	71.48	73.37	74.46	74.72
11	68.08	70.18	71.47	71.83
12	64.81	67.01	68.35	68.95
13	61.62	64.15	65.53	66.22
14	58.69	61.28	62.78	63.49
15	55.80	58.58	60.11	60.93
16	53.21	55.93	57.58	58.40
17	50.73	53.47	55.16	56.12
18	48.30	51.06	52.82	53.70

The relation between depth and PDD is also similar to the 6MeV in relation to the decrease PDD with depth and an increase in PDD with field size for the same depth, but the increase is less then what we have seen for 6MeV Fig.(2), and table (2). Probably because of more penetration power and forward scatter for the 10MeV energy. Apart from the PDD before the maximum PDD which is expected for the higher energy with higher penetration power, an increase in the total dose at all depths was also observed this is also related to the increase in the penetration power of the beam leading to deliver more dose at depth and to the increase in x-ray intensity with the increase in energy.

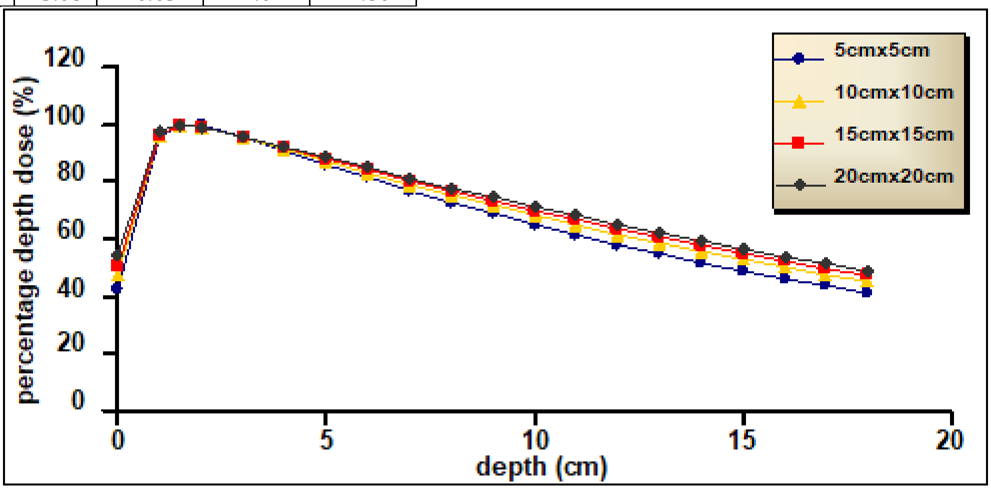

Figure 2: The change of 10MeV, depth dose with a different field size.

Table 3: Surface doses of 6MeV and 10 MeV photon beams for different field sizes

square field size (cm^2)	6 MeV (%)	10 MeV (%)
5×5	42.43	29.95
10×10	47.62	38.01
15×15	50.36	40.88
20×20	54.75	46.47

In figure (3) show that when the energy increase the dose surface decrease because the contribution of retro-diffused electrons decrease.

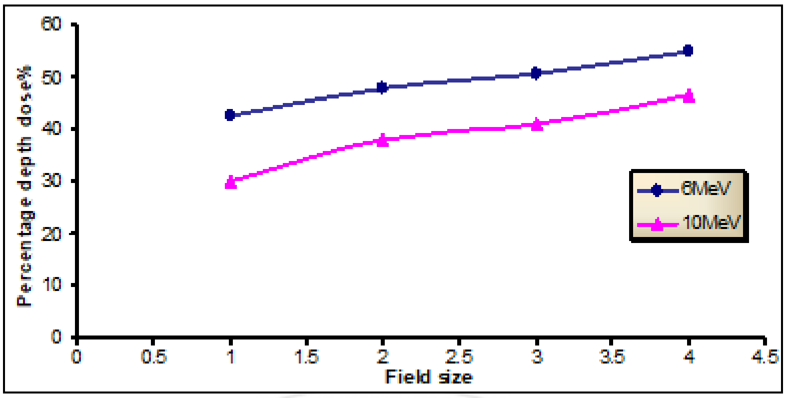

Figure 3: The change of surface dose percent with for different field size

4. Discussion

Attenuation of a photon beam by an absorbing material is caused by five major types of interactions. This reaction between photon and nucleus is only important at very high photon energies (> 10 MeV). The other four processes are coherent scattering, the photoelectric effect, the Compton effect, and the pair production. Each of these processes can be represented by its own attenuation coefficient, which varies in its particular way with the energy of the photon and with the atomic number of the absorbing material (8).

High energy medical linear accelerators are used for the treatment of cancer in radiotherapy. X-ray beams are used to deposit absorbed dose at depth within a patient at the site of the tumor (9).

As the human tissue has low atomic number the photoelectric interaction will not happen only at low radiation energy as the photoelectric interaction directly proportional with $Z^3$and inversely proportional with the radiation energy i.e. with $(1/hf)^3$. for this reason this type of interaction not found in the therapeutic range of radiation.

In higher energy Compton interaction increases, compared to photoelectric absorption. The probability of Compton interaction also depends on the electron density (number of electrons/g × density), With the exception of hydrogen, the total number of electrons/g is fairly constant in tissue; thus, the probability of Compton scattering per unit mass is nearly independent of Z, and the probability of Compton scattering per unit volume is approximately proportional to the density of the material. Compared to other elements, the absence of neutrons in the hydrogen atom results in an approximate doubling of electron density. Thus, hydrogenous materials have a higher probability of Compton scattering than anhydrogenous material of equal mass (10). The pair production interaction threshold it is still unimportant as it needs higher nuclear charge , in general pair production is not important only at high MeV about 30MeV photons(11).

From figure (1 and 2) above, as the energy increases from 6 MeV to 10 MeV, the PDD value also increases. This is due to the effect of secondary electrons begins to build up at the surface and the number of secondary electrons passing through the medium gradually increases with depth, that leads to increase in the deposited energy (the dose) up to a distance from the surface approximately equal to the range of the secondary electrons. In case of electron beams, the stream of ionizing particles (i.e. electrons) is already produced (by a Linac) so, they deposit their energy starting from the surface.

In the figure (3), show that when the energy increase the dose surface decrease because the photons deposit their energy through the secondary electrons and there is a dose build-up region before d_{max}. Perhaps one may also examine the mean free path of photons. The mean free path of photons is the average distance that they will travel before interactions. Only the very low energy photons will be likely to deposit the energy near the surface. The higher energy ones will not.

References

[1] Bomford C.K., Kunkler I.H., Walter J. ,2003Textbook of Radiotherapy . 6th ed. Walter and Miller's.
[2] David I. T., John B. T., 2006Back to the future: the history and development of the clinical linear accelerator. Phys. Med. Biol. 51 , R343–R362.
[3] Butson, M., Mathur, J., Perez, M., Metcalfe, P., 1996. 6 MV X-ray dose in the build up region: empirical model and the incident angle . Aust. Phys. Eng. Sci. 19, 74–82.
[4] Kato H., Fujii S., Suzuki S. ,2004. Method of calculating percentage depth dose for diagnostic X-rays. Nippon Hoshasen Gijutsu Gakkai Zasshi .60,1107–15.
[5] Jordan T.J., 1996. Megavoltage x-ray beams: 2–50 MV. Br J Rariol Suppl. 25,62–109.
[6] Ravikumar M., Ravichandran R., 2002. Dose measurements in the build-up region for the photon beams from Clinac-1800 dual energy medical linear accelerator. Strahlenther Onkol. 176,223–8.

[7] Saeed A.B., Muhammad A. R., Aalia N., 2009. AN ANALYSIS OF DEPTH DOSE CHARACTERISTICS OF PHOTON IN WATER. J Ayub Med Coll Abbottabad. 21(4), 41-45
[8] Faiz M. K., 1984. Physics of radiation therapy. Williams and Wilkins.
[9] Sixel K., Podgorsak E., 1994. Build up region and depth of dose maximum of megavoltage X-ray beams. Med. Phys.21 (3), 411–416.
[10] Jerrold T.B., Anthony J.S., Edwin M.L.,et al., 2012The Essential Physics of Medical Imaging, Third Edition. Lippincott Williams and Wilkins.
[11] Trauernicht C. J., " Characterization of a 6 MV photon beam in terms of primary and scattered dose components, (Ms.C) Thesis, University of Cape Town, (2009).