Investigation of tube voltage dependence on CT number and its effect on dose calculation algorithms using thorax phantom in Monaco treatment planning system for external beam radiation therapy

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Abstract

Introduction: The accuracy of dose calculation algorithms depends on the electron density and computed tomography (CT) number of medium scanned. Our study aimed to verify the impact of different CT scanning protocols on Hounsfield unit (HU) and effect on dose calculation algorithms. **Materials and Methods:** CIRS thorax phantom with different density material plugs was scanned at varying tube voltages from CT scanner and HU values were measured in treatment planning system (TPS). Calibration curves of electron density at different tube voltages were plotted and used for dose calculation with different calculation algorithms at varying high energy megavoltage photon energies. **Results:** Insignificant difference is obtained in electron density curves plotted at different tube voltages. The mean variation in HU values was found at different tube voltages for bone, lung, and water are 896.75 (standard deviation [SD] 122.88), −799.25 (SD 5.74), and −17.5 (SD 0.57), respectively. The estimated P values for change in HU values were 0.089, 0.258, and 0.121 for bone, lung, and water, respectively. Pencil beam (PB) convolution and collapsed cone algorithms show no significant dose difference, i.e., <1% variation and Monte Carlo (MC) shows maximum dose difference up to 1.4%. **Conclusion:** Third-generation algorithms such as MC shows dependence on varying tube voltages in dose calculation. Calibration curves plotted at different kVp in TPS advised to be chosen wisely to avoid any dosimetric errors in different medium.

Keywords: Collapsed cone convolution, Hounsfield unit, Monte Carlo, pencil beam convolution

INTRODUCTION

Radiotherapy treatment process is very complex in nature consisting of various steps and each step is prone to systematic and random errors. Each component of chain influences the radiotherapy outcome and must be handled cautiously. After diagnosis, patients are advised to start the radiotherapy procedure with computed tomography (CT) Simulation. CT is a vital diagnostic imaging modality that can produce images of patient geometry in axial, sagittal, and coronal planes used for radiation treatment planning. It provides high-resolution transverse images and gives information about the Hounsfield Units (HUs) and electron densities (ED) of medium scanned which influences the delineation of target volumes and the surrounding normal tissues. Before using treatment planning system (TPS) for clinical needs, a relation between CT numbers

Access this article online

Quick Response Code:

Website:
www.jmp.org.in

DOI:
10.4103/jmp.JMP_124_20

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and relative electron density (RED) has to be established in the form of calibration curve as described by Schneider *et al.*[1] TPS usually equipped with the standard calibration curve provided by the vendors to use in clinical environment but its usage without validation may result into the dosimetric inaccuracy.

Treatment planning workstation is important component in the radiotherapy chain procedure. Many factors are present in TPS which can influence the dosimetric treatment outcome and calibration curve is one such parameter present in the system. In most clinics, only the calibration curve plotted at 120 kVp for all scans ignoring the potential benefit of dose and noise reduction with an applied tube voltage. Constantinou *et al.*[2] suggested having more than a standard 120кVp calibration curve, for example, an additional 80kVp curve for pediatric patients which may potentially decreases the CT dose and offers a high contrast to noise ratio. Although having many calibration curve increases the amount of workload and degree of error are also increases.

The accuracy of new generation dose calculation algorithms present in radiotherapy primarily depends on the electron density obtained from CT data and calibration of CT-HU to RED.[3] The CT number depends on the linear attenuation coefficient of the material with the formula mentioned in equation (1):

$$HU_{tissue} = 1000 \times ([\mu_{tissue} - \mu_{water}]/\mu_{water}) \qquad (1)$$

Where HU is Hounsfield Unit (HU) and μ is the linear attenuation coefficient of the medium.

The change in HU values due to kVp settings and geometric distribution of various tissue substitute materials has also been studied by Nobha *et al.*[4] and found to be well within 2%. The CT number of medium depends on the spectra of CT scanner and CT-RED curve may lead to determination of material composition of medium. A slight change in CT number may lead to relative change in electron density of the medium and in turn leads to the variation in dose calculation. Accurate calculation of dose distribution in an inhomogeneous medium like human body is a complex phenomenon, especially for tumors located in the lung. Till date, only the Monte Carlo (MC) method is considered to be the most accurate algorithm for dose calculation, but it requires the greatest processing time. Apart from MC method, all other methods make different degrees of approximation and simplification which lead to much faster calculation speed but also result in less accurate dose distribution compared with the MC calculation algorithm. The applied voltage was reported to be most relevant parameter leading to errors in the reconstructed Hounsfield numbers of about 300 units for high density.[5]

In previous studies, dose variation within 1% has been reported when varying the kVp.[6] In addition, the HU measurements of solid water and air had found to differ significantly between scanners from different manufacturers.[7,8] The highest variation was noted in case of high density materials and CT scan at lowest kVp. Many authors have conducted the evaluation of dose calculation algorithms for external beam radiation therapy. Rana *et al.*[9] investigated the dose prediction accuracy of Acuros XB algorithm and anisotropic analytical algorithm (AAA) for different field sizes and air gap thickness. The results from that study revealed that dose prediction errors are up to 3.8% for Acuros XB and up to 10.9% for AAA, respectively. Furthermore, the study by Rana *et al.*[10] demonstrated the limitation of dose calculation algorithms when treating a smaller size of tumor, especially when large air gaps are created by immobilization devices.

In the current study, the effect of different CT scanning protocols, i.e., varying kVp settings on HU number variation and their dosimetric impact on dose calculation in TPS using different algorithms, for example, MC, collapsed cone convolution and pencil beam (PB) were investigated using CIRS thorax phantom.

MATERIALS AND METHODS

CIRS thorax phantom
The CIRS Model 002 LFC (Norfolk, VA) intensity-modulated radiotherapy (IMRT) thorax phantom shown in Figure 1 were used in the present study, which is designed to address the comparison of calculation algorithms TPS and complete system QA from CT imaging to dose verification. The 002 LFC is an elliptical in shape and properly represents an average human torso in proportion, density, and two-dimensional structure. It measures 30 cm long × 30 cm wide ×20 cm thick. The phantom is constructed of proprietary tissue equivalent epoxy materials. Linear attenuations of the simulated tissues are within 1% of actual attenuation for water and bone, and within 3% for lung from 50 keV to 15 MeV. Tissue equivalent interchangeable rod inserts accommodate ionization chambers allowing for point dose measurements in multiple planes within the phantom shown in Figure 1. Adapter placement allows verification in the most critical areas of the chest. One half of the phantom is divided into 12 sections, each 1 cm thick, to support radiographic or Gafchromic film. Three kinds of materials are contained in this phantom to emulate different tissues in human body, including soft tissue, lung, and bone. Dose measurements can be performed by placing ionization chamber and Gafchromic film inside grooves provided in phantom.

Validation of kV and mAs in computed tomography scanner
kV and mAs stability was validated before starting the study. Stability of scanner was measured from the last 6 year readings to till now which are given below [Tables 1-3].

Plotting calibration curves
The relationship between the RED and CT number plotted in TPS is known as calibration curve. In radiotherapy treatment planning simulation, different kVp settings are used depending on the different body sections to be imaged. To calculate the mean HU values, the phantom was scanned using CT scanner (GE optima 580 w, USA) with the different kVp protocols (tube voltage: 80 kVp, 100 kVp, 120 kVp, and 140

kVp with slice thickness of 2.5 mm) to acquire CT images. In order to obtain the calibration curve, a CIRS phantom was used at different tube voltages protocol scanning. The phantom was placed on CT couch, leveling of the phantom was ensured. The HU values obtained from the systems were plotted against RED of the materials. Figure 2 illustrates CT-RED calibration curves obtained for different tube voltages from the Monaco TPS.

Formula used:

Mean Percentage Dose Difference

$$= \frac{(\text{Individual Value- Mean Value})}{\text{Mean Value}} \times 100 \qquad (2)$$

Treatment planning system dose calculation measurement

TPS was commissioned as per the standard guidelines of TRS 430. Several end-to-end tests were performed in phantom with all photon energies to found out the percentage of variation in between chamber measured and TPS calculated dose are given below. These values are baseline TPS data found during commissioning of TPS.

Table 4a shows the dose difference in TPS data and measured data with Farmer type chamber, Fc-65 for different calculation algorithms in a homogeneous medium in central axis measurement. All the percentage values obtained for MC, Collapsed cone (CC), and PB were well within the tolerance limit. At depth 10 cm, the full scattering condition is satisfied and measurement showing the similarity in dose obtained as in planning system. For 6 mega voltage (MV) flattening filter free (FFF) and 10MV FFF CC algorithm is not available into the Monaco TPS, so no variation was found out.

Table 4b shows the percentage variation in dose differences calculated in an inhomogeneous medium through ionization chamber and TPS for all energies in central axis measurement. The field sizes were chosen were 10 cm × 10 cm and 15 cm × 15 cm for evaluation in dose difference as field size 10 cm × 10 cm includes mostly water equivalent part of thorax Phantom, whereas 15 cm × 15 cm includes both lungs partially with dose measurement in water equivalent material.

In the present study, three different calculation algorithms of the Monaco TPS (Version 5.11, Elekta AB, Stockholm, Sweden) were evaluated. These three algorithms were MC, Collapse Cone, and PB dose calculation engine. The acquired CT images of CIRS Phantom were sent to the TPS through digital Imaging and communication in medicine networking system. Circular region of interest of diameter 1.5 cm was defined on the CT images of the phantom and mean CT numbers for different materials were obtained from TPS as shown in Figure 3, using Monaco TPS Version 5.11, Elekta AB, Stockholm, Sweden). Lungs, Bone, and water medium were properly contoured, and the point of interest was marked kept same in all three calculations algorithm cases.

A single anterior beam plan using 6 MV, 10 MV. 15 MV, 6MV FFF and 10 MVFFF at gantry angle 0° with isocenter placed at the center of the phantom for dose of 100 centiGray (cGy) was planned and calculated as shown in Figure 4. Dose calculations

Figure 1: CIRS thorax phantom
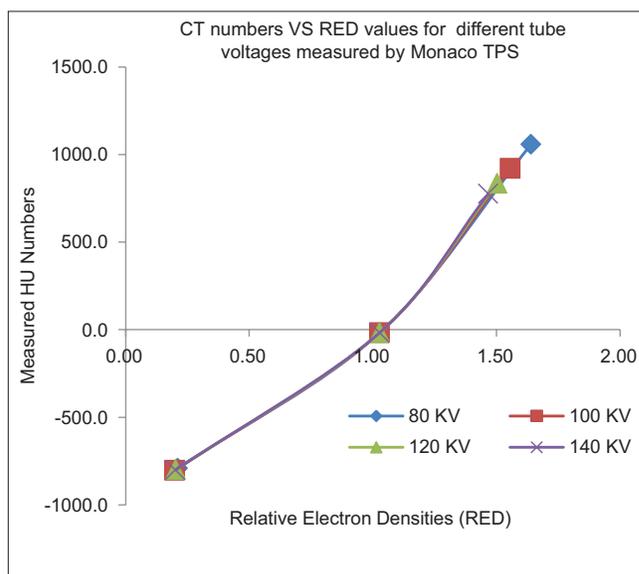
Figure 2: Computed tomography to relative electron density calibration curve obtained from Monaco treatment planning system
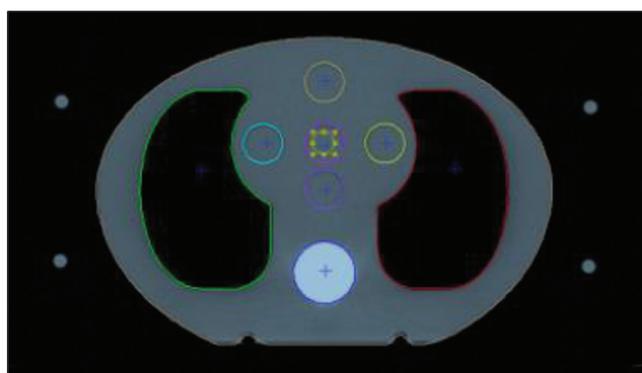
Figure 3: Computed tomography scan of thorax phantom with different densities

were performed with PB, CC, and MC algorithm on Monaco TPS version 5.11 with calculation grid size of 3 mm and statistical uncertainty of 0.5%. Dosimetric comparison was performed in between for various CT to RED calibration curves obtained from various tube voltages, i.e., 80 kVp, 100 kVp, 120 kVp, and 140kVp on Monaco TPS.
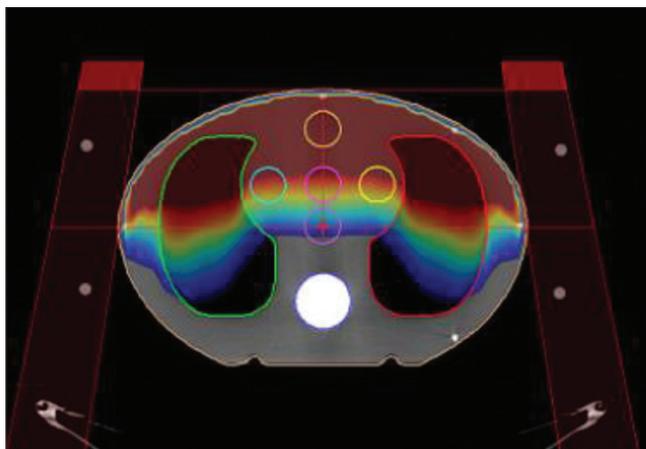
Figure 4: Axial plane of dose measurement in Monaco treatment planning system with different density contours of lung, bone and water

RESULTS

Evaluation of variation in Hounsfield unit-relative electron density calibration curves

Here, we have evaluated the variation in HU-RED calibration curves for different tube voltages used in CT scanning protocols.

In Table 5, the HU numbers of different materials for various kVp values on Monaco TPS were presented. The mean variation in HU values were found at different tube voltages for bone, lung, and water are 896.75 (standard deviation [SD]: 122.88), −799.25 (SD: 5.74), and −17.5 (SD: 0.57), respectively. No significant difference was found between the HU numbers obtained from different tube voltages. The estimated *P* values were 0.089, 0.258, and 0.121 for bone, lung, and water, respectively.

Maximum CT number difference was observed in bone. However, the measured CT numbers for lung and water were in agreement with one another with in 20 MU.

Impact of different kVp on dose calculation algorithm

Dose calculated with the CT images on Monaco TPS for different algorithms in four different contour of bone, bilateral lungs, and water in cGy for a plan created for 100 cGy dose at the isocenter using 6 MV, 10 MV, 15 MV, 6FFF, and 10FFF from various tube voltages of 80 kVp, 100 kVp, 120 kVp, and 140 kVp. Phantom CT scan with 120 kVp scanning protocol was considered to be reference raw data for comparative analysis of dose in three different density contours created on CT images of the phantom, and100 MU was delivered on different density cavities in thorax phantom using 3 mm grid size and 0.5% statistical uncertainty in dose calculation with 10 cm × 10 cm field size.

$$\text{Total Error} = \int_{min}^{max} f(tubevoltage) + \int_{min}^{max} f(Uncertainity) \quad (3)$$

For any fixed grid size, G

On rearranging above equation, the error contribution due to tube voltage difference may be obtained as:

$$\int_{min}^{max} f(tubevoltage) = \text{Total Error} - \int_{min}^{max} f(Uncertainity) \quad (4)$$

In case of MC Calculation Algorithm, for any fixed grid size and maximum uncertainty window of 0.5%, the equation will become,

$$\int_{min}^{max} f(tubevoltage) = \text{Total Error} - \mathbf{0.5} \quad (5)$$

For FFF beam, CC algorithm calculation is not available in TPS system and Calculation performed only for 6 MV, 10 MV, and 15 MV flattened beam. In flattened and unflattened Beams, dose calculation difference with the PB calculation algorithms in different density medium at varying tube voltages were found insignificant and <1%. Similarly, for flattened beams, CC calculation algorithm shows <1% variation at varying kVp scans in different medium. MC calculation algorithm shows impact in dose calculations in different medium for varying tube voltage scans at different photon energy which is discussed below.

Monte Carlo dose calculation variation in different density medium at varying tube voltages

Using equation no. 5, the actual error due to tube voltage was evaluated making other parameters such as grid size constant and the resultant error obtained is only due to the tube voltage impact in dosimetric spectra. Table 6 shows the percentage variation evaluated for all beam energies in different density medium using MC calculation engine. The maximum uncertainty introduced in a calculation window was subtracted from the total error obtained to get exactly the contribution of error due to varying tube voltage. The percentage variation in doses was <1% in all cases at different energy and density medium except for 6FF and 6FFF beams whose values were 1.2% and 1.4%, respectively, for high density medium of bone.

DISCUSSION

Computed tomography relative electron density calibration curves

At all the points in a medium of contoured circular diameter, different HU values were shown by the TPS. For the purpose of simplicity and plotting, the curve mean values of all circular medium were considered against the RED. The spectral changes in scanning beam ultimately result into the changes in the HU values of medium. The highest variation in CT number was observed in bone material with respect to different CT scanning protocols (tube voltages) than followed by lung equivalent material and least in the water equivalent material. In lung equivalent material, small CT number was observed as the electron density is extremely low, thus, becomes more sensitive to the imaging noise over variation in tube voltages, causing more variation in HU values for air-like materials. The HU-RED curves as shown in Figure 2 reflected no specific difference in the curves obtained using HU and ED values from different tube voltages scans. A very small deviation of curve is shown near to bone HU values. The reported HU variations may be explained due to nonuniform beam filtration

Table 1: kVp stability of computed tomography scanner

Set kV	1st year measured kV	2nd year measured kV	3rd year measured kV	4th year measured kV	5th year measured kV	6th year measured kV	SD	Relative SD (%)
80	81.05	80.7	80.28	80.74	80.97	79.34	0.579	80.51±0.72
100	99.22	99.5	99.92	99.82	99.86	98.48	0.503	99.63±0.50
120	119.33	120.33	119.62	120.05	120.87	118.83	0.667	119.83±0.55
140	139.5	140	140.82	141.3	141.82	139.15	0.959	140.43±0.68

ALL of the above results are within tolerance limit of±2 kV. SD: Standard deviation

Table 2: Coefficient of mA linearity was also found out from last 6 years

mA setting	COL (1st year)	COL (2nd year)	COL (3rd year)	COL (4th year)	COL (5th year)	COL (6th year)
100, 150, 200	0.0018	0.004	0.00024	0.0003	0.0014	0.0013

All of the above results are within tolerance limit=±0.1. COL: Coefficient of mA linearity

Table 3: Output constancy was also checked and found the coefficient of variation

kV setting	mAs	COV (1st year)	COV (2nd year)	COV (3rd year)	COV (4th year)	COV (5th year)	COV (6th year)
80, 100, 120, 140	100	0.003	0.006	0.000663	0.000329	0.002156	0.039

All above results were within tolerance limit=±0.05. COV: Coefficient of variation

Table 4: Dose measurement in Homogeneous and CIRS Thorax Phantom

(a) TPS dose comparison in homogeneous phantom using Fc-65 ionization chamber for cental axis measurement

Energy (MV)	Chamber dose (cGy)	TPS dose (cGy)			Percentage variation		
		PB	CC	MC	PB	CC	MC
6	164.08	163.7	163.4	165.7	0.23	0.41	−0.987
10	175.38	173.9	174.3	174.5	0.84	0.62	0.50
15	183.95	185.4	184.9	183.9	−0.78	−0.51	0.027
6FFF	162.22	161.0	-	162.2	0.75	-	0.012
10FFF	174.52	172.9	-	174.2	0.93	-	0.18

(b) Dose measurement in water equivalent groove in CIRS thorax phantom and TPS

Field size	Energy												
	6 MV (dose cGy)			10 MV (dose cGy)			15 MV (dose cGy)			6 MV FFF (dose cGy)		10 MV FFF (dose cGy)	
	Algorithms			Algorithms			Algorithms			Algorithms		Algorithms	
	MC	CC	PB	MC	CC	PB	MC	CC	PB	MC	PB	MC	PB
10×10	91.8	90.3	90.7	96.1	93.9	94.4	100.8	98.6	99.4	91.5	89.7	96.2	94.2
Chamber dose		90.3			97.4			100.3			88.2		92.4
Percentage variation	1.66	0	0.44	−1.33	−3.6	−3.1	0.49	−1.69	−0.89	3.74	1.7	4.1	1.94
15×15	95.3	94.2	94.4	98.6	97.3	96.8	103.8	102.2	102.1	93.5	91.8	97.7	95.6
Chamber dose		94.91			100.91			103.71			92.64		95.32
Percentage variation	0.41	−0.75	−0.54	−2.29	−3.57	−4.07	0.087	−1.45	−1.55	0.92	−0.91	2.49	0.29

SAD setup, Depth=10 cm, FS=10 cm×10 cm, MU=200. MC: Monte carlo, PB: Pencil beam, CC: Collapsed cone, TPS: Treatment planning system, FFF: Flattening filter free, CIRS: Computerized Imaging Reference Systems, SAD: Source to Axis Distance, FS: Field Size, MU: Monitor Units

of scanning beam passing different densities inserts. The CT number variation depends on the scanner-specific factors such as spectral energy, filtration of radiation, and reconstruction algorithm used. Many researchers have reported large deviations of HU values in high-density material like Teflon.[11] According to the different guidelines, there are range of RED values for air, soft–tissue, and bone, and different tolerances of HU values are reported for different materials.[12] Every

planning system equipped with the standard calibration curves which could be used with some errors for clinical purposes, but it is always recommended to plot the curve of existing CT simulator for regular clinical usage. To reduce any uncertainties during radiotherapy procedure protocol, it is recommended to validate the CT-RED curves of the CT unit whose images would be used for clinical planning and treatment.

Dosimetric comparison of different calculation algorithms versus tube voltage

Monaco TPS showing variation in calculated doses with different dose calculation algorithms with respect to different CT scanning protocols (tube voltages) in Tables 7 and 8. Out of three algorithms, only MC shows the variation in the calculated doses at different tube voltages scan, whereas PB and CC show very negligible variations. This may be due to the reason that MC calculation algorithm even takes care of small changes in electron density of the medium and includes in dose calculation. The PB and CC algorithms failed to calculate dose variations very accurately and results very small difference of <1% at different scanning protocols. The MC algorithm simulates the transport of millions of particles and photons through matter. It utilizes the law of probability distribution of individual interactions of particles and photons. The CC algorithm is a convolution superposition model-based engine. It consists of a convolution equation that separately considers the transport of primary photons and scattered electrons. A point kernel convolution/superposition model accounts for inhomogeneity correction in patients. In CC algorithm, variations in lateral photon and electron transport are approximately modeled. In PB algorithm, the beam is divided into infinitesimal into narrow PBs and into the field grid PBs is calculated altogether. Finally, the dose at any point is obtained by summation of the dose contribution of all PBs into point of interest. However, in the PB algorithm, variations in photon transport and lateral electrons are not modeled.

A patient's body contains different densities, requiring a correction factor for each beam causing beam attenuation. The PB algorithm is very fast due to its use of a one-dimensional density correction, which does not accurately model the distribution of secondary electrons in heterogeneous media. The CC algorithm was developed to model the physical processes involved instead of semi empirically tabulated measurements in water. The implementation of such a model,

however, uses some approximations that make the model perform better under certain irradiation geometries (lung) and worse in others such as with bone heterogeneities.[13-15] This shows that MC simulation algorithm is a powerful tool for quality assurance in radiotherapy as it is only calculation method that can account for all the physical phenomena that take place in the interaction of radiation beams with inhomogeneous media. MC can even describe the dose deposition in the vicinity of high-Z interfaces. Figure 4 clearly shows the distribution of dose in different density materials by MC calculation algorithm.

Table 6 clearly shows the maximum dose variation in different density materials with varying photon energies calculated with MC calculation algorithms. This variation depicts the MC while calculating considers the electron density and energy spectrum factors while performing three-dimensional calculations in a medium. At higher energy, the scattering distribution will be maximum in bone medium results in maximum variation in dose calculation with varying electron density obtained at different tube voltages. For lower energy in the low density medium, maximum variation is observed due to consideration of scattering contribution with varying electron density at different tube voltages. According to Chen *et al.*,[16] MC method is considered to be the most accurate algorithm for dose calculation, but it requires the greatest processing time. Apart from MC, all other methods make different degrees of approximation and simplification which lead to much faster calculation speed but also result in less accurate dose distribution comparing with the MC simulation.[17-20] Variation of 1% was reported in the previous study on CATPHAN phantom for MC calculation algorithm,[1] in contrast our study reveals maximum difference of 1.2%–1.4% when MC calculation algorithm is used which may be due to sufficient density materials availability in CIRS phantom and the contribution of scattered beam and energy spectrum could be sufficiently dissipate in the medium. In CATPHAN phantom, only different density materials of small sizes are present which could provide insufficient scattering contribution effect of different density. CIRS phantom mimics the torso of human body and suitable for the dose distribution study providing sufficient medium for incorporating scattering contribution.

At different scanning protocol, the spectra of tube voltages changes and offers different electron density to the medium.

Table 5: Mean Hounsfield unit values from Monaco treatment planning system for different tube voltages

Density material	80 kVp	100 kVp	120 kVp	140 kVp
Bone	1058	920	834	775
Lung	−791	−804	−800	−802
Water	−17	−17	−18	−18

Table 6: Actual error for different energy in density medium using Monte Carlo calculation at varying tube voltage using 0.5% uncertainty window

Energy (MV)	Bone (%)	Left lung (%)	Right lung (%)	Water (%)
6FF	1.2	0.37	0.37	0.53
6FFF	1.4	0.65	0.65	0.73
10FF	0.45	0.05	-	0.12
10FFF	0.42	-	-	-
15FF	0.6	-	-	0.3

FFF: Flattening filter free

Table 7: Percentage dose variation calculated using beams for pencil beam convolution and collapsed cone convolution algorithms at different tube voltages

| Energy (MV) | Density material | PB (cGy) | | | | SD | Mean | Maximum mean percentage dose difference |
		80 kVp	100 kVp	120 kVp	140 kVp			
6FF	Bone	73.7	74.5	74.1	73.7	0.38	74	0.67
	Left lung	108.4	108.7	108.4	108.7	0.17	108.5	0.18
	Right lung	108.8	108.9	108.6	109.1	0.21	108.8	0.27
	Water	114.5	114.7	114.7	114.9	0.16	114.7	0.17
6FFF	Bone	59.6	60.1	60	60.2	0.26	59.9	0.5
	Left lung	76.6	76.7	76.6	76.8	0.09	76.7	0.13
	Right lung	76.9	77	76.8	77.1	0.13	76.9	0.26
	Water	92.9	93	93	93.2	0.12	93.0	0.21
10FF	Bone	74.4	75	74.9	75.1	0.31	74.8	0.53
	Left lung	106.7	106.8	106.7	106.9	0.09	106.8	0.09
	Right lung	107	107	106.8	107.2	0.16	107	0.18
	Water	112.2	112.4	112.4	112.6	0.16	112.4	0.18
10FFF	Bone	58.4	58.9	58.8	59	0.26	58.8	0.68
	Left lung	74.9	74.9	74.9	75	0.05	74.9	0.13
	Right lung	75.1	75.1	75	75.3	0.12	75.1	0.26
	Water	89.3	89.4	89.4	89.5	0.08	89.4	0.11
15FF	Bone	76.9	77.5	77.3	77.6	0.31	77.3	0.52
	Left lung	108.8	108.9	108.7	109	0.13	108.8	0.18
	Right lung	109.1	109.1	108.9	109.3	0.16	109.1	0.18
	Water	112.9	113	113	113.2	0.12	113.0	0.17
Energy (MV)	**Density material**	**CC (cGy)**				**SD**	**Mean**	**Maximum mean percentage dose difference**
		80 kVp	100 kVp	120 kVp	140 kVp			
6FF	Bone	69.7	70.3	70.8	69.7	0.53	70.1	0.99
	Left lung	109.1	109.2	109.5	109.6	0.24	109.3	0.27
	Right lung	109.3	109.3	109.5	109.7	0.19	109.5	0.18
	Water	115.8	116.2	116.2	116.5	0.28	116.2	0.34
6FFF	Bone							
	Left lung							
	Right lung							
	Water							
10FF	Bone	73.1	73.6	74	74.3	0.52	73.75	0.88
	Left lung	104.7	104.7	105	105	0.17	104.85	0.14
	Right lung	105.1	105	105.3	105.4	0.18	105.2	0.19
	Water	112.9	113.1	113.2	113.4	0.21	113.15	0.22
10FFF	Bone							
	Left lung							
	Right lung							
	Water							
15FF	Bone	76.1	76.4	76.8	77	0.4	76.6	0.65
	Left lung	106.6	106.6	106.9	106.8	0.15	106.7	0.18
	Right lung	106.5	106.4	106.6	106.7	0.13	106.55	0.14
	Water	114.1	114.3	114.3	114.5	0.16	114.3	0.17

PB: Pencil beam, CC: Collapsed cone, FFF: Flattening filter free, SD: Standard deviation

The circle of diameter chosen for study itself shows variation in value of HU at different points in a circular diameter, and for study purpose, the mean values were considered, and curve of HU and RED was plotted, but in reality, every point in a medium was offered different energy spectra by CT and results into different ED in a medium. The circular diameter for different materials shows variation in electron density in a selected grid size for calculation and total dose when calculated for selected diameter's for different algorithm depicts variation.[21]

Figures 5 and 6 show the expected behavior of beam energies 6ff and 6fff in different density materials at varying tube voltages and depicts the maximum variation in bone material due to scattering contribution and changes in energy spectra. It is recommended to use the system cautiously

Table 8: Dose calculated using beams for Monte Carlo calculation algorithms at different tube voltages

Energy (MV)	Density material	MC (cGy)				SD	Mean	Maximum mean percentage dose difference
		80 kVp	100 kVp	120 kVp	140 kVp			
6FF	Bone	69.2	70.3	70.8	71.2	0.86	70.4	1.7
	Left lung	113.4	114.2	114.8	115.2	0.78	114.4	0.87
	Right lung	113.6	114.4	114.9	115.4	0.78	114.6	0.87
	Water	114.6	115.9	116.1	116.6	0.85	115.8	1.03
6FFF	Bone	56.3	57.3	57.6	58.3	0.83	57.4	1.9
	Left lung	77.1	77.9	78.2	78.8	0.71	78.0	1.15
	Right lung	77.4	78.2	78.4	79.1	0.70	78.3	1.15
	Water	92.1	93.4	93.3	94.2	0.86	93.25	1.23
10FF	Bone	72.7	73.5	73.6	73.9	0.51	73.4	0.95
	Left lung	109.2	109.9	109.9	110.1	0.39	109.8	0.55
	Right lung	109.4	110.1	110.0	110.2	0.36	109.9	0.45
	Water	112.5	113.5	113.2	113.5	0.47	113.2	0.62
10FFF	Bone	53.6	54.0	54.4	54.5	0.41	54.1	0.92
	Left lung	63.9	64.1	64.5	64.4	0.27	64.2	0.46
	Right lung	64.2	64.4	64.7	64.7	0.24	64.5	0.46
	Water	82.3	82.8	82.9	83.0	0.31	82.75	0.30
15FF	Bone	75.2	76.0	76.6	76.5	0.64	76.07	1.1
	Left lung	110.4	111.0	111.7	111.5	0.58	111.15	0.49
	Right lung	110.5	111.2	111.8	111.5	0.55	111.25	0.49
	Water	112.9	113.9	114.3	114.1	0.62	113.8	0.80

MC: Monte Carlo, FFF: Flattening filter free, SD: Standard deviation
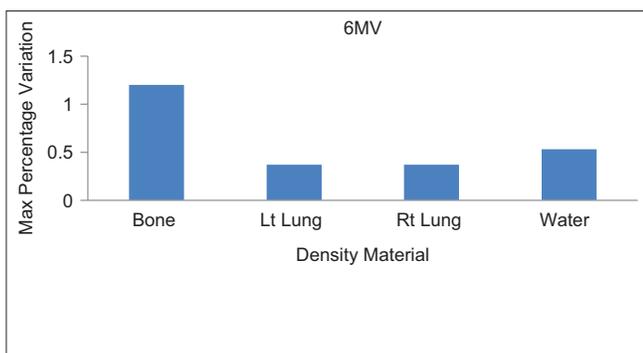
Figure 5: Maximum percentage variation of dose in different density materials for 6 mega voltage photon energy calculated with 0.5% uncertainty Monte Carlo calculation algorithm

Figure 6: Maximum percentage variation of dose in different density materials for 6 mega voltage flattening filter free photon energy calculated with 0.5% uncertainty Monte Carlo calculation algorithm

when different tube voltage scan is used different from baseline calibration curve, and judiciously, the uncertainty window level needs to be set for optimum result; otherwise, there are chances to obtained wrong doses in clinical environment.

CONCLUSION

We have found that different kVp setting shows no statistically significant variation in the measured HU values. The highest variation was observed in case of high-density bone material at the lowest kVp tube voltage. PB and CC convolution algorithms show <1% variation in dose distribution in all cases of varying tube voltage electron density obtained but MC calculation algorithm shows deviation up to 1.2%–1.4%.

The study can be validated through other dosimeters such as Thermoluminescent dosimeters, optically stimulated luminescence, three-dimensional gel dosimeter, and MC Simulation method as in the present study only ionization chamber-based assessment has performed. Institutions using third-generation algorithms like MC shall use it cautiously when for high-end techniques such as intensity-modulated radiation therapy and volumetric-modulated radiation therapy considering tube voltage effect on electron density. Every center shall plot calibration curves of different tube voltages before using for dose calculation and same energy calibration curve shall be used on which simulation was performed. TPS quality assurance needs to be performed with different calibration curves of different tube voltage energy before using for clinical treatment with third-generation calculation algorithms. In future, impact of tube voltage on gamma

index needs to be explored for MC calculation algorithm on high-end techniques such as IMRT and volumetric-modulated arc therapy.

Financial support and sponsorship

Nil.

Conflicts of interest

There are no conflicts of interest.

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