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Dosimetric Evaluation for Effect of Calibration Different CT Machines Parameters on dose Calculation Accuracy for Advance Radiotherapy Techniques

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Abstract: Data from a computed tomography (CT) scan serves as the foundational input for the radiotherapy treatment planning system (TPS), which incorporates the effects of inhomogeneities into dosage calculations. The factors unique to each scanner may have an impact on how CT numbers are measured. Thus, it's critical to confirm how different CT scanning techniques affect the Hounsfield unit (HU) and how it affects dosage estimation. In this study, different three types of CT machines with the voltage application 80KVp to 140 KVp were examined for their effects on HU for various tissue replacements in phantoms as well as their dosimetric effects on dose computation in TPS due to variations in HU-relative electron density (RED) calibration curves. Using three CT machines scans of the phantom taken at various tube voltages, HU for various densities of materials was calculated Calibration curves for HU-RED. The same 120kVp values, no appreciable difference in HU of various density materials was found significant difference for three CT machines with 0.85% for dose calculation according to different types of CT machine. Doses that were computed using various HU-RED calibration curves were accurate to within 0.85%. The range of dose variations determined by algorithms using different HU-RED calibration curves was discovered to be well within 1.08%. Hence, it can be said that adopting a 120 KVp CT acquisition approach in clinical practice while using the standard HU-RED calibration curve is feasible. at different tube voltages.
Keywords: CT calibration carve -Dose calculation-radiotherapy technique – SBRT.

1 Introduction

The human body consists of a variety of tissues and cavities with different physical and radiological properties. Most important among these, from a radiation dosimetry perspective, are tissues and cavities that are radiologically different from water, including lungs, oral cavities, teeth, nasal passages, sinuses, and bones. The dose distribution is affected by these tissues in homogeneities and since treatments are becoming increasingly conformal, the opportunity for geo-graphic misses of the target due to incorrect isodose coverage increases. In view of the inconsistent use of inhomogeneity correction Optimization of therapeutic benefit is dependent on maximizing the dose to the planning target volume while minimizing the dose to normal tissues. This optimization requires the accurate, three-dimensional localization of both the diseased target tissues and the sensitive normal tissues. In the last two decades, major progress in imaging technology has improved our ability to identify and to localize these critical volumes and to determine their densities in vivo on a voxel-by-voxel basic.

Accuracy of treatment planning system (TPS)-based planning is highly hinged upon computed tomography (CT) images. The quality of these CT images influences the recognition and delineation of target volumes and the surrounding normal organs. Substandard image quality may result in improper delineation of the target volume and normal organs by omission or over-inclusion of a portion of normal organ volume and significant misconception. Thus, it is essential to sustain the optimal image quality of CT

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scanners used for simulation of radiotherapy patients. The accuracy of dose calculation using these radiotherapy TPS, taking into account the effect of tissue inhomogeneities, is based on such CT data and calibration of CT Hounsfield units (HU) to relative electron density (RED). CT number or HU from CT images provide information on the attenuation characteristics of X-ray beam in a particular volume element in patient

- Adjustments for inhomogeneity were also discussed, as clinicians were reluctant to use them without clinical outcome data. However, correction of inhomogeneity has become an essential part of treatment planning in modern therapy and is necessary. for intensity modulated radiation therapy (IMRT). [12] Recent advances in dose estimation using advanced Monte Carlo simulation algorithms such as pencil beam, convolution/overlap, and shrunk cone have improved dosimetry and dose calculation accuracy. [13-16] However, advanced dose algorithms require the electron density (ED) of the CT data to account for inhomogeneity effects, rather than physical density scaling as advocated by previous algorithms such as equivalent path length. (EPL). In order to correlate the CT counts reported in a patient's CT scan with the corresponding ED values, it is necessary to establish a CT count - ED calibration curve. Below is the CT number for each voxel, expressed in Hounsfield units (HU):

$$\text{CT Number}(HU(x,y,z)) = 1000\left(\frac{\mu_t(x,y,z)-\mu_w}{\mu_w}\right) \quad (1)$$

Several researchers[1-2] have proposed tissue characterization based on CT count and ED calibration used in TPS using a commercial phantom. The calibration curve (CT-ED) is stored in the TPS database for dose calculation purposes. The CT-ED curve and its effect on dosimetry were captured in the context of older dose calculation algorithms as 1. [1] used advanced TPS and quantified dosimetric effects very similar to those previously reported by Jones et al. CT scanner selection and technical aspects of TPS are presented [2] Each CT scanner manufacturer optimizes CT images based on the choice of body part to be scanned; however, depending on the scanning protocol, different methods may be used. Because technique selection on a CT scanner can result in the same tissue with a different CT number, the treatment planner must be aware of the impact of such changes.

Many researchers have observed variation in CT counts due to different scanning parameters [5-8], and some studies have been conducted to investigate its dosimetric effect using heterogeneous cubic or anthropomorphic phantoms. The effect of kiloamp setting on low Z-scanner inhomogeneity has been found to be clinically insignificant [2], but there is limited variation between CT scanner studies. [3] compared CT parameters and showed that up to 20% variation can be observed in HU; however, the dose effect is only 1. Thus, the effects of CT number on Monte Carlo calculations with photons and electron beams were observed to be different and require attention. [4] allowable variability in GE scanner CT count at different kV settings and tube currents. Tube current (mAs) has been shown to play no role and only kV causes changes in CT count.

2 Materials and Methods

CT number versus relative ED for different tube voltages and reconstructed FOVs were plotted for Philips CT , Siemens Somatom 4D scanners and GE CT Scanner . Phantom using for Computed Tomography Number to Electron Density Calibration

To review the CT number calibration, a tissue characterization phantom (RMI, Gammex, Middleton, WI, USA) was used to evaluate various scanning conditions. The Phantom consists of a hard disc of water approximating the size of an average and is equipped with interchangeable rods made of various tissue-equivalent materials. The physical density (G/cm3) ranges from 0.3 (LN-300 lung) to 1.84 (cortical bone) and the corresponding ED relative to water ranges from 0.292 to 1.707. The CT MRI phantom is widely used in radiation therapy clinics in the United States. The manufacturing quality assurance of these tissue-equivalent plugs is very accurate (<1% variation), which was tested on five phantoms The phantom was placed in the center of the CT stand in careful alignment with the lasers and scanned using different imaging protocols using different tube voltages (80–140 kvp) for each scanner. Two reconstruction fields of view (33 cm and 48 cm) were selected for image reconstruction using a 512 x 512 matrix with a contiguous slice thickness of 5 mm. After image reconstruction, a circular region of interest (ROI) with a diameter of 1.5 cm was placed on each density pin and the mean CT counts of the ROI were recorded. To minimize the effects of image artifacts and beam hardening, multiple phantom CT scans were acquired with different combinations of insertion sites, and the resulting average CT numbers were calculated. The same process was repeated on several scanners, including the Philips (85 cm) and 28-inch (72 cm) scanner (Philips Healthcare, Andover, MA, USA) and the Somatom 4 scanner (Siemens Medical Solutions, Malvern). , PA, USA).) and GE PET-CT Scanner as shown in fig (1)

Fig. 1: PET CT Scanner as example for using in current study (Model 3)

The 1.1 density bump (bump) is characteristic of the RMI phantom and has been observed by other investigators. This is most likely due to the fact that the artifact in the lid has a different chemical composition but the same physical density. Differences between CT number and tube voltages are minimal in the density range of 0.3 (lung) to 1.0 (water). This discrepancy becomes significant in high-density materials and can be as high as 43% of cortical bone (1668 HU at 80 kVp vs. 1167 at 140 kVp), with a trend toward higher kVp leading to lower CT counts. This is likely due to increased photoelectric damping due to lower photon energy, leading to an increase in the number of CTs. Full and half FOV reconstructions have little effect on CT readings for all materials in any scanner; the only exception was an 11% difference (1869.4 HU vs. 1686.4 HU) for cortical bones at 80 kVp Somatom 4 CT, and GE PET CT . The CT–ED number calibration tables were imported into Eclipse TPS (Varian Medical Systems, Palo Alto, CA, USA) and used to examine their effect on dose calculations version 13.8 . According to institutional review board waiver status, two representative cases (liver small lesion for SBRT techniques) were selected for current study study. Treatment planning was performed using an analytical anisotropic algorithm(AAA) , which provides excellent inhomogeneity correction as reported by many investigators. designed to provide optimal coverage of a typical tumor lesion in the center of a patient's liver cancer on 6 MV photon beam with true beam machines varian model . Each plan used a different CT number for inhomogeneity correction – ED calibration table for a given tube voltage (80 kVp–140 kVp).

3 Results and Discussions

- The results of standard CT volume determination in this study were in reasonable agreement with previously reported HU changes associated with different CT scan protocols (tube voltage) and reported the greatest change for high-density tissue substitutes [9,10]. In this study, the largest difference in CT number from the nominal value was observed in Teflon.

The reported differences in HU can be explained by non-uniform beam filtering of scan beams passing through inserts of different densities. Many researchers have reported large differences between nominal and measured CT counts for different CT scanners [11–13].

As in air, a deviation from the nominal value of CT number is observed because the electron density of air is very low and therefore becomes more sensitive to image noise as the tube voltage changes, resulting in different air HU values. Differences between measured CT counts and nominal CT counts have been reported in the literature for different CT scanners, particularly for the lowest and highest density sensitometry inserts, and this difference depends on scanner-specific factors such as energy spectrum, CT algorithms. reconstruction and radiation filtering [9].

Table 1: The HU , RED and ED for CT model 1.

Medium	Density	HU	RED	ED
lung inh	**0.2**	**0.2**	**0.05988**	**0.634**
lung exh	**0.5**	**0.5**	**0.149701**	**1.632**
Adipose	**0.96**	**0.96**	**0.287425**	**3.17**
bone 1750	**2.15**	**2.15**	**0.643713**	**6.6**
bone 800	**1.53**	**1.53**	**0.458084**	**4.862**
breast 50/50	**0.99**	**0.99**	**0.296407**	**3.261**
bone 200	**1.16**	**1.16**	**0.347305**	**3.73**
muscle	**1.06**	**1.06**	**0.317365**	**3.483**
bone 1250	**1.82**	**1.82**	**0.54491**	**5.663**
liver	**1.07**	**1.07**	**0.320359**	**3.516**
bone core	2.15	2.15	0.643713	6.6

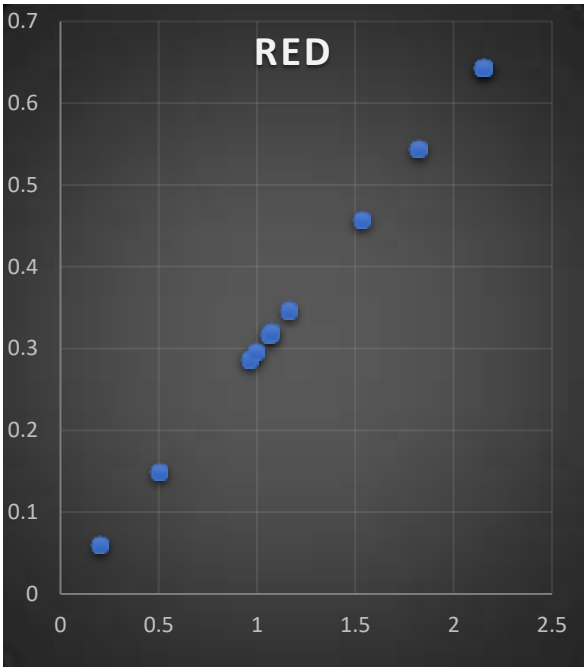


Fig .2: The RED for CT scanner model 1

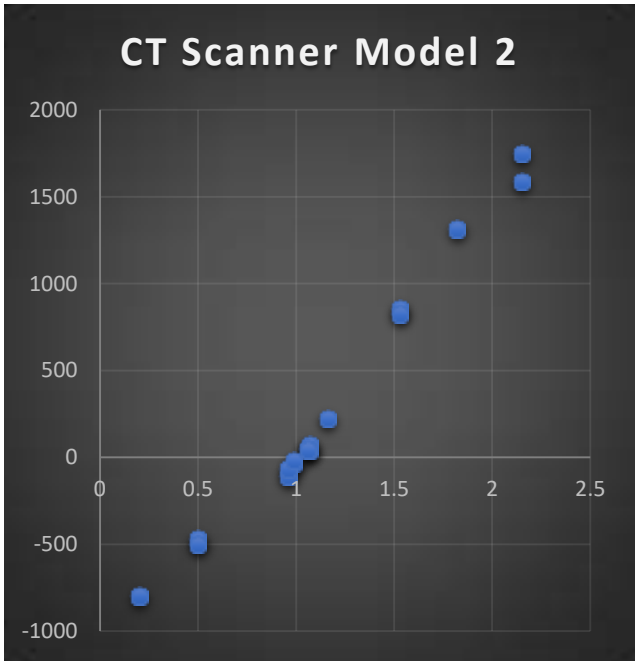


Fig .3 : The RED for CT scanner model 2.

Table 2:. The HU , RED and ED for CT model 2:

outer				
medium	density	ED *10^23	HU	RED=
Bone	2.15	6.6	1747	1.980198
lung	0.5	1.632	-470	0.489649
bone	1.82	5.663	1315	1.69907
bone	1.53	4.862	856	1.458746
adipose	0.96	3.17	-110	0.951095
liver	1.07	3.516	41	1.054905
liver	1.07	3.516	71	1.054905
breast	0.99	3.26	-37	0.978098
bone	1.16	3.73	225	1.119112
muscle	1.06	3.483	53	1.045005
lung in	0.2	0.634	-797	0.190219
lung ex	0.5	1.632	-499	0.489649
adipose	0.96	3.171	-66	0.951395
bone	2.15	6.6	1590	1.980198
bone	1.53	4.862	824	1.458746
breast	0.99	3.261	-19	0.978398
muscle	1.06	3.483	43	1.045005

Table 3: The HU, RED and ED for CT model 3:-

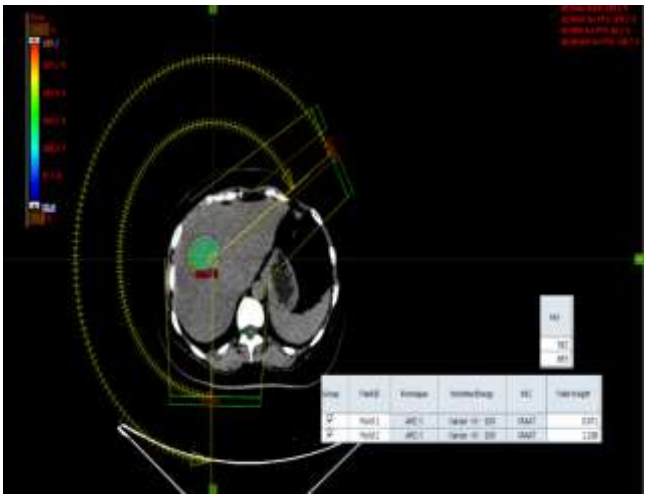
Medium	HU	□ (g/cm^3)	ED (e/cm^3) EXP23	RED = ED_m/ED_w
0.62MA-21	1819	2.15	6.6	1.976048
0.62MA-05	-494	0.5	1.632	0.488623
0.62MA-27	1315	1.82	5.663	1.695509
0.62MA-15	822	1.53	4.862	1.455689
0.62MA-11	-75	0.96	3.171	0.949401
0.62MA-09	33	1.07	3.516	1.052695
0.62MA-09	50	1.07	3.516	1.052695
0.62MA-06	-62	0.99	3.261	0.976347
0.62MA-08	235	1.16	3.73	1.116766
0.62MA-10	20	1.06	3.483	1.042814
0.62MA-04	-767	0.2	0.634	0.18982
0.62MA-05	-493	0.5	1.632	0.488623
0.62MA-	-45	0.96	3.171	0.949401

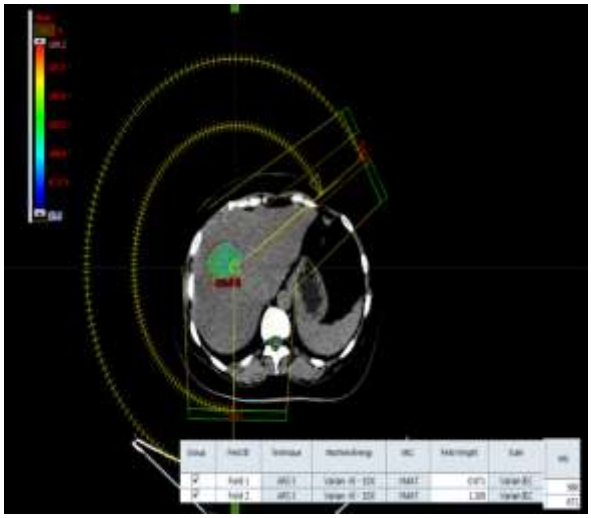
11				
0.62MA-29	1688	2.15	6.6	1.976048
0.62MA-15	221	1.53	4.862	1.455689
0.62MA-06	-83	0.99	3.261	0.976347
0.62MA-10	58	1.06	3.483	1.042814
WATER	0	1	3.34	1

The HU-RED curves, as shown in Fig. 2(a) and (b) , do not show any specific differences between the curves obtained with HU from both systems. CT number linearity increased with kVp, except for Teflon. The measurement results of both systems were consistent. The differences were between 990-892 HU at 140 kVp and 944 HU at 80 kVp.

in current results the geometric arrangement of plugs with electron density and variation of kVp parameters during CT simulation does not cause significant errors in the calculation of heterogeneity-based high-energy photon beam dose. In current study, the dose distributions obtained using different CT-RED calibration curves at different kVp parameters and evaluated at three different CT scanners [10] also showed that even when changing kVp, the change in calculated doses after applying several corrections for tissue heterogeneity for high values - energy photon beams remained a good 1.08 %. The same trend for low-energy (6 MV) photon beams appeared in our study.

-Example for advance case for SBRT for liver and abdomen for three different types of CT machines with different standard Electron Density Calibration





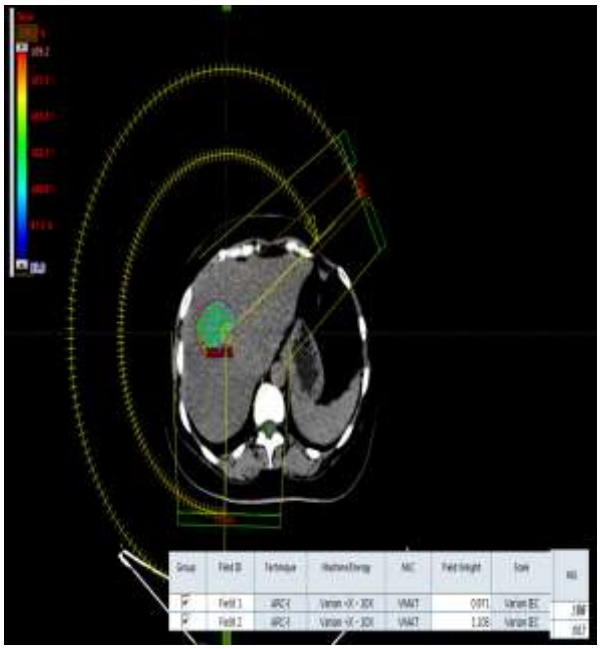


Fig.4: Three SBRT plan using different model of CT correction and difference in Monitors units output.

Dosimetric Impact in liver Cases were treated used SBRT techniques

-The dosimetric effect of ED variation was reviewed to evaluate clinically relevant cases (liver cancer patient). For the liver, differences between three SBRT techniques were minimal in PTV coverage in all ED tables and diifert monitors units for different plans according different in CT scanner as in Figure 1-3.

SBRT plans may have slight differences between different plans due to the plan optimization process. Overall, differences in DVH due to different numbers of CT compared to ED calibration were insignificant (<1%) in all 3 plans (for **Electron Density Calibration**). In order to

better investigate and quantify the slight deviation of DVH, the PTV volume coverage of all ED calibrations was compared to 140 kV. Differences between dose levels (90-110%) were minimal. In all plans, the 80 kVp calibration resulted in the largest deviation from the 140 kVp calibration with the lowest volume. This can be caused by the high-density material in the bone and soft tissues like bowel as OAR . However, the largest difference was only 1.08% between the three plans and could be considered clinically significant due to small volume and high precision plan.

4 Discussion

We reevaluated and quantified the variation in CT number - ED calibration among vendors, tube voltages, and FOVs, as well as its impact on radiation therapy planning and dose computation, as demonstrated by other researchers. [11] discovered a more than 200 HU variance in cortical bone between scanner manufacturers after scanning an ED calibration phantom with the identical scanning parameters on six different scanners. Other publishes papers demonstrated that there was no significant difference in scanning performance by analyzing published data for a variety of scanners. Some paper publishes demonstrated that there was no significant difference in the connection between CT number and relative ED for low-density materials across manufacturers and calibration procedures. Significant discrepancies in data sets from different equipment and measuring procedures were reported for high-density materials.

Analytic calculations based on effective depth revealed that for a 10% change in CT number, increases in inhomogeneity correction factors were less than 1. In a comparable investigation, the CT number was found to be consistent with regard to diverse acquisition parameters, with the exception of the tube voltage setting, which can result in inaccuracies of up to 300 HU for high-density materials. [12]

And evaluated the dosimetric impact of several CT number - ED curves for the entire liver as well as three typical bone locations when exposed to single beam irradiation. For high-density bones, the dose per MU was found to be 2% higher for 80 kV than for 130 kV at a depth just beyond bone. For low-density bones and lungs, the difference in kV is only 1% or less. The tube voltage was discovered to be the most influential component, with other scanner parameters having little influence.

We also discovered that CT number discrepancies are minor in low-density materials but significant in high-density materials. We studied the impact on dose-volume coverage in real patient plans rather than comparing single point dosage or MU/Gy in simplified phantoms.

The precision with which the CT number and dose are calculated. Aside from bones, contrast agents and metal implants are two high-density materials typically found in patient CT scans. evaluated the effect of CT contrast agents on dose calculation. It has been claimed that contrast agents with CT numbers less than 500 HU and volumes less than 5 cm in diameter will not produce significant variations in dosage calculation (1-3%). It is unfortunate that none of the scanners can provide artifact-free CT data as well as none of the TPS can give accurate dose distribution with high-Z materials. Some TPS provide corrections by inserting electron/physical density up to Z = 22. [14]

In most of the studies reported so far, the dosimetric effects of various conversions of CT values to ED have focused primarily on photon beams. Variation in CT values may be large, but the effect on dose in low-density media or breast and pelvic malignancies is limited (<2%). The effects of scanning parameters on CT values and the corresponding effects of dosimetry on electron and proton dose calculations require further investigation, but are not discussed here due to distance and stopping power issues [15].

5 Conclusion

Based on previously published articles and looking at this question from an individual perspective, it is concluded that the variation in CT count varies according to scan parameters and CT scanner vendors. However, in low-density media, variations in CT count are minimal with scanners and X-ray energy, but deviations can be significant for high-density materials. A higher tube voltage gives a lower CT number, while other parameters such as the reconstruction FOV and scanner aperture have little effect on the CT number. A greater difference in dose coverage was observed between different tube voltages for higher density tissues.

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