High-dose MVCT image guidance for stereotactic body radiation therapy

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Purpose: Stereotactic body radiation therapy (SBRT) is a potent treatment for early stage primary and limited metastatic disease. Accurate tumor localization is essential to administer SBRT safely and effectively. Tomotherapy combines helical IMRT with onboard megavoltage CT (MVCT) imaging and is well suited for SBRT; however, MVCT results in reduced soft tissue contrast and increased image noise compared with kilovoltage CT. The goal of this work was to investigate the use of increased imaging doses on a clinical tomotherapy machine to improve image quality for SBRT image guidance.

Methods: Two nonstandard, high-dose imaging modes were created on a tomotherapy machine by increasing the linear accelerator (LINAC) pulse rate from the nominal setting of 80 Hz, to 160 Hz and 300 Hz, respectively. Weighted CT dose indexes (wCTDIs) were measured for the standard, medium, and high-dose modes in a 30 cm solid water phantom using a calibrated A1SL ion chamber. Image quality was assessed from scans of a customized image quality phantom. Metrics evaluated include: contrast-to-noise ratios (CNRs), high-contrast spatial resolution, image uniformity, and percent image noise. In addition, two patients receiving SBRT were localized using high-dose MVCT scans. Raw detector data collected after each scan were used to reconstruct standard-dose images for comparison.

Results: MVCT scans acquired using a pitch of 1.0 resulted in wCTDI values of 2.2, 4.7, and 8.5 cGy for the standard, medium, and high-dose modes respectively. CNR values for both low and high-contrast materials were found to increase with the square root of dose. Axial high-contrast spatial resolution was comparable for all imaging modes at 0.5 lp/mm. Image uniformity was improved and percent noise decreased as the imaging dose increased. Similar improvements in image quality were observed in patient images, with decreases in image noise being the most notable.

Conclusions: High-dose imaging modes are made possible on a clinical tomotherapy machine by increasing the LINAC pulse rate. Increasing the imaging dose results in increased CNRs; making it easier to distinguish the boundaries of low contrast objects. The imaging dose levels observed in this work are considered acceptable at our institution for SBRT treatments delivered in 3–5 fractions. © 2012 American Association of Physicists in Medicine. [http://dx.doi.org/10.1118/1.4736416]

Key words: tomotherapy, MVCT, high-dose imaging, SBRT

I. INTRODUCTION

Stereotactic body radiation therapy (SBRT) is an efficacious treatment for both early stage primary cancer as well as limited metastatic disease. SBRT delivers ablative doses of radiation, typically in 3–5 treatment fractions, and has been shown to result in excellent local control rates,1,2 similar to those seen with surgery for selected patients.3 From a dosimetric standpoint, SBRT necessitates the use of highly conformal planning techniques to produce steep dose gradients and thereby limit the dose received by critical structures adjacent to the target. However, given the steep dose fall-off and reduced number of fractions, the risk of local failure and/or normal tissue injury due to a geometric miss is greater with SBRT since the averaging of random positioning errors observed with more conventional fractionation is reduced.4 In order to ensure the most accurate treatment delivery possible, daily image guidance is required.4,5

Today a number of image guided radiation therapy (IGRT) systems are commercially available. The TomoTherapy Hi-ArtTM (Accuray, Inc., Madison, WI) comprises one such system.6 Originally designed to deliver helical tomotherapy, the Hi-Art device uses a 6 MV linear accelerator (LINAC) mounted on a computed tomography (CT) ring gantry to deliver intensity modulated fan-beams of radiation, while the patient is translated through the machine on a moving treatment couch.7,8 Image guidance is provided via a single slice CT detector mounted opposite the radiation source, which is used for helical fan-beam megavoltage CT (MVCT) imaging.9
Helical tomotherapy is well suited for SBRT. The use of up to 51 beam angles for treatment planning allows for highly conformal dose distributions with improved sparing of normal tissues compared to more conventional 3D approaches. In addition, the binary nature of the multileaf collimator makes blocking of critical structures both straightforward and effective. One of the challenges with TomoTherapy is the lack of low contrast resolution observed with the MVCT imaging system relative to kilovoltage CT and kilovoltage cone beam CT, particularly when localizing treatment sites in the abdomen. This is of special concern for SBRT treatments, where doses of up to 20 Gy are delivered in a single fraction. This issue is not unique to TomoTherapy, rather it is a common problem among all MV imaging systems.

The ability to resolve contrast differences in an image is fundamentally limited by the number of photons detected during acquisition. If the number of photons is too small, noise will dominate the image—making the borders between different contrast regions indistinguishable. MV imaging systems are at a disadvantage compared to kV systems because the fraction of energy deposited per photon is similar. For doses between systems to remain comparable, far fewer photons can be used to acquire the MV images. In addition, percent differences in physical contrast between two objects are easier to detect when the objects are more attenuating. Though the dominant mode of interaction for photons used in both kV and MV imaging systems is Compton scattering (Compton events make up almost 90% of interactions at 60 keV in water), the attenuation coefficient in tissue is greater at kV energies. This results in greater soft tissue contrast with kV imaging; even when using the same number of photons. 

For these reasons, efforts to improve MV image quality tend to focus on decreasing image noise by increasing detector signal-to-noise ratio (SNR). A straightforward way of doing this is to increase the imaging dose. Gayou et al. have shown considerable improvements in MV cone-beam CT (CBCT) image quality as the imaging dose is increased from 2 cGy to 12 cGy, while Ruchala et al. demonstrated that 1.1 cm objects with contrast levels of 2% could be resolved with MVCT using imaging doses of 8 cGy.

For MV CBCT systems like the one described by Pouliot et al., increasing the imaging dose is straightforward and requires only that the number of monitor units used during the image acquisition be increased. The TomoTherapy system on the other hand uses a fixed dose rate for the imaging beam, making it difficult to modify the imaging dose level. In this work, we have created high-dose imaging modes on a clinical tomotherapy machine specifically for SBRT applications. Details of the implementation as well as results of preliminary testing are presented.

II. METHODS AND MATERIALS

II.A. The TomoTherapy imaging system

The TomoTherapy imaging system is similar to a helical CT scanner in that during image acquisition, the couch translation and gantry rotation occur simultaneously. The CT detector in this case is an arc-shaped, xenon detector comparable to those used in older style general electric CT scanners. The Hi-Art style detector has 738 channels, 640 of which are read out by the current data acquisition system (DAS). Each channel consists of two ionization cavities separated by metallic septa and filled with xenon gas pressureized to approximately 25 atm. A potential difference of 200 V is applied to every other cavity wall and the charge collected in adjacent cavities is summed to give the signal from each channel. The dimensions of a given channel measured at the detector are 1.2 mm (IEC transverse) by 2.5 cm (IEC longitudinal). The transverse field-of-view (FOV) of the machine is fixed at just over 39 cm (measured at iso-center, 85 cm from the source) and the central 540 channels of the detector are sufficient to cover the full lateral extent of the fan-beam. The detector has a radius of curvature of 99 cm but is mounted 132 cm from the source. This places the arc-detector out of focus with respect to the source and has been shown to increase detector efficiency for off-axis detector elements.

The x-ray source used for imaging is the same as that used for treatment; however, the energy of the beam is reduced to approximately 3.5 MV in order to improve image contrast and detector SNR. In addition, the pulse rate of the beam is decreased from 300 Hz for treatment down to 80 Hz for imaging. Unlike conventional CT scanners where these parameters can be adjusted, the energy and output of the beam are fixed at the time of commissioning on the Hi-Art device.

The TomoTherapy imaging system operates as a helical CT scanner. As such, the helical pitch is an important parameter for characterizing scanning protocols. The pitch is defined as the couch translation per gantry rotation normalized to the field width, and fundamentally influences the spatial resolution in the longitudinal direction as well as the absorbed dose. The default configuration at the time of testing uses a 4 mm collimator setting with pitches of 1.0, 2.0, and 3.0. These pitch settings correspond to the fine, normal and coarse scan selections offered in the TomoTherapy operator station software and are used to reconstruct CT slices at intervals of 2, 4, and 6 mm, respectively. Note that the 4 mm collimator setting is the nominal value. Because the collimating jaws are located close to the source, the imaging beam projects a full-width at half-maximum field width of 6-7 mm at iso-center.

II.B. High-dose imaging modes

At the University of Colorado, two high-dose imaging modes were created on our TomoTherapy machine for testing purposes. This was done by cloning the standard imaging beam parameters and increasing the LINAC pulse rate. For one of the modes, the pulse rate was increased to 160 Hz, while for the other it was increased to the treatment pulse rate of 300 Hz. Switching between modes requires the machine to be restarted and an air scan performed. The air scan is a 60 s rotational procedure delivered using the imaging beam with the treatment couch fully retracted. MVCT detector data collected during this procedure are used to correct subsequent scans for daily variations in detector response and must be
acquired at the same sampling rate as data acquired during normal pretreatment scans.

II.C. Imaging dose measurements

Ionization measurements were obtained in a 30 cm diameter solid water phantom using a calibrated A1SL ion-chamber (Standard Imaging, Middleton, WI) to characterize the doses delivered using the different imaging modes. Central plane average doses were taken as the weighted CT dose index\textsuperscript{23} given by

\[
\text{wCTDI} = \frac{1}{3} D_0 + \frac{2}{3} D_p, \tag{1}
\]

where \(D_0\) is the dose measured at the center of the phantom and \(D_p\) corresponds to the dose measured 1 cm from the phantom periphery. For all dose measurements, the scanned length of the phantom on either side of the ion chamber collecting volume was large enough so that scanning additional slices did not affect the measured values within our ability to measure. For each imaging mode, \(D_0\) and \(D_p\) were measured using the three default pitch settings corresponding to the fine, normal, and coarse slice reconstruction intervals.

II.D. Image quality characterization

Image quality for the three dose modes was assessed from MVCT images of a commercial image quality phantom (Siemens Medical Solutions, Concord, CA) surrounded by a 35 cm diameter acrylic annulus. The latter was added to mimic the attenuation properties of a patient undergoing an abdominal CT scan. The image quality phantom is a 20 cm diameter cylindrical phantom with a standard, medium, and high-dose imaging modes. The latter was added to mimic the attenuation properties of a patient undergoing an abdominal CT scan. The image quality phantom is a 20 cm diameter cylindrical phantom with a standard, medium, and high-dose imaging modes. The latter was added to mimic the attenuation properties of a patient undergoing an abdominal CT scan. The image quality phantom is a 20 cm diameter cylindrical phantom with a standard, medium, and high-dose imaging modes. The latter was added to mimic the attenuation properties of a patient undergoing an abdominal CT scan. The image quality phantom is a 20 cm diameter cylindrical phantom with a standard, medium, and high-dose imaging modes. The latter was added to mimic the attenuation properties of a patient undergoing an abdominal CT scan. The image quality phantom is a 20 cm diameter cylindrical phantom with a standard, medium, and high-dose imaging modes. The latter was added to mimic the attenuation properties of a patient undergoing an abdominal CT scan. The image quality phantom is a 20 cm diameter cylindrical phantom with a standard, medium, and high-dose imaging modes. The latter was added to mimic the attenuation properties of a patient undergoing an abdominal CT scan. The image quality phantom is a 20 cm diameter cylindrical phantom with a standard, medium, and high-dose imaging modes. The latter was added to mimic the attenuation properties of a patient undergoing an abdominal CT scan. The image quality phantom is a 20 cm diameter cylindrical phantom with a standard, medium, and high-dose imaging modes.
scan. When the high-dose images are reconstructed, the additional data collected for each projection are combined; resulting in improved photon statistics. For this work, the exported detector data were sampled offline (i.e., detector data for only one in four pulses were used for each projection) to create a standard-dose projection set for each scan. These standard-dose projection sets were used to reconstruct images of the same patient anatomy under conditions similar to those used with the standard-dose imaging mode for comparison purposes.

III. RESULTS

III.A. Dose measurements

Results of dose measurements and computed wCTDI values for scans performed using the standard, medium, and high-dose imaging modes with the fine scan setting are listed in Table I. In addition, wCTDI values for the three imaging modes using the fine, normal, and coarse scan settings are displayed graphically in Fig. 2. Measurements indicate that the imaging dose increases linearly with LINAC pulse rate and that the highest pulse rate tested results in a wCTDI of approximately 8.5 cGy to the 30 cm diameter solid water phantom. Additionally, Fig. 2 shows an increase in the wCTDI with decreasing pitch. This is caused by a greater overlap of the radiation beam with itself on adjacent gantry rotations with decreasing pitch and is consistent with the findings of Meeks et al.20

Because the imaging doses being measured are quite small, the charge leakage in the ion-chamber/electrometer setup is non-negligible. To account for this, we made use of the linear relationship observed between the uncorrected dose measurements and the LINAC pulse rate. We estimated the offset due to charge leakage from the intersection of a best fit line to this data with the dose axis. This offset was then added to the raw dose measurements to obtain the reported values.

To estimate the random uncertainty associated with our measurements, a series of ten 271 s scans were performed using the standard-dose mode and fine pitch setting. Central axis ionization measurements for these scans were acquired and converted to dose. The standard deviation as a percentage of the mean dose for these measurements was found to be 5.2%. This percentage was used to estimate the random uncertainty in each dose value. wCTDI uncertainty estimates were computed as the sum in quadrature of the error associated with the central axis and peripheral dose readings, respectively. This uncertainty is listed in Table I and illustrated by the error bars in Fig. 2.

III.B. Image quality

MVCT scans of the three image quality modules acquired using the various imaging modes are shown in Fig. 3. The most notable improvement in image quality was a progressive decrease in image noise with increasing dose. As illustrated in the top row of Fig. 3, this reduction in noise makes it easier to identify borders of the smaller, low contrast inserts.

Average CNRs computed for the various material inserts are plotted as a function of wCTDI in Fig. 4. Error bars (when visible) indicate a single standard deviation. The dashed lines represent a square root power law fit to the data for each material. CNRs are seen to increase with the square root of dose for each of the material inserts—which is in good agreement with theoretical predictions.27

Figure 5 shows average MTF values calculated for the different imaging modes as a function of spatial frequency. Again error bars indicate one standard deviation. High-contrast spatial resolution was similar for all imaging modes tested with a limiting resolution of 0.5 lp/mm; corresponding to a MTF cutoff value of 0.1. This result was in line with our expectations since the MTF measures the efficiency of the system in transferring the object to image contrast and as such, should not be influenced by the delivered dose.

Finally, average image uniformity and percent image noise values are listed in Table II. As expected, image noise was seen to decrease with the higher dose modes; however image uniformity was also observed to improve. Since image uniformity (or the lack thereof) is usually characterized by spectral changes due to beam hardening, this result was unexpected. From Fig. 3, it is apparent that the improvement in image uniformity stems from a reduction in the central artifact seen in all images. This arises from a difference in the signal collected by the central detectors with the different pulse rates, which is subsequently transformed by a spectral calibration prior to

<table>
<thead>
<tr>
<th>Imaging modes</th>
<th>$D_0$ [cGy]</th>
<th>$D_p$ [cGy]</th>
<th>wCTDI [cGy]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard—80 Hz</td>
<td>1.6 ± 0.1</td>
<td>2.5 ± 0.1</td>
<td>2.2 ± 0.1</td>
</tr>
<tr>
<td>Medium—160 Hz</td>
<td>3.4 ± 0.2</td>
<td>5.4 ± 0.3</td>
<td>4.7 ± 0.2</td>
</tr>
<tr>
<td>High—300 Hz</td>
<td>6.2 ± 0.3</td>
<td>9.7 ± 0.5</td>
<td>8.5 ± 0.4</td>
</tr>
</tbody>
</table>

Fig. 2. Weighted CT dose index values computed from phantom dose measurements made for the standard, medium, and high-dose imaging modes vs helical pitch. Pitch values correspond to the fine, normal, and coarse scan settings in the TomoTherapy operator station software.
reconstruction. Because the dose per pulse with the different imaging modes is the same, it is not clear why this change in the detected signal occurs. Regardless, the observed differences in uniformity could easily be corrected by applying a different spectral calibration for each imaging mode.

III.C. Clinical images

Figure 6 shows representative images acquired for one of the two liver SBRT patients imaged with the high-dose imaging mode using the normal scan setting. Also shown is the same patient anatomy reconstructed from a standard-dose projection set. As with the phantom scans of Fig. 3, the most notable improvement in image quality is an overall reduction in image noise observed with the high-dose image compared to the standard image. This noise reduction makes it easier to distinguish the edges of soft tissue boundaries.

IV. DISCUSSION

Previous investigations looking at imaging doses delivered with the TomoTherapy Hi-Art system have focused mainly on the effects of helical pitch, with numerous authors

<table>
<thead>
<tr>
<th>Imaging modes</th>
<th>WCTDI [cGy]</th>
<th>U [HU]</th>
<th>σ% [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard—80 Hz</td>
<td>2.2 ± 0.1</td>
<td>28.8</td>
<td>5.2</td>
</tr>
<tr>
<td>Medium—160 Hz</td>
<td>4.7 ± 0.2</td>
<td>14.7</td>
<td>3.5</td>
</tr>
<tr>
<td>High—300 Hz</td>
<td>8.5 ± 0.4</td>
<td>6.0</td>
<td>2.6</td>
</tr>
</tbody>
</table>
noting an increase in imaging dose with decreasing pitch.\textsuperscript{13,20} Given the single slice nature of the TomoTherapy MVCT detector, this effect does not impact image quality except to increase slightly the image resolution in the longitudinal direction. Likewise, in the latest upgrade to the TomoTherapy system (version 4), the imaging beam has been modified to use a nominal 1 mm collimator setting. While this also acts to improve the longitudinal image resolution, it does little to improve the contrast-to-noise characteristics in the axial plane. In multiple slice scanners, the use of multislice CT detectors allows for signal averaging, which can be used to improve noise statistics for a given image at the expense of decreased longitudinal resolution. A similar procedure could potentially be performed with the TomoTherapy system by averaging projection data acquired with a low pitch scan to reconstruct larger slice thickness images; however, this would come at the cost of longer scan times—which for larger volumes can already exceed several minutes.

In this work, we investigate increasing the imaging dose rate on a clinical TomoTherapy machine to reduce detector noise and thereby increase the SNR. This was done by increasing the LINAC pulse rate during image acquisition. The current Hi-Art system uses a decreased energy beam for imaging to improve soft tissue contrast and increase the detector SNR. This energy reduction is accomplished primarily via beam loading, whereby the injector current is increased so that the RF power from the magnetron is distributed among a larger number of electrons. The increased injector current with the imaging beam potentially limits the LINAC pulse rate because of additional stress placed on various beam line components including the injector, electron gun circuit board, and target. While extensive testing would be required to determine the long-term effects of using higher pulse rates with the standard imaging beam settings, initial testing performed at Accuray, Inc. demonstrated that the pulse rates utilized in this work were operationally safe for limited use.

An alternative method of increasing the imaging dose rate would be to modify the RF system by adjusting the pulser forming network (PFN) voltage. Because the TomoTherapy beam line does not employ a bending magnet, the machine is tuned away from the peak output configuration in order to improve machine stability. Instead, the RF configuration is set to lie within a range where the output and energy of the beam can be adjusted independently by modifying either the PFN voltage or injector current respectively. Because this range is limited, attempting to significantly increase the output of the machine by adjusting the PFN voltage would have the undesired effect of also changing the beam energy. This in turn would require adjustment of other RF parameters (such as the injector current, magnetron voltage, and/or magnetron current) and would necessitate the use of an oscilloscope to verify that a given set of tuning parameters is stable. Another alternative would be to keep the dose rate of the linac fixed but reduce the gantry speed (i.e., increase the gantry period). This however would also result in longer scan times.

Using the high-dose imaging mode, CNR values for the various phantom materials tested were observed to increase in proportion to the square root of the imaging dose. For the lowest contrast insert (relative electron density 1.09), this yielded a mean CNR of 3.43 ± 0.16 compared to 1.90 ± 0.15 with the standard-dose mode. This is somewhat lower than values typically observed with kV cone-beam CT systems. Mail et al. measured a CNR of 4.22 ± 0.41 with the Elekta Synergy XVI kV CBCT system (Elekta Oncology Systems, Crawley, UK) for a polystyrene insert (−100 HU) relative to low-density polyethylene (−35 HU) in an irregularly shaped body phantom using a bowtie filter and imaging dose of approximately 2 cGy.\textsuperscript{28} Similar CNR values were also reported by Stützel et al. using the Siemens Artiste kV CBCT prototype (Siemens OCS, Erlangen, Germany) with wCTDI values ranging from 1.5 to 3.0 cGy.\textsuperscript{13}

While use of the high-dose imaging mode undoubtedly offers some improvement in image quality over the standard configuration, it is reasonable to question whether the improvements seen are necessary. Using Rose’s model,\textsuperscript{29} Reitz et al. have suggested that a CNR of 2–5 would allow a human to detect an object with an error probability of 10\% down to approximately 0\%, respectively.\textsuperscript{30} Phantom measurements performed in this work indicate that with the standard-dose imaging mode, the 10\% error probability is exceeded when object contrast falls below 9\%. For the same object contrast using the high-dose mode, the CNR is increased by almost a factor of 2, resulting in a calculated error probability of less than 1\%, assuming a Gaussian error distribution.

Beyond this type of analysis, it is difficult to quantitatively link improvements in image quality to improved image guidance since the accuracy of image guidance is ultimately observer dependent. In general, it is expected that better...
image quality will lead to improved image guidance, depending on which anatomical features are used to guide the image registration. The improvements in image quality seen with the high-dose imaging mode are most likely of benefit when imaging low contrast regions in the pelvis and abdomen (e.g., prostate and liver) or when imaging larger patients where increased attenuation results in a reduced SNR.

Increasing imaging dose to improve image quality must be weighed against the risks associated with additional exposure to the patient. Recently, there has been heightened concern over the increased use of medical imaging and the potential risks this poses to the patient population for secondary malignancies.31, 32 These concerns are typically raised in regard to diagnostic imaging; however, they can also be viewed in the context of IGRT.23 Assessing the risks posed by imaging is difficult in both cases because the risks that come with the additional radiation exposure are offset by the potential benefits of early detection in the case of diagnostic x-ray imaging and more precise targeting in the case of IGRT.

In principle, MVCT imaging dose would be straightforward to calculate and can thus be easily managed by incorporating the dose into the treatment planning. This method minimizes risk to the patient and has previously been demonstrated with megavoltage cone beam CT.33–35

In SBRT, treatments are given in only a few fractions so that the total imaging dose with the high-dose mode is still likely to be less than the imaging dose delivered to patients receiving conventionally fractionated treatments localized with the standard-dose mode. In this work, doses on the order of 4 cGy were measured with the high-dose mode using the normal scan setting. Even at this relatively modest dose level, we observed improvements in image quality compared to the standard imaging mode in two liver SBRT patients. Furthermore, the imaging dose for these two patients was still a factor of 2 lower than doses reported for MV cone-beam systems being used for daily imaging in conventionally fractionated treatments.33

V. CONCLUSION

High-dose imaging modes are made possible on a clinical TomoTherapy machine by increasing the LINAC pulse rate. Dose measurements performed in a 30 cm diameter solid water phantom indicated that increasing the imaging pulse rate to 300 Hz results in average doses of 8.5 cGy when scanning with the fine (lowest pitch) scan setting. High-dose scans of a customized image quality phantom showed that increasing the imaging dose results in reduced image noise and increased CNRs; making it easier to distinguish the boundaries of low contrast objects. These results were also observed in localization images acquired for two liver patients undergoing SBRT treatments.

The benefits of improved image quality with increased imaging dose must be carefully weighed against the risks associated with added exposure to the patient; especially in conventionally fractionated treatment regimens. For SBRT treatments given in only a few fractions, the increase in imaging dose using the high-dose mode is relatively small (<10 cGy with the dose levels observed in this work). This increase compared to the risks associated with missing the target in hypo-fractionated therapy is considered acceptable at our institution. Furthermore, even with the high-dose mode, the total imaging dose is still likely to be less than the imaging dose given in a conventionally fractionated course of treatment using the standard-dose mode.

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