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REVIEW ARTICLE

Can CT scan protocols used for radiotherapy treatment planning be adjusted to optimize image quality and patient dose? A systematic review

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ABSTRACT

This article reviews publications related to the use of CT scans for radiotherapy treatment planning, specifically the impact of scan protocol changes on CT number and treatment planning dosimetry and on CT image quality. A search on PubMed and EMBASE and a subsequent review of references yielded 53 relevant articles. CT scan parameters significantly affect image quality. Some will also affect Hounsfield unit (HU) values, though this is not comprehensively reported on. Changes in tube kilovoltage and, on some scanners, field of view and reconstruction algorithms have been found to produce notable HU changes. The degree of HU change which can be tolerated without changing planning dose by >1% depends on the body region and size, planning algorithms, treatment beam energy and type of plan. A change in soft-tissue HU value has a greater impact than changes in HU for bone and air. The use of anthropomorphic phantoms is recommended when assessing HU changes. There is limited published work on CT scan protocol optimization in radiotherapy. Publications suggest that HU tolerances of ±20 HU for soft tissue and of ±50 HU for the lung and bone would restrict dose changes in the treatment plan to <1%. Literature related to the use of CT images in radiotherapy planning has been reviewed to establish the acceptable level of HU change and the impact on image quality of scan protocol adjustment. Conclusions have been presented and further work identified.

INTRODUCTION

CT images used in radiotherapy treatment planning must serve two key purposes: to allow, with high geometric fidelity, the position of the tumour and surrounding tissues along with organs at risk to be accurately identified and to provide a map of the electron density information for the various tissues to be used in the treatment planning system (TPS) dose calculation. Most radiotherapy centres now have access to dedicated CT scanners which are designed solely for radiotherapy. The opportunity therefore exists to optimize scan protocols to best support imaging objectives for radiotherapy.

CT scan protocols used in diagnostic imaging departments routinely vary reconstruction algorithm, slice width, tube current, field of view (FOV) and other parameters to produce high-quality images to match the imaging task. On radiotherapy CT scanners, a "one-size fits all" approach is sometimes taken with minimal variation of scan parameters.[1,2] This conservatism relates to the concern that varying scan parameters will change HU values in the images and subsequently introduce inaccuracies to the dosimetric information produced in the TPS. The disadvantage of this approach is that the quality of the images can be compromised, meaning that the identification and outlining of key structures is performed on a suboptimal image. Inaccuracies and variability in the outlining process are well known and can represent a significant source of uncertainty in the radiotherapy process.[3–6] Their causes include the level of expertize and training of the clinician in anatomical and image interpretation; pathological variation in the patient; decision making with regard to the level of likely spread of disease; and difficulties with distinguishing between tissues of similar densities.[7–9] For the last point, poor-quality CT images will certainly be detrimental to the process. Additionally, the use of autocontouring systems might require the adjustment of CT image acquisition to allow the autocontouring systems to work effectively. Whitfield et al,[10] in their review article on automatic delineation, comment that the definition of a minimum standard for image quality is necessary.

The technological developments in CT are advancing rapidly and new features such as metal artefact reduction, dual-energy imaging, iterative reconstruction and automatic kilovoltage selection are becoming common on CT scanners.[11–16] Some of these developments could help to improve the quality of radiotherapy planning CT (PCT) scans. Additionally, adjustment of more fundamental scan parameters such as reconstruction algorithms and FOV to better match the body region imaged would deliver higher image quality and potentially improve accuracy at the outlining stage. If these new techniques are to be considered or existing scan protocols optimized, it is important that there is a good understanding about the level of HU variation which can be tolerated for different CT imaging techniques, without adversely affecting the dose distribution in the planning process. The objective of this review is two-fold: to review the literature so as to establish the accuracy of HUs required in CT images when used for radiotherapy and to summarize the work that has already been published related to CT scan protocol adjustment to ensure good quality imaging for radiotherapy at reasonable levels of dose.

It should be noted that the focus of this review is voxel-by-voxel-based CT planning. Alternative planning methods used involve bulk electron density allocation. This categorizes tissues into typically three or four types such as bone, air, soft tissue and muscle and allocates them pre-defined electron density values. Studies have shown that planning dosimetry using this method can match the voxel method to within a few percentages.[17–19] It is often used in MRI and cone beam imaging where HU data are unavailable or unreliable. This area of work is beyond the scope of this article.

METHODS AND MATERIALS

Search strategy

Searches were carried out using PubMed and EMBASE. The search was restricted to articles in English and initially to articles published between 2000 and 2016. Key terms used were radiotherapy planning, computed tomography, calibration, phantoms, electron density and image quality. The search was narrowed by positively excluding articles including the following terms: PET, SPECT, ultrasound, 4D gated and brachytherapy. The search was subsequently supplemented by reviewing the lists of references in the articles which were read in detail. Additionally, summary articles related to the use of imaging in radiotherapy and published in the Institute of Physics and Engineering in Medicine's professional magazine SCOPE were reviewed for further references. Only publications which discussed the use of CT for radiotherapy planning and related sources of inaccuracy were selected for detailed review.

RESULTS

165 articles were identified and, after the title and abstracts were reviewed, 53 were selected as relevant. 19 of these discussed aspects of image quality in CT, the rest focused predominately on commissioning the TPS or dosimetry changes in planning due to variations in the CT image. No review articles were found.

Acceptable variation for Hounsfield units

Tolerances defined in guidance documents

A TPS needs a CT calibration curve to convert the HU values of different tissue types or materials in the planning scan to electron density. The TPS models the interactions of the treatment beams within the patient and through use of a dose calculation algorithm produces density-corrected dose calculations. Different types of planing algorithms are used by commercial TPSs. They differ in complexity and the methodology used to model the beam interactions.[20] The choice of the algorithm affects the accuracy of the dose calculation for different treatment regimes and the speed of calculation.[21,22] In practice, the CT calibration is a plot of HU *vs* relative electron density (RED) values for a range of different materials. The RED is the electron density of a material relative to water. Typical RED values are 0.2 for lung, 1.0 for water and 1.5 for bone.[23] The relationship is generally bilinear with different linear equations describing the relationship between RED and HU for different materials above and below approximately 100 HU.[24–26] The reason for this change for high atomic number materials is the proportion of compton *vs* photoelectric interactions of the X-ray.[27] The curve is usually defined when a TPS is commissioned.[28] Some planning systems allow the use of several curves to accommodate information from different CT protocols. Some TPSs use physical density instead of electron density.

A number of TPS-commissioning guidance documents discuss the CT calibration curve and tolerances for accuracy. The International Atomic Energy Agency (IAEA) quotes a requirement of 3% accuracy for calculated doses.[23] Other authors have quoted 1–2%.[22,29] The IAEA tolerance for accuracy of HU is given as ± 20 HU, corresponding to RED variation of ±0.02.[23,30,31] This is used for materials of different densities ranging from air to water and up to bone. Example data in IAEA document Techdoc-1583 shows that the CT calibration curve may vary for different CT scanners, particularly for materials which are denser than water.[31] Data also show the variation of CT values measured on the CRIS 002LFC thorax phantom (CIRS Treatment; Simulation and Phantom Technology, Norfolk, VA). For bone-equivalent material, HU values varied from 780 to 900 for different scanners. The authors indicate that in radiotherapy treatment, this would result in a 2% error for a 6-MV photon beam passing through 5 cm of the bone-equivalent material. This equates to a variation of ±60 HU producing a ±1% error in calculated dose for these beam conditions. This appears to imply, though it is not stated, that a tolerance wider than ±20 HU is acceptable at the higher density end of the CT calibration curve.

Guidance has been produced by several professional bodies and is summarized in Table 1. Referencing the 2% tolerance of RED for lung given in Institute of Physics and Engineering in Medicine Report 88, Kilby et al[33] later commented that the tolerance is tight and demonstrated that, for routine quality control results collected over a year, a CT scanner struggled to meet it.[32] The European Society for Radiotherapy and Oncology and the Swiss Society for Radiobiology and Medical Physics have set tolerances for quality control constancy tests for non–intensity-modulated radiation therapy beams.[33,34] Both the American Association of Physicists in Medicine and the Netherlands Commission on Radiation Dosimetry have produced detailed test protocols but no specific tolerances for this parameter.[28,36,37] It is possible to calculate an HU tolerance for the quoted RED tolerance using the appropriate equation of the calibration curve.[38] Typical

equations published by Thomas[24] were RED = (HU/1000) + 1.00 for materials with HU <100 and RED = (HU/1950) + 1.00 for HU ≥100. Calculated values of HU are included in Table 1.

Experimental investigations related to planning CT

Rutonjski et al[39] audited Elekta CMS XiO TPSs (Elekta Instrument AB, Stockholm, Sweden) in three radiotherapy centres using the IAEA protocol.[30] Using the CRIS 002LFC thorax phantom, they generated CT calibration curves and compared the results against the manufacturer-supplied or generic curves which were in the TPSs. No information was provided about the origin of those curves. The biggest differences between the measured HU values and those already in the TPSs were seen at the upper end of the calibration curve (RED = 1.5). For one centre, the measured HU was 790 compared with the TPS value of 890. For the other two centres, the differences were smaller. Planning calculations were carried out using a point kernel convolution/superposition planning algorithm. The conclusion was that this variation would impact on dose accuracy by <2% for 6-MV photons with 5-cm-thick bone-like material. Although the study noted that the results at the high-density end of the calibration curve exceeded the IAEA criteria, the decision was made not to change the information in the TPS due to the small impact on dose accuracy. Tests had also been made on a pencil beam convolution, equivalent path length algorithm which gave significantly different results. Further work and review of the literature allowed the authors to conclude that this type of algorithm was not appropriate for lung treatments.

The choice of phantom used when collecting data for the CT calibration curve is important.[26] The phantom size and shape, volume of scattering material and the position of the different inserts within the phantom will all affect the HU values. Anthropomorphic phantoms which mimic patient size and shape are recommended.[31,40] Craig et al[41] used a new design of phantom to assess three radiotherapy TPSs and found an error in the CT calibration curve used in two of them. An incorrect assumption had been made when the calibration curve was initially defined in the TPS that Teflon could be used as a substitute for cortical bone. The impact was that at the upper end of the curve, the estimated RED for bone was approximately 40% too high. At 1500 HU, the RED used was 2.4 instead of 1.7. The authors calculated that using 1-cm-thick bone material at a depth of 2 cm in water, the corresponding error in dosimetry for an 18-MV X-ray beam was <3% at a depth of 5–15 cm in water along the central axis.

Cozzi et al[42] discussed a scenario where the CT calibration curve in the TMS Helax TPS (MDS Nordion Therapy Systems, Uppsala, Sweden) could not be edited to account for calibration values from the local CT scanner. The difference between the measured data and the default CT calibration data was the greatest for the higher density materials. Where RED = 1.3, the HU difference was approximately 100 HU and where RED = 1.5, the difference was approximately 150 HU. The two calibration curves were used to produce plans for 6- and 15-MV photon beam treatments. The CT values used in the fixed TPS calibration curve were higher than the measured values resulting in a degree of underdosing. The worst case dose difference was 1.9% for 10-cm bone at 6 MV. At a more realistic 3-cm bone thickness at 6 MV, the error was <0.5% and deemed acceptable. A summary is given in Table 2.

Chu et al[43] looked at HU variation with FOV on a conventional CT scanner and a simulator. They established that using equivalent path length a change of 20 HU would result in a 2%

Table 1. Summary of tolerances in guidance documents

Tissue type	References	RED value	Defined RED or HU tolerance	Corresponding HU[a]
Lung	ESTRO, SGSMP[34,35]	0.2	±0.05 (±25%)	±50
	IPEM[32]	0.2	±0.004 (±2%)	±4
	IPEM[32]	0.4	±0.008 (±2%)	±8
	IAEA[23,30,31]	0.21	±0.02 (±10%) or 20 HU	±20
	AAPM[46]	0.2	±50 HU	–
Soft tissue	ESTRO, SGSMP[34,35]	1.0	±0.05 (±5%)	±50
	IPEM[32]	1.0	±0.01 (±1%)	±10
	IAEA[23,30,31]	1.06	±0.02 (±2%) or 20 HU	±20
	AAPM[46]	1.0	±30 HU	±30
Bone	ESTRO, SGSMP[34,35]	1.5	±0.1 (±7%)	±170
	IPEM[34]	1.3	±0.03 (±2%)	±50
	IPEM[34]	1.8	±0.04 (±2%)	±70
	IAEA[23,30,31]	1.6	±0.02 (±1%) or 20 HU	±34
	AAPM[46]	1.3	±50 HU	–

AAPM, American Association of Physicists in Medicine; ESTRO, European Society for Radiotherapy and Oncology; HU, Hounsfield unit; IAEA, International Atomic Energy Agency; IPEM, Institute of Physics and Engineering in Medicine; RED, relative electron density; SGSMP, Swiss Society for Radiobiology and Medical Physics.
[a]HU tolerance calculated using Thomas[24] equations.

uncertainty in electron density for soft tissue and a change of 250 HU would result in a 5% uncertainty for the cortical bone. Considering different depths of tissue, they concluded that at 6 MV, there was uncertainty in dose of 2% at depths up to 20 cm of soft tissue. With the addition of 1 cm of bone, this increased by <0.5%. For higher energy beams, the uncertainties were lower. At 18 MV, the 2% dose uncertainty corresponded to soft-tissue thickness of up to 30-cm depth. Finally, the authors looked at clinical cases for brain, lung and pelvis plans. The results are given in Table 3.

Kilby et al[33] aimed to produce electron density tolerances with a clear link to the dosimetric error under different treatment conditions. Electron density tolerances were generated for 6-MV photon beams, with a maximum dose error of 2% and for maximum tissue thicknesses of 20 cm of water, 10 cm of lung and 7 cm of bone. The TPS used was Nucletron® system (Nucletron BV, Veenendaal, Netherlands) using a Hogstrom model.[44] The electron density tolerances were calculated as ±0.03 for water, ±0.05 for lung and ±0.08 for bone. The article also presented tolerances for 15-MV photons which were broader than 6 MV.

Kirwin et al[38] looked at the HU variation produced by different head and body reconstruction algorithms on a Siemens Emotion Duo CT scanner (Siemens Healthcare, Erlangen, Germany). They reviewed the articles by Thomas,[24] Kilby et al[33] and Knoos et al[45] and developed HU tolerances for water (RED value = 1.002), lung (RED value = 0.190) and bone (RED = 1.117 for trabecular bone and RED = 1.5 for dense bone). The electron density tolerances developed by Kilby et al and the different formulae by Thomas and Knoos et al were used to produce HU tolerances which were 160 (Thomas method) or 210 (Kilby method) for bone, 30 for water and 50 for lung. The authors chose to use the tighter 160 tolerance for bone for their work.

Recently, there has been extensive investigation into the usefulness of on-board cone beam CT (CBCT) systems to support image-guided radiotherapy.[46,47] Some studies have assessed the accuracy of HU values produced by these systems and the

Table 2. Summary of Hounsfield unit (HU) change and resultant dose change from experimental investigations on a single tissue type

Energy (MV)	Tissue type and thickness	Planning algorithm (TPS)	PCT or CBCT	Air or lung, HU change	Soft tissue, HU change	Bone, HU change	Dose change (%)	Reference
6	Lung, 10 cm	Hogstrom model (Nucletron®)	PCT	25 (RED change 0.025)	–	–	0.9	Kilby et al[33]
6	Lung, 8 cm	Tissue maximum ratios	PCT	35	–	–	1.0	Thomas[24]
6	Soft tissue, 10 cm	Helax	PCT	–	20	–	0.7	Cozzi et al[42]
15	Soft tissue, 10 cm	Helax	PCT	–	20	–	0.3	Cozzi et al[42]
6	Soft tissue, 3 cm	Helax	PCT	–	20	–	0.1	Cozzi et al[42]
6	Water, 10 cm	Hogstrom model (Nucletron)	PCT	–	30[a] RED change 0.03	–	1.1	Kilby et al[33]
6	Liver, 8 cm	Tissue maximum ratios	PCT	–	30	–	1.0	Thomas[24]
6	Bone, 3 cm	Helax	PCT	–	–	100	0.3	Cozzi et al[42]
6	Bone, 10 cm	Hogstrom model (Nucletron)	PCT	–	–	107[a] RED change 0.055	2.0	Kilby et al [33]
6	Bone, 10 cm	Helax	PCT	–	–	100	1.6	Cozzi et al[42]
6	Cranium bone, 1.5 cm	Tissue maximum ratios	PCT	–	–	540	1.0	Thomas[24]

CBCT, cone beam CT; PCT, planning CT; RED, relative electron density; TPS, treatment planning system.
The Nucletron system was obtained from Nucletron BV, Veenendaal, Netherlands.
[a]HU tolerance calculated using Thomas[24] equations.

Table 3. Summary of Hounsfield unit (HU) change and resultant dose change on clinical plans

Energy (MV)	Type of plan	Planning algorithm (TPS)	PCT or CBCT	Air or lung, HU change	Soft tissue, HU change	Bone, HU change	Dose change in plan (%)	Reference
6	Clinical brain	Collapsed cone convolution (Pinnacle; Philips, Amsterdam, Netherlands)	CBCT	Not given	20	250	<1	Chu et al[43]
6	Clinical brain, five wedged fields	Modified Batho method (Eclipse™; Varian, CA)	CBCT *vs* PCT	Not given	45	Not given	<1	Yoo et al[51]
6	Clinical brain, four conformal fields	Modified Batho power law (Varian Cadplan®; Varian, CA)	PCT	50[a] (RED change 0.05)	30[a] (RED change 0.03)	156[a] (RED change 0.08)	1.0	Kilby et al[33]
6	Clinical lung	Collapsed cone convolution (Pinnacle)	CBCT	Not given	20	250 HU	<2	Chu et al[43]
6	Clinical lung, three field	Modified Batho power law (Varian Cadplan)	PCT	50[a] (RED change 0.05)	30[a] (RED change 0.03)	156[a] (RED change 0.08)	1.3	Kilby et al[33]
6	Clinical lung, VMAT 225°	Collapsed cone convolution (Pinnacle)	CBCT *vs* PCT	−300 to −100 HU	Not given	Not given	−10	Disher et al[50]
6	Clinical lung, VMAT 225°	Collapsed cone convolution (Pinnacle)	CBCT *vs* PCT	−200 to +200	Not given	Not given	+10	Disher et al[50]
6	Clinical lung, VMAT 225°	Collapsed cone convolution (Pinnacle)	CBCT *vs* PCT	200 to +100	Not given	Not given	Close match	Disher et al[50]
6	Clinical pelvis	Collapsed cone convolution (Pinnacle)	CBCT	Not given	20	250	<2	Chu et al[43]
6	Clinical pelvis, five field	Anisotropic analytic algorithm (Eclipse)	CBCT *vs* PCT	100	0	100	<1	Hatton et al[49]
6	Clinical pelvis, seven field IMRT	Anisotropic analytic algorithm (Eclipse)	CBCT *vs* PCT	20	20	500	3.4	Guan and Dong[48]
6	Clinical pelvis, seven field IMRT	Anisotropic analytic algorithm (Eclipse)	CBCT *vs* PCT	20	20	200	0.6	Guan and Dong[48]
16	Clinical prostate, three field conformal	Modified Batho power law (Varian Cadplan)	PCT	50[a] (RED change 0.05)	30[a] (RED change 0.03)	156[a] (RED change 0.08)	1.7	Kilby et al[33]
18	Clinical pelvis, four field	Anisotropic analytic algorithm (Eclipse)	CBCT *vs* PCT	20	20	500	2.4	Guan and Dong[48]
18	Clinical pelvis, four field	Anisotropic analytic algorithm (Eclipse)	CBCT *vs* PCT	2	20	200	0.3	Guan and Dong[48]

CBCT, cone beam CT; IMRT, intensity-modulated radiation therapy; PCT, planning CT; RED, relative electron density; TPS, treatment planning system; VMAT, volumetric modulated arc therapy.
[a]HU tolerance calculated using Thomas[24] equations.

associated impact on planning dose accuracy. Guan and Dong[48] planned on CBCT images of a pelvis phantom using a number of different RED-HU curves. Several 18-MV four-field pelvis treatment plans were produced using the Eclipse TPS with the anisotropic analytic algorithm. Firstly, a RED-HU curve was used which had been acquired on the CBCT system and then a second curve which had been acquired on the PCT. The difference between the two RED-HU curves was that the bone HU was

400 lower for the CBCT curve than the PCT curve. Soft tissue HU values were within 20 for both curves. In the resultant plans for the CBCT image, the central axis dose (D_{cax}) value was 2.3% lower for the one using the CBCT RED-HU curve. Further work was carried out using a third RED-HU curve which had been acquired on the CBCT with a different phantom. Here, the bone HU compared with the PCT RED-HU curve was 200 higher, and in this case, the D_{cax} dose was 0.3% higher. For a seven-field 6-MV intensity-modulated radiation therapy plan, there were larger dose differences, with D_{cax} 3.1% lower where bone HU was 400 lower in the CBCT RED-HU curve and 0.3% higher where bone HU was 200 higher in the CBCT RED-HU curve.

Another study, also using the Eclipse TPS with anisotropic analytic algorithm, used a pelvis phantom and a four-field plan. The work investigated the dose difference at the PTV centre between plans produced with different TPS calibration curves.[49] The baseline plan for the CBCT pelvis image used the RED-HU calibration curve obtained on the PCT. Another RED-HU curve was obtained on the CBCT system and a second plan produced. HU differences on the two calibration curves were, for the CBCT curve, −100 HU for air, 0 HU for soft tissue and +100 HU for bone. This curve gave a dose difference of less than +0.5% at a reference point in the planning target volume centre compared with when the PCT curve was used to plan the CBCT image. Planning on a brain was also investigated, comparing CBCT and PCT plans.[51] The TPS was Eclipse using a Modified Batho Method planning algorithm. Patients had scans acquired on both CBCT and PCT scanners. A single RED-HU calibration curve was used for both sets of plans. The HU values for brain in the CBCT image were 45 HU higher than those in the PCT. This resulted in up to 1% dose difference in the two treatment plans.

Disher et al[50] investigated a number of different ways to modify CBCT HU values for patients with lung cancer. For a 6-MV volume-modulated arc therapy plan on a Rando Phantom, using the collapsed cone convolution algorithm on a Pinnacle TPS, a CBCT image had lung tissue CT values which differed from the PCT image by between −300 and −100 HU. A plan was produced using the RED-HU curve collected on the PCT scanner and also another using a second curve collected on the CBCT system. The dose difference between the plans, based on mapping doses levels across the planning target volume, was −10% on the CBCT plan. A correction method manually assigned a pre-determined HU value to the lung tissue in the CBCT image and changed the range of HU value differences for lung tissue to between −200 and +200 when compared with the PCT scan. The dose difference was then +10%. Another correction method manually assigned a HU value to only pixel values at the lower end of that typically found in the lungs, below −882. This reduced the HU difference to −200 to +100 and resulted in a much improved dose match with negligible difference between the CBCT and PCT plans.

Tables 2 and 3 show the HU and associated changes in radiotherapy dose calculations for lung, soft tissue and bone. A mix of clinical plans and experimental scenarios based on known tissue thicknesses are covered by the articles reviewed.

Scanner variables which affect Hounsfield units

There are many variables in a CT scan protocol, and there is evidence that some but not all parameters will change HU values. Ebert et al[52] investigated the use of the X-ray tube current modulation software on a GE LightSpeed-RT scanner (GE Healthcare, Milwaukee, WI). They looked at different materials including lung and water plus high-density metals such as titanium and stainless steel which are used for prostheses. This study and others concluded that varying the delivered tube current, either manually or by automatic modulation during scanning, resulted in only minimal variation in HU.[27,41,46] None of these publications state the degree of HU variation with tube current. The only exception was in the study by Ebert where a 300 × 160-mm block of solid water containing a stainless steel insert was scanned.[41] Tube current was 100 mA and tube voltage was 120. RED was very high at 6.7. For 200–400 mA, the measured HU was 16,500; for 100 mA, the measured HU was 18,000. It should be noted that 100 mA is an untypical tube current setting for this thickness of tissue. The high degree of noise measured in the stainless steel insert indicates underexposure. The CT number at 80 kV was 22,000 ± 8000 HU compared with 16,500 ± 500 at 120 kV. This study highlighted the need to carefully review exposure settings when high-density materials are scanned but otherwise concluded that tube current modulation could be used.

Studies have generally identified that varying CT tube voltage produces one of the biggest variations of HU. This has been seen on GE LightSpeed-RT, Siemens Somatom Sensation Open and Siemens Somatom AR scanners (Siemens, Erlangen, Germany) when the tube voltage settings were varied between 80 and 140 kV.[2,53,54] For a bone-like material (RED = 1.2), Ebert et al[52] measured approximately 450 HU at 80 kV compared with 280 HU at 140 kV, a difference of 170 HU. For metals, the differences were much greater at >5000 HU. The same degree of HU variation with varying tube voltage has been found on Philips (Philips Healthcare, Netherlands) and Toshiba (Toshiba Medical Systems Co. Ltd, Japan) scanners.[42,56] Kearns and McJury[2] measured HU values of 895, 960 and 1320 for dense bone at 80, 120 and 140 kV, respectively. After processing the images in their TPS, and with reference to electron density tolerances from Kilby et al,[33] they concluded that 80 kV should not be used for planning scans. In practice, 80 kV is only likely to be of use when imaging paediatric patients since the lower beam energy would result in under exposure with adult-sized patients.[55] The use of 80 kV could be accommodated where the planning system allowed more than one calibration curve, provided an appropriately sized phantom was used to produce the RED-HU calibration curve.

On some scanners, the acquisition FOV used can affect the HUs. On a GE Hi Speed DX/i CT scanner, the HU values changed when the switch from a large to small FOV forced a change in the physical filter in the X-ray beam.[54] The degree of change of HUs was not stated. On a Toshiba Aquilion One scanner, there was a change of only 2 HU for water between the small, medium and large FOVs.[57] For cortical bone material of HU value = 1400, another study using a Toshiba Aquilion 16 scanner found the variation to be 2% when switching from 240- to 400-mm FOV.[58] The reason for this was also suggested to be the physical filter which changed at 320-mm FOV. Negligible changes in the HUs

were seen with different FOVs for materials where HU was <100. Understanding the mechanism by which the physical filter is changed in the scan protocol is important because it is not always FOV dependent. Some scanners provide an extended reconstruction FOV to allow imaging of regions of the body which are outside the scan FOV. Evaluations of the Philips Brilliance Big Bore scanner and the GE LightSpeed wide-bore scanner found large differences of approximately 500 HU for bone for the extended FOV compared with the standard FOV.[59,60]

Only minimal change in HU values arising from changes in slice thickness, X-ray tube rotation time and spiral *vs* sequential scanning was noted on a Toshiba Aquilion scanner.[56] The degree of variation was not given. Special reconstruction algorithms FC23 and FC64 on that scanner which used beam-hardening filters did, however, produce variability in the CT values, though the extent of variation is not clearly stated. On a Siemens Emotion Duo scanner, a range of head and body reconstruction algorithms, H10s to H80s, B10s to B90s and U90s were tested.[38] Across the different head algorithms, the maximum difference seen was 25 HU for air and 50 HU for bone when considering 110- or 130-kV scans. The maximum variation for body algorithms was less at <12 HU. Some algorithms produced very little HU difference.

The findings from the literature are summarized in Table 4. The literature does not comprehensively cover all makes and models of CT scanner used in radiotherapy nor the wide variety of possible settings. Scanner performance will always depend on the design, calibration and the settings used.

Although publications related to CBCT have been reviewed in the section on HU change, it is not intended that this review includes an in-depth review of CBCT settings and image quality because the focus is on PCT. CBCT, by nature of the fact that it is a wide-beam imaging technique with high scatter levels when compared with standard CT, produces images with reduced contrast and higher noise levels.[46,62] The current CBCT systems used in radiotherapy also suffer from significant non-uniformity of HU values across the axial plane and artefacts when compared with conventional CT.[63] This variation of HU needs to be considered and appropriately allowed for when collecting HU values from CBCT images.

Impact of scan parameter changes on image quality

Tube voltage, tube current, slice thickness and interval, pitch, reconstruction algorithm, scan time, acquisition and reconstruction FOV are all parameters which will influence image quality in CT.[36] It is well known that reducing tube current will increase noise and reduce the signal-to-noise ratio which can reduce the visibility of low-contrast details.[61,64–69] A reduction of slice thickness has the same effect, though the positive impact of using smaller slice thicknesses is reduced partial volume effect and potentially increased resolution of small details in the longitudinal direction.[61,64,65] An increase in pitch or feed per rotation has the benefit of reducing patient dose but will also result in increased noise, unless tube current is increased to compensate.[55] Varying tube voltage to match the size of the patient will improve X-ray beam penetration and reduce image noise and absorbed dose.[17,55]

The reconstruction algorithms selected affected the image slice width on the Toshiba Aquilion One scanner with smoother algorithms broadening the slice width.[57] Similarly, the resolution in the axial plane also varied significantly depending on the reconstruction algorithm used, with sharper algorithms producing increased noise but improved high-contrast resolution. This has also been seen on other scanner makes and models.[64,66–69] The reconstruction FOV and related matrix will have a significant impact on the visibility of small details.[60,65] A smaller FOV will improve visibility of fine details compared with that seen with a larger FOV. The selection of X-ray tube focal spot size will also slightly influence how well a fine detail is seen in the image with a smaller focus giving improved detail visibility.[64–69] Tube voltage, current and pitch generally influence HU and image quality in a similar manner when varied irrespective of the make or model of the scanner. The changes introduced when changing FOV and the reconstruction algorithms, however, vary considerably from one manufacturer to

Table 4. Summary of scan parameters and level of Hounsfield unit (HU) change in articles reviewed

CT scan parameter	Impact on HU and scanner manufacturers covered by review
Tube current	No change unless very low current used—GE, Toshiba (Toshiba Medical, Zoetermeer, Netherlands)[42,52,53,57]
Kilovoltage	Significant level of HU change—Philips, Toshiba, GE, Siemens (Philips, Amsterdam, Netherlands)[2,42,52,53,57]
Acquisition FOV	Depends on CT scanner make/model and which FOV is selected—GE, Toshiba (GE Healthcare, Milwaukee)[53,57,58]
Reconstruction FOV	Standard FOVs—no information in articles reviewed
	Extended FOVs—significant change across FOV—Philips, GE[59,60]
Slice thickness	Minimal change—Toshiba[56]
X-ray tube rotation time	Minimal change—Toshiba[56]
Spiral *vs* sequential	Minimal change—Toshiba[56]
Reconstruction algorithms	Depends on CT scanner make/model and which algorithm is selected—Siemens, Toshiba (Siemens Healthcare, Erlangen, Germany)[38,56]

FOV, field of view.

another. Reconstruction algorithms, in particular, are generally less well understood by clinical users.

Where the CT data set is used to produce digitally reconstructed radiographs (DRRs), the DRR image quality is determined by both the CT scan parameters and the DRR calculation algorithm.[70] The parameters which will affect DRR image quality are primarily image slice thickness, FOV diameter, total exposure time and focal spot size.[36,69,70] Pitch factor has been shown to affect the ability to see low-contrast objects in the DRR even when effective mAs was kept constant.[71,72] Higher pitches reduce the DRR contrast but also reduce patient dose. Increasingly, the use of DRRs is being replaced by three-dimensional matching, therefore only a few relevant references are included here.

CONCLUSION

From the publications related to planning dose change arising from RED or HU change, the following conclusions can be drawn: a given change of HU or RED will result in a larger change in dose for a greater thickness of tissue than for reduced tissue thickness, therefore the impact of HU change will vary for different body regions; a single-field treatment plan will deliver a greater dose change for a specific HU change than a multiple field plan; the use of lower energy treatment beam results in a higher dose change for a given HU change than the use of higher energy treatment beam. Owing to the proportion of soft tissue in the body compared with bone or air, a change in HU for soft tissue has a greater impact than a change in HU for bone or air. It is well known that the planning algorithm used has an influence on the accuracy of the planning dose. Some are more accurate for treatment of different body regions than others.[23,73] Therefore, any attempts to link HU change to TPS dose change must consider the algorithm used and also the body region. The articles reviewed, however, would suggest that the following HU tolerances could be set to achieve a 1% dose change limit: ±20 HU for soft tissue and ±50 HU for lung and bone. Some publications suggest that it may be possible to allow a higher change than this for bone and still remain within 1% for dose change. It is important to remember that effects of changes must be considered for all tissue types (air, bone, soft tissue) together when present in the clinical plan. These proposed tolerances match the American Association of Physicists in Medicine tolerances for use of CBCT for bone and air and the IAEA tolerances for soft tissue.[30,31,46]

There are clear advantages of having appropriately defined tolerances for HU variation. When adjusting CT scan protocols, it is helpful to know quickly whether changes to scan protocols are likely to be detrimental to the dosimetric aspects of the planning scan. HUs can be easily measured with a phantom on the scanner, thereby allowing early exclusion of inappropriate adjustment to scan parameters. Both image quality and HU changes could be assessed with a multipurpose phantom before undertaking a more detailed check to assess the level of dose change in the TPS with an anthropomorphic phantom. This review has highlighted the need to use phantoms which approximately match the size and shape of patients when measuring HUs.[31,40]

When reviewing the influence of scan protocol settings, published data is sparse. Considering the number of scanners and the variety of settings within CT protocols, the impact of scan parameters in radiotherapy CT is not well detailed in the literature. Publications tend to look at a limited set of scan parameters and only give detailed information on variability when it is considered significant. No publications were found which fully assessed the performance of a radiotherapy CT scanner based on variation in both image quality parameters and HU or RED. The high number of publications supporting optimization in diagnostic CT underlines the fact that scan protocol settings affect image quality.[74,75] The radiation dose delivered from CT imaging must also be considered and justified.[76] The use of scan protocols in radiotherapy CT which are tailored for specific disease sites should, where possible, be used to ensure good-quality imaging with careful assessment made of the dosimetric impact for clinical treatment conditions.[27]

This area of work would benefit from more publications related to the adjustment of CT protocols used in radiotherapy. This should include the assessment of image quality changes in CT planning scans alongside changes in HU values and subsequent dose changes in the TPS. It would also be interesting to investigate whether the effectiveness of autocontouring packages can be improved by CT scan protocol adjustment.

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