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Optimisation of CT scan parameters to increase the accuracy of gross tumour volume identification in brain radiotherapy

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Abstract

Aim: This study aimed to optimise computed tomography (CT) simulation scan parameters to increase the accuracy for gross tumour volume identification in brain radiotherapy. For this purpose, high-contrast scan protocols were assessed.
Materials and methods: A CT accreditation phantom (ACR Gammex 464) was used to optimise brain CT scan parameters on a Toshiba Alexion 16-row multislice CT scanner. Dose, tube voltage, tube current–time and CT dose index (CTDI) were varied to create five image quality enhancement (IQE) protocols. They were assessed in terms of contrast-to-noise ratio (CNR), signal-to-noise ratio (SNR) and noise level and compared with a standard clinical protocol. Finally, the ability of the selected protocols to identify low-contrast objects was examined based on a subjective method.
Results: Among the five IQE protocols, the one with the highest tube current–time product (250 mA) and lowest tube voltage (100 kVp) showed higher CNR, while another with a tube current–time product of 150 mA and a tube voltage of 135 kVp had improved SNR and lower noise level compared to the standard protocol. In contouring low-contrast objects, the protocol with the highest milliampere and lowest peak kilovoltage exhibited the lowest error rate (1%) compared to the standard protocol (25%).
Findings: CT image quality should be optimised using the high-dose parameters created in this study to provide better soft tissue contrast. This could lead to an accurate identification of gross tumour volume recognition in the planning of radiotherapy treatment.

Introduction

Computed tomography (CT) has become a standard imaging modality for the identification of target volumes in treatment planning due to the high incidence of cancers.[1] Target volume contouring on CT images during radiotherapy is the first and vital step towards proper treatment. This CT modality is used not only for volume contouring but also for dose calculations. The position and volume of the tumour and surrounding tissues should be accurately identified during radiotherapy.[1,2] Nowadays, with the introduction of precise radiotherapy techniques such as intensity-modulated radiation therapy (IMRT), image-guided radiation therapy (IGRT) and stereotactic body radiotherapy (SBRT), the demand for high-quality CT images for accurate contouring has increased, although CT imaging suffers from poor soft tissue contrast.[1,3–5] To overcome this limitation, the patient has to undergo two imaging modalities, one with high soft tissue contrast such as magnetic resonance imaging (MRI)[6,7] or positron emission tomography (PET) for an accurate identification of target volumes, and other method such as CT imaging for dose calculation. However, image fusion is rarely performed and sometimes can itself cause errors in target volume identification.[8] The long duration of MRI scanning and its uncomfortable conditions could lead to artifacts. Besides, MRI is not allowed in patients with cardiac pacemakers, metallic cochlear implants or aneurysm.[9] In PET, on the other hand, there is still no consensus regarding accuracy in the identification of tumour region and its relation to radionuclide uptake volume. Moreover, there is still debate over the appropriate radionuclide for each lesion and the proper technique for an accurate recognition of active target volumes.[10–12] Furthermore, as a general rule, as the number of steps between imaging and the beginning of treatment increases, more error is added to each subsequent step.[1] Therefore, it seems that the improvement of CT image quality alone might reduce the need for additional imaging modalities. For radiotherapy of brain tumours, CT images are often fused with MRI images because of a low contrast of CT images, which has its challenges.

In CT scanning and other digital imaging techniques, there is a direct correlation between image quality and radiation dose. Excessive doses reduce noise and eventually decrease image quality.[10] Although CT dose reduction strategies have been used for diagnostic purposes, they may not apply to all patients who need CT scans.[14–16] The radiation dose to patients should be as low as reasonably achievable (ALARA principle) while still providing adequate image quality to enable an accurate diagnosis and treatment.[17] The ALARA principle does not necessarily mean the lowest radiation dose to the patient. Under some clinical conditions, it may be necessary to slightly increase the dosage to obtain appropriate image quality.[16,18,19] The anti-scatter grid in radiography is a device that improves the contrast of images, but it might increase the image receptor dose by a factor of 2–4.[19,20] Moreover, patients who undergo CT imaging for radiotherapy planning may not benefit from dose reduction strategies. The radiation dose to these patients during treatment is much higher than the CT imaging dose (50 Gy vs. 10 mGy).[14,21] The IGRT technique uses additional portal imaging approaches to modify the patient's position during radiotherapy. Hence, CT imaging doses cannot be of much concern to patients even though the scan parameters are optimised to a high-dose scanning protocol in order to improve image quality.[14,16,21] For image quality improvement, parameters such as contrast-to-noise ratio (CNR) and signal-to-noise ratio (SNR) should be modified.[22–24]

Previous studies related to CT imaging in radiotherapy have focused on the impact of scan parameters on the level of Hounsfield unit (HU) alterations that can eventually vary dose calculations.[25,26] Few studies have investigated the effect of image quality improvement on target volume identification. Tomic et al.[24] studied image quality parameters such as SNR, CNR and resolution in various CT simulators. Davis et al. assessed CT image quality using various scan parameters. They concluded that collimation and reconstruction algorithms have considerable impact on CNR.[25] In IGRT, according to the guidelines of the American Association of Physicists in Medicine (AAPM) TG-75 report, CT dosage should be taken into account with regard to the total treatment dose, and efforts should be made to avoid erroneous delivery of a high dose to normal tissues and a low dose to the tumor.[14,15] Furthermore, the high-quality image of IGRT allows fast contouring to be performed accurately during treatment.[5,15] Li et al. introduced a new strategy to optimise CT simulation images based on patient size, treatment planning task and radiation dose.[27] Westerly et al. concluded that an increase in CT imaging dose could lead to amplified CNR, and hence contouring can be accomplished with considerable accuracy.[28]

In order to accurately identify a brain tumour border as well as to reduce dose delivery uncertainties in radiation therapy, the present study is an attempt to optimise CT scan protocols to accurately identify gross tumour volumes (GTVs) of the brain. The reason for choosing brain tumours is that diagnostic CT scans hardly produce acceptable soft tissue contrast, so most patients are referred to MRI fusion. The fusion has its own several challenges. In this regard, the aims of this study was to optimise CT scan protocols to accurately identify GTVs and detect and contour the soft tissue as well as reduce dose delivery uncertainties in radiation therapy

Method

In this study, measurements were undertaken on a similar head-and-neck phantom. A CT accreditation phantom (ACR Gammex 464; Sun Nuclear Corporation, Melbourne, Florida, USA) was used to optimise CT scan parameters and create image quality enhancement (IQE) protocols. This phantom consists of four modules that can test positioning and alignment, CT number accuracy, slice thickness, low-contrast detectability, image uniformity and spatial resolution.[29] Optimisation was performed on a Toshiba Alexion 16-row multislice CT scanner (Canon Medical Systems, Los Angeles, California, USA). This system had already been calibrated according to quality control instructions at local and international levels.[26]

Among a wide range of soft tissue CT scan protocols, five IQE protocols produce high-contrast CT images of the soft tissue with a slightly higher dose compared to standard clinical protocols. Details of these protocols are summarised in Table 1. For all protocols, the matrix size and slice thickness were 512×512 and 3 mm, respectively, and all scans were done in a spiral mode with a soft tissue FC26 kernel. An Adaptive Iterative Dose Reduction 3D (AIDR 3D) algorithm was applied for all protocols. The automatic exposure control system (CARE Dose4D) was turned off in the IQE protocols. CT dose index (CTDI), dose–length product (DLP) and effective dose for each protocol were recorded.[27] The standard protocol had the lowest CTDI (11.1 mGy) and effective dose (0.5 mSv), while protocol 5 had the highest values (50 mGy and 2.5 mSv, respectively). Each CT scan was repeated three times to reduce statistical uncertainties. Furthermore, soft tissue convolution kernels (FC26), with a window width of 100 HU and a window level of 100 HU, were selected for use in the protocols.

The image quality of IQE protocols was compared with the standard protocol to identify the protocol with most optimal scan parameters. Low-contrast resolution, percentage of contrast and high-contrast spatial resolution were directly extracted from the Automated CT Software (ACTS) v.21, and CNR, SNR and noise level were obtained from the images. CNR was measured from two low-contrast regions of interest (ROIs) in objects using Equation (1),[13,31] where A and SD_{inside} are the mean signal and standard deviation of ROIs inside the object, and B and $SD_{outside}$ are the mean signal and standard deviation of ROIs outside the object. For both ROIs, a circle with 20 mm diameter was used for marking the area.

$$\mathrm{CNR} = \frac{2(A^2 - B^2)}{SD^2_{inside} + SD^2_{outside}} \tag{1}$$

To measure SNR and noise level, a series of ROIs in the centre and peripheral areas (top, bottom, left, right, centre) were selected and then measured by Equations (2) and (3), in which $\overline{CT}_i$ and SD_i denote the CT number and standard deviation of peripheral and central ROIs, respectively.[13,31]

$$\mathrm{SNR} = \frac{1}{5}\sum_i \frac{\overline{CT}_i}{SD_i} \tag{2}$$

$$\mathrm{Noise} = \frac{1}{5}\sum_i SD_i \tag{3}$$

To determine the accuracy of low-contrast object identification, the border of low-contrast circular objects with 25 mm diameter and 0.6% (6 HU) difference from the background density were contoured using the standard and optimised high-contrast protocols by an expert radiation oncologist who was blinded to the research objectives. Finally, the contouring areas in the standard and optimised protocols were compared with each other.

Table 1. Parameters of different protocols used in the ACR phantom

Protocol	Voltage (kVp)	Current–time product (mA)	Pitch	CARE Dose4D	$CTDI_{vol}$(mGy)	DLP (mGy.cm)	Effective dose (mSv)
Standard	120	140	0.938	On	11.1	207	0.48
1	135	120	1	Off	34.8	638	1.48
2	120	120	0.7	Off	39.2	681	1.57
3	100	250	0.781	Off	54.5	970	2.25
4	135	150	0.731	Off	59.5	1041	2.40
5	100	250	0.7	Off	60.8	1058	2.45

Results

Among the five high-contrast scan protocols, low-contrast resolution was 2 mm at 0.6 HU difference from the background, and high-contrast resolution was 0.7 lp/cm for all. Changes in mean CNR and SNR for the protocols are illustrated in Figures 1 and 2, respectively. The highest mean CNR (76.7%) for protocol 5 was more than that of the standard protocol, while the highest mean SNR for protocol 4 was slightly higher than that of protocol 5 (0.69 vs. 0.59) whose value was approximately 66.5% higher than the standard protocol. Furthermore, as depicted in Figure 3, maximum noise reduction (59.2%) was also observed in protocol 4, although the noise level of protocols 4 and 5 was approximately identical (13.26 vs. 13.53). Figure 4 shows a comparison of image quality of protocols 4 and 5 with that of the standard protocol, and Figure 5 shows the identification of low-contrast objects in protocol 5 and the standard protocol, and their contouring pattern. The real area of the objects was about 490.5 mm^2, while the areas identified by the standard and No. 5 protocols were about 370 and 485 mm^2, respectively. Thus, the maximum error rates in the identification of low-contrast objects by the standard and No. 5 protocols were about 25% and 0.1%, respectively.

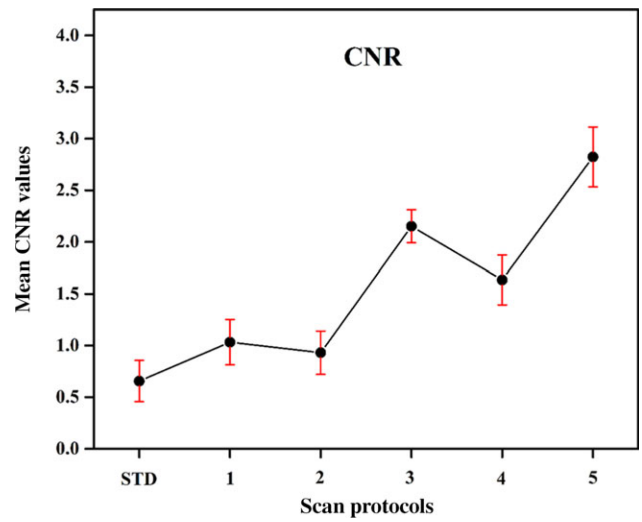

Figure 1. Mean contrast-to-noise ratios of protocols using the low-contrast detectability module of the ACR phantom.

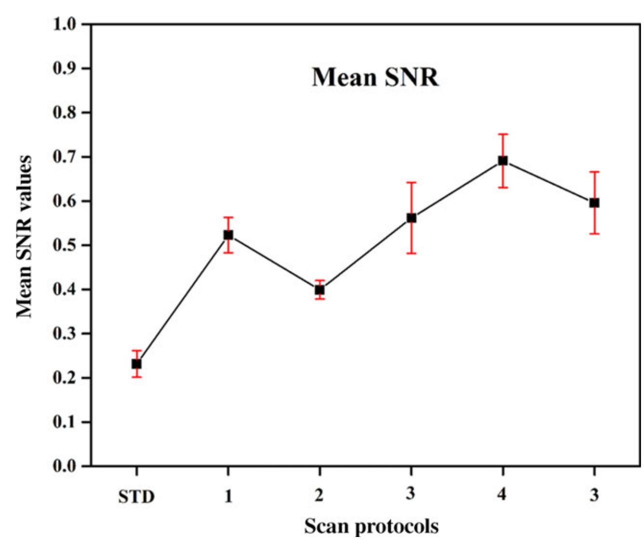

Figure 2. Mean signal-to-noise ratios of protocols using the low-contrast detectability module of the ACR phantom.

Discussion

To increase the accuracy of GTV contouring, CT scan parameters in radiotherapy could be optimised to improve the overall treatment response.[32] In routine CT scanning for brain tumours, the surrounding normal tissue and oedema are barely discriminated from the tumour, so in treatment planning for brain radiotherapy, the accurate identification of tumour border remains a challenging issue. To ensure sufficient irradiation inside the subclinical microscopic areas of regional infiltration, as well as taking into account patient positioning errors, clinical target volume and planning target volume are added to the GTV, where the volume depends on the type and anatomical site of the tumour, clinical systematic and random errors and also image quality. Any error in GTV contouring could lead to underdosing to the tumour and overdosing to normal tissues. As a result, the patient might suffer severe complications or recurrence after a few months. In the present study, the main aim was, firstly, to reduce dose delivery uncertainties in radiation therapy. For this purpose, the study was implemented in two steps. First, the effects of various CT scan parameters on low-contrast detectability were evaluated, and then the accuracy of tumour volume identification and contouring using a standard image quality protocol was compared with that of five IQE protocols capable of producing high-contrast CT images of soft tissues. In our study, among the IQE protocols, protocol 5 had the highest effective dose (2.5 mSv) compared to the standard protocol (0.5 mSv). Using high-dose parameters in CT simulation can improve not only tumour identification but also fusion accuracy of CT with MRI or PET images.[5,14] In IGRT where fast modification of volume changes may be necessary during treatment, a high-quality image plays a vital role in rapid, real-time tumour contouring.[5] At first glance, it may seem that we barely increased the standard scan parameters to higher dose parameters, but it should be noted that

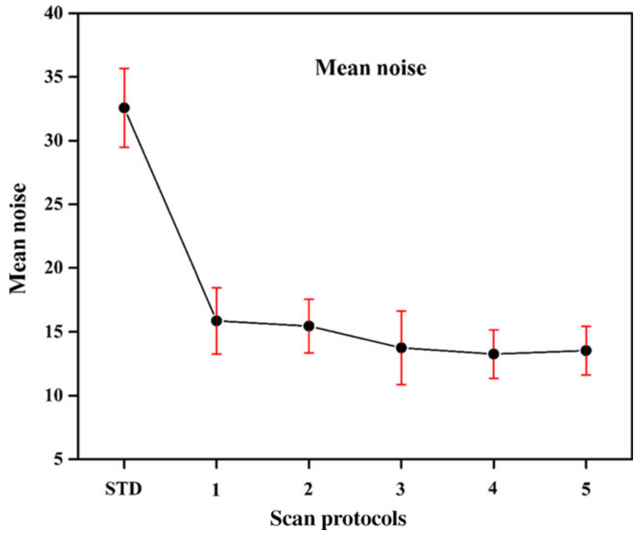

Figure 3. Mean noise levels of protocols using the image uniformity module of the ACR phantom.

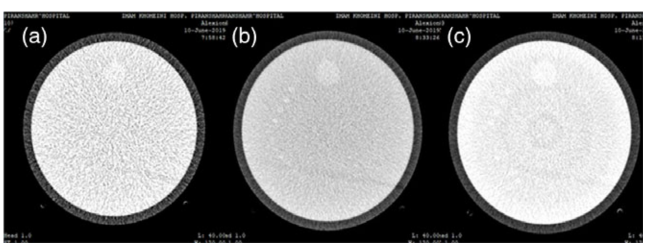

Figure 4. Reconstructed images with kernel FC26 using the low-contrast detectability module of the ACR phantom. (a) Standard protocol, (b) protocol 5 with CT dose index (CTDI) = 60.8 mGy, and (c) protocol 4 with CTDI = 59.5 mGy.

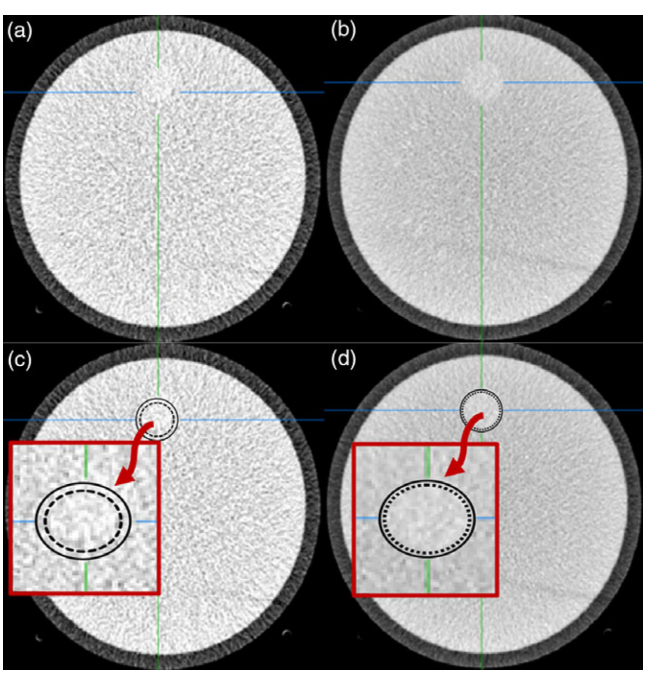

Figure 5. Image quality of a low-contrast object for (a) the standard protocol and (b) protocol 5. The solid line in (c) and (d) represents the real border of the object. The dashed line in (c) represents contouring of the object using the standard protocol. The dotted line in (d) represents contouring of the object using protocol 5.

the ALARA principle considers imaging and treatment simultaneously in radiotherapy.[15] Therefore, when the total dose of both imaging and radiotherapy is considered, the benefits of a slight increase in imaging dose (which is negligible in a very high-dose radiotherapy process) may be clearly justified.[5]

Recently, studies have concluded that CNR and SNR are critical image quality parameters for contouring accuracy and precision.[5] However, few studies have reported CNR improvement in high-dose scan parameters for radiotherapy, and the effect of enhanced CNR and SNR on GTV contouring has not been clearly demonstrated. In our study, the maximum CNR of 2.8 was reported in protocol 5, while the SNR of protocol 4 was slightly higher than that of protocol 5. Noise level is another important parameter that can affect the details of low-contrast objects, eventually leading to a decline in overall image quality.[33] In the present study, the noise level of protocol 4 was reduced by 59.2% compared to the standard protocol, and spatial resolution in all the protocols was 0.7 lp/mm. In general, it is clear that the noise level barely affects spatial resolution and contrast, which can consequently ruin the diagnostic perception of the image. It has been reported that image noise decreases inversely by the square root of the dosage,[23] but in our study, we observed that spatial resolution was rarely affected by dosage. It could be concluded that an objective evaluation of image noise effect on spatial resolution in phantom studies is deficient. The soft tissue reconstruction kernels of CT algorithms eliminate information concerning high spatial frequency and thus create images with low levels of noise and spatial resolution.[23] The low tube voltage and high tube current–time product of protocol 5, compared to those of the standard protocol, caused a high increase in CNR. Chen et al. proposed a moderate tube voltage of 100–120 kVp and an extremely high tube current–time of 1000 mA to improve CNR for abdominal and pelvic CT scans, but high tube loadability (high milliamperes) is not possible for some CT systems and, if implemented, is rarely applicable in clinics due to early depreciation of the system.[5] The maximum SNR in our study was achieved using protocol 4, which may have been due to a slight increase in tube voltage leading to noise reduction and, eventually, SNR improvement. Consistent with this result, Karmazyn et al. found that noise level increased from 0.08 to 0.11 when tube voltage was decreased from 120 to 100 kVp.[34] In the study of Boas and Fleischmann who focused on CT artifacts, by increasing the tube current from 60 to 440 mA, the noise level reduced 2.7 times. Hence, the authors concluded that by gradually increasing the noise level, the images become increasingly blurred, obscuring soft tissue boundaries.[35] Hence, increasing the tube current could be a useful way to cope with image blurring, because of a possible reduction in Poisson noise. Hernandez-Giron et al. investigated the effect of different tube voltages and demonstrated that a tube voltage of 80 kVp could greatly improve low-contrast detectability.[22] The drastic effect of tube voltage on low-contrast detectability improvement was also observed in our study. Although the noise levels of protocols 4 and 5 were approximately the same, the CNR of protocol 5 with low tube voltage was about 42% higher than that of protocol 4.

The identification of GTV border and distinguishing it from OARs, especially in soft tissues such as the brain, is one of the serious challenges in treatment planning. In the present study, the contouring performance of protocol 5 was decided to be evaluated because of its highest CNR. In the subjective analysis, the accuracy of low-contrast circular object contouring was tested using the images created by the standard protocol and protocol 5. The error rate in contouring using the standard protocol was 25%, while for protocol 5 it was only 1%. This may be because of the high CNR and low noise level of images produced by protocol 5. Although CT plays a big role in cancer diagnosis,[36] extensive research is needed to address and eliminate its potential faults.

This pilot study has some limitations. First, image quality parameters were assessed by a phantom based on an objective analysis. Then, a subjective analysis was performed to evaluate the contouring accuracy of low-contrast circular objects. Tube loadability, in order to create high-exposure conditions, is another limitation of this study. Exposure conditions were selected so that they can be practically implemented in the clinic. We used an iterative algorithm in this study. Although a filtered back-projection algorithm is known to improve CNR significantly, the procedure may be time-consuming. For a patient immobilised because of fixation, long-term CT imaging may not be possible. For verification of our findings, further studies should be conducted using the filtered back-projection algorithm.

Conclusion

CT image quality can be optimised using high-dose parameters to provide better soft tissue contrast for treatment planning in radiotherapy. We demonstrated an improvement in the accuracy of GTV identification by this approach.

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Conflict of Interest. None.

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