Clinical implementation of kVCT-guided tomotherapy with ClearRT

Bin Yang1 · Hui Geng1 · Tien Yee Amy Chang2 · Mei Yan Tse1 · Wai Wang Lam1 · Chen-Yu Huang1 · Tungho Wu4 · Ka Ki Lau1 · Kin Yin Cheung1 · George Chiu3 · Siu Ki Yu1

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Abstract
A helical fan-beam kilovoltage computed tomography (kVCT) was recently introduced into Tomotherapy units. This study aims to share the initial experience of kVCT in clinical workflow, compare its performance with that of the existing megavoltage computed tomography (MVCT), and explore its potential in adaptive planning. We retrospectively enrolled 23 patients who underwent both MVCT and kVCT scans. The clinical performance data regarding image acquisition time, nominal dose length product (DLP), registration time and registration corrections were extracted and compared. Image quality was scored by six experienced radiation therapists and quantified based on phantom measurements. CT number stability and the implementation of adaptive radiotherapy were dosimetrically evaluated by performing the dose recalculation on kVCT. Compared to MVCT, kVCT significantly reduced DLP (except the highest kVp protocol), image acquisition and registration time. KVCT obtained higher scores than MVCT on all criteria except artifacts. Phantom measurements also revealed a better image performance characterization of kVCT except for image uniformity. The CT number variation could lead to a dose difference of 0.5% for D95% of target and Dmean of organ-at-risk. For the treatment planning with kVCT, a systematic dose difference (> 1%) in PTV dose metrics was observed at regions with large longitudinal density discontinuities compared to the reference plans. The new kVCT imaging provides enhanced soft-tissue visualization. The improved efficiency with kVCT-guided treatment will allow more patients to be treated each day. In most cases, the dose calculation accuracy of kVCT images is acceptable except for regions with severe artifacts.

Keywords ClearRT · Radixact · kVCT · Tomotherapy · Image quality · Dosimetric evaluation

Introduction

Helical Tomotherapy is an innovative radiotherapy modality utilizing a rotational delivery of intensity-modulated radiation therapy (IMRT) treatment [6, 12]. A Tomotherapy unit combines a linear accelerator with an integrated xenon-filled detector for megavoltage computed tomography (MVCT). Prior to treatment delivery, patient positioning is verified using the integrated MVCT to correct for inter-fractional setup errors. MVCT uses a 3.5 MV imaging beam that is lower than the nominal 6 MV treatment beam. All MVCT scans are acquired with the same collimation width of 4 mm in the isocenter plane regardless of the pitch selection. The imaging field of view (FOV) of MVCT is limited by the width of the multileaf collimator that projects to 400 mm at the isocenter [7].

Meeks et al. [7] conducted a comprehensive study on the performance characterization of MVCT. They concluded that the relatively low-dose MVCT scans provided sufficient contrast for delineating a range of soft-tissue structures, although MVCT was inferior to the diagnostic kVCT scanners in terms of noise and low-contrast resolution. De Marco et al. [3] evaluated the image quality and imaging dose of MVCT with varying acquisition and reconstruction parameters and concluded negligible differences in image quality among different pitch and reconstruction intervals. Apart from the application in image guidance, some studies reported using
MVCT in radiation treatment simulation for patients with bilateral femoral prostheses and extensive dental fillings [1, 2, 13]. The feasibility of dose recalculation based on MVCT images has been explored [7, 11]. The performance of slow MVCT for delineating the excursion boundary of a moving target was also evaluated by Smeenk et al. [9].

Radixact Treatment Delivery System (v3.0.0, Accuray Incorporated, Sunnyvale, USA), the next generation Tomotherapy platform, can now be retrofitted with a helical fan-beam kilovoltage computed tomography (kVCT) imaging system, ClearRT, for kVCT-guided treatment. This new kVCT system consists of a kV x-ray tube installed at 267.5 degree and a CsI: Ti detector panel containing 2880 × 2880 pixels (0.15 mm per pixel). The source-to-axis distance (SAD) is 1040 mm, and the source-to-panel distance (SPD) is 1550 mm. The dynamic panel position allows for a maximum FOV of 500 mm.

Velten et al. [10] has recently assessed the image quality metrics (IQMs) of ClearRT and compared it to that of a kVCT simulator and linear accelerator-based cone beam CTs (CBCTs). They concluded that ClearRT had comparable performance to the existing kVCBCT and clinical acquisitions showed improved clarity in ClearRT over MVCT. However, a quantitative comparison between this new kVCT technique and MVCT is lacking, and there is no dosimetric evaluation in their assessment. The purpose of this study is to report the initial experience of ClearRT in clinical routine and compare its clinical performance with that of the existing MVCT on the Radixact system. We also evaluate the daily patient dose recalculation based on kVCT and give an insight into its potential for inter-fraction dose adaptation.

Materials and methods

Patient characteristics

This retrospective study was reviewed and approved by HKSH Medical Group Research Committee (RC-2021-24). Our institution has been implementing ClearRT for clinical use since it was installed on a Radixact system at the end of June 2021. A total of 23 patients treated between June and August 2021 were retrospectively enrolled in this study. Each patient underwent at least one MVCT scan and one kVCT scan during the treatment course. Only one single scan per fraction (MVCT or kVCT) was performed for ethical reasons. Among them, tumors in the head (6), thorax (9) and pelvis (8) regions were treated. No patient had metal implants or prostheses.

Workflow performance metrics

The kVCT and MVCT were compared by examining scanning time, nominal dose length product (DLP), CT dose index (CTDI), registration time and translational (IEC X, Y and Z) and rotational (IEC Roll only) corrections in patient positioning. In a helical scan, DLP was the product of CTDIvol and the couch travel during the scan. Because there was an over-scan distance on each end of the scan range in kVCT, the selected slice range in the system interface could not be directly used for deriving the DLP. For each scan, both DLP and CTDIvol were indicated by the system. MVCT and kVCT data were extracted from two consecutive fractions in order to minimize the inter-fractional anatomic change, and the same radiation therapist performed the registration on both fractions. Although an iterative MVCT reconstruction algorithm was introduced at the launch of the Radixact system, the Standard reconstruction algorithm was adopted in this study for a general comparison. Key parameters of the most used kVCT and MVCT imaging protocols in our institution are summarized in Table 1. Unlike the data in Velten et al. [10] that mainly focused on Fine and Coarse mode, our institution preferred Normal mode for the clinical workflow. Wilcoxon signed-rank tests were performed to determine whether significant differences existed between the two cohorts.

We also conducted a survey. Six experienced radiation therapists were invited to score the performance of kVCT and MVCT (where a score of 1 represented the worst performance, and 5 indicated the best performance) regarding signal-to-noise ratio (SNR), the frequency and extent of artifacts including beam hardening, motion, aliasing, ring, scatter artifacts [8], soft-tissue differentiation and overall experience of the image registration. The scoring of soft-tissue differentiation was mainly based on the low-contrast visibility, particularly in regions close to the targets and critical organs.

Image performance characterization

The CTP515 low-contrast module in a Catphan phantom (Model 504, The Phantom Laboratory, Greenwich, USA), which provided nominal target contrast levels of 0.3%, 0.5% and 1.0%, was scanned to check the visualization of low-contrast targets. The discernible circle with the smallest diameter was recorded.

Image uniformity and noise were evaluated by scanning the uniform region of a 30 cm diameter cylindrical Solid Water Tomo-Phantom HE (‘Cheese Phantom’, Sun Nuclear, Melbourne, USA). Image uniformity is defined as the maximum CT number difference (in Hounsfield units) between the peripheral and central regions of interest (ROIs) (20 cm²). Noise is expressed as the pixels’ standard deviation (SD) in a large central ROI (400 cm²). In addition, the dark horizontal stripe where the two halves of the Cheese Phantom abutted was intentionally excluded in the central ROIs.
Geometrical distortion was evaluated by measuring the distances between the fiducial markers inside the Cheese Phantom. The kV to MV alignment accuracy was assessed by comparing the IEC X, Y and Z offsets of kVCT and MVCT after manually aligning a dedicated fiducial with the laser center during image registration.

In addition, spatial resolution was evaluated by scanning a high contrast resolution plug inside the Cheese Phantom, which has seven sets of five holes with a diameter ranging from 2 to 0.8 mm. The diameter of the smallest pinholes visible in the image, with each hole distinguishable from neighboring holes, represented spatial resolution.

Another phantom with different density plugs (Model 062 M, CIRS, Norfolk, VA) was scanned to create the CT number versus electron density calibration curves and evaluate the SNR and the contrast-to-noise ratio (CNR) for all protocols. SNR is defined as the mean divided by the SD of the pixels in a small ROI (2 cm²), and CNR is calculated by dividing the pixel difference between soft-tissue and water ROIs with the SD of water ROI.

All the phantoms mentioned above and examples of their kVCT images are shown in Fig. 1.

CT number stability and its dosimetric impact

The CT number stability was monitored by scanning the CIRS phantom weekly with different imaging protocols for over two months. The routine machine quality assurance (QA) of CT number calibration (water and air only) was also performed weekly using the Cheese Phantom as recommended by the vendor. The purpose of running this procedure regularly was to monitor and improve the stability of Hounsfield Unit (HU) values. In order to quantify the dosimetric impact of the CT number variation, three clinical plans with treatments sites in the head & neck, thorax and pelvis regions were recalculated using the upper and lower range of CT calibration curves collected over time. Dose metrics of these plans were recorded and compared after that.

Dosimetric evaluation of planning with kVCT

For further exploring the possibility of dose assessment and plan adaptation using kVCT, as illustrated in Fig. 2, six clinical plans were created based on images of an anthropomorphic phantom (RANDO Woman Phantom, The Phantom Laboratory, Salem, USA) scanned with a CT-simulator (Siemens Somatom Confidence, Siemens Healthineers, Forchheim, Germany). Treatments sites were located in the brain, nasopharynx, lower neck, lung, T-spine, and pelvis. The prescription of all plans was 60 Gy to 95% of the planning target volume (PTV) in 30 fractions, and plan criteria followed institutional
dosimetric protocols. Especially, PTVs were contoured in regions where kVCT artifacts might occur for the lower neck, T-spine, and brain plans. The same RANDO phantom was scanned on the treatment position using different kVCT imaging protocols (Table 1). A dose recalculation and dose metrics comparison were carried out using the original plans as a reference after these kVCT images were imported into the treatment planning system.
Results

Workflow performance metrics

As shown in Fig. 3a, the scanning time of kVCT (mean < 35 s) is significantly reduced by 78% compared with MVCT. The average scanning time saving is 86.4, 135.6 and 131.0 s for the Head, Thorax and Pelvis protocol, respectively. Compared with the registration using MVCT, the mean registration time with kVCT were shortened by 26.3, 60.5 and 94.6 s for the three protocols, respectively. Although the registration strategy was to align the bony anatomy that had good contrast in both kVCT and MVCT, the inferior image quality of MVCT in the superior–inferior direction increased the registration time as reported by radiation therapists. The average registration corrections of the whole group were 0.3 (−9.5–9.9), 0.6 (−3.9–5.0), 0.5 (−6.7–11.0) mm and 0.1 (−2.0–2.1)° in IEC X, Y, Z directions and Roll angle rotation, respectively. The differences in registration corrections between kVCT and MVCT on the same patient were −0.3 (−4.8–4.2), 0.2 (−5.6–4.2), 0.1 (−4.7–5.4) mm and −0.1 (−2.5–2.5)°. The kVCT and MVCT registration corrections did not differ statistically (all p > 0.3), indicating no significant difference between the two cohorts other than the inter-fractional setup error.

Figure 3b indicates that kVCT’s nominal DLP is less than the MVCT except for the Pelvis protocol that uses the highest kV setting.

The survey results (Fig. 3c) show that, except on imaging artifacts, all observers reach a consensus that kVCT is superior to MVCT concerning SNR, soft-tissue differentiation and overall experience of the image registration. Examples of the most common artifacts observed in clinical scans, including gas motion streak, CT number reduction caused by large longitudinal density discontinuities and photon starvation, are presented in Fig. 4a–c. It is worth noting that extensive CT number reduction was observed at the transition from air to shoulder/cranium, which agrees with the findings by Velten et al. [10]. As ClearRT image reconstruction uses the data ~2 cm superior and inferior to the primary beam as an input of the scatter correction, such artifact may occur at regions with large longitudinal density discontinuities. Figure 4d–f and g–i show examples of kVCT and MVCT scans in the abdomen region of the same patient. MVCT presents inferior image quality in the superior-inferior direction compared with kVCT.

Image performance characterization

Key IQMs obtained from phantom measurements are tabulated in Table 2. KVCT shows better imaging performance than MVCT except for the uniformity test, where all three kVCT protocols exhibit worse uniformity than MVCT. The inferior result of the low-contrast test for MVCT was mainly attributed to the dominant Compton scattering effect. Both imaging modalities showed minor geometrical distortion (<0.8 mm) and the kV to MV alignment was better than 0.5 mm.

CT number stability and its dosimetric impact

Figure 5 shows the variations of CT calibration curves for different imaging protocols. The CT number of MVCT is more stable than that of kVCT. For non-bone materials (mass density < 1.1 g/cc), the maximum CT number variation is ±15, ±10, ±13 and ±5 HU for the Head, Thorax, Pelvis and MVCT protocols, respectively. The CT number variation of kVCT has an increasing trend with the mass density of bone materials, resulting in the most significant variation of ±46 HU for the bone 1250 plug (mass density = 1.82 g/cc) with the Pelvis protocol. Throughout the study period, our results comply with the tolerance of the CT number constancy test as recommended in AAPM TG-148.
Table 3 summarizes the changes in dose metrics of the three clinical plans due to the CT number variations. D95% (dose to 95% of the volume) and Dmax (the maximum dose) of the PTV differ by no more than 0.5% and 0.6%, respectively. The largest variation in Dmax and Dmean (mean dose) of organ-at-risk (OAR) is 0.6% and 0.5%, respectively. Our findings agree with the calculation by Kilby et al. [4] and are less than the estimation in AAPM TG-148 [5]. If the CT calibration curve is not created based on the mean CT number, the worst scenario could double the dosimetric variation.

**Dosimetric evaluation of planning with kVCT**

Differences in dose metrics between kVCT recalculated and reference plans are summarized in Table 4. Except for the brain and T-spine plans, the dose metric variations of both PTV and OAR were within 1%. The difference between ΔD95% and ΔDmax in the lung plan is probably due to the residual registration error as the lung PTV has the smallest volume of 1.8 cc among all plans. Artifacts in brain and T-spine regions could result in a systematic dose difference (> 1%) in PTV dose metrics. For lesions situated in these two regions, care should be taken when evaluating the dose.

**Fig. 4** Examples of the most common artifacts as indicated by red arrows. a Gas motion streak; b CT number reduction caused by large longitudinal density discontinuities; c Photon starvation and longitudinal density discontinuities. Examples of kVCT (d–f) and MVCT (g–i) images in the abdomen region.

**Fig. 5** Variations of CT calibration curves for different imaging protocols over 2 months.
Discussion

This study highlighted the initial clinical experience and image performance characterization of ClearRT, the new kVCT in Tomotherapy. Compared with MVCT, the scanning time and the registration time of kVCT were significantly reduced, resulting in improved treatment throughput and efficiency. Compared with MVCT, the improved image quality of kVCT in noise and soft tissue contrast facilitated precise dose delivery, leading to more preferable and practical image-guided adaptive radiotherapy. A more comprehensive and systematic dosimetric evaluation of kVCT in adaptive radiotherapy is beyond the scope of this note and will be the focus of our future work. In spite of the improved soft-tissue visualization, some artifacts, such as beam hardening and metallic artifacts, become more prominent in kVCT than in MVCT. Currently, metal artifact reduction was not included in the ClearRT reconstruction algorithms. The vendor also warned that attention should be paid to metal artifacts when performing dose calculations on ClearRT images. In the energy range of MVCT, the Compton scattering effect dominates in attenuating the beams, which is almost independent of the atomic number (Z). In contrast, the photoelectric effect dominates in the kVCT energy range, largely dependent on the atomic number by \( Z^3 \) for high Z material. As a result, metal artifacts can be significantly reduced with MVCT. As reported by Aubin et al. [2, 13], the considerable metal artifacts on kVCT images obscured the anatomical boundaries of targets and OARs. Therefore, for patients with metal implants or prostheses, workflow with MVCT is preferred.

The mode setting in kVCT protocols (Fine, Normal and Coarse) determines the longitudinal beam width, couch speed and views per gantry rotation. Velten et al. [10] has revealed some differences in IQMs between Fine and Coarse mode, and claimed Coarse mode yields similar IQMs to MVCT. We also found mode selection in kVCT significantly impacts CT number uniformity and stability between Fine and Normal mode. Particularly, the uniformity for Fine mode in kVCT is comparable to that of MVCT, but the CT number variation in Fine mode is half of that in Normal mode. Such results can be explained by the relatively large residual scatter error of Normal mode that utilizes a wider beam width (100 mm) than Fine mode (50 mm).

Dose calculation inaccuracy due to the longitudinal density discontinuities occurs from air to shoulder/cranium. These artifacts are caused by the intrinsic reconstruction algorithm of ClearRT, and a simple solution does not exist at the current stage. Therefore, dose evaluation or adaptation based on MVCT is recommended for lesions located in such areas. In addition, the phantom size also

| Protocol | Uniformity (HU) | Noise (HU) | Spatial resolution (mm) | Low-contrast SNR | Supra-slice CNR | Subslice CNR | Lung inhale | Lung exhale | Water | Lung initial | Spleen-slice | Stomach | Liver | Liver | Plastic water | Adipose | Breast | Brain | Brain 1250 | Brain 200 |
|----------|----------------|------------|------------------------|------------------|-----------------|--------------|-------------|------------|-------|-------------|------------|---------|-------|-------|----------------|--------|--------|-------|-------------|----------|
| kVCT Head | 21.1           | 22.3       | 2.0                    | 0.5              | 0.3             | 7.0          | Lowest      | Lowest     | Lowest | Largest     | Lowest     | Lowest | Lowest | Lowest | Lowest    | Lowest | Lowest | Lowest | Lowest  | Lowest  |
| kVCT Thorax | 20.4          | 18.0       | 1.2                    | 0.5              | 0.3             | 5.0          | Lowest      | Lowest     | Lowest | Lowest     | Lowest     | Lowest | Lowest | Lowest | Lowest    | Lowest | Lowest | Lowest | Lowest  | Lowest  |
| kVCT Pelvis | 23.0          | 18.2       | 1.2                    | 0.5              | 0.3             | 4.0          | Lowest      | Lowest     | Lowest | Lowest     | Lowest     | Lowest | Lowest | Lowest | Lowest    | Lowest | Lowest | Lowest | Lowest  | Lowest  |
| MVCT      | 8.4            | 44.8       | 1.4                    | 0.5              | 0.3             | None         | None        | None       | None  | None        | None       | None   | None   | None   | None      | None    | None   | None   | None    | None    |
showed a nonnegligible impact on the CT number accuracy and stability. In scans with a half (semicircle) Cheese Phantom, the CT number variation of bone material could be half of the presented value in this study, which could also be interpreted by the change of scatter condition and the beam hardening in phantoms with different sizes. Concerning the possibility of implementing adaptive therapy using kVCT for a large-size patient, Fine mode protocols could be a better option for reducing the dosimetric uncertainty.

To benchmark and compare the imaging dose, the multiple scan average dose (MSAD) was also measured for each imaging protocol using a calibrated ion chamber (Model A1SL, Standard Imaging, Middleton, WI) placed at the center of the Cheese Phantom with a scan range covering the whole phantom. The measured MSAD was 0.30, 0.86,
1.24 and 1.32 cGy for the Head, Thorax, Pelvis and MVCT protocols, respectively. Except for the Head protocol, the difference between the MSAD and the reference CTDI$_{vol}$ (Table 1) was less than 0.1 cGy. It should be noted that the nominal DLP of the Head protocol was derived from measurements in a 16 cm CTDI phantom. The mean DLP of Head protocol in a 32 cm CTDI phantom was close to 6.6 cGy•cm based on the in-phantom MSAD measurements.

There is so far no intentional guideline regarding the QA of kVCT in Tomotherapy. Nevertheless, those recommendations in AAPM TG-148 [5] for MVCT apply to kVCT. Especially regarding the CT number stability, it is still recommended to follow a tolerance of less than 30 HU for waterlike materials and less than 50 HU for lung and bone-like materials, considering the dose calculations on kVCT images. Apart from the QA items of geometric distortion, noise & CT number uniformity, spatial resolution and contrast, the routine QA of kV to MV alignment accuracy is recommended. In addition to MSAD, the in free air measurement with an A1SL ionization chamber can also be performed to benchmark or measure the imaging dose of kVCT.

**Conclusions**

Compared with MVCT, the recently introduced kVCT in a Radixact system improves clinical workflow with significantly reduced image acquisition and registration time. The nominal DLP is also reduced for kVCT protocols with low and intermediate kV. The quality of kVCT images was scored superior to that of MVCT in clinical routine except for the frequency and extent of imaging artifacts. Despite CT number uniformity and stability that heavily depend on scatter conditions, kVCT shows a better image performance characterization than MVCT based on phantom measurements. The CT number variation could result in dosimetric errors of up to 0.5% concerning D95% of target and D$_{mean}$ of OAR if kVCT images were used for dose calculation. In most cases, the accuracy of the dose calculation can be assured, but artifacts may cause significant dose deviations at the region adjacent to the transition from air to shoulder/cranium.

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**Declarations**

**Conflict of interest** The authors have no relevant conflicts of interest to disclose.

**Ethical approval** This retrospective study involving human participants was reviewed and approved by HKSH Medical Group Research Committee (RC-2021-24).

**Consent to participate** Written informed consent for participation was not required for this study in accordance with the national legislation and the institutional requirements.

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