Original Article

Megavoltage cone beam computed tomography: Commissioning and evaluation of patient dose

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ABSTRACT

The improvement in conformal radiotherapy techniques enables us to achieve steep dose gradients around the target which allows the delivery of higher doses to a tumor volume while maintaining the sparing of surrounding normal tissue. One of the reasons for this improvement was the implementation of intensity-modulated radio therapy (IMRT) by using linear accelerators fitted with multi-leaf collimator (MLC), Tomo therapy and Rapid arc. In this situation, verification of patient set-up and evaluation of internal organ motion just prior to radiation delivery become important. To this end, several volumetric image-guided techniques have been developed for patient localization, such as Siemens OPTIVUE/MVCB and MVision megavoltage cone beam CT (MV-CBCT) system. Quality assurance for MV-CBCT is important to insure that the performance of the Electronic portal image device (EPID) and MV-CBCT is suitable for the required treatment accuracy. In this work, the commissioning and clinical implementation of the OPTIVUE/MVCB system was presented. The geometry and gain calibration procedures for the system were described. The image quality characteristics of the OPTIVUE/MVCB system were measured and assessed qualitatively and quantitatively, including the image noise and uniformity, low-contrast resolution, and spatial resolution. The image reconstruction and registration software were evaluated. Dose at isocenter from CBCT and the EPID were evaluated using ionization chamber and thermo-luminescent dosimeters; then compared with that calculated by the treatment planning system (TPS- XiO 4.4). The results showed that there are no offsets greater than 1 mm in the flat panel alignment in the lateral and longitudinal direction over 18 months of the study. The image quality tests showed that the image noise and uniformity were within the acceptable range, and that a 2 cm large object with 1% electron density contrast can be detected with the OPTIVUE/MVCB system with 5 monitor units (MU) protocol. The registration software was accurate within 2 mm in the anterior-posterior, left-right, and superior-inferior directions. The additional dose to the patient from MV-CBCT study set with 5 MU at the isocenter of the treatment plan was 5 cGy. For Electronic portal image device (EPID) verification using two orthogonal images with 2 MU per image the additional dose to the patient was 3.8 cGy. These measured dose values were matched with that calculated by the TPS-XiO, where the calculated doses were 5.2 cGy and 3.9 cGy for MVCT and EPID respectively.

Key words: Dosimetry, EPID, IMRT, IGRT, image quality, MV-CBCT, Thermo-luminescent dosimeters

Introduction

A number of studies have proposed that greater doses, in the range of 70-74Gy, for prostate cancer patients in the postoperative setting would potentially achieve greater disease free control rates and are likely to be safe, if image guided radio therapy (IGRT) techniques are used.[1] Intensity-modulated radio therapy (IMRT) with daily localization for salvage or adjuvant RT would allow for reducing the PTV margin and so for dose escalation. This margin reduction will minimize toxicity and spare more organs at risk. A number of online IGRT methods have been developed to correct for positional variations of the prostate-bed, rectum, and bladder volumes. They include ultrasound (US) imaging, implanted fiducial markers, and cone-beam CT (CBCT).[1] Kilo voltage (kV) and megavoltage cone beam computed tomography (MV-CBCT) fitted on a standard linear accelerator are promising and can provide ultra-fast online volume image guidance with low imaging dose and sufficient image quality. These IGRT devices can be safely applied for patients with lung cancer under breath hold.[2]

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The development of image-guidance tools and techniques in radiotherapy has been greatly motivated by the continuous advances in external beam radiation delivery. With 3D conformal radiotherapy and IMRT, it is now possible to deliver radiation doses that confirm to the tumor volume. Many clinical studies and simulations indicate that these more precisely conformal, higher dose treatments can decrease both the spread of disease and normal tissue complications.[3-7]

The technology of the online MV-CBCT imaging is currently used in many institutions to generate a 3D anatomical dataset of the patient in treatment position and to account for organ motion and set-up variations. It utilizes an accelerator therapy beam, delivered with 200° gantry rotation, and captured by an electronic portal imager.

To make IGRT practical in a busy clinic and to use it for the reduction of treatment toxicity by margin reduction, it must be smoothly integrated into the patient set-up process. Quality control of these imaging modalities is a newly added task for physicists and therapists. Manufacturer's QA guidelines must be followed, and test tools must be used regularly.

To estimate the patient dose in the online MV-CBCT imaging protocol, a comprehensive series of absolute dose measurements from the OPTIVUE/MVCB delivery on cylindrical and anthropomorphic phantoms were performed and analyzed, using both ionization chambers and TLD. Dose delivery were simulated in the XiO treatment planning system (Computerized Medical Systems, St Louis, MO), by creating 40 fields and distribute the 5 monitor units (MU) of the imaging protocol on the 40 fields to make an arc beam mimicking the CB delivery and placing interest points at the isocenter detector locations. The goal was to evaluate the reproducibility of the cone-beam delivery in a treatment planning system, thereby allowing the incorporation of the MV-CBCT dose into the treatment plan.[8,9]

In this work, the procedures of commissioning of the MV-CBCT system are presented. The additional doses from MV-CBCT have been measured using TLD and ionization chamber. An assessment of was made to determine if it is necessary to subtract the measured dose from the prescribed dose of the patient or ignore it.

Materials and Methods

Linear accelerator and imaging system

The linear accelerator used in this study was ONCOR, (Siemens Medical Solutions, Malvern, PA) with double focused MLC delivery system. The Linear accelerator is equipped with an amorphous silicon flat panel fabricated for MV photons. The $41 \times 41\ cm^2$ flat panel X-ray detector (AG9-ES, PerkinElmer, Optoelectronics) consists of a 1 mm copper plate and a Kodak Lanex Fast scintillator plate (Gd2O2S: Tb) overlaid on top of light-sensing and charge integrating thin-film transistor (TFT) array. The flat panel has 1024×1024 TFT detector elements with a pixel pitch of 0.4 mm. The detector is mounted on a retractable support which deploys in less than 10 seconds with a positional reproducibility of 1 mm in any direction (7). The entire imaging system operates under a prototype SYNGO™ based COHERENCE™ therapist workspace, which communicates to the control console, the linac and a local patient database. The workspace contains applications allowing for the automatic acquisition of projection images, image reconstruction, CT to CBCT image registration, and couch position adjustment. Each projection of the CBCT acquisition was corrected for defective pixels as well as for pixel-to pixel offset and gain variations before 3D reconstruction. The CT data set from MV-CBCT system equipped on the ONCOR linear accelerator can be acquired with one of three protocols. The first one delivers 4MU in arc of 200° to performing the CT data set. The second and third one use 8 and 15 M, respectively.

Flat panel commissioning tests

Flat panel safety interlocks

The tests covered the EPID system interlocks and imaging arm touch-guard interlock check. For proper operation of the imaging system, many of the safety tests should be performed. The most important tests are those for mechanical alignment stability and accuracy of the EPID and MV-CBCT. These were measured using electronic radiographs taken on daily basis. System interlocks were checked on room door; beam on, and with termination key. EPID imaging arm-touch guard-interlock switch was checked by applying low pressure to each of the four corners of the touch guard. Action on each touch guard corner activates interlock, and the corresponding movement stops. Gantry movements were checked when the touch guard is activated. System movement interlock was checked by opening and closing EPID panel laterally and longitudinally.

Flat panel calibration

Alignment calibration

The accuracy of patient position measurement strongly relies on the precision with which the flat panel was aligned. The physical position of the detector was carefully set with respect to the mechanical isocenter of the machine. However, residual longitudinal, vertical, and rotational misalignments must be accounted for. For this purpose, a reticule was delivered with the machine for quality assurance. The reticule (named "Xretic") which consists of two orthogonal tungsten wires, was inserted in its slots in the gantry head, so that the crossing of the two wires corresponded exactly to the beam central axis [Figure 1a] (10). A series of portal images at four gantry angles (0, 90, 180, and 270 deg) and four different sources to image

distances (SIDs), (130, 140, 150, and 160 cm) were acquired [Figure 1b]. For each gantry angle/SID combination, the position of the projection of the wires on the flat panel was compared to the position of the central row and column of pixels of the detector, and the residual misalignments were calculated and stored. Additionally, a second series of portal images at SID of 145 cm was acquired at eight different gantry angles (0, 45, 90, 135, 180, 225, 270, and 315 deg). The residual offsets from this series were used specifically for MV-CBCT images. This alignment test was performed as the daily EPID QA.

Pixel calibration

The pixel offsets are automatically updated every 5 min by averaging ten images for each integration time in steps of 200 ms between 285 and 1485 ms, when the beam is off. To obtain the gains for portal imaging, an image was acquired with a non-attenuated beam with a known dose at low and high dose rates for the 6 and 10 MV photon energies at seven different SIDs: 120, 130, 140, 145, 150, 155, and 160 cm. The gains were derived from the known dose deposition in each pixel and the offset-corrected pixel values. When the panel is used at a different SID, the gains are linearly interpolated between the two closest calibration distances. A dead pixel map was then created using 100 images acquired with a non-attenuated beam, accounting for defective pixels. A pixel was marked as defective if its white noise (as measured by the standard deviation of its value over 100 white images) was more than six times higher than the overall standard deviation over the entire panel; if its dark noise was more than six times higher than the overall standard deviation over the entire panel; or if its corrected value for offset and gain deviated by more than 1% from the median value of its 9×9 neighbors. The value for a dead pixel was replaced by the average value of the active adjacent pixels. An additional gain calibration procedure was performed specifically for the cone beam mode: A non-attenuated cone beam arc was acquired with the 15 and 60 MU protocols, and the gains were derived for each of the 200 projected frames.

Mega voltage cone beam ct (mv-cbct) commissioning

The manufacturer had defined the commissioning procedures which include the following:[10]

- Setting the cone beam 15MU protocol kernel to "Smoothing-head and neck" filtering and verifying the other protocol parameters
- Checking cone beam calibration test using calibration phantom.
- Creating a cone beam field for image quality checks
- Positioning the image quality phantom and acquiring the image quality image set
- Checking cone beam geometry correctness
- Checking cone beam noise and artifacts
- Checking cone beam spatial resolution
- Checking cone beam low contrast resolution

MV-CBCT calibration

The Geometrical phantom manufactured by Siemens contains 108 X-ray-opaque tungsten ball bearings (BB's) of two sizes, small and large (3.2 and 6 mm diameter, respectively), which are embedded to form a single helix as shown in Figure 2a. Their known coordinates are used to calculate the transformation matrices projecting the three-dimensional voxels in the reconstruction volume to a two-dimensional pixel on the flat panel. For geometrical calibration accuracy test, a CT set for the Geometrical phantom was acquired and the MV-CBCT system was calibrated to the manufacturer's specifications. In this test, the software makes the registration of the acquired CT set of the geometrical phantom by fusing

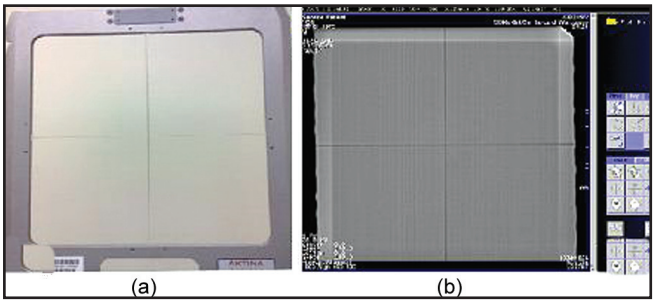

Figure 1a,b: The reticule and its electrical radiograph showing the alignment between the EPID isocenter and the linear accelerator isocente

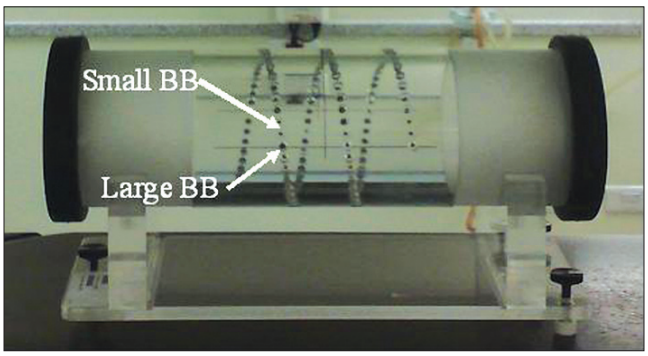

Figure 2a: The geometrical phantom of the MVCB-CT

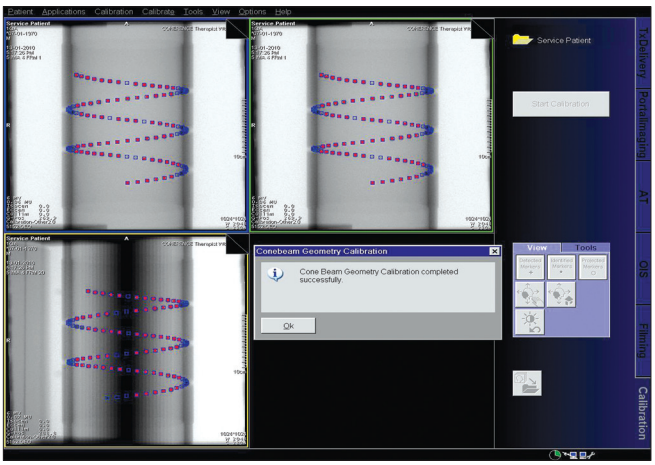

Figure 2b: The reference calibration phantom on the left top, the calibration phantom image on the right top, the overlapping of the acquired geometrical phantom image with the reference image on the left bottom

with the standard image of that phantom. A quality factor representing the mean geometric error in BB positions was used to classify the projection matrix as valid or invalid, and a correction for failing projections was used by interpolation. Figure 2b shows the overlapping of the acquired geometrical phantom image with the standard image from the software data base. The software gives a message about the performance of the calibration.

Image quality

Slice and point position accuracy

The image quality phantom was carefully aligned on the tabletop using the Xretic as shown in Figure 1a, by matching the projection of the two orthogonal metal wires of the Xretic with the reference lines of the phantom in the anterior and the two lateral directions [Figure 3a], and a MV-CBCT image data set was acquired using the 15 MU protocol and a longitudinal field size set to its maximum of 27.4 cm. The MV-CBCT quality phantom has three sets of four beads that used to check correct geometry. The beads are distributed evenly around the circumference of the phantom with Z coordinates of 100 mm, 0 mm, and -100 mm for the head (superior), center and feet (inferior) slices respectively. In each set, the beads are located at the 3, 6, 9, and 12 o'clock positions. In the geometry check the position accuracy was checked by placing a reference point at the center of each bead in the MVCB image of the phantom and ensuring that the recorded position of that reference point in the reconstruction software was within 2 mm of the physically known bead position in the phantom. This procedure was repeated several times, on five consecutive images of the phantom in the same position, to verify stability of the reconstruction software. Figure 3b shows an axial view for the head slice (10 cm from the center) with four new reference points created and moved to the center of the four beads of this slice. Reference point properties seen at the left bottom of this Figure displays the x, y, and z coordinates for the center of the bead of interest:

Uniformity, noise and spatial resolution

In the MVCB QA phantom (starting from the end facing toward the gantry), Section 1 is a 4 cm uniform solid water cylinder that is used to check image noise and uniformity. MV-CBCT study set was taken for the QA phantom. The CBCT image set was reconstructed with 1 mm slice thickness, using the "smoothing head and neck" filter.

Figure 3: (a) The image quality phantom, (b) an axial view for the head slice of the image quality phantom showing the 4 superior beads

The transverse slices were then displayed on the Siemens Coherence workstation using the adaptive targeting window. On the central slice of section 1, five circular regions of interest (ROI) were drawn; one in the center and four in the periphery at 0, 90, 180, and 270 degrees.

Image noise and uniformity were analyzed according to the procedures specified by the manufacturer as following:

a) The center slice of section 1 is loaded on the 2D Viewer.
b) Automatically 5 ROIs are generated on the image in the axial view.
c) On the display at the right of the ROI label "Mean/SD" are displayed where the first number is the mean pixel value and the second number is the standard deviation
d) The mean and standard deviation values of each ROI are checked and the expected results will show below the measured values.
e) In this check, the center ROI is used as a reference.
f) The difference between the mean pixel value of each ROI and the mean pixel value of the center ROI is calculated and determined if that value falls within the expected range value from the following data.

Expected results for 6 mv acquisitions

- The center ROI which takes number 2 should have:
  a) a standard deviation between +26 and +42
  b) a mean value of pixels between -30 and +42
- The mean value of the peripheral ROIS should be between -80 to +80

Image reconstruction artifacts due to dead pixels or wrong gantry rotation speed can also be visually checked on each slice of Section (1). Sections (2) and (4) which contain inserts of different materials inside solid water background, as shown in Figures 4a and b. The relative electron density of each material with respect to the background is reported in Table 1. The low contrast resolution was qualitatively checked by adjusting the window and level to preset values and counting the number of inserts of each material that are visible on the image. Table 2 lists the circles that should be

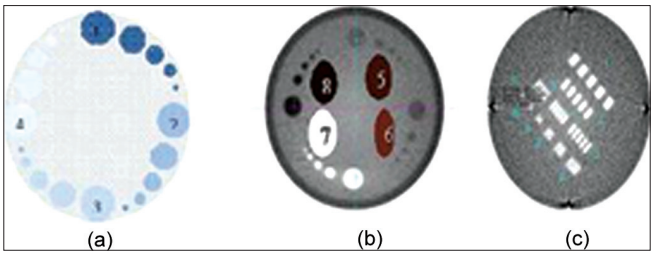

Figure 4: (a) Low contrast section of the image quality phantom. Each section has inserts of four different materials: (1) 1% SIG, (2) 3% SIG, (3) brain, and (4) liver. (b) High contrast section of the image quality phantom: (5) inner bone, (6) acrylic, (7) air, and (8) CB2 (bone-50% mineral). For each material, there are five inserts of diameters 2, 1, 0.7, 0.5, and 0.3 cm. (c) Spatial resolution section of the image quality phantom where eleven bar groups with different numbers of line pairs per millimeter are inserted in this section

Table 1: The relative electron density of each material with respect to the background

Phantom section	*Material*	*Physical density (g/cm³)*	*Electron density relative to solid water background*	*Size of the smallest visible insert (mm)*
2, 4	*SIG solid water background	1.02	1.00	–
2	1% SIG	1.03	1.01	20
2	3% SIG	1.05	1.03	20
2	Brain	1.05	1.05	10
2	Liver	1.09	1.07	5
4	Inner bone	1.14	1.09	5
4	Acrylic	1.18	1.17	5
4	CB2 (bone-50% mineral)	1.56	1.48	3
4	Air	0.0012	0.001	3

*Standard imaging grade

Table 2: Number of circles that should be visible in each of the 8 groups for 6 MV image acquisitions

Group	1	2	3	4	5	6	7	8
No. of visible circles	0	0	1	2	4	4	5	5

visible in each of the 8 groups in sections 2 and 4 for 6MV image acquisitions.

Section (3) of the MVCB QA is used to analyze spatial resolution of the image.

This section contains 11 bar groups, each group containing five bars, arranged so that each group has a different resolution, as shown in Figure 4c. This is a qualitative analysis based on the number of bars that are visible on the image where we determine how many groups (each with five line pairs) are visible. The expected results for this test using 6 MV image acquisitions are:

- The largest line group to the sixth largest group (corresponding to 0.30 lp/mm) should be visible with all five dark lines distinctly visible.
- Lines 7 to 16 are not resolvable with the current technology.

In the Figure 4c the lines in groups numbered:

- 1 through 6 should be distinctly visible.
- 7 through 11 should not be distinctly visible.

Doses from MV-CBCT and 2D imaging

The additional doses to patient from MV-CBCT and EPID images, which were acquired to check the correct patient position before treatment delivery, were evaluated and measured with two different techniques. In the first technique, a 0.6cc farmer ionization chamber was placed in the center of a 3D solid phantom of dimensions $30 \times 30 \times 30\,cm^3$. The phantom was placed in the radiation beam such that the center of the ionization chamber was located at the treatment machine isocenter. MV-CBCT study set with 5MU protocol; and pair of orthogonal EPIs (Electronic portal images) were acquired. Each image was acquired with 2MU and a dose from each set-up verification technique was measured. In The second technique; TLDs were placed in a human pediatric phantom and the images were acquired for the MV-CBCT study. The dose at the plan isocenter was measured either from MV-CBCT or EPIDs by using the TLDs. The dose distribution from the two orthogonal EPIs in a left breast case was measured. The doses were reported and compared with that calculated by the treatment planning system.

Table 3: The differences between the measured and the expected positions for the bead group in the inferior slice

Bead group in the inferior slice	*Bead number*	*Difference in (mm)*		
		X ± 2 mm	*Y ± 2 mm*	*Z ± –2 mm*
12 o'clock	9	-0.09	-0.8	–1.0
3 o'clock	10	0.25	0	–0.5
6 o'clock	11	0.050	0.4	–0.2
9 o'clock	12	0.85	0	–0.9

Results

Image quality

Slice and point position accuracy

Table 3 reports the differences between the measured and the expected positions for the four beads in the inferior slice of the QA phantom. As shown in this Table the maximum differences between the expected and measured beads positions was 1 mm and so for all other beads in the superior and middle slices, the differences were within the recommended ±2 mm position precision in all three directions. Variations of up to 0.3 mm in the reconstructed position of the beads over the five consecutive scans for the same phantom position were observed. However, uncertainties in the subjective, user-dependent placement of the reference point at the center of the bead could contribute to these variations. The registration of the MV-CBCT image to the planning CT yielded offsets that were all within ±2 mm of their nominal value.

Noise and uniformity

Figure 5 shows the middle slice of section 1 (uniform solid water) of the MV-CB QA phantom and the five ROIs along with their mean pixel values and standard deviation. The standard deviation in the center ROI was 28.8 where the acceptable range according to Siemens protocol is a standard deviation between +26 and +42. That indicates current results are within acceptable limits. The measured mean value of pixels for the central ROI was 19.5 which indicates that this value is within acceptable range of –30 to +42. The maximum difference in the measured mean pixel values between the central ROI and the peripheral ROI was 49.5 and this value is also in the acceptable range of ±80.

Contrast and spatial resolution

The commissioning results with the 15 MU protocol, showed that the low-contrast resolution did not allow visualizing a 2 cm object with 1% and 3% contrast. However, for high contrast, a 5 mm object with 18% contrast, and a 3 mm object with 50% contrast, were visualized.

Figures 6a and b shows slices of Section 2 and 4 of the image quality phantom. In Section 2, as in Figure 6a for low contrast medium; four, three, zero, and zero inserts for the liver, brain, 3% Standard Imaging Grade (SIG) solid water, and 1% SIG, respectively were counted. In Section 4, five, five, four, and four, inserts for air, CB2 (bone -50% mineral), inner bone, and acrylic, respectively was counted. According to the manufacture specifications the spatial resolution for Siemens Mvison MV-CBCT is accepted where bars in group 6 which corresponding to 0.3 1p/mm are clearly visible [Figure 6c].

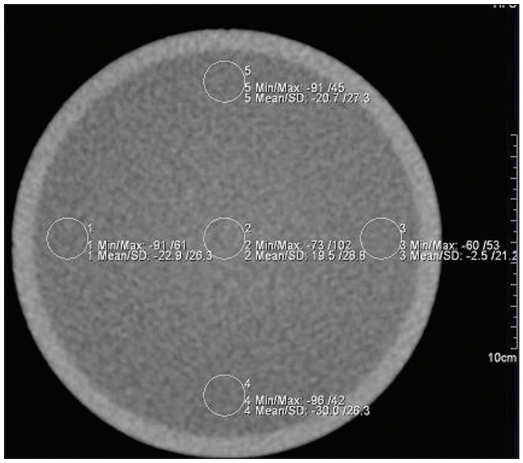

Figure 5: The center slice of region I in the image QA phantom showing 5 ROIs on the image in the axial view; on the display at the right of the ROI label "Mean/SD are displayed

The results showed that for MV-CBCT with 5MU protocol the dose to plan isocenter was 5.1 cGy for head and neck where the isocenter was located at the middle plane of the head. The dose for mid-plane of the chest was 4.8 cGy while for the pelvis sites the dose was 4cGy. These doses were larger for superficial organs which may be critical organs as the lens in the head and neck sites. The dose to eye during set-up verification for the head and neck case using MV-CBCT with 5MU protocol was 6.7 cGy and the dose to lung was 6 cGy during set-up verification of the Mediastinum.

For EPID the dose at plan isocenter from the two images was 3.8, 3.5 and 3 cGy for head and neck, chest and pelvis respectively, where the EPID set-up verification was acquired with 2 MU for each field. As in MV-CBCT the dose for superficial organs which located in the entrance of the two orthogonal verification beams was larger than the dose at isocenter. Figure 7 shows the calculated dose distribution from the two orthogonal fields of EPID set-up verification for a left breast case. This study case showed that in case of superficial and one sided tumors these two orthogonal fields give homogeneous dose to the target so these two fields can be included in the original treatment plan of such cases. The calculated dose at plan isocenter was 3.9 cGy which matched with measurement results using TLD in the human phantom where the average measured dose were 3.8 cGy.

Discussion

Intensity modulated radiotherapy with steep dose gradients has allowed the delivery of higher doses to

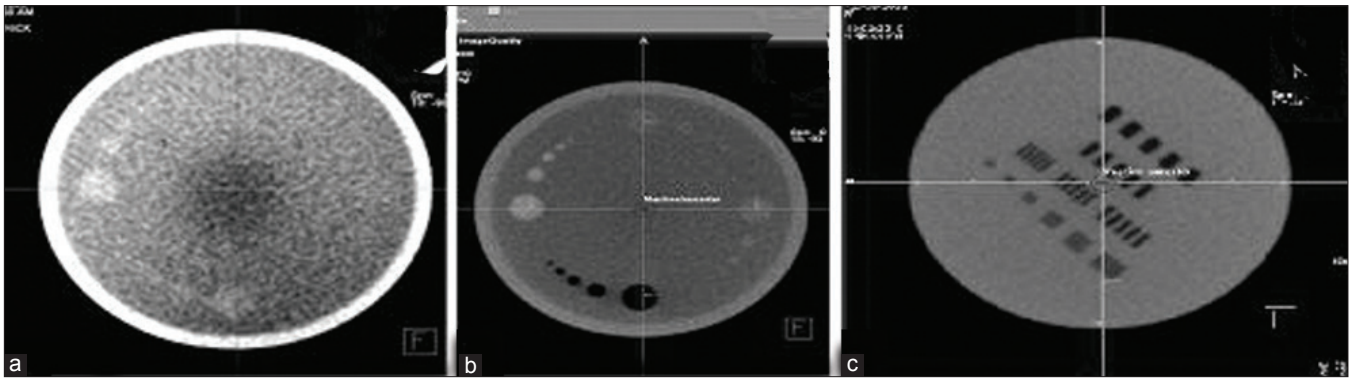

Figure 6 (a, b): Slices of Section 2 and 4 of the image quality phantom, and (c) spatial resolution for Siemens Mvison MV-CBCT

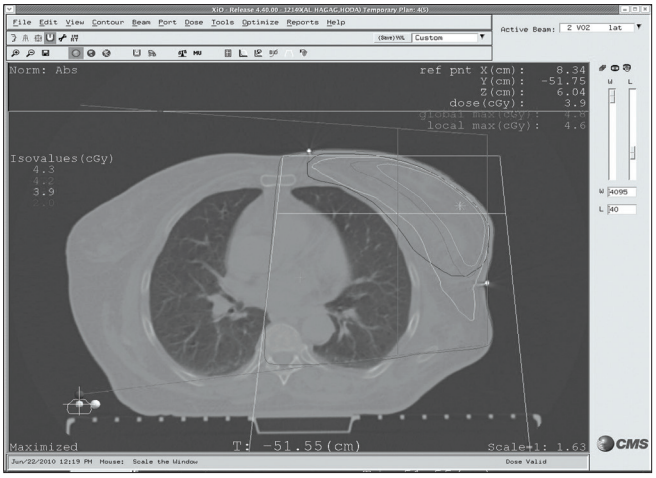

Figure 7: Calculated dose distribution from the two orthogonal fields of EPID set-up verification for a left breast case

the tumor volume while maintaining the sparing of the surrounding normal tissue. In this situation, verification of patient set-up just prior to radiation delivery is a crucial step.[11] The mega voltage EPID has the ability to adjust display contrast to assess the target position and to help the radiotherapist to adjust the patient promptly.

Formal commissioning of the volumetric X-ray image-guided radiotherapy system and the associated hardware and image-guided processes necessitates the development and use of an appropriate quality assurance program, before the system is used clinically. Regular QA provides confidence on the system ability to manage geometric variations in the patient set-up.[12]

In some cases of pelvic tumors it is easy to use the high contrast bony landmarks for set-up verification but in case of cancer prostate, where the difference of electron density between the prostate and the surrounding tissue is 1-4% and is a mobile organ, it is difficult to use this system to distinguish between the prostate and the surrounding normal tissue. Current resolution is inadequate to detect low contrast structures such as prostate. The amount of scatter also affects the contrast of the image.[13] It contributes significantly to the cupping effect, which is a progressive decrease in the pixel density towards the center of the object, due to a higher scatter to primary photon ratio. This effect is bigger for large anatomical sites, such as the pelvis, and increases for large patients. This poses a problem, especially when the target located at the center of the image. For example, in a patient treated for prostate cancer, the contrast around the prostate volume can be so low that the target is nearly invisible. This effect is corrected by the smoothing filter.[11]

The difference in spatial resolution between the MV-CBCT and the planning CT data sets is small compared to that between noise and contrast. This allows the use of fixed small objects when registering two data sets, such as surgical clips, if they are located around a tumor, or implanted fiducially.[9] It should be noted that the resolution is limited by the pixel size on the image. Under standard conditions, the CBCT images use 256×256 pixels. At the fixed SID of 145 cm, the field size is fixed at 27.4 cm^2. Since the possible reduction of the longitudinal field size by the collimator does not affect the pixel size, it is fixed at 1.07 mm per pixel. The use of 512×512 reconstructed images decreases the pixel size by a factor 2 which can be increased to factor 4 in case of $1,024 \times 1,024$ pixels, thus increasing the spatial resolution; however using them for localization purposes, increases the reconstruction time of the CB image, it also decreases the signal strength drastically and needs more MU.

All image quality characterization was performed with the 15 MU protocol. However, for patient localization, different lower MU protocols were used for different anatomical sites, resulting in lower dose delivered to the patient. It was shown that the image quality from the 3 MU protocol (2.5 cGy at isocenter) was sufficient for bony registration, but that a higher dose (6-10 cGy, typically corresponding to 8 to 15 MU protocols depending on patient size) was necessary to distinguish soft tissue contrast. The contrast to noise ratio decreases by about 20% when the field size increases from 5 to 27.4 cm^2.[14]

Conclusion

A MV-CBCT system for patient localization was clinically commissioned at our institution. The initial geometry calibration to reconstruct volumetric images from 200 two-dimensional projections was performed, and the position accuracy of the system has been shown to be within 2 mm in the AP, LR, and SI directions. The image quality, parameters such as image noise, uniformity, low contrast, and high contrast resolutions were verified using an image quality phantom. The results showed that current resolution is inadequate to detect low contrast structures such as prostate. Daily and monthly quality assurance programs are important to insure that the EPID and the CBCT are working in a proper manner. According to the set-up verification protocol the measurements, the additional doses from MV-CBCT with 5 MU and EPID with 4 MU are not surprising, but in case of the higher MU protocol, dose from the set-up verification fields or the CBCT study should be added to the patient treatment plan.

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