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Research Article

Accuracy of a Novel Head and Neck Phantom for Heterogeneous Media Verification Using an Irregular Field Algorithm

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Abstract

Objectives: The treatment outcome in patients can be improved with a fast and accurate treatment planning system (TPS) algorithm. The aim of this study was to design a novel head and neck phantom and to use it to test whether the accuracy of the irregular field algorithm of the Precise Plan 2.16 (Elekta Instrument AB, Stockholm, Sweden) TPS was within ±5% of the International Commission on Radiation Units and Measurements (ICRU) limit for homogenous and inhomogeneous media by rotating the Elekta Precise linear accelerator gantry angle using 2 fields.
Methods: A locally designed acrylic phantom was constructed in the shape of a block with 5 inserts. Acquisition of images was performed using a HiSpeed NX/i computed tomography scanner (GE Healthcare, Inc. Chicago, IL, USA); the Precise Plan 2.16 TPS was used to determine the beam application setup parameters and an Elekta Precise linear accelerator was used for radiation dose delivery. A pre-calibrated NE 2570/1 Farmer-type ion chamber with an electrometer was used to measure the dose. The mimicked organs were the brain, temporal bone, trachea, and skull.
Results: The maximum percentage deviation for 10×10 cm and 5×5 cm inhomogeneous inserts was 1.62 and 4.6, respectively, at a gantry angle of 180°, and that of the 10×10 cm homogeneous insert was 3.41 at a gantry angle of 270°. The percentage deviation for only the bone insert (homogeneous) and for all inserts (inhomogeneous) using parallel opposed beams was 2.89 and 2.07, respectively. Also, the percentage deviation between the locally designed head and neck phantom and the solid water phantom of the linear accelerator was 0.3%.
Conclusion: The validation result of our novel phantom in comparison with the solid water phantom was good. The maximum percentage deviations were below the ICRU limit of ±5%, irrespective of gantry angles and field sizes.
Keywords: Irregular field algorithm, ionization chamber, phantom, Plexiglas, solid water phantom, treatment planning system
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The International Commission on Radiation Units and Measurements (ICRU) has provided recommendations that the radiation dose must be delivered within ±5% of prescribed dose. When commissioning treatment planning dose calculation algorithms for radiotherapy, often, the aim is to achieve good agreement between the calculated dose (D_c) and measured dose (D_m), within 1%–2%, for open and wedge (block or compensators) fields in water.[1-8] This is possible using measurement and model-based algorithms in water phantoms; however, such an agreement is usually not possible for measurement-based algorithms in phantoms with heterogeneities. This can be explained based on

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the fact that measurement-based models can account for the effect of tissue inhomogeneities on the primary radiation. However, correcting for scatter radiation is difficult because it depends on field size, beam energy and shape, and location and density of inhomogeneities.[9] In contrast, model-based algorithms can account for the effect of tissue inhomogeneities on scatter radiation using the density scaling method or other approaches.[10-13]

Several techniques for performing quality assurance of TPS have been proposed.[14-17] Similarly, reduction in errors and uncertainties during dose calculation plays an important role in the success of a treatment procedure. The performance and quality of any treatment planning system (TPS) is dependent on the type of algorithm used.[18-26]

Treatment planning requires the ability to calculate the dose to any arbitrary point within the patient for any beam orientation. The irregular field program, also known as the area integration algorithm, is well suited for this purpose. Patient tissue inhomogeneities, beam blocks, and beam compensators are included in the calculation model. The irregular field program requires separation of the dose into primary and scatter components. The primary component is usually computed and includes transmission through any blocks and blocking trays, beam compensators, and patient inhomogeneities. The scatter component is usually computed and includes presence of blocks, beam compensators, and curvature of the patient, but not patient inhomogeneities.[27-29]

The concept of this dosimetry of the irregular field program involves the use of tissue-maximum ratio and scatter-maximum ratio, which are analogous to tissue-air ratio (TAR) and scatter-air ratio (SAR) concepts, respectively. The underlining program equation of the area integration (irregular field program), which is similar to the external beam program, is as follows:

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

where TRAY and TRAY2=tray factors

OUTPUT=the output factor normalized to a (0×0) field size at a distance SAD+DMAX

FSC=the air field size correction dependency factor, which is computed for equivalent square of the collimator opening

SSD=source-to-surface distance

DMAX=maximum dose

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

X and Y=co-ordinates at the depth of the point of calculation

c=correction for the virtual location of the source; c is determined from a plot of the inverse of the square root of Dm versus distance from the source

QF=the off-axis beam quality factor

OCR=the in-air off-central-axis ratio value

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

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

p=value of the penumbra, calculated using the Wilkinson Source Model

This study will focus on verifying the percentage dose accuracy of the irregular field algorithm using homogeneous and inhomogeneous inserts by varying gantry angles for given field sizes of 5×5 cm and 10×10 cm. The reason for designing this novel phantom was to compensate for a ready-made phantom like the Rando Alderson Phantom which is not available in most radiotherapy centers in Nigeria.

Methods

The in-house designed phantom was made from Plexiglas with a thickness of 0.33 mm. A plastic-based hardener (allplast) was used for holding one slab to another to form a cube. Plexiglas (dimension, 4×8 feet) was purchased from a local plastic shop; a part of which was cut using a plastic cutter into six slabs each of a dimension of 20×20 cm. Five holes were drilled on one face. Each drilled hole had a diameter of 2.5 cm gummed together using the plastic-based hardener "allplast". The phantom block was drilled to hold a cylindrical rod (13.5 cm) made of Plexiglas to accommodate a 0.6 cm^3 graphite ionization chamber (NE 2570/1) and also four holes for the tissue-equivalent mixed chemicals. The center of the chamber was 10 cm from the end of the block and diagonally displaced by 7 cm from the other holes. The inferior block of the phantom was drilled to allow water flow (Fig. 1a). Percentage compositions by mass of the tissue equivalent materials were determined at the Pharmaceutical Technology Laboratory of Lagos University Teaching Hospital and was used to mimic each biological tissue (Table 1). The in-house designed phantom was filled

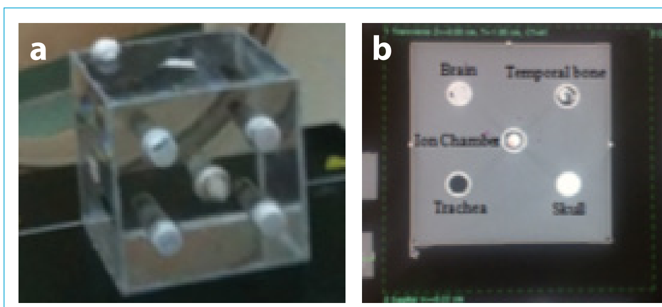

Figure 1 (a, b). Locally designed head and neck phantom with mimicked inserts (LT) and CT image (RT).

Table 1. Mimicked organ and chemical compositions

Mimicked tissue equivalent organ	Chemical Compositions (% by mass)
Brain	C, H_2O, and MgO (52, 25, and 23)
Temporal bone	C, H_2O, MgO, and Ca (40, 27, 23, and 10)
Trachea	C, H_2O, and MgO (14, 15, and 71)
Skull	C, H_2O, MgO, and Ca (35, 30, 20, and 15)

with water and loaded with the tissue-equivalent materials and scanned using a HiSpeed NX/i computed tomography (CT) scanner. Slices of images were acquired for four different tissue-equivalent materials including the ion-chamber port (Fig. 1b). Images were transported through the Digital Imaging and Communications in Medicine (DICOM) to the Precise PLAN Release 2.16 TPS, where 12 field technique, denoted as beam (BM) 1–BM 12, was used with field sizes of 10×10 cm covering the four inserts and no wedge was used. Gantry angles (in degrees) for the 12 fields were 0°, 22.5°, 45°, 90°, 135°, 157.5°, 180°, 197.5°, 215°, 270°, 315°, and 337.5°, respectively. The total dose for the 12 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, source-to-axis distance (SAD) was 100 cm, and SSD was approximately 85 cm. The collimator angle was 0°; the upper SAD of the diaphragm was approximately 10 cm and the lower SAD of the diaphragm was 10 cm, giving a total area diaphragm size of 10×10 cm. Under modifiers, the tray factor was 1 and no multileaf collimator (MLC) was present.

A second scan was performed following the same protocol using 5×5 cm fields. The six-field technique, denoted as BM 1–BM 6, was used covering the four inserts and no wedge was used. Gantry angles for the six fields were 0°, 45°, 90°, 180°, 225°, and 270°, respectively. The total dose for the six fields was 100 cGy and a total MU making up 100 MU was prescribed. The type of beam used was "simple." A photon energy of 6 MV was used. SAD was 100 cm and SSD was approximately 85 cm. The collimator angle was 0°; the upper SAD of the diaphragm was approximately 5 cm and lower SAD of the diaphragm was 5 cm, giving a total area diaphragm size of 5×5 cm. Under modifiers, the tray factor was 1, and no MLC was present.

A third scan was performed using the same protocol with 10×10 cm field sizes, but the insert was bone only equivalent material (assumed to be a homogenous medium). Acquired images from the CT simulator were also transferred to the Precise PLAN 2.16 TPS through the DICOM. A six-field technique was used, denoted as BM 1–BM 6, covering the four inserts which were uniformly homogeneous with no wedge used. Gantry angles for the six fields were 0°, 45°, 90°, 180°, 225°, and 270°, respectively. The total dose for the 6 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. The collimator angle was 0°; the upper SAD of the diaphragm was approximately 10 cm and the lower SAD was 10 cm, giving a total area diaphragm size of 10×10 cm. Under modifiers, the 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.

Comparison using the Elekta-Precise clinical linear accelerator for bone only (homogenous) and all four (inhomogeneous) inserts was measured using SSD of 85 cm with a 6-MV photon beam to determine variations in percentage deviation.

Similarly, a simple experimental protocol for validation of the algorithm was also performed between the locally designed head and neck phantom and the solid water phantom with SSD of 85 cm. The Elekta-Precise clinical linear accelerator was initially calibrated using a large water phantom, with 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 dependent on the ionization chamber response. Absorbed dose at the reference depth was calculated as follows:

$$D_{W,Q} = M_Q \times N_{D,W} \times K_{QQ_0} \qquad (2)$$

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 International Atomic Energy Agency (IAEA).[1, 30]

Deviation between DC and Dm was obtained using the following equation:

$$\% \text{ Deviation} = \left| \frac{D_c - D_m}{D_m} \right| \times 100 \qquad (3)$$

where D_c=calculated dose

D_m=measured dose

Statistical Analysis

Data analysis was conducted using SPSS 16.0 (SPSS Inc, Chicago, IL, USA) Descriptive statistics, one-sample t-test, and unpaired t-test was implored at a 95% level of significance. A $p<0.05$ was considered to be statistically significant.

Results

The measured absorbed doses (Gy) for the 12 beams with four inhomogeneous inserts with a field size of 10×10 cm at a gantry angle of 0°, 22.5°, 45°, 90°, 135°, 157.5°, 180°, 197.5°, 215°, 270°, 315°, and 337.5° were 1.000, 1.008, 1.007, 1.012, 1.008, 1.000, 0.984, 1.001, 1.007, 1.009, 1.008, and 0.987, respectively, and the corresponding percentage deviation were 0.00, 0.79, 0.70, 1.19, 0.79, 0.00, 1.62, 0.10, 0.70, 0.89, 0.79, and 1.31, respectively (Table 2).

The measured absorbed doses (Gy) for the six beams with four inhomogeneous inserts with a field size of 5×5 cm at a gantry angle of 0°, 45°, 90°, 180°, 225°, and 270° were 0.9894, 0.9920, 0.9694, 0.9560, 0.9864, and 0.9588, respectively, and the corresponding percentage deviation were 1.07, 0.81, 3.61, 4.60, 1.38, and 4.30, respectively (Table 3).

The measured absorbed doses (Gy) for the six beams with the bone only homogeneous for the four inserts with a field size of 10×10 cm at a gantry angle of 0°, 45°, 90°, 180°, 225°, and 270° were 0.9870, 0.9802, 0.9740, 0.9760, 0.9740, and 0.9670, respectively, and the corresponding percentage deviation were 1.31, 2.02, 2.67, 2.46, 2.67, and 3.41, respectively (Table 4).

Using the linear accelerator, a comparison was performed to define the extent of deviation when the irregular field algorithm was computed using only bone for the four inserts with parallel opposed beams (90° and 270°) and using different tissue materials for the four inserts with parallel opposed beams (90° and 270°). The mean dose computed for the two inserts were 0.972±3.16E-4 and 1.021±5.16E-4, respectively. Deviation from the initially calibrated dose of 1 Gy by the water phantom were 2.89% and 2.07%, respectively (Table 5).

Validation was made between the mean dose (Gy) calculated for the locally designed head and neck phantom and that calculated for solid water phantom by directly using the linear accelerator at a gantry angle of 0°. The mean doses were 0.744±5.48E-4 and 0.746±5.16E-4 Gy respectively, and percentage deviation between them was 0.3% (Table 6).

Table 2. Measured absorbed dose (Gy) and % deviation for 12 fields with four inhomogeneous inserts with a field size of 10×10 cm

Beam	Gantry angle (°)	Mean absorbed dose±SD (Gy)	% Dev
BM 1	0°	1.000±0	0.00
BM 2	22.5°	1.008±5.774E-4	0.79
BM 3	45°	1.007±0	0.70
BM 4	90°	1.012±5.774E-4	1.19
BM 5	135°	1.008±11.547E-4	0.79
BM 6	157.5°	1.000±5.774E-4	0.00
BM 7	180°	0.984±51.962E-4	1.62
BM 8	197.5°	1.001±5.774E-4	0.10
BM 9	215°	1.007±64.291E-4	0.70
BM 10	270°	1.009±0	0.89
BM 11	315°	1.008±63.509E-4	0.79
BM 12	337.5°	0.987±0	1.31

BM=Beam, Gy=Gray, SD=Standard deviation, E=Exponential, % Dev=Percentage deviation

Table 3. Measured absorbed dose (Gy) and % deviation for six fields with four inhomogeneous inserts with a field size of 5×5 cm

Beam	Gantry angle (°)	Mean absorbed dose±SD (Gy)	% Dev
BM 1	0°	0.9894±5.744E-4	1.07
BM 2	45°	0.9920±7.071E-4	0.81
BM 3	90°	0.9694±5.744E-4	3.61
BM 4	180°	0.9560±0	4.60
BM 5	225°	0.9864±8.944E-4	1.38
BM 6	270°	0.9588±10.954E-4	4.30

BM=Beam, Gy=Gray, SD=Standard deviation, E=Exponential, % Dev=Percentage deviation

Discussion

The measured absorbed dose (Gy) for the 12 beams with four inhomogeneous inserts with a field size of 10×10 cm at BM 1 (0°) and BM 6 (157.5°) was 1 Gy with deviation of 0, indicating that D_m values at these beams were accurate and similar to D_c value of 1 Gy. There was no significant difference in D_c and D_m values (p=0.086). The maximum percentage deviation was 1.62 with BM 7 at a gantry angle of 180° (Table 2).

The minimum and maximum percentage deviations with the six beams with four inhomogeneous inserts with a field size of 5×5 cm were 0.81 with BM 2 at a gantry angle of 45°and and 4.60 with BM 4 at a gantry angle of and 180°, respectively. There was no significant difference in D_c and D_m values (p=0.002) (Table 3).

There was a significant difference in the dose value for the

Table 4. Measured absorbed dose (Gy) and % deviation with six beams for bone homogeneous inserts with a field size of 10×10 cm

Beam	Gantry angle (°)	Mean absorbed dose±SD (Gy)	% Dev
BM 1	0°	0.9870±0	1.31
BM 2	45°	0.9802±4.083E-4	2.02
BM 3	90°	0.9740±0	2.67
BM 4	180°	0.9760±0	2.46
BM 5	225°	0.9740±0	2.67
BM 6	270°	0.9670±0	3.41

BM=Beam, Gy=Gray, SD=Standard deviation, E=Exponential, % Dev=Percentage deviation

12 beams with four inhomogeneous inserts with a field size of 10×10 cm and that for six beams with four inhomogeneous inserts with a field size of 5×5 cm ($p<0.001$). The results obtained show that there was more deviation in accuracy with the 5×5 cm field size.

The minimum and maximum percentage deviation with the six beams with bone homogeneous inserts with a field size of 10×10 cm was 1.31 with BM 1 at a gantry angle of 0° and 3.41 with BM 6 at a gantry angle of 270°, respectively. Parallel opposed fields had the maximum dose of 2.67 and 3.41 with gantry angles of 90° and 270°, respectively; the result was similar to that reported by Akpochafor et al.,[31] whose maximum percentage deviation at BM 10 (270°) was 3.95 using similar algorithm with field size of 25×25 cm. There was no significant difference in D_C and D_m values for bone homogeneous inserts with a field size of 10×10 cm ($p=0.002$) (Table 3).

There was significant difference between bone insert and all insert with opposed field beams ($p<0.001$). However, % accuracy was better with all inserts (2.07) than with bone insert (2.88). This result also confirms the reason for the better accuracy noticed with all insert with 12 beams and bone insert with 6 beams. This shows that the phantom gave better accuracy with inhomogeneous inserts than homogeneous insert (Table 4).

Validation of our locally designed phantom and the standard water phantom showed a 0.3% deviation. This proves that the designed head and neck phantom was good, although there was significant difference in the D_m value ($p<0.001$) (Table 5).

Generally, the results measured were within the range of ±5%, as recommended by ICRU,[6] and were consistent with the results of Van Dyk, whose variation was within ±4%, except that for six fields with four inhomogeneous inserts with a field size of 5×5 cm which was higher (4.60%). [26] Results by Mijnheer et al.[23] and Brahme et al.[32] were within 3%–3.5%, whereas those in this study were higher and in the range of 0%–4.6%. Akpochafor et al. used a locally designed pelvic phantom with the same algorithm with a maximum percentage deviation of 4%, against 4.6 % which proves to be better than that determined in this study.[31] This deviation could be attributed to different densities of organs within the head and neck region and associated error using small field size.

Table 6. Validation result comparing the mean dose and % deviation of the designed H &N phantom and linear accelerator's solid water phantom

	Solid Water Phantom	Designed H&N Phantom
	(Gy)	(Gy)
	0.743	0.746
	0.743	0.745
	0.744	0.746
	0.744	0.746
	0.744	0.746
	0.743	0.745
Mean±SD	0.744±5.48E-4	0.746±5.16E-4
% Deviation=0.3		

H&N=Head and Neck, Gy=Grays, SD=Standard Deviation, E=Exponential

Table 5. Results of absorbed dose for bone (homogeneous) and all organs (inhomogeneous) with opposed beam plan

	Bone Insert with opposed field beam	All Insert with opposed field beam
	(Gy)	(Gy)
	0.971	1.021
	0.972	1.020
	0.972	1.020
	0.972	1.021
	0.971	1.021
	0.972	1.021
Mean±SD	0.972±5.16E-4	1.021±5.16E-4

Gy=Gray, SD=Standard Deviation, E=Exponential

Conclusion

The locally designed phantom showed good accuracy for the 10×10 cm field for different inserts. Deviation was higher with the 5×5 cm for different material media (inhomogeneities). The designed phantom will be suitable in a region like the head where various tissue densities are noticed. The locally designed phantom has proven to be suitable for quality control test in determining the accuracy of the TPS algorithm during radiotherapy. It will most likely be applicable in places in Nigeria where readymade phantoms are not available.

Disclosures

Ethics Committee Approval: The study was approved by the Local Ethics Committee.

Peer-review: Externally peer-reviewed.

Conflict of Interest: None declared.

Authorship contributions: Concept – M.O.A., A.D.O., M.Y.H.; Design – M.O.A., A.D.O., M.Y.H.; Supervision – M.O.A., A.D.O., M.Y.H.; Materials – M.O.A., A.D.O., M.Y.H.; Data collection &/or processing – S.O.A., M.A.A., C.C.I., T.A.O., A.E.O.; Analysis and/or interpretation – S.O.A., M.A.A., C.C.I., T.A.O., A.E.O.; Literature search – S.O.A., M.A.A., C.C.I., T.A.O., A.E.O.; Writing – M.O.A., A.D.O., M.Y.H., S.O.A., M.A.A., C.C.I., T.A.O., A.E.O.; Critical review – S.O.A., M.A.A., C.C.I., T.A.O., A.E.O.

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