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An Assessment to Determine Total Scatter Factors for Photon Beam Linear Accelerators

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Abstract

Objectives: To explore a simple and effective way to determine the total scatter of a C- Series linear accelerator.

Methodology: Measurements for this study were acquired using a Varian C-Series linear accelerator, with a 6MV photon beam, a Blue water phantom, 2 IBA CC13 ion chambers and an IBA CCU electrometer. Measurements were acquired for field sizes ranging from 5cm x 5cm to 40cm x 40cm, increasing field size by an increment of 5cm. Three readings were collected for each field size and averaged. All readings were done at a depth of 10cm to reduce the probability of measuring electron contamination in the photon beam. Measurements for Sc calculation were acquired in air using a build-up cap with a 3cm diameter.

Results: The average radiation dose measured increased as the field sizes increased. The maximum dose recorded for Sc was 2.33cGy while that for Scp was7.96cGy. There was a non-linear direct relationship between radiation dose measured, Scp and Sc calculated and field size. The maximum standard deviation in charge readings for Sc measurement was 1.18% which was recorded for the 35cm x 35cm field. The minimum standard deviation was 0.70% obtained with the 20cm x 20cm field. the proportion of the phantom scatter contribution to the total scatter, decreased exponentially with increases in field size. The largest contribution was identified with the 10cm x 10cm field while the lowest was identified with the 40cm x 40cm field.

Conclusion: It can be concluded that this method was effective in assessing the total scatter factor and its derivatives for field sizes ranging from 5cm x 5cm to 40cm x 40cm.

Keywords Collimator Scatter; Phantom Scatter; Beam Fluence; Percentage Depth Dose (PDD); Dmax; Planning Target Volume (PTV); Gross Tumor Volume (GTV)

Introduction

Modern medical linear accelerators can be operated in two modes; expelling either electrons or photons for cancer treatment. While treating with photons, contaminants may be created through a photonuclear reaction from elements in the head of the gantry and other hardware such as the collimators, filters and target. In photon mode, high energy photons possess energy exceeding the threshold of the photonuclear reaction of elements such as lead (used for shielding in the head of the gantry). A variety of photon energies are employed in linear accelerators for cancer treatment, which may enhance the liberation of electrons which are undesirable as they contaminate the treatment beam, contributing to the patients' skin dose. Optimal tumour control with limited side effects requires delivering the maximum prescription dose to the Gross Tumor Volume (GTV) while simultaneously reducing the dose to the surrounding structures. To account for target movement, a Clinical Target Volume (CTV) is created by adding a margin around the GTV. Systematic uncertainties are accounted for by the Planning Target Volume (PTV) which is a 5mm margin around the CTV [1].

The quality of radiation therapy delivery as a direct impact on the dose delivered to the patient. This takes into account measures to alleviate failures, and being keen on dosimetric guidelines as they influence the clinical outcome. Treatment verification aims to measure and assess the accuracy of the radiation expelled during treatment and the fluence of dose distribution. These dose verification checks are acquired through the delivery of hypothetical or actual patient treatment to phantoms. Heterogeneous and homogeneous phantoms coupled with ion chambers have been employed to monitor dose delivery for multi-field plans, and Thermo-luminescent dosimeters (TLDs) and ion chambers can also be used to achieve point dose measurements in areas of low dose gradient complementing relative dose measurements with other devices (Figure 1a,b) [2].

It is imperative to measure organ dose in radiotherapy, however, due to practical difficulties in vivo measurements on patients are limited. To overcome this issue, anthropomorphic phantoms are employed. Due to the significance of the Photoelectric and Compton effects, consideration has to be given to the effective atomic number and electron density of the material used to construct the phantoms. Ideally, the materials used should have the same density, electron density, effective atomic number, and tissue inhomogeneity of the human body.

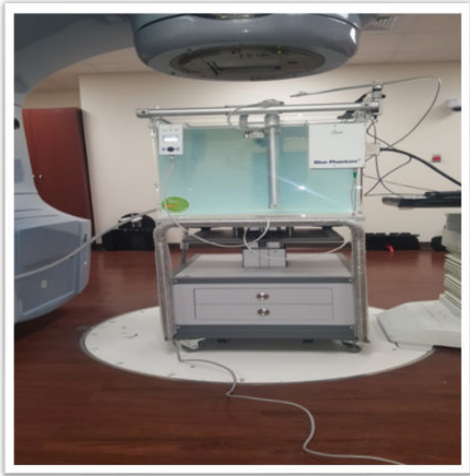

Figure 1a: Showing the set-up of phantom and ion chambers

Figure 1b: Showing set-up of phantom and ion chambers.

The phantom should possess tissue-equivalent materials such as bones, muscles and organs while simulating the physical shape of the body. The spatial distribution of phantom materials should also simulate that of the human body and be reproducible to enable accurate dosimetric acquisitions. Further, the design ought to avoid materials such as metals or other elements that will affect imaging or treatment of the phantom. It is import to verify the accuracy of the heterogeneity corrected dose calculated from the treatment planning system. The algorithm used should be verified in a clinically representative manner before clinical use, where consistency becomes paramount. This process is conducted through a dose-response curve employed to convert TLD/ion chamber response into absorbed dose after irradiation, using a linear trend line equation.

The determination of the monitor units (MUs) for treatment delivery of the prescribed dose is heavily dependent on the total scatter factor of the treatment system. The total scatter factor of the system is a measure of the scatter contributed from the collimator leaves (Sc) also known as the output factor. This is the ratio of the output in air for a given field to that for the reference field, with a build-up cap to maintain charged particle equilibrium. The other deciding factor is the phantom scatter (Sp), which is the ratio of absorbed dose at depth of maximum dose (Dmax) for a given field to that in the reference field at the same depth and collimator opening [3,4]. The Dmax for the 6MV photon beam utilized for these measurement was 1.5cm. Measurements with a phantom have scattering effects from both the phantom and collimators; therefore, the collimator opening should remain constant during measurements to eliminate the effects of the collimator. This paper seeks to explore a simple and effective way to determine the total scatter of a C- Series linear accelerator, as a precursor to converting TLD readouts to absorb dose in External Beam Radiation Therapy (EBRT) [5,6].

Methodology

Measurements for this study were acquired using a Varian C-Series linear accelerator, with a 6MV photon beam. A Blue Water Phantom with dimensions (65cm x 60cm x 54cm) was employed and charge was measured using 2 IBA CC13 ion chambers (inner diameter 6 mm and active volume 0.13cm3), for field and reference readings respectively. The ion chambers were connected to an IBA CCU electrometer (where correction factors were applied for temperature and pressure readings), and measurements were acquired for field sizes ranging from 5cm x 5cm to 40cm x 40cm, increasing field size by an increment of 5cm. Three readings were collected for each field size and averaged, after which the standard deviation and maximum deviation were calculated. Due care was exercised to ensure the buildup cap was completely covered when taking the 5cm x 5cm field measurements. However, measurements needed to calculate the Scp were acquired using the Blue Water Phantom without the buildup cap [7,8].

Results and Discussion

A correction factor was used to account for the variations in the temperature and pressure of the assessment site to that of standard atmospheric pressure and room temperature. As seen in table 1, the average radiation dose measured increased as the field sizes increased. This increase was very subtle when measurements were obtained in air (Sc).

Temperature	23.5oC
Pressure	100.75 kPa
Electrometer Bias	300V
Beam Energy	6MV
Dose Rate	600 mu/min

Table 1: Showing set up parameters for measurements

The maximum dose recorded here was 2.33cGy. This subtle increase can be attributed to the photon beam interacting with the surface of the collimator leaves as they conform the photon beam. As a result, there was a direct relationship between both variables. This relationship, however, was non-linear as the increases in radiation dose were not progressing consistently. When measurements were obtained with the phantom (Sp) the radiation dose measured were two to three times more than measurements for Sc.

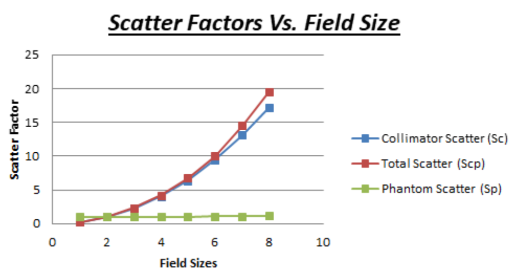

Figure 2: Showing scatter factors vs. field size

The maximum radiation dose obtained for Scp was 7.96cGy. This is a direct result of the photon beam interacting with a larger and larger volume of the phantom, thus creating more charge in the active volume of the ion chamber. A direct relationship between the radiation dose and field size was also discovered here and similarly the relationship was non-linear [9,10].

	Average Radiation Dose measured (cGy)	
Field Size (cm)	Collimator Scatter (Sc)	Total Scatter (Scp)
5 x 5	2.07	5.9
10 x 10	2.15	6.61
15 x 15	2.18	7.03
20 x 20	2.18	7.31
25 x 25	2.22	7.54
30 x 30	2.25	7.69
35 x 35	2.31	7.84
40 x 40	2.33	7.96

Table 2: Showing the mean radiation dose measured at each field size for both variables

Field Size (cm)	Collimator Scatter (Sc)	Standard Deviation	Total Scatter (Scp)	Standard Deviation	Phantom Scatter (Sp)
5 x5	0.24	0.73%	0.26	0%	0.06
10 x 10	1	1.12%	1	0%	1
15 x 15	2.28	0.80%	2.32	0%	1.0176
20 x 20	4.05	0.70%	4.21	0%	1.0395
25 x 25	6.4	0.74%	6.72	0%	1.05
30 x 30	9.44	0.78%	10.01	0%	1.0603
35 x 35	13.15	1.18%	14.53	0%	1.1049
40 x 40	17.17	0.81%	19.51	0%	1.1363

Table 3: Showing the calculated Sc, Scp with corresponding standard deviations and Sp values.

The total scatter and its derivatives were calculated using the following equations(4):

$$Sp \approx \frac{Scp}{Sc} \qquad (1)$$

$$Sp(L)=\frac{Scp(L)\text{ given field}}{Sc(L)\text{ reference field}} \qquad (2)$$

$$Sc(L)=\frac{\frac{{}_{cap}R(c_{L,100},L)}{M}}{\frac{{}_{cap}R(c_{0,100},L_0)}{M}} \qquad (3)$$

$$Sp(L)=\frac{{}_{d}R(c_{L,100},L)}{R} \qquad (4)$$

Where: R-Ion chamber readings (charge); M- Monitor units (mu); L- Field dimensions; L0- Reference Field; 100- SSD; C- Collimator opening; PDD-Percentage depth dose

The maximum standard deviation in charge readings for Sc measurement was 1.18% which was recorded for the 35cm x 35cm field. The minimum standard deviation was 0.70% obtained with the 20cm x 20cm field. It was discovered that at field sizes 10cm x 10cm and 5cm x 5cm the Sc and the Scp values were the same which equate to an Sp value of 1. The Sc and Scp values all increased exponentially with increases in field size. The measurements obtained for field sizes above 10cm x 10cm had a maximum deviation of 2.34 between Sc and Scp. The minimum deviation was 0.04 obtained with the 15cm x 15cm field. As shown in figure 2 the proportion of the phantom scatter contribution to the total scatter, decreased exponentially with increases in field size. The largest contribution was identified with the 10cm x 10cm field while the lowest was identified with the 40cm x 40cm field. It can be concluded that this method was effective in assessing the total scatter factor and its derivatives for field sizes ranging from 5cm x 5cm to 40cm x 40cm [11,12].

Summary

The total scatter factor of the system is a measure of the scatter contributed from the collimator leaves (Sc) also known as the output factor. This is the ratio of the output in air for a given field to that for the reference field, with a build-up cap to provide electronic stability. The other deciding factor is the phantom scatter (Sp), which is the ratio of absorbed dose at depth of maximum dose (Dmax) for a given field to that in the reference field at the same depth and collimator opening. This paper seeks to explore a simple and effective way to determine the total scatter of a C- Series linear accelerator.

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Authors Contribution

Barrington Brevitt envisaged paper, conducted data collection and analysis, prepared manuscript and approved the final version for submission. Barnswell Dennis conducted data collection and analysis and approved the final version of the manuscript. Dr Pankaj Patel and Professor Mitko Voutchkov participated in data analysis and interpretation, study design and revision of manuscript and approval of final version. The authors declare that there is no conflict of interest.

References

1. Podgorsak EB (2005) Radiation oncology physics. Vienna: International Atomic Energy Agency 123-271.
2. Bjärngard B (1980) Thermoluminescence Dosimetry in the μGy Range, by P. Spanne. Medical Physics 7: 267-267.
3. Khan F M (2010). The Physics of Radiation Therapy. 4th edn. Lippincott, Williams & Wilkins.
4. Birgani MJT, Chegeni N, Behrooz MA, BagheriM, Danyaei A, et al. (2017) An analytical method to calculate phantom scatter factor for photon beam accelerators. Electronic physician 9: 3523.
5. Griffith RL, Levi MR, Cartano D. (2014) The Photoelectric Effect.
6. Bushong SC (2013) Radiologic science for technologists-E-book: physics, biology, and protection. 10th edn. Elsevier Health Sciences.
7. Systems VM. Admin and Physics. Varian Medical System; 2018. p. 1028.
8. Bedford JL, Childs PJ, Warrington AP (2004) Verification of inverse planning and treatment delivery for segmental IMRT. J Appl Clin Med Phys 5:1–17.
9. Bedford JL, Warrington AP (2009). Commissioning of Volumetric Modulated Arc Therapy (VMAT). Int J Radiat Oncol Biol Phys 73: 537-545.
10. Ezzell GA, Galvin JM, Low D, Palta JR, Rosen I, et al. (2003) Guidance Document on Delivery, Treatment Planning & Clinical Implementation of IMRT: Report of the IMRT subcommittee of the AAPM Radiation Therapy Committee. Med Phys 30: 2089-2115.
11. Hansen VN, Evans PM, Budgell GJ, Mott JH, Williams PC, et al.(1998). Quality Assurance of the Dose Delivered by Small Radiation Segments. Phys Med Biol 43: 2665-2675.
12. Low DA, Harms WB, Mutic S, Purdy JA (1998). A Technique for the Quantitative Evaluation of Dose Distributions. Med. Phys 25: 656-661.