



Dosimetric Effect of the Gantry Rotations of a Novel Trunk Phantom Using an Area Integration Algorithm

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Background: Treatment planning systems (TPSs) have proved to be a useful tool in predetermining how a treatment outcome will be in radiotherapy. The accuracy of any TPS to calculate dose to any arbitrary point within a material is largely dependent on the mathematical algorithm used. **Aims:** The purpose of this study was to design a local trunk phantom and use the phantom to check the percentage dose accuracy of the Area Integration Algorithm of a Precise PLAN 2.16 TPS if it is in agreement with results obtained from manufacturer's verification by varying the gantry angle and whether it is within $\pm 5\%$ International Commission on Radiation Units and Measurements (ICRU) minimal limit. **Materials and Methods:** The study was executed with a locally designed phantom made of Plexiglas with six insert and an ionization chamber port. The phantom was simulated using a HiSpeed NX/i computed tomography scanner and Precise PLAN 2.16 TPS for application of beam setup parameters. The mimicked organs for the inserts were: 25%–75% Glycerol-Water for liver, pure carboxyl methyl cellulose was used for lungs, 30%–70% Glycerol-Water for muscle, 40%–60% Glycerol-Water was used for adipose, pure Sodium hypochlorite was used for bone and pure sodium laureth sulfate (Texapon) for kidney. **Results:** The maximum percentage (%) deviation with a large field for six inhomogeneous inserts and with bone only homogeneous inserts were 3.4% and 2.9%, respectively. The maximum % deviation with a small field for six inhomogeneous inserts was 3.2%. The % deviation between the solid water phantom and the locally designed phantom was 3.5%. **Conclusion:** The Area Integration Algorithm has shown an overall accuracy of 4% below 5% ICRU minimal limit. There was no statistically significant difference in field sizes and in inhomogeneity/homogeneity, respectively. Variation exists in % deviation for small field size with parallel opposed field between our verification and the manufacturers.

Key words: Phantom, treatment planning system, area integration algorithm, irregular field program, computed tomography

INTRODUCTION

The International Commission on Radiation Units and Measurements (ICRU) has recommended that radiation dose must be delivered to within $\pm 5\%$ of the prescribed dose. Although ICRU report 24 also states that recommended uncertainty in the delivered dose to a phantom (mimicked human subject) at “optimal model” should be $\pm 2.5\%$ and at “minimal” or “lowest acceptable” model be $\pm 5\%$ and International Atomic Energy Agency (IAEA) dosimetry limit be $\pm 3.0\%$.¹⁻³ When

commissioning treatment planning dose calculation algorithms in radiotherapy, the aim is often to achieve good agreement between calculated and measured doses within 1%–2% for open and wedge (block or compensators) fields in water. While this is possible to achieve using both measurement- and model-based algorithms in water phantoms, such an agreement is usually not possible for measurement-based algorithms in

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phantoms with heterogeneities. This can be explained based on the fact that measurement-based models are able to account for the effect of tissue inhomogeneities on the primary radiation. However, correcting for the scatter radiation is difficult since it depends on field size, beam energy and shape, location, and density of the inhomogeneities.⁴ In contrast, model-based algorithms can account for the effect of tissue inhomogeneities on the scatter radiation using the density scaling method or other approaches.⁵⁻⁹

Several technique of carrying out the quality assurance of treatment planning system (TPS) has been proposed by the various authors.¹⁰⁻¹⁴ Likewise, the reduction of errors and uncertainties in the dose calculation plays an important role in the success of a treatment procedure. The performance and quality of any TPS is dependent on the type of algorithm used.¹⁵⁻²²

Treatment planning requires the ability to calculate dose to any arbitrary point, within the patient, for any beam orientation. The Area Integration Algorithm (Irregular Field program) is well suited for this type of general problem. Patient tissue inhomogeneities, beam blocks, and beam compensators are included in the calculation model. The Area Integration Algorithm (Irregular Field program) requires the separation of the dose into primary and scatter components. The primary component is usually computed, and it includes the transmission through any blocks and blocking trays, beam compensators, and patient inhomogeneities. The scatter component is usually computed, and it includes the presence of blocks, beam compensators, and curvature of the patient but not patient inhomogeneities.²³⁻²⁵

The concept of this dosimetry of Area Integration Algorithm (Irregular Field program) involves the use of tissue-maximum ratio and scatter-maximum ratio which is analogous to tissue-air ratio (TAR) and scatter-air ratio concepts. The underlining program equation of the Area Integration (Irregular Field program) which is similar to external beam program is as follows:

$$\text{Dose Rate} = \text{TRAY TRAY2 OUTPUT FSC} \frac{(P \text{ OCR } QF \text{ TAR0} + SC)}{\frac{(SSD + DMAX + c)^2}{X^2 + Y^2 + (SPD + c)^2}} \quad (1)$$

Where:

TRAY and TRAY2 = are the tray factors.

OUTPUT = the output factor normalized to a (0 × 0) field size at a distance source to axis distance + dose at maximum (SAD + Dmax).

FSC = the airfield size correction dependency factor, computed for equivalent square of the collimator opening

SSD = source to surface distance

Dmax = Dose at maximum

SPD = is the source to point distance of the point of calculation

X and Y = coordinates at a depth of the point of calculation

c = is the correction for the virtual location of the source.

c is found from a plot of the inverse of the square root of measured dose versus distance from the source.

QF = is the off-axis beam quality factor

OCR = is the in air off central axis ratio value

TAR0 = is the zero (0 × 0) field TAR for the slant depth

SC = is defined as scatter contribution computed from the field size and block contours at the level of the point of calculation.

P = is the value of the penumbra, calculated by Wilkinson Source Model

Note: The above program equation does not support the use of wedge since wedge field was not used in this study.

This study will focus on verifying the percentage dose accuracy of the Area Integration Algorithm using both homogeneous and inhomogeneous insert by varying gantry angles for given field sizes of 5 cm × 5 cm and 22 cm × 25 cm, respectively.

MATERIALS AND METHODS

The Plexiglas used was purchased from a local plastic shop and was machined together with a collaborative effect from the Department of Radiation Biology and Radiotherapy and the Department of Biomedical Engineering from College of Medicine of the University of Lagos. The glass dimension was 4 by 8 feet, a part of which was cut using a plastic cutter into six slabs each of dimension 30 cm × 30 cm. A plastic-based hardener (allplast) was used for holding one slab to another to form a cube. Seven holes were drilled on one face. Each drilled hole had a diameter of 2.5 cm gummed together using plastic-based hardener called “allplast.” Before the holes were drilled, the distance from the surface of the designed in-house phantom to the ionization chamber was 15 cm, while the distance between two diagonal insert were approximately 22 cm. The distances from one insert to the other (horizontally) was 7 cm and vertically were 18 cm. Furthermore, additional drilled hole was made at the top of the locally designed phantom to allow for easy filling of water and evacuation of water from the phantom. After these holes and distance have been marked out, another cylindrical rod made of Plexiglas material of thickness 0.2 mm, length 14.3 cm, and diameter 2.5 cm were fitted into the seven drilled holes and were held together at the tip by the “allplast” gum to avoid leakage [Figure 1]. Furthermore, the locally designed phantom was loaded with tissue-equivalent

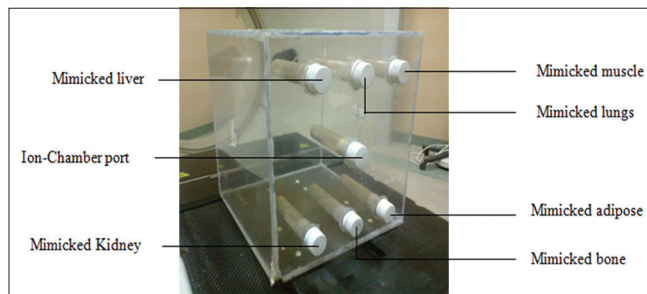


Figure 1: Locally designed phantom with mimicked inserts

material putting into consideration the attenuation coefficients, electron densities, and the effective atomic numbers of each chemical composition. The reason for using chemical component with our local phantom was because of lack of availability of commercial readymade phantom [Table 1]. The locally designed phantom was scanned under a Hi-Speed NX/i computed tomography (CT)-scanner. Slices of images were acquired for six different tissue-equivalent materials. The scanned images were transferred to the precise PLAN Release 2.16 TPS for beam setup. The CT number (is also expressed in a standardized and convenient form as the Hounsfield unit [HU]) for the six different tissue equivalent materials was determined from the CT monitor [Table 2]. Images were transported through the Digital Imaging and Communications in Medicine (DICOM) to the Precise PLAN Release 2.16 TPS where twelve-field technique denoted as Beam (BM) 1–BM 12 were used with large field sizes covering the six inserts with no wedge used. The gantry angles in degree were: 0°, 22.5°, 45°, 90°, 135°, 157.5°, 180°, 197.5°, 215°, 270°, 315°, and 337.5°. The total dose for the twelve fields was 100 cGy, and the total monitor unit (MU) was 100 MU. The type of beam used was “simple.” The photon energy used was 6 MV, SAD was 100 cm and SSD was approximately 84 cm. Collimator angle was 0°, the diaphragm upper SAD was approximately 25 cm and lower SAD was 22 cm, giving at total area diaphragm size of 22 cm × 25 cm. Under modifiers, tray factor was 1, and no multileaf collimator (MLC) was present.

A second scan was carried using the same protocol using large field sizes, but the insert was only bone equivalent material (assumed homogeneous medium). The acquired images from the CT simulator were also transferred to the Precise PLAN 2.16 TPS through a DICOM. Six-field technique was used denoted as BM 1–BM 6 covering the six inserts which were uniformly homogeneous with no wedge used. The gantry angles were: 0°, 45°, 90°, 180°, 225°, and 270°, respectively. The total dose for the six fields was 100 cGy, and total MU was 100 MU. The type of beam used was “simple.” The photon energy used was 6 MV; SAD was 100 cm and SSD was approximately 84 cm collimator angle

Table 1: Equivalent tissue materials, elemental composition with their densities in grams per cubic centimetres

Tissue equivalent material	Elemental composition	Density (g/cm ³)
Liver	25% glycerol: 75% water	1.06
Lungs	Pure CMC	1.60
Muscle	30% glycerol: 70% water	1.07
Adipose	40% glycerol: 60% water	1.09
Bone	Pure sodium hypochlorite (soda bleach)	2.13
Kidney	100% sodium laureth sulphate	1.05

CMC=Carboxyl methyl cellulose

Table 2: Mimicked tissue equivalent material and their measured mean computed tomography number in Hounsfield unit

Mimicked tissue material	Mean CT number (HU)
Liver	42
Lung	–840
Muscle	47
Adipose	–104
Bone	1480
Kidney	38

HU=Hounsfield unit; CT=Computed tomography

was 0°, the diaphragm upper SAD was approximately 25 cm, and lower SAD was 22 cm, giving at total area diaphragm size of 22 cm × 25 cm. Under modifiers, tray factor was 1, and no MLC was present.

A third scan was done following the same protocol with small fields. Six-field technique denoted as BM 1–BM 6 was used covering the six inserts with no wedge used. The gantry angles were: 0°, 45°, 90°, 180°, 225°, and 270°, respectively. The total dose for the six fields was 100 cGy, and total MU making up 100 MU was prescribed. The type of beam used was “simple.” Photon energy of 6 MV was used. SAD was 100 cm, and SSD was approximately 84 cm collimator angle was 0°, the diaphragm upper SAD was approximately 5 cm and lower SAD was 5 cm, giving at total area diaphragm size of 5 cm × 5 cm. Under modifiers, tray factor was 1, and no MLC was present. All planned images from the Precise PLAN 2.16 TPS were transferred to the Elekta Precise Linear Accelerator for treatment.

Furthermore, a simple experimental protocol was done to validate the accuracy of the designed trunk phantom. This was done by comparing dose values (in Gy) between the locally designed trunk phantom and solid water phantom (SWP) which is also a standard phantom in place of the nonavailable Rando Anderson Phantom which was used by the manufacturer. Photon energy of 6MV with SSD of 100 cm was used for the

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setup of the designed trunk phantom and the SWP. Six readings were also observed with the gantry angle at 0°.

The Elekta Precise Clinical Linear Accelerator was calibrated using a large water phantom before this study was done using a 6 MV photon beam to give 100 cGy (1 Gy) at 100 MU with a pre-calibrated NE 2570/1 Farmer-Type Ionization Chamber to determine the absorbed dose. Necessary corrections for temperature, pressure, polarization, recombination, etc., were effected on the Ionization Chamber response. Absorbed dose at reference depth was calculated as follows:

$$D_{w,Q} = M_Q \times N_{D,w} \times K_{Q,Q_0} \quad (1)$$

Where M_Q is the electrometer reading (charge) corrected for temperature and pressure, $N_{D,w}$ is the chamber calibration factor and K_{Q,Q_0} is the factor which corrects for difference in the response of the dosimeter at the calibration quality Q and at quality Q_0 of the clinical X-ray beam according to the TRS 398 protocol of the IAEA.^{1,26}

The deviation between the calculated and measured dose was obtained using the equation as:

$$\text{Percentage (\%) Deviation} = \left| \frac{D_c - D_m}{D_m} \right| \times 100 \quad (2)$$

Where:

D_c = Calculated dose

D_m = Measured dose

In addition to the equation 2, the introduction of the absolute value sign is to turn all negative values from measured absorbed dose into positive values.

Statistical analysis

Data analysis value was done using data analysis was done using SPSS 16.0 (SPSS Inc, Chicago, IL, USA). Descriptive statistics and unpaired *t*-test were implored at 95% level of significance. $P < 0.05$ was considered statistically significant.

RESULTS

A summary of the CT numbers also known as the HU for the mimicked chemical composition for liver, lung, muscle, adipose, bone, and kidney was 42 HU, -840 HU, 47 HU, -104 HU, 1480 HU, and 38 HU, respectively [Table 2].

The gantry angle, measured absorbed dose (Gy) and % deviation for twelve beams with six inhomogeneous inserts with field size of 22 cm × 25 cm at 0° was 1.0004, 22.5° was 0.9981, 45° was 0.9974, 90° was 1.0034, 135° was 0.976, 157.5° was 1.0082, 180° was 0.9693, 197.5° was 0.9722, 215° was 0.9820, 270° was 0.9661, 315° was 0.9708 and 337.5° was 0.9799 Gy respectively and corresponding % deviation was 0.04, 0.19, 0.26, 0.34, 2.40, 0.82, 3.07, 2.78, 1.80, 3.39, 2.92, and 2.01, respectively [Table 3].

The gantry angle, measured absorbed dose (Gy), and % deviation for six beams with bone only homogeneous

inserts with field size of 22 cm × 25 cm at 0° was 1.0293, 45° was 1.0174, 90° was 1.0234, 180° was 1.0113, 225° was 1.0127, 270° was 0.9915 Gy respectively and corresponding % deviation was 2.93, 1.74, 2.34, 1.13, 1.27, and 0.85, respectively [Table 4].

The gantry angle, measured absorbed dose (Gy), and % deviation for six beams with six inhomogeneous inserts with field size of 5 cm × 5 cm at 0° was 1.0056, 45° was 0.9996, 90° was 0.9844, 180° was 0.9684, 225° was 0.9723, 270° was 0.9755 Gy respectively and corresponding % deviation was 0.56, 0.04, 1.56, 3.16, 2.77, and 2.45, respectively [Table 5].

The comparison was made between the mean dose (Gy) of the locally designed pelvic phantom and SWP directly from the linear accelerator at a gantry angle of zero degree (0°) to further validate our result. The mean doses were 0.6065 and 0.6274 Gy, respectively, and % deviation that exists between them was 3.50% [Table 6].

Table 3: Measured absorbed dose (Gy) and percentage deviation for twelve beams with six in homogeneous inserts with field size of 22 cm × 25 cm

BM	Gantry angle (°)	Mean absorbed dose±SD (Gy)	Percentage deviation
1	0	1.0004±1.33E-3	0.04
2	22.5	0.9981±1.33E-3	0.19
3	45	0.9974±2.30E-3	0.26
4	90	1.0034±1.30E-3	0.34
5	135	0.9760±4.56E-3	2.40
6	157.5	1.0082±2.37E-3	0.82
7	180	0.9693±1.13E-2	3.07
8	197.5	0.9722±6.00E-5	2.78
9	215	0.9820±4.79E-3	1.80
10	270	0.9661±9.24E-3	3.39
11	315	0.9708±1.06E-2	2.92
12	337.5	0.9799±4.71E-3	2.01

BM=Beam; SD=Standard deviation; E=Exponential

Table 4: Measured absorbed dose (Gy) and percentage deviation for six beams with bone only in homogeneous inserts with field size of 22 cm × 25 cm

BM	Gantry angle (°)	Mean absorbed dose±SD (Gy)	Percentage deviation
1	0	1.0293±1.15E-3	2.93
2	45	1.0174±1.33E-3	1.74
3	90	1.0234±2.30E-3	2.34
4	180	1.0113±5.74E-3	1.13
5	225	1.0127±1.30E-3	1.27
6	270	0.9915±0	0.85

BM=Beam; SD=Standard deviation; E=Exponential

DISCUSSION

A study has been carried out to verify the performance and accuracy of a TPS which uses an Area Integration Algorithm (Irregular Field Program) based on Clarkson Integration.^{15-17,23-25} The results were within $\pm 5\%$ as recommended by ICRU¹³ and were consistent with Van Dyk whose variation was within $\pm 4\%$,²² Mijnheer and Brahme results were within 3%–3.5% which is consistent with our study.^{19,27} This study was also in line with Akpochafor *et al.* whose maximum % deviation was 4%, against 3.5% which was determined in this study.²⁸

Accuracy and verification of % deviation by the manufacturer using Rando Phantom (Alderson Research Laboratories, 390 Ludlow St., Stamford Conn. 06904) showed an overall accuracy 3.3% using parallel opposed fields in regions of uniform dose distribution without beam compensators, this result was closely in line with this study's parallel opposed field (at 90° and 270°) % deviation, which showed an overall accuracy 3.4%, with inhomogeneous inserts. Better accuracy was seen with bone only homogeneous inserts with uniform dose distribution. Our result was lower (1.7%) compared to the manufacturer verification (3.3%) in terms of accuracy. This variation might be attributed to the comparison made using homogeneity against inhomogeneity of the manufacturer material for verification. In the same vein, using small field sizes, verification of % deviation by the manufacturer showed an overall accuracy 1.8% using parallel opposed field in regions of uniform dose distribution with inhomogeneous insert, this result was better than that obtained in this study whose overall accuracy was 2.5%.

Maximum % deviation of 3.39% was noticed at a gantry angle of 270° (opposed field) and the least deviation 0.04 was noticed at a gantry angle of 0° (anterior field) for six inhomogeneous inserts at field size of 22 cm × 25 cm. In addition, maximum % deviation of 2.93% was noticed for bone only homogeneous inserts at gantry angle of 0° (anterior field) and least deviation 0.85% was noticed at a gantry angle of 270° (opposed field) with a large field size of 22 cm × 25 cm. Furthermore, the maximum % deviation 3.16% was noticed at a gantry angle of 180° and least the maximum % deviation 0.04% was noticed at a gantry angle of 45° with a small field size of 5 cm × 5 cm. This result indicated that with small field at 180°, the accelerator couch will contribute and affect percentage deviation. There was no statistically significant difference in % deviation between inhomogeneous and homogeneous inserts ($P = 0.592$). Similarly, there was no statistically significant difference in % deviation between large and small field sizes ($P = 0.891$) [Tables 3-5].

In conclusion, the highest deviation was noticed when the plastic water phantom was compared to the local trunk phantom. The % deviation between both materials was 3.5% at a gantry angle of 0°. It was observed that one side of the

Table 5: Measured absorbed dose (Gy) and percentage deviation for six beams with six in homogeneous inserts with field size of 5 cm × 5 cm

BM	Gantry angle (°)	Mean absorbed dose $\pm$ SD (Gy)	Percentage deviation
1	0	1.0056 $\pm$ 1.33E-3	0.56
2	45	0.9996 $\pm$ 0	0.04
3	90	0.9844 $\pm$ 1.33E-3	1.56
4	180	0.9684 $\pm$ 1.27E-3	3.16
5	225	0.9723 $\pm$ 6.35E-3	2.77
6	270	0.9755 $\pm$ 1.79E-3	2.45

BM=Beam; SD=Standard deviation; E=Exponential

Table 6: Percentage deviation between solid water phantom and locally designed trunk phantom at gantry angle of 0°

Gantry angle (0°)	PWP (Gy)	LTP (Gy)	Mean dose (Gy) of PWP	Mean dose (Gy) of LTP	Percentage deviation
0	0.6282	0.6076			
0	0.6259	0.6076			
0	0.6282	0.6054	0.6274	0.6065	3.50
0	0.6282	0.6054			
0	0.6259	0.6076			
0	0.6282	0.6054			

PWP=Plastic water phantom; LTP=Local trunk phantom

parallel opposed field (270°) for six inhomogeneous insert with large field size, anterior field (0°) for bone only homogeneous insert with large field size and posterior field (180°) for six inhomogeneous insert with small field size showed the highest deviations. An overview of this study showed that the locally designed phantom was within approximately $\pm 4\%$ [Table 6].

CONCLUSION

A low cost locally designed trunk phantom has been made for use in radiotherapy centers in Nigeria with about six centers using the same Precise PLAN 2.16 TPS. The validation of our result using a standard SWP in place of Rando Alderson Phantom which was not available was within acceptable range; it was seen to be below ICRU minimal limit. The designed trunk phantom showed an overall accuracy of $\pm 4\%$ with the Area Integration algorithm of the TPS which also falls within the acceptable range of $\pm 5\%$ set by ICRU. The designed phantom was made of inexpensive materials that are readily available with almost equal density with some organs in the body. The phantom proves highly useful for quality assurance and control test in radiotherapy.

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Conflicts of interest

There are no conflicts of interest.

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